

KASE
BJÄRNGÅRD
ATTIX

THE DOSIMETRY
OF IONIZING RADIATION

I

THE DOSIMETRY OF IONIZING RADIATION

VOLUME I

Edited by

Kenneth R. Kase

Bengt E. Bjärngård

Frank H. Attix

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Contributors

Gudrun Alm Carlsson
André Bridier
Johan J. Broerse
Andrée Dutreix
Elis Holm
John R. Johnson
Albrecht M. Kellerer
Kurt Lidén
John T. Lyman
Johannes Zoetelief

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Volume I

Edited by KENNETH R. KASE

*Joint Center for Radiation Therapy
Harvard Medical School
Boston, Massachusetts*

BENGT E. BJÄRNGÅRD

*Joint Center for Radiation Therapy
Harvard Medical School
Boston, Massachusetts*

FRANK H. ATTIX

*Department of Medical Physics
University of Wisconsin Medical School
Madison, Wisconsin*



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Contents

LIST OF CONTRIBUTORS
PREFACE

vii
ix

1. Theoretical Basis for Dosimetry

Gudrun Alm Carlsson

I. Introduction	2
II. Energy Imparted: The Fundamental Quantity of Radiation Dosimetry	4
III. Absorbed Dose	13
IV. Detector Response	15
V. Radiometric Quantities: Mean Energy Imparted and Absorbed Dose in Terms of Vectorial Energy Fluence	19
VI. Absorbed Dose in Terms of Scalar Radiometric Quantities and Interaction Coefficients	36
VII. Cavity Theory	67
References	71

2. Fundamentals of Microdosimetry

Albrecht M. Kellerer

I. Introduction	78
II. General Concepts and Basic Quantities	79
III. The Compound Poisson Process in Microdosimetry	97
IV. Determination and Utilization of the Microdosimetric Parameters	113
V. The Straggling Problem and the Single-Event Spectrum	125
VI. Geometric Aspects of the Inchoate Distribution	137
Appendix: Algorithm for the Compound Poisson Process	154
References	158

3. Dosimetry for External Beams of Photon and Electron Radiation

Andrée Dutreix and André Bridier

I. The Detector	164
II. Dosimetry for Photon Beams	181

III. Dosimetry for Electron Beams	202
IV. Uncertainties	216
References	223
4. Dosimetry of External Beams of Nuclear Particles	
<i>Johan J. Broerse, John T. Lyman, and Johannes Zoetelief</i>	
I. Introduction	230
II. Beam Characteristics	232
III. Dosimetry Methods	238
IV. Radiation Quality	250
V. Determination of Absorbed Dose at a Reference Point	252
VI. Treatment Planning for External Beam Therapy	264
VII. Response of Biological Dosimeters	268
VIII. <i>In Vivo</i> Dosimetry	275
IX. Conclusions	278
References	281
5. Measurement and Dosimetry of Radioactivity in the Environment	
<i>Kurt Lidén and Elis Holm</i>	
I. Introduction	292
II. Sources of Environmental Radioactivity	294
III. General Aspects of Measurement Methods	305
IV. Analytical Procedures and Measurement of Radioactivity	310
V. Nonconventional Methods for Assessment of Radionuclides	351
VI. Discussion of Errors and the Need for Uniformity in Gathering and Reporting Data	353
VII. Estimation of Radiation Levels Received by Human Beings	357
References	364
6. Internal Dosimetry for Radiation Protection	
<i>John R. Johnson</i>	
I. Introduction	369
II. Basic Concepts	370
III. Specific Effective Energy	376
IV. Models of Radionuclide Transport in Humans	380
V. Individual Monitoring and Dose Assessment	388
VI. Summary	406
References	406
INDEX	411

List of Contributors

Numbers in parentheses indicate the pages on which the authors' contributions begin.

- GUDRUN ALM CARLSSON (1), Department of Radiation Physics, The Medical School, University of Linköping, S-581 85 Linköping, Sweden
 ANDRE BRIDIER (163), Service de Radiophysique, Institut Gustave-Roussy, 94800 Villejuif, France
 JOHAN J. BROERSE (229), Radiobiological Institute TNO, 2280 HV Rijswijk, The Netherlands
 ANDREE DUTREIX (163), Service de Radiophysique, Institut Gustave-Roussy, 94800 Villejuif, France
 ELIS HOLM (291), Department of Radiation Physics, Lund University, Lund, Sweden
 JOHN R. JOHNSON (369), Biomedical Research Branch, Atomic Energy of Canada Limited Research Company, Chalk River Nuclear Laboratories, Chalk River, Ontario KOJ 1JO, Canada
 ALBRECHT M. KELLERER (77), Institut für Medizinische Strahlenkunde, Julius-Maximilians-Universität Würzburg, D-8700 Würzburg, Federal Republic of Germany
 KURT LIDEN (291), Department of Radiation Physics, Lasarettet, Lund University, S-22185 Lund, Sweden
 JOHN T. LYMAN (229), Lawrence Berkeley Laboratory, University of California, Berkeley, California 94720
 JOHANNES ZOETELIEF (229), Radiobiological Institute TNO, 2280 HV Rijswijk, The Netherlands

Preface

Since 1956 Academic Press has published some of the most authoritative, respected, and widely used references in the field of radiation dosimetry. These have included "Radiation Dosimetry," First Edition (1956), edited by Gerald J. Hine and Gordon L. Brownell; Second Edition (1966-1969, 3 volumes), edited by Frank H. Attix, William C. Roesch, and Eugene Tochilin; and "Topics in Radiation Dosimetry," Supplement 1 (1972), edited by Frank H. Attix. It is the goal of the present work to fill the need for newer reference material of comparable quality.

There were two major questions that faced the editors and publisher when this project was being considered: whether to undertake a full-scale third edition, and if not, what to call the new books.

It was decided that this would not be considered a third edition, because much of the material in the second edition of "Radiation Dosimetry" is still timely and useful, and because of the extraordinary editorial difficulties encountered in producing a very large multivolume, multiauthor work that is fully comprehensive and thoroughly cross-referenced. Instead, the new work is planned to be at least three volumes that will cover a variety of dosimetry subjects, including theory, instrumentation, methods, and applications.

The new books are titled "The Dosimetry of Ionizing Radiation;" this title is certainly descriptive of their content. We do not mean to imply, however, that this set will cover all possible subjects under such a broad heading. It is also assumed that the second edition of "Radiation Dosimetry" (hereafter referred to as "Radiation Dosimetry") will continue to be available for some time to come, so that authors of the present work may freely make reference to it.

This first volume has been planned and organized to present several broad topics in dosimetry that can serve as foundations for what follows in later volumes. There are two chapters dealing with theoretical aspects of dosimetry. These are followed by two chapters concerning measurement of radiation fields, which will be applicable to radiation protection as well as to research, medical, and industrial uses of radiation beams. The final two chapters discuss determination of radioactivity in the environment and analysis of internal dose.

The first chapter by Gudrun Alm Carlsson presents her unique theoretical treatment of dosimetry. Albrecht M. Kellerer's chapter on microdosimetry updates and extends the earlier work of Harald Rossi in Volume I of "Radiation Dosimetry," and of William A. Glass and William A. Gross in "Topics in Radiation Dosimetry."

The chapter by Andrée Dutreix and André Bridier presents a timely summary of photon and electron beam dosimetry, taking into account the recent wave of protocols published in several countries. It supplements the chapters by John S. Laughlin on electron beam dosimetry and H. E. Johns on photon beam dosimetry in Volume III of "Radiation Dosimetry."

Johan Broerse, John Lyman, and Johannes Zoetelief have provided a chapter dealing with heavy-particle dosimetry. It provides important up-to-date information in the areas covered by the chapters by Mudundi R. Raju *et al.*, E. Tochilin and B. W. Shumway, and J. DePangher and Eugene Tochilin in Volume III of "Radiation Dosimetry."

The chapter by Kurt Lidén and Elis Holm is a thorough and well-documented reference on environmental radioactivity that largely replaces the chapter by W. V. Mayneord and C. R. Hill in "Radiation Dosimetry," Volume III.

Finally, John R. Johnson has written an excellent and much-needed chapter on internal dosimetry in reference to radiation protection.

Clearly this book is a valuable collection of work by outstanding authorities in their fields of specialty. In quality it measures up to the best of what has been published before. In any event, the editors cannot be accused of a parochial choice of authors: Canada, France, Germany, The Netherlands, Sweden, and the United States are represented!

Finally, we express our appreciation to our wives for their continued support and encouragement during the preparation of this book. In particular, Herb Attix thanks his wife, Shirley, for her unflagging patience and understanding, and for outstanding secretarial assistance in carrying out this and other writing and editing projects.

Kenneth R. Kase is now affiliated with the Department of Radiation Oncology, University of Massachusetts Medical Center, Worcester, Massachusetts.

1

Theoretical Basis for Dosimetry

GUDRUN ALM CARLSSON

DEPARTMENT OF RADIATION PHYSICS
THE MEDICAL SCHOOL
UNIVERSITY OF LINKÖPING
LINKÖPING, SWEDEN

I. Introduction	2
II. Energy Imparted: The Fundamental Quantity of Radiation Dosimetry	4
A. Basic Processes of Energy Impartation	4
B. The Imparted Energy $\delta\epsilon$ in a Basic Process	6
C. The Energy Imparted to the Matter in a Volume	11
III. Absorbed Dose	13
IV. Detector Response	15
A. Statistical Fluctuations in the Energy Imparted to the Detector: Stochastic Dosimetry	15
B. Detector Efficiency: Nonstochastic Dosimetry	16
V. Radiometric Quantities: Mean Energy Imparted and Absorbed Dose in Terms of Vectorial Energy Fluence	19
A. Scalar Radiometric Quantities	19
B. Vectorial Radiometric Quantities	22
C. Applicability of Scalar and Vectorial Fluence	23
D. Measurements of Fluence	25
E. Mean Energy Imparted and Absorbed Dose in Terms of the Vectorial Energy Fluence: Connection to Radiation Transport Theory	26
F. Examples of the Usefulness of the Vector Formalism	30
VI. Absorbed Dose in Terms of Scalar Radiometric Quantities and Interaction Coefficients	36
A. Definition of Absorbed Dose	36
B. Practical Calculations of Absorbed Dose in Nonequilibrium Requiring Complete Knowledge of the Field of Charged Particles	39
C. Radiation Equilibrium	45
D. Fano's Theorem	65
VII. Cavity Theory	67
References	71

I. Introduction

Radiation dosimetry has its origin in the medical application of ionizing radiation starting with the discovery of x rays by Röntgen (1895). In particular, the application in radiation therapy called for methods to predict and reproduce clinical results. Physical techniques soon proved superior to biological ones for executing reproducible radiation measurements. Radiation dosimetry has developed into a pure physical science.

The concept "dose" was introduced as the quantity of interest in relating physical measurements with the biological effect of radiation. "Dose" was meant to be a measure of the radiation emitted from an x-ray tube and given to the patient. The instrument used to measure x-ray output was called a "dosage meter." The connection to the medical profession is obvious. The term "dose" was used in a pharmacological sense analogously to its meaning when used in prescribing a dose of medicine.

Although "dose" was not rigorously defined it was evidently thought of as something like radiant energy, i.e., a quantity of the radiation field (radiometric quantity). It was soon recognized that the effects of radiation on biological tissues are not correlated to the radiant energy incident on but rather to that actually imparted to it. Christen (1914) defined a concept of "dose" which comes close to what is now called absorbed dose. It was defined as the "roentgen energy absorbed by a unit volume" and could be calculated as the product of a radiometric quantity and coefficients describing the interactions between radiation and matter.

Use of the concept of "dose" with two different meanings, as a radiometric quantity and as absorbed dose, has been a continuing source of confusion. A contributing factor is that the air ionization chamber has been the instrument of choice for determining the output from x-ray machines. The quantity measured is called exposure (earlier, exposure dose). Exposure is proportional to the absorbed dose in air under electronic equilibrium conditions and is thus related to the product of a radiometric quantity, the energy fluence of photons, and the mass energy absorption coefficient describing the interactions of the photons and their secondary electrons with air. As long as determinations of absorbed dose in airlike materials such as soft tissues are the main interest, and not specification of the radiation field itself, the fact that exposure is not a true radiometric quantity has no serious practical consequences. In connection with determinations of absorbed dose or mean energy imparted to high-atomic-number detectors such as those frequently used in, e.g., diagnostic radiology, ambiguities are more likely to appear as a result of interpreting exposure as a parameter of the radiation field.

Radiation dosimetry is fundamental to all fields of science dealing with radiation effects and is concerned with problems which are often intricate as hinted

above. A firm scientific basis is needed to face increasing demands on accurate dosimetry. This chapter is an attempt to review and to elucidate the elements for such a basis. Quantities suitable for radiation dosimetry have been defined in the unique work to coordinate radiation terminology and usage by the International Commission on Radiation Units and Measurements, ICRU. Basic definitions and terminology used in this chapter conform with the recent "Radiation Quantities and Units, Report 33" of the ICRU (1980).

A striking feature in early definitions of quantities for radiation measurements was their close connection to a particular experimental arrangement used in the measurements. With time the quantities have been given more general definitions such that they can be determined with a variety of experimental methods. Excellent reviews of the development of dosimetric quantities as reflected in the "Radiation Quantities and Units" reports of the ICRU have been given by Roesch and Attix (1968) and Wyckoff (1980). Quantities may also be defined which cannot be measured but which are helpful in calculations of other quantities or for the understanding of dosimetric problems. For instance, the vectorial quantities used extensively in radiation transport theory—a field of science closely related to radiation dosimetry—can be exploited to develop a vector formalism which is forceful in clarifying basic dosimetric quantities and experimental arrangements for their determination. It seems to be a trend that modern textbooks in radiation dosimetry contain a presentation of vectorial quantities and the connection to radiation transport theory (see, e.g., Stolz and Bernhardt, 1981), a trend which will be followed here.

The energy imparted by ionizing radiation to the matter in a volume (ICRU, 1980) is the fundamental quantity of radiation dosimetry, on the basis of which the quantity absorbed dose is derived to allow specification of the spatial distribution of the energy imparted to an irradiated medium. The definitions and physical significance of energy imparted and absorbed dose are treated in detail in Sections II and III.

A radiation detector responds to irradiation with a signal which is basically related to the energy imparted to the detector volume. This relation is treated in Section IV, the statistical fluctuations in the energy imparted to the detector in Section IV,A, and factors determining the conversion of the imparted energy into a detectable signal in Section IV,B.

Section V deals with the definitions of nonstochastic radiometric quantities, both scalar and vectorial. Expectation values of the energy imparted to a body and absorbed dose can by means of vectorial quantities be expressed in a way that clearly demonstrates the close connection between radiation dosimetry and radiation transport theory. Examples are given to demonstrate the usefulness of the vector formalism.

In Section VI, the quantity absorbed dose is defined in terms of the scalar quantity fluence (or energy fluence) and interaction coefficients allowing numer-

ical calculations to be performed in specified conditions. When radiation equilibrium can be presumed, the calculations are considerably simplified. In particular, calculations of absorbed dose are relevant to the field of cavity theory dealing with determinations of the absorbed dose at a specified point in a medium by means of the signal from a radiation detector. The concept of radiation equilibrium is treated in detail (Section VI,C) and the geometrical arrangements needed to establish various kinds of approximate radiation equilibrium at a point in a medium and at all points of a detector volume are discussed. Problems of nonequilibrium inevitably arise when, e.g., a bare Bragg-Gray detector is used in a medium of differing atomic composition or when the ranges of secondary radiations are comparable with the ranges (or mean free paths) of the primary radiation. Calculations of absorbed dose in nonequilibrium require complete knowledge of the field of charged ionizing particles and are considered in the particular light of cavity theory.

In the final section, Section VII, some problems of cavity theory are touched on to explore the significance of the absorbed dose equations derived in Section VI to this field.

II. Energy Imparted: The Fundamental Quantity of Radiation Dosimetry

Most physical, chemical, and biological effects yielding detectable signals from a radiation detector are basically correlated to the physical quantity: the energy imparted (ICRU, 1980) to the detector. The signal may in addition depend on the microscopic distribution in space and time of the imparted energy. The signal from the detector is, for instance, temperature rise, the number of ions produced, the amount of light emitted upon subsequent heating (thermoluminescence), etc. In biological objects, ionizations are sometimes thought to be more effective than excitations in causing the biological changes; both effects are correlated to the energy imparted. The quantity "the energy imparted by ionizing radiation to the matter in a volume" is regarded as the fundamental quantity of radiation dosimetry.

A. BASIC PROCESSES OF ENERGY IMPARTATION

The concept of "energy imparted by ionizing radiation to the matter in a volume" has a well-defined meaning as stated in definitions by the ICRU (1980). More familiar expressions like "energy absorbed, deposited in, or transferred to" are frequently used as synonyms for energy imparted. This will be avoided here since the latter expressions are sometimes also used with other meanings. In the following, emphasis will be given to the specific expression of "energy imparted," which will be used exclusively in the sense of its ICRU definition.

1. THEORETICAL BASIS FOR DOSIMETRY

The energy imparted ϵ to the matter in a volume is, due to the quantum nature of both radiation and matter, composed of discrete contributions $\delta\epsilon$ from a number of basic processes occurring in the volume (Alm Carlsson, 1979). The physical significance of the energy imparted is most clearly demonstrated by analyzing the contributions from the basic processes representing those elementary mechanisms by which energy is imparted to matter. One basic process of energy impartation is an interaction by an ionizing particle with the atomic constituents of matter; another is spontaneous nuclear or elementary particle transformation at which ionizing particles are created (source processes). The fundamental character of the basic processes of energy impartation arises from the fact that the *imparted energy* $\delta\epsilon$ in such a process can be determined from a detailed knowledge about interaction processes and spontaneous nuclear and elementary particle transformations only.* In contrast, "the *energy imparted* to the matter in a volume" requires that a volume and time interval be specified for its determination.

The imparted energy $\delta\epsilon$ in a basic process of energy impartation is given by

$$\delta\epsilon = T_b - \sum_i T_{a,i} + Q \quad (1)$$

where T_b is the kinetic energy (energy excluding rest-mass energy: $T = E - m_0c^2$) of the interacting ionizing particle immediately before the interaction (in a spontaneous process such as nuclear decay, $T_b = 0$), $\sum_i T_{a,i}$ the sum of the kinetic energies of all ionizing particles created in the process (including the residual kinetic energy of the primary particle if this is still an ionizing particle after interaction), and Q the release of rest-mass energy of nuclei and elementary particles in those basic processes in which transformations of nuclei and elementary particles occur.†

Secondary particles liberated in a basic process contribute to $\sum_i T_{a,i}$ only if they are considered to be ionizing particles. The value of the imparted energy $\delta\epsilon$ depends critically on the definition of an ionizing particle. *Ionizing particles* are charged or uncharged particles capable of causing ionization by primary or secondary processes [ICRU (1980): see also Alm Carlsson (1978) for a detailed discussion]. Ionization may occur directly through interactions with the atomic

*A distinction is made between "energy imparted," which refers to the "energy imparted to the matter in a volume," and "imparted energy," which refers to the fundamental processes of energy impartation and can be considered regardless of any volume. This is in accordance with a terminology suggested by Kellerer and Chmelevsky (1975a).

† Q is identical to the Q values used in nuclear physics to classify nuclear reactions as exoergic ($Q > 0$) or endoergic ($Q < 0$) (see, e.g., Evans, 1955).

electrons (primary processes) or indirectly through a nuclear reaction or spontaneous elementary particle transformation (e.g., neutron capture or π -meson decay) in which particles capable of ionizing directly are created (secondary processes). All charged particles can be characterized as ionizing on the basis of their ability to produce ionization through Coulomb collision with atomic electrons. However, when their kinetic energy drops below a certain cutoff value, T_{cut} (depending on particle type and the material considered), they cease to be ionizing unless they are still able to initiate nuclear or elementary particle reactions. For instance, positrons remain ionizing below the cutoff energy T_{cut} for ionizing through Coulomb collision because they can cause ionization by annihilation. Negatrons, protons, and other heavier nuclei, however, are important examples of charged particles which will cease to be ionizing when their kinetic energies drop below the cutoff energy T_{cut} . The same applies to photons but not to neutrons, which are able to produce ionization through secondary processes upon being captured by a nucleus.

B. THE IMPARTED ENERGY $\delta\epsilon$ IN A BASIC PROCESS

Determination of the imparted energy $\delta\epsilon$ in a basic process requires a detailed knowledge about the outcomes of interaction processes and spontaneous nuclear and elementary particle transformations. No complete treatment of the different processes which may occur will be given; the important interactions between ionizing particles and matter are described in a number of existing textbooks on radiation dosimetry (see, e.g., Whyte, 1959; Attix *et al.*, 1968, 1969; Kase and Nelson, 1978; ICRU, 1978; Greening, 1981). Instead, we focus attention in more general terms on those products of a basic process which constitute the imparted energy. It is instructive to distinguish between basic processes on the basis of their Q -values: $Q = 0$, $Q > 0$, $Q < 0$; cf. Eq. (1).

1. Interactions with $Q = 0$

Coulomb collision with atomic electrons, giving rise to excitations and ionizations, is the most important interaction of this kind. With $Q = 0$, the imparted energy $\delta\epsilon$ is given by

$$\delta\epsilon = T_b - \sum_i T_{a,i} \quad (2)$$

In words, the imparted energy is that part of the kinetic energy of the interacting particle which is converted into energy forms other than the kinetic energy of ionizing particles. For a more concrete exposition, consider the following example, Fig. 1.

1. THEORETICAL BASIS FOR DOSIMETRY

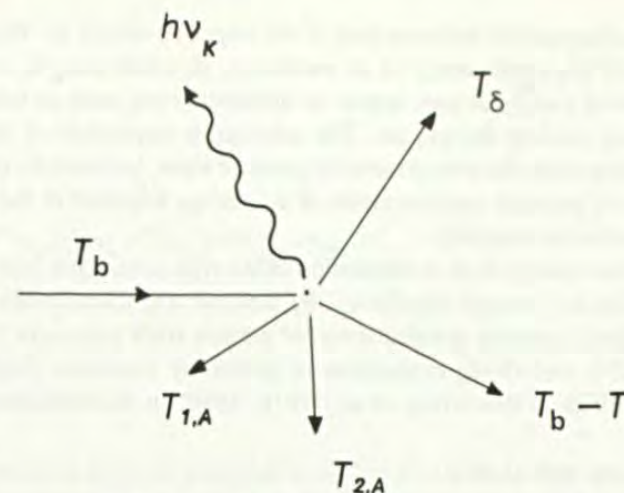


Fig. 1. A charged ionizing particle with kinetic energy T_b collides with an atomic electron and loses kinetic energy T . The atomic electron is knocked out and escapes with kinetic energy T_δ (δ particle). In the subsequent deexcitation of the electronic structure, one fluorescence photon with energy $h\nu_k$ and two Auger electrons with kinetic energies $T_{1,A}$ and $T_{2,A}$ are emitted. [After Alm Carlsson (1979).]

In Fig. 1, all particles which emerge from the interaction and the subsequent deexcitation processes as ionizing particles are depicted as waves (photons) or lines. Since the deexcitations of the electronic structure follow very closely in time (within $\sim 10^{-14}$ s) upon the interaction of the incident particle with the atomic electron, it is convenient to include them as a part of this interaction [treated as separate spontaneous ($T_b = 0$) basic processes they would yield negative values of imparted energy].* The imparted energy $\delta\epsilon$ is given by

$$\begin{aligned} \delta\epsilon &= T_b - (T_b - T + T_\delta + T_{1,A} + T_{2,A} + h\nu_k) \\ &= T - T_\delta - T_{1,A} - T_{2,A} - h\nu_k \end{aligned} \quad (3)$$

and may deviate appreciably from the kinetic energy T lost by the interacting particle. Note that if the interacting particle loses so much energy that its residual kinetic energy drops below the cutoff energy of an ionizing particle, i.e., $(T_b - T) < T_{\text{cut}}$, the term $T_b - T$ in the parentheses of Eq. (3) disappears and T_b replaces T in the last equality. In this case, the residual kinetic energy

*As a result of multiple Auger and Coster-Kronig transitions, the atom may become highly charged. The energies of ionizing particles, mostly low-energy photons, emitted in subsequent (occurring within 10^{-13} to 10^{-11} s) neutralization processes (Charlton *et al.*, 1983) are also included in $\sum_i T_{a,i}$.

of the interacting particle becomes part of the imparted energy $\delta\epsilon$. Equation (3) also yields the imparted energy of an excitation, in which case $T_b = 0$.

The imparted energy $\delta\epsilon$ may appear in different ways, such as heat, visible light, chemical binding energy, etc. The subsequent disposition of this energy is not of interest from the pure dosimetric point of view, but must be considered in the design of practical measurements of the energy imparted to the matter in a volume (radiation detector).

The imparted energy $\delta\epsilon$ in an interaction of the type considered here ($Q = 0$) has been called an "energy transfer ϵ_i " by Kellerer and Chmelevsky (1975b). It forms the basic quantity in calculations of particle track structures (see, e.g., Paretzke, 1974) underlying evaluations of proximity functions (Kellerer and Chmelevsky 1975c; Chmelevsky *et al.*, 1978, 1980) in microdosimetry.*

2. Interactions with $Q < 0$

A negative value of Q means that part $|Q|$ of the kinetic energy of the interacting particle is expended to increase the rest mass of nuclei and elementary particles. The imparted energy $\delta\epsilon$ can be written

$$\delta\epsilon = T_b - \sum_i T_{a,i} - |Q| \quad (4)$$

Kinetic energy of ionizing particles (radiant energy) which is expended to increase the rest mass of nuclei and elementary particles is not counted as imparted energy. The imparted energy is defined in terms of conversions of radiant energy related to changes in the electronic structure only. Most physical, chemical, and biological effects of radiation are considered to depend on such changes (transmutations of nuclei may under some circumstances, however, be of significance). This is important and is probably the reason to adopt the specific expression "imparted energy" instead of simply "absorbed energy." There exists a definite distinction between "imparted energy" and "absorbed energy" in the sense of radiant energy absorbed, i.e., removed from the radiation field and converted into other energy forms.

As an example of an interaction with $Q < 0$, consider pair production of a photon in the field of an atomic nucleus, Fig. 2.

The imparted energy of the pair production process is

$$\delta\epsilon = h\nu - (T_+ + T_-) - 2m_0c^2 \quad (5)$$

*In recent track calculations of Zaider *et al.* (1983), electrons are followed down to a kinetic energy of only 0.4 eV in water, considerably below the cutoff energy (12.6 eV) of ionizing electrons. Their track pattern of "energy depositions" is not strictly a pattern of imparted energies $\delta\epsilon$. In radiobiology, the interesting cutoff energy depends on the biological model considered.

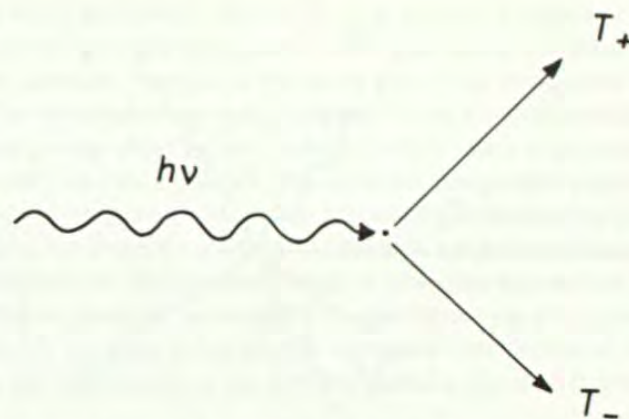


Fig. 2. A photon of energy $h\nu$ is absorbed in the field of an atomic nucleus by emission of a positron and a negatron (pair production) with kinetic energy T_+ and T_- . [After Alm Carlsson (1979).]

where $(T_+ + T_-)$ is obtained from the energy relation

$$T_+ + T_- = h\nu - 2m_0c^2 - T_R \quad (6)$$

and m_0c^2 is the rest-mass energy of an electron. Substituting Eq. (6) into Eq. (5) yields $\delta\epsilon = T_R$, the kinetic energy of the recoiling nucleus ($T_R \approx 0$). If the negatron is created with $T_- < T_{cut}$, T_- constitutes part of $\delta\epsilon$. The positron remains ionizing at all values of T_+ .

3. Processes with $Q > 0$

A positive value of Q means that rest-mass energy of nuclei and elementary particles is released into other energy forms, such as the kinetic energy of ionizing particles. These processes—interactions by ionizing particles or spontaneous nuclear and elementary particle transformations—act as radiation sources.

It is convenient to write the imparted energy $\delta\epsilon$ of a process with $Q > 0$ in the form

$$\delta\epsilon = Q - \left(\sum_i T_{a,i} - T_b \right) \quad (7)$$

Here, $\sum_i T_{a,i} - T_b$ is that part of the released rest-mass energy Q that appears as kinetic or quantum energy of ionizing particles. The imparted energy $\delta\epsilon$ is the remaining part of Q , i.e., that which is not converted into the kinetic or quantum energy of ionizing particles.

As an example, consider a two-quantum annihilation of a positron, Fig. 3.

The imparted energy $\delta\epsilon$ in the annihilation process is

$$\delta\epsilon = 2m_0c^2 - (h\nu_1 + h\nu_2 + h\nu_k + T_{1,A} + T_{2,A} - T_+) \quad (8)$$

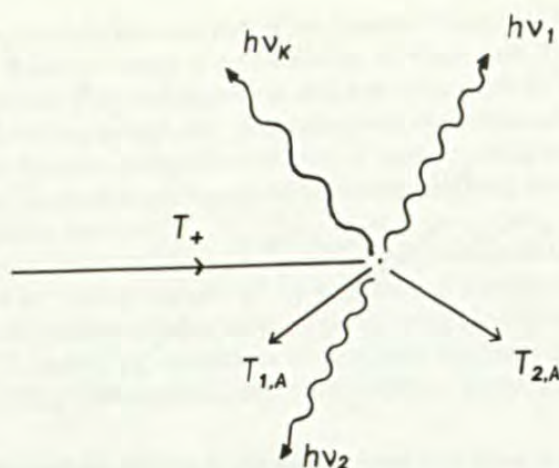


Fig. 3. A positron with kinetic energy T_+ annihilates with an atomic electron giving rise to two annihilation photons with energies $h\nu_1$ and $h\nu_2$. In subsequent deexcitations of the electronic structure, a characteristic roentgen ray with energy $h\nu_k$ and two Auger electrons with kinetic energies $T_{1,A}$ and $T_{2,A}$ are emitted.

The energies of the annihilation photons are given by the energy relation (E_B is the binding energy of the annihilated atomic electron)

$$h\nu_1 + h\nu_2 = 2m_0c^2 - E_B + T_+ \quad (9)$$

Substituting Eq. (9) into Eq. (8) yields

$$\delta\epsilon = E_B - T_{1,A} - T_{2,A} - h\nu_k \quad (10)$$

which is identical to the $\delta\epsilon$ in the example illustrated in Fig. 1, since $T - T_\delta$ in Eq. (3) can be identified with E_B . The imparted energy in an annihilation process is independent of whether the annihilation occurs in flight ($T_+ > 0$) or at rest ($T_+ = 0$) and equals the imparted energy of other processes causing the same type of ionization.

Spontaneous [$T_b = 0$ in Eqs. (1) and (7)] nuclear decay or isomeric nuclear transition are other examples of processes with $Q > 0$. Following the scheme of calculations in the preceding examples, it can be shown that the imparted energy in a nuclear decay equals the kinetic energy of the recoiling daughter nucleus plus the decrease in the total binding energy of the electrons bound to the mother nucleus before decay. The recoil energy of the daughter nucleus may be sufficient to cause some atomic electrons to be stripped off and the nucleus (possibly surrounded by some of its atomic electrons) to proceed as an ionizing particle. The imparted energy in the decay then reduces to the decrease in the total binding energy of atomic electrons. The same reasoning applies to nuclear reactions induced by ionizing particles.

The fact that a spontaneous nuclear decay is by itself a process of energy impartation apart from those subsequently occurring during the slowing down of the ionizing particles liberated in the decay may seem strange but is a consequence of the definition of the energy imparted to the matter in a volume. A conceptually interesting detail is that in decays which result in an increase in the atomic number, as in a β^- -decay, the imparted energy may become negative due to the resulting increase in binding energy of the surrounding electrons. In the same way, the imparted energy of an initiated nuclear reaction may become negative. Practically, the imparted energy in processes like nuclear decays and initiated nuclear reactions is neglected in calculations of, e.g., absorbed dose (see Section VI, A), since it is negligible compared with the sum of the imparted energies in the interactions of the ionizing particles liberated in the processes.

C. THE ENERGY IMPARTED TO THE MATTER IN A VOLUME

The energy imparted ϵ to the matter in a volume is the sum of the imparted energies in all those basic processes which have occurred in the volume during the time interval considered:

$$\epsilon = \sum_i \delta\epsilon_i \quad (11)$$

It may be convenient (cf. Section V, C) to express ϵ in an alternative way [identical to the formulation used by the ICRU (1980) in defining ϵ]. Consider an ionizing particle of kinetic energy T_{in} incident on a volume, Fig. 4.

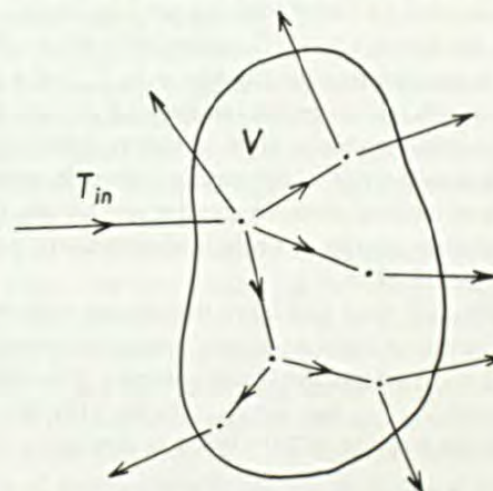


Fig. 4. An ionizing particle with kinetic energy T_{in} enters the volume V and gives rise to a series of correlated basic processes in it. Points indicate basic processes occurring during the time interval considered. In addition to interactions by ionizing particles, decays of radioactive nuclei created in a preceding interaction have to be considered. Freely moving ionizing particles are represented by lines, and the arrows indicate directions of motion. [After Alm Carlsson (1979).]

Ionizing particles interacting in the volume have, except for the incoming particle, been created in a preceding basic process in it and are represented by lines connecting two points. Ionizing particles created within the volume and escaping from it without further interactions are represented by lines pointing out of the volume. An ionizing particle escaping and reentering after one or more interactions outside is considered as a new incoming particle. Then, if K ionizing particles enter and interact in the volume during the time interval considered and N radioactive nuclei existing in the volume before this time interval decay during it, the energy imparted to the matter in the volume is given by

$$\begin{aligned}\epsilon &= \sum_{j=1}^{K+N} \left(\sum_{i=1}^{n_j} \delta\epsilon_{ij} \right) = \sum_{j=1}^{K+N} \left(\sum_{i=1}^{n_j} \left\{ T_b - \sum_k T_{a,k} + Q \right\}_{ij} \right) \\ &= \sum_{j=1}^K T_{in,j} - \sum_{m=1}^L T_{out,m} + \sum_{n=1}^M Q_n\end{aligned}\quad (12)$$

where n_j is the total number of correlated basic processes initiated in the volume by the j th incoming particle (alternatively the j th decaying nucleus).

The first term in the last equality is the sum of the kinetic energies of all those incident ionizing particles which interact in the volume, the second is the sum of the kinetic energies of all ionizing particles which have been created in a basic process in the volume and which escape without further interactions, and the last term is the net release of rest-mass energy of nuclei and elementary particles in those basic processes which have occurred within the volume. In calculating the sum $\sum_i \delta\epsilon_{ij}$ one notices that for every ionizing particle created and interacting again in the volume, the kinetic energy T_b immediately before the interaction is exactly canceled by a negative term in the sum $-\sum_k T_{a,k}$ of a preceding basic process, viz., that in which it was created (an ionizing particle does not change its kinetic energy in moving from one point of energy impartation to the next, Fig. 4). Thus, the remaining terms in calculating ϵ are only concerned with the kinetic energies of those ionizing particles entering and leaving the volume plus the net release of rest-mass energy of nuclei and elementary particles in those M basic processes in which $Q \neq 0$.

An ionizing particle may enter and leave the volume without interacting in it. It does not contribute to ϵ since its kinetic energy on entering the volume equals that on leaving it. By adding the kinetic energies of noninteracting ionizing particles to the sums $\sum_j T_{in,j}$ and $\sum_m T_{out,m}$ in Eq. (12), the definition of ϵ in the formulation adopted by the ICRU (1980) is obtained:

$$\epsilon = R_{in} - R_{out} + \sum_n Q_n \quad (13)$$

where $R = \sum T$ (the sum of the kinetic or quantum energies of ionizing particles) is called *radiant energy*.

The value of the energy imparted ϵ depends, as does the value of the imparted energy $\delta\epsilon$ in a single basic process, on the choice of a cutoff energy T_{cut} for ionizing particles. Values of $\delta\epsilon$ are more critically dependent on this choice than values of ϵ , which may seem contradictory. However, only interactions by *ionizing* particles are considered processes of energy impartation. For instance, lowering the value for T_{cut} means that $\delta\epsilon$ of a particular interaction decreases. At the same time, the number of particles which are characterized as ionizing increases, as does the total number of basic processes of energy impartation in the volume (the number of lines and points in Fig. 4 increases). The uncertainty in determining the exact value for T_{cut} seems, in practice, not to be crucial with respect to determinations of energy imparted.

III. Absorbed Dose

It is of interest to ask for the spatial variation of the energy imparted to an irradiated body, i.e., to divide it into smaller volume elements and to determine the energy imparted to these as a function of position.

Due to the stochastic nature of the emission of ionizing particles from radiation sources, as well as the stochastic nature of the interactions by ionizing particles, ϵ is a stochastic quantity. The value taken by ϵ in an irradiation cannot be predicted. In repeated irradiations under given conditions, it will fluctuate statistically according to a probability distribution resulting from statistical fluctuations in both the number and the type of basic processes (Section IV, A). Descriptions of spatial variations of the energy imparted are simplified by considering expectation values only.

Complete geometrical resolution of the spatial distribution of the energy imparted requires that the body be subdivided into very small, infinitesimal volume elements. Conceptual difficulties regarding the meaning of the energy imparted to an infinitesimal volume can be resolved by turning to expectation values and may form an additional reason to restrict descriptions of spatial variations of energy imparted to such values.

In proceeding to very small volumes of dimensions comparable to the ranges of interaction forces, the very concept of the energy imparted to the matter in the volume becomes obscured. An ionizing particle passing outside the volume may interact with an atomic constituent inside the volume and, conversely, an ionizing particle passing through the volume may interact with an atomic constituent outside it. Moreover, de-excitation and neutralization processes in the electronic structure imply that more than one atom or molecule may be involved in a basic process of energy impartation. On an atomic scale, the basic processes cannot be regarded as confined to points in space. The equivalence between the definitions, Eqs. (11) and (13), of the energy imparted to the matter in a volume is questionable; in deriving Eq. (13) from Eq. (11) it was tacitly assumed that interaction forces have a zero range and basic processes occur at well-defined

points (Fig. 4), which is justified when volumes with large dimensions compared to atomic dimensions are treated.

On an atomic scale, particles—both free ionizing particles and the constituents of matter such as the atomic electrons—are quantum-mechanically described by wave functions, which means that their localization in space cannot be determined exactly; the wave function of a particle contains information of the probability of finding it at a given point only. Free ionizing particles of a given energy are described by plane waves, meaning that they have a uniform probability to be anywhere in a beam incident on a volume of atomic dimensions. Provided the atomic composition and the density (the expectation value of the number of atoms per unit volume) are homogeneous in the volume and its surroundings up to distances equal to the maximum range of interaction forces, it is meaningful to state that *on the average* there will, for each particle passing outside the volume and interacting with an atomic constituent inside, exist a particle passing inside the volume and interacting with an atomic constituent outside it. In terms of expectation values, the definitions of the energy imparted, Eqs. (11) and (13), remain equivalent also with respect to infinitesimal volumes. However, for small volumes situated in an inhomogeneous surrounding, as at an interface between media of differing atomic composition and density, the significance of the concept "the energy imparted to the matter in the volume" is still obscure and cannot be resolved on the basis of expectation values. This complication is disregarded in the following.

When expectation values are considered, a volume may contain fractions of an atom, molecule, or crystal lattice (or whatever structure determines the type of interactions occurring) as well as fractions of interactions. The expectation value $\bar{\epsilon}$ (the mean energy imparted to the matter in an infinitesimal volume ΔV) takes a definite value, which, however, approaches zero as ΔV approaches zero. Values of zero can be avoided if instead the quotient $\bar{\epsilon}/\Delta V$ is determined; $\bar{\epsilon}/\Delta V$ approaches a finite value in the limit of $\Delta V \rightarrow 0$. The quantity the absorbed dose D is defined as (ICRU, 1980)

$$D = \lim_{m \rightarrow 0} \frac{\bar{\epsilon}}{m} = \lim_{V \rightarrow 0} \frac{1}{\rho V} \bar{\epsilon} \quad (14)$$

where m is the mass contained in the volume V .*

The absorbed dose is the expectation value of the energy imparted per unit mass of an infinitesimal volume (the condition $m \rightarrow 0$ meaning that $V \rightarrow 0$ and not $\rho \rightarrow 0$, as expressed in the second equality). It takes a value at each *point*

*The definition of D is sometimes written $D = d\bar{\epsilon}/dm$. Here, $d\bar{\epsilon}$ means the mean energy imparted to the matter in an infinitesimal volume of mass dm . The quotient $d\bar{\epsilon}/dm$ could, however, be confused with the derivative of a function $\bar{\epsilon}(m)$, the mean energy imparted to the matter in a volume of mass m ; D is the derivative of this function at $m = 0$. The formulation in Eq. (14) prevents confusion and is suitable in exploring the vector formalism (Section V).

of an irradiated medium and is a quantity that offers a way to describe spatial distributions of energy imparted.

Physical measurements are basically concerned with determinations of the stochastic quantity ϵ to the matter in a finite volume (detector). Determination of absorbed dose at a point can not, in principle, be performed through a single measurement but requires the use of averaging and extrapolation procedures. The expectation value $\bar{\epsilon}/m$ is obtainable as the mean of a number of repeated measurements. In determining $\bar{\epsilon}/m$ for a range of volumes, an extrapolation to infinitesimal volumes can be made. Often, the absorbed dose can be considered to be practically constant throughout a finite volume. In this case, a value for absorbed dose can be determined in a single measurement provided the mean energy imparted $\bar{\epsilon}$ to the volume in which D is constant is large enough to render the statistical fluctuations in ϵ negligible.

The fundamental connection between the quantities ϵ and D is often overlooked, which is probably promoted by calling D "absorbed dose." The concept "absorbed" is easy to grasp intuitively; that of "imparted" appears to be stranger. In keeping strictly to the well-defined concept of "energy imparted" as distinct from "absorbed energy," the quantity D could as well be called "imparted dose."

IV. Detector Response

A. STATISTICAL FLUCTUATIONS IN THE ENERGY IMPARTED TO THE DETECTOR: STOCHASTIC DOSIMETRY

A radiation detector responds to irradiation with a signal M which is basically related to the energy imparted ϵ to the detector volume, Eq. (11). Some of the contributions $\delta\epsilon$ are statistically correlated since they are imparted by statistically correlated particles; a primary particle emitted from a radiation source is statistically correlated to the secondary particles of all generations that it generates when slowing down. The primary particles emitted from the radiation sources are statistically independent.

The imparted energies $\delta\epsilon$ in a volume from statistically correlated particles can be lumped together, yielding the magnitude of energy imparted in an energy-impartation event [energy-deposition event in the terminology of ICRU (1980)]. The energy imparted to the matter in a volume can alternatively be obtained by summing the imparted energies in a number of statistically independent energy-impartation events. The stochastics of the quantity ϵ are determined by the combined statistical fluctuations in the number and in the size of such events:

$$\frac{V(\epsilon)}{(\bar{\epsilon})^2} = \frac{V(n)}{E^2(n)} + \frac{1}{E(n)} \frac{V_1(\epsilon)}{E_1^2(\epsilon)} \quad (15)$$

where $V(n)$ and $E(n)$ are the variance and the expectation value of the number n of energy-impartation events occurring in the detector, and $V_1(\epsilon)$ and $E_1(\epsilon)$

are those of the energy imparted in a single energy-impartation event, being determined by a frequency function $f_i(\epsilon)$. A proof of the relation in Eq. (15) is given by Kellerer (1968).

When n is distributed according to the Poisson distribution, $V(n) = E(n)$ and Eq. (15) gives the relative variance of the compound Poisson process.

The frequency function $f_i(\epsilon)$ is of interest in a variety of situations. In microdosimetry, determinations of $f_i(\epsilon)$ with respect to very small detector volumes (cell structures with dimensions of micrometers or smaller) form the basis of a description of radiation quality with respect to biological effectiveness. The frequency function $f_i(\epsilon)$ is also of interest in connection with detectors of much larger size. A particle energy spectrometer, e.g., a crystal scintillator used for γ -ray spectroscopy, responds with a pulse-height distribution $f(v)$ which is related to $f_i(\epsilon)$; $f_i(\epsilon)$ is usually called the detector response function and can be calculated by means of statistical methods such as analog Monte Carlo calculations (see, e.g., Seltzer, 1981; Rogers, 1982). The stochastic nature of the processes following energy impartation (the emission of light quanta in the scintillator, the generation and multiplication of photoelectrons in the PM-tube, etc.) means that there is a larger relative spread in the values taken by the final electrical signal v than by the energy imparted in an energy-impartation event. The quantum noise in, e.g., radiographic screens is directly related to the fluctuations in the energy imparted to the image elements of the screen and can be analyzed with the aid of Eq. (15). Here $f_i(\epsilon)$ for an image element must be known to determine $E_i(\epsilon)$ and $V_i(\epsilon)$.

B. DETECTOR EFFICIENCY: NONSTOCHASTIC DOSIMETRY

In most conventional dosimetry applications, the relative statistical fluctuations in the energy imparted to the detector (often called dosimeter or dose-meter) are small due to the integrated effect of a large number of energy-impartation events. The relative variance approaches zero when $E(n)$ gets large, Eq. (15). The fluctuations in the energy imparted to the detector are small compared to other uncertainties in registering the signal M . Then

$$M = \bar{\eta} \bar{\epsilon}_{\text{det}} \quad (16)$$

where $\bar{\eta}$ is the efficiency of the detector. (A bar is used with η to designate that the efficiency is here a nonstochastic quantity. There exists a stochastic quantity η , defined by the relation $M = \eta \epsilon_{\text{det}}$, that is of significance in stochastic dosimetry.)

In a pure, elementary substance such as carbon in which no chemical changes or lattice defects are produced, the imparted energy is ultimately converted into heat energy. The efficiency $\bar{\eta} = 1$ is independent of particle type and energy; M is induced heat energy. Usually the signal M is registered as the temperature rise in the detector material. The registered signal is in this case primarily a

measure of the mean specific energy imparted (ICRU, 1980) or the spatial average of absorbed dose $\bar{D} = \bar{\epsilon}/m$ in the detector.

When other effects are considered, the efficiency $\bar{\eta}$ depends on particle type and energy. For a limited number of detector systems, generally applicable values for $\bar{\eta}$ are known with an accuracy such that they can be used for absolute determinations of the mean energy imparted to the detector.

Collection of the number M of ion pairs formed in a gaseous detector volume is a well-established technique of measurement. The efficiency $\bar{\eta}$ equals $1/\bar{W}$, where $\bar{W}$ is the mean energy imparted per ion pair formed. The ion pairs formed are essentially primary products of the physical interactions between charged particles and the gas molecules in the detector. Values of $\bar{W}$ depend on the relative occurrence of ionizations and excitations in the interactions and vary with particle type and energy. Ion pairs may additionally be formed when excited molecules collide thermally with other molecules in the gas resulting in a (weak) dependence of $\bar{W}$ on temperature and pressure (ICRU, 1979).

As a result of chemical reactions initiated by the primary products of interactions, an amount M of a particular substance x may be produced, destroyed, or changed. The efficiency $\bar{\eta}$ is in this case identical to the radiation chemical yield $G(x)$ (ICRU, 1980). Values of $G(x)$ are influenced not only by the relative occurrence of ionizations and excitations among the primary products of interaction but also by the spatial proximity of the excitations and ionizations along the charged particle tracks, crudely described by the concept of linear energy transfer, LET (ICRU, 1980), and by intertrack interferences. Contrary to $\bar{W}$ values, values of $G(x)$ depend in an intricate manner on the microscopic distribution in both space and time of the imparted energy; they depend on levels of absorbed dose and absorbed dose rates.

Values of the mean energy imparted required to form an ion pair are frequently determined by stopping charged particles of known kinetic energy T in the detector volume. From such measurements, W values are determined according to

$$W = T/\bar{N} \quad (17)$$

where $\bar{N}$ is the mean number of ion pairs produced when the primary charged particle with initial kinetic energy T , together with all its secondary particles, is completely stopped in the gas. As defined by ICRU (1980) W , Eq. (17), is the mean energy expended in a gas per ion pair formed.

Commonly, the detector is traversed by charged particles which are not completely stopped in it. Values of the efficiency $1/\bar{W} = M/\bar{\epsilon}$ cannot be evaluated from W values determined according to Eq. (17) except for the special case when charged particle equilibrium (Section VI,C,3) prevails at all points in the detector and bremsstrahlung losses are negligible, or in the case when W is practically independent of energy as for electrons. In other cases, the efficiency must be

evaluated by means of differential values that are valid along segments of the charged particle tracks. Differential W values defined by ICRU (1979) are applicable in cases when δ -particle equilibrium (Section VI,C,4) prevails and the δ particles do not lose energy to bremsstrahlung. For a still more general situation when no kind of equilibrium can be assumed in the detector (Section VI,B), differential values accounting separately for the ion pairs formed by δ particles may have to be considered (Carlsson, 1972; ICRU, 1979).

The W values, Eq. (17), are essentially independent of T for electrons with $T \geq 1$ keV (Combecher, 1980). For $T \leq 1$ keV, W increases with decreasing kinetic energy T , indicating that a decreasing fraction of the energy lost by an electron goes into ionization. The same trend applies to protons and other heavy charged particles such as α particles. For energies T such that W is essentially independent of energy ($T \geq 1$ MeV for, e.g., protons), W values for different types of charged particles are contained within a narrow energy interval, indicating that the partition between ionizations and excitations in the primary interactions is relatively independent of the type of charged particle.

The $G(x)$ values are usually determined according to the definition of $\bar{\eta} = G(x) = M/\bar{\epsilon}$ as are $\bar{W}$ values at irradiations with high-energy photons (>1 MeV). Uncertainties arise in determining $\bar{\epsilon}$ which must be deduced indirectly with the aid of other detector systems involving cavity theory (Section VII). This source of error is avoided in determinations of W values according to Eq. (17). Discrepancies of as much as 4% in values of $G(x)$ obtained with incident high-energy electrons and with secondary electrons generated by photons with similar LET distributions of the electrons in the detector solution could be traced back to errors in cavity theory (Nahum and Greening, 1978).

For electrons in air W values are well established. ICRU (1979) recommends a value of 33.85 ± 0.15 eV per ion pair (or J/C) formed in dry air ($T \geq 1$ keV). The inverse of this can be used for the efficiency $\bar{\eta}$ irrespective of the distribution in energy of the electrons traversing the detector. There exists some indication that W values for strongly relativistic electrons with $T \geq 10$ MeV may increase somewhat with increasing energy (ICRU, 1984a). The efficiency increases (W decreases) by addition of water vapor to dry air (ICRU, 1979), giving the conventional value of 33.7 J/C for usual conditions.

The ferrous sulfate (Fricke) dosimeter is the only aqueous chemical system that has reached such a degree of standardization that accurate values of $G(x)$ for electrons (including secondary electrons generated by photons) can be recommended for general use. The value of $G(x)$ depends on the actual distribution in LET of the electrons imparting energy to the solution (ICRU, 1969, 1970, 1984a). For example, the value of $G(x)$ for the amount of Fe^{3+} produced in electronic equilibrium from ^{60}Co gamma rays is 1.61 ± 0.02 $\mu\text{mole/J}$ (ICRU, 1969, 1980).

In neutron dosimetry, the efficiency $\bar{\eta} = 1/\bar{W}$ depends in a sensitive manner on the generation and slowing down of different kinds of heavy charged particles (protons, α particles, heavy nuclei). It is advisable that surrounding walls of an ion chamber be of identical atomic composition to get essentially charged particle equilibrium in the gas (see Sections VI,C and D on Fano's theorem) to avoid a dependence of $\bar{W}$ values on the size of the gas volume. ICRU (1979) recommends a value of 31 ± 1.5 eV for neutron energies between 1 and 14 MeV and tissue-equivalent ion chambers. The $G(x)$ values for use in neutron dosimetry are, like $\bar{W}$ values, known to a lower degree of accuracy (ICRU, 1977) than corresponding values for use in electron and photon dosimetry.

V. Radiometric Quantities: Mean Energy Imparted and Absorbed Dose in Terms of Vectorial Energy Fluence

In terms of expectation values or the average behavior of radiation, it is meaningful to talk about a radiation field, varying continuously in space and time. Radiometric quantities are used to describe the radiation field. Since expectation values are considered, the radiometric quantities are nonstochastic and their definitions are given in differential form (ICRU, 1980). In the following, expectation values are presumed even though, for brevity, this is not always explicitly stated.

A. SCALAR RADIOMETRIC QUANTITIES

A complete description of the radiation field requires that at any point $\mathbf{r}$ and at any time t the number N and the kind j of ionizing particles, their kinetic energies T , and directions of motion Ω be specified. The most detailed description of the radiation field is contained in the quantity p_T , called the spectral distribution with respect to kinetic energy of the particle radiance, p :

$$p_{T,j}(\mathbf{r}) = \frac{dp_j(\mathbf{r})}{dT} = \frac{d^4N_j(t, T, \Omega; \mathbf{r})}{dt dT d\Omega da_\perp} \quad (18)$$

with meaning that $p_{T,j}(\mathbf{r}) dt dT d\Omega da_\perp$ is the expectation value of the number of particles N of type j with kinetic energy in the interval dT around T and direction of motion in the solid angle element $d\Omega$ around Ω , which during the time interval dt around t at the point $\mathbf{r}$ pass through the area element $da_\perp$ perpendicular to the direction of motion Ω .

A field quantity takes a value at each point of space. This is stressed by writing $p_T(\mathbf{r})$ instead of simply p_T .

The fluence $\Phi(\mathbf{r})$ is the integral (including a summation over the particle types j) of $p_T(\mathbf{r})$ over time, kinetic energy, and solid angle:

$$\begin{aligned}\Phi(\mathbf{r}) &= \int_{t_1}^{t_2} dt \int_0^\infty dT \int_{4\pi} d\Omega \frac{d^4 N(t, T, \Omega; \mathbf{r})}{dt dT d\Omega da_\perp} \\ &= \int_{4\pi} d\Omega \frac{d^2 N(\Omega; \mathbf{r})}{d\Omega da_\perp} = \frac{dN(\mathbf{r})}{da}\end{aligned}\quad (19)$$

where in the last equality dN can be interpreted as the number of particles striking a sphere of cross-sectional area da centered at $\mathbf{r}$.

The sphere, commonly employed in defining particle fluence (ICRU, 1980), is convenient but not essential for the definition of fluence. As seen from Eq. (19), the fluence yields the number of particles passing through per unit area at point $\mathbf{r}$ under the particular constraint that for each direction of motion of the particles the area element considered is perpendicular to that direction. The sphere has the convenient property that its cross-sectional area is equal in all directions.

The quantity $p_T(\mathbf{r})$ in Eq. (18) is displayed as a derivative with respect to time t , kinetic energy T , direction of motion Ω , and area $a_\perp$ while the function $N(t, T, \Omega; \mathbf{r})$ being differentiated is described as a function of t , T , and Ω only. Also, $N(t, T, \Omega; \mathbf{r})$ is a sum distribution (ICRU, 1980) and designates the number N of all particles which pass through the area $a_\perp$ perpendicular to Ω during all time intervals between 0 and t , with kinetic energies between 0 and T and with directions of motion $\Omega = (\theta, \phi)$ contained in the interval between $(0, 0)$ and (θ, ϕ) .^{*} Since $p_T(\mathbf{r})$ describes the field at the particular point $\mathbf{r}$, the derivative of the function $N(t, T, \Omega; \mathbf{r})$ with respect to area is considered at $a_\perp = 0$ only [analogous to the definition of absorbed dose, Eq. (14) and the footnote on p. 14]:

$$\frac{dN(t, T, \Omega; \mathbf{r})}{da_\perp} = \lim_{a_\perp \rightarrow 0} \frac{N(t, T, \Omega; \mathbf{r})}{a_\perp} \quad (20)$$

For the variables t , T , and Ω it is of interest to know the rate of change at varying values to allow complete specification of the particles passing through per unit area at point $\mathbf{r}$.

In defining the scalar radiometric quantities, ICRU (1980) starts with the quantity fluence as defined by the last equality in Eq. (19). The more descriptive quantities are derived from the fluence by successive derivations. Consequently, the definition of $p_T(\mathbf{r})$ differs from the formulation in Eq. (18) in that da (the cross-sectional area of the sphere) is substituted for $da_\perp$, which does not,

^{*}The angles θ and ϕ are the spherical coordinates most commonly used to specify direction of motion. A detailed treatment of distributions with respect to directions of motion expressed by means of θ and ϕ are given by, e.g., Roesch and Attix (1968).

however, change the physical significance of $p_T(\mathbf{r})$. Here, Eq. (18) is preferred since it draws attention to essential features of the quantity fluence which are relevant to fluence measurements which do not, as commonly assumed, require use of a spherical ("isotropic") detector.

An important aspect of the quantity fluence is that it can be interpreted as the track-length density (Weinberg and Wigner, 1958) of the particles at a point in space, i.e.,

$$\Phi(\mathbf{r}) = dL(\mathbf{r})/dV \quad (21)$$

where dL is the sum of the track lengths of all the particles traversing the (infinitesimal) volume element dV around $\mathbf{r}$.

This relation is of considerable practical usefulness in, e.g., Monte Carlo calculations of particle fluences. The track lengths of the particles traversing finite volumes (e.g., slices) of the medium are summed and divided by the volume. A mean value $\bar{\Phi}$ of the fluence over that volume is obtained from

$$\bar{\Phi} = \frac{1}{V} \int_V \frac{dL(\mathbf{r})}{dV} dV \quad (22)$$

Various approaches to prove the identity between fluence and track-length density have been used by, e.g., Rossi (1967), Kellerer (1971), and Chilton (1978). Here, a derivation will be given that takes still another approach starting from the most descriptive quantity $p_T(\mathbf{r})$ (Alm Carlsson and Carlsson, 1982).

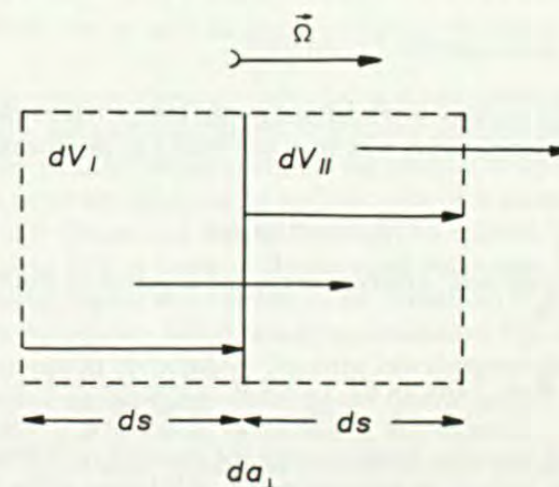


Fig. 5. Illustration to show that fluence is identical to total track length per unit volume. A plane area element $da_\perp$ perpendicular to the direction of motion Ω is traversed by particles with velocity v traveling the distance $ds = v dt$ during the time interval dt . Volume elements dV_I and dV_{II} are defined by $dV = da_\perp ds$. Arrows indicate track elements ds of four particles with various positions at time t . (Vectors indicated by boldface in text, arrows in illustrations.)

The area element $da_{\perp}$ and the time interval dt are assumed to be so small that the quantity $p_T(\mathbf{r})$, Eq. (18), remains constant throughout the volume elements dV_1 and dV_{II} , Fig. 5. It is also assumed that the distance ds defined by $ds = v dt$ is so short that the majority of the particles will proceed along this distance without interacting. During the time interval dt a number of $p_T(\mathbf{r}) dt dT d\Omega da_{\perp}$ particles with kinetic energies in the interval dT around T and directions of motion in $d\Omega$ around Ω pass through the area $da_{\perp}$. Some particles which pass through $da_{\perp}$ after time t (but before $t + dt$) do not travel all the distance ds in volume dV_{II} . However, the lacking part of ds is compensated for by the track of a particle moving in the interior of dV_{II} at time t and escaping from it during dt , such that the total track length dL of all particles moving in dV_{II} during t and $t + dt$ equals $dL = p_T(\mathbf{r}) dt dT d\Omega da_{\perp} ds$. The intervals dT and $d\Omega$ are chosen small enough to have negligible influence upon the distance ds traveled during dt , and upon the fraction of particles escaping through the upper and lower surfaces of the volume. Since $dV = da_{\perp} ds$, the total track length per unit volume and per unit interval of time, kinetic energy, and direction of motion is

$$\frac{d^4L(t, T, \Omega; \mathbf{r})}{dt dT d\Omega dV} = p_T(\mathbf{r}) \quad (23)$$

Integrating Eq. (23) over time, kinetic energy, and direction of motion yields the relation in Eq. (21).

B. VECTORIAL RADIOMETRIC QUANTITIES

The basic scalar quantity $p_T(\mathbf{r})$ defined in Eq. (18) is represented in vectorial form through multiplication with Ω , a unit vector in the direction of motion of the particles:

$$\mathbf{p}_T(\mathbf{r}) = \Omega p_T(\mathbf{r}) \quad (24)$$

Integrating $\mathbf{p}_T(\mathbf{r})$ over time, kinetic energy, and direction of motion defines the vectorial fluence Φ :

$$\Phi = \int_{t_1}^{t_2} dt \int_0^{\infty} dT \int_{4\pi} d\Omega \Omega \frac{d^4N(t, T, \Omega; \mathbf{r})}{dt dT d\Omega da_{\perp}} \quad (25)$$

If the integration is limited to kinetic energy and direction of motion, the vectorial fluence rate ϕ is derived. In transport theory, ϕ is known as the "current density, $\mathbf{J}$ " [see, e.g., Duderstadt and Martin (1979); see also Alm Carlsson and Carlsson (1982) for a detailed discussion of quantities from the radiation transport theory and their connection to the scalar radiometric quantities defined in ICRU (1980)].

1. THEORETICAL BASIS FOR DOSIMETRY

Since the vectorial fluence Φ is a sum of vectors, Eq. (25), while the fluence Φ is the scalar sum of the vector lengths, Eq. (19), one has

$$|\Phi| \leq \Phi \quad (26)$$

where $|\Phi|$ is the length of the vector Φ ($|\Phi| = \sqrt{\Phi \cdot \Phi}$).

Equality in Eq. (26) holds only for the case of monodirectional radiation. With isotropic radiation, $|\Phi|$ equals zero. It also equals zero in the case of two parallel opposed identical beams of particles.

The energy fluence Ψ is a useful scalar quantity in dosimetry and is obtained by multiplying $p_T(\mathbf{r})$ with the kinetic energy T of the particles and integrating:

$$\Psi(\mathbf{r}) = \int_{t_1}^{t_2} dt \int_0^{\infty} dT \int_{4\pi} d\Omega T p_T(\mathbf{r}) \quad (27)$$

The corresponding vectorial energy fluence Ψ is obtained by vectorial integration of the quantity $T p_T(\mathbf{r})$

$$\Psi(\mathbf{r}) = \int_{t_1}^{t_2} dt \int_0^{\infty} dT \int_{4\pi} d\Omega \Omega T p_T(\mathbf{r}) \quad (28)$$

and analogously to the comparison of fluence and vectorial fluence

$$|\Psi| \leq \Psi \quad (29)$$

C. APPLICABILITY OF SCALAR AND VECTORIAL FLUENCE

The scalar quantity fluence is relevant to determinations of absorbed dose. The fluence is the total track length of ionizing particles per unit volume, Eq. (21). Ionizing particles impart energy to the medium in interactions along their tracks. The mean energy imparted per unit volume is related to the total track length per unit volume and thus to fluence. This relation is treated in Section VI on absorbed dose in terms of fluences and interaction coefficients.

The vectorial fluence Φ is relevant to determinations of the net flow of ionizing particles through any area A of a given orientation, Fig. 6. It is also relevant to determinations of mean energy imparted and absorbed dose since the energy imparted ϵ can be defined in terms of the net flow of radiant energy into a body, Eq. (13). This will be shown at the end of this section.

In Fig. 6 the number of particles passing through the area element dA with directions of motion Ω is the same as that passing through the area $dA \cos \theta$ perpendicular to Ω . The number of particles (with directions of motion Ω) passing through per unit area perpendicular to Ω is given by the fluence differential with respect to Ω , Φ_{Ω} (obtained through integration of p_T over time and kinetic energy). Thus, the number of particles passing through dA is given by $\Phi_{\Omega} dA \cos \theta$,

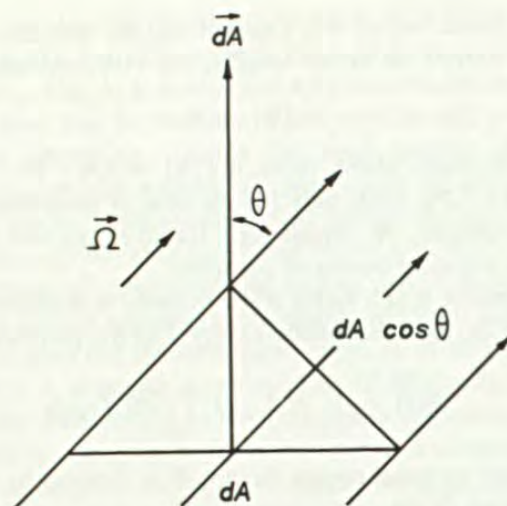


Fig. 6. The number of particles with directions of motion Ω passing through the area element dA is the same as that passing through the area $dA \cos \theta$ perpendicular to Ω . dA is so small that the fluence can be considered to be constant over it. [After Carlsson (1979).]

which can formally be written as a scalar product:

$$\Phi_{\Omega} dA \cos \theta = \Phi_{\Omega} \cdot d\mathbf{A} \quad (30)$$

where $d\mathbf{A}$ is the area vector perpendicular to dA and with its length $|d\mathbf{A}| = dA$. Note that when the angle between Ω and $d\mathbf{A}$ is acute, the scalar product is positive, while it is negative when the angle is obtuse. The transport of particles through the given area element dA is considered positive when the particles move in the same direction as the area vector $d\mathbf{A}$.

Integrating over all directions of motion, the *net number* of particles passing through dA is

$$\int_{4\pi} d\Omega \int \Phi_{\Omega} \cdot d\mathbf{A} = \int_{4\pi} d\Omega \int_{t_1}^{t_2} dt \int_0^{\infty} dT \Omega p_T \cdot d\mathbf{A} = d\mathbf{A} \cdot \Phi \quad (31)$$

The last equality in Eq. (31) follows since $d\mathbf{A}$ is a fixed direction independent of the directions of motion of the particles.

The *net number* of particles passing through a given oriented area per unit area $\Phi \cdot d\mathbf{A}/dA$ has been called the *plane fluence*, Φ_p (Whyte, 1959; Carlsson, 1979):

$$\Phi_p = \frac{\Phi \cdot d\mathbf{A}}{dA} \quad (32)$$

The vectorial fluence Φ can be represented as the sum of Φ_+ and Φ_- , where Φ_+ and Φ_- are the vectorial fluences of particles passing with acute ($\theta \leq \pi/2$) and obtuse ($\theta \geq \pi/2$) angles to $d\mathbf{A}$ (Fig. 6), respectively.

The *total* number of particles passing through the area element dA , independent of the directions of motion of the particles, is $(\Phi_+ - \Phi_-) \cdot d\mathbf{A}$ and has been called the *planar fluence* by Roesch and Attix (1968) (note the difference between "plane" and "planar" fluence).

An important relation, extensively used in the examples below (Section V,F), is obtained by combining Eqs. (26) and (31):

$$d\mathbf{A} \cdot \Phi = |d\mathbf{A}| |\Phi| \cos \theta \leq dA \Phi \quad (33)$$

where θ is the angle between the vectors $d\mathbf{A}$ and Φ . With multidirectional particles, the net number of particles passing through the area element dA is always less than the fluence times area dA . This is true also when the total number of particles is substituted for the net number, in which case equality holds for perpendicularly incident particles.

D. MEASUREMENTS OF FLUENCE

Fluence measurements can in principle be performed using a spherical

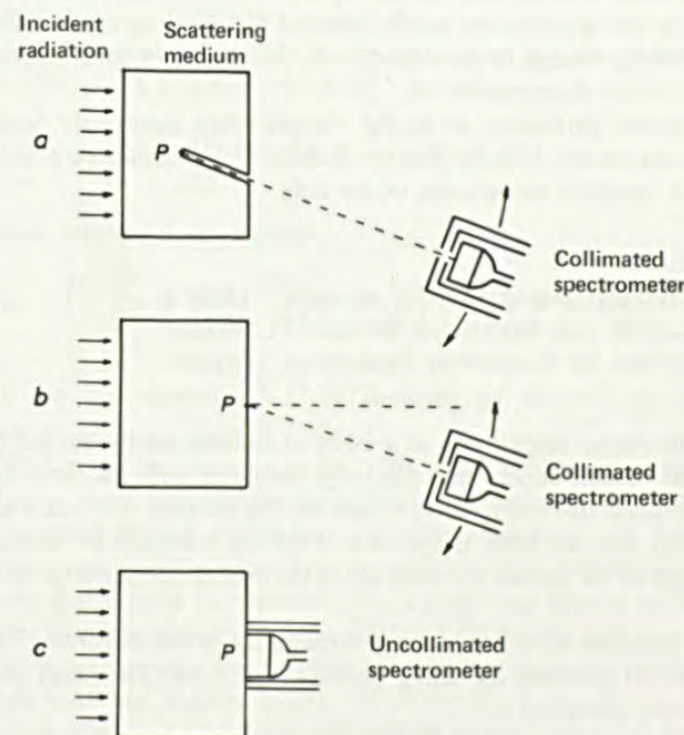


Fig. 7. Experimental arrangements for fluence measurements (a) in and (b) outside a scattering medium. The collimated detector and the channel [of zero depth in (b)] are moved to detect particles with different directions of motion. The design in (c) means that the plane fluence of particles escaping from the medium is determined (totally absorbing detector).

("isotropic") detector. In practice, this method seems to be difficult to execute. For instance, measuring the fluence at a point within a scattering medium, the dimensions of the sphere must be small compared to the ranges or mean free paths of the particles being measured. With penetrating particles such as photons, problems in finding a detector with good counting characteristics are apparent. However, the definition of fluence does not require the use of a spherical detector, which is evident from the definition in Eq. (19). Suitable arrangements (Hettinger and Starfelt, 1959; Bjärngård and Hettinger, 1961) for measurements inside and behind a scattering medium are shown in Fig. 7a, b.

In Fig. 7a a small channel in the scattering medium allows the particles passing the point of interest in a given direction to escape and to reach the counter. The collimated detector and the channel (of zero length in Fig. 7b) are moved to count particles with different directions of motion. The fluence is derived from an integration of the results obtained in the different directions. In Fig. 7c an uncollimated (totally absorbing) detector accepting the particles irrespective of their directions of motion means that the plane fluence of particles emerging from the medium (emitted into one half space) is determined. When the counter is operated in the spectrometer mode, spectral distributions of the fluence with respect to kinetic energy of the particles $\Phi_T = d\Phi/dT$, of great importance in dosimetry, can be determined.

Measurements performed as in Fig. 7c are often incorrectly presented as fluence measurements. In using data on fluences from the literature, it is advisable to check carefully the sources of the data.

E. MEAN ENERGY IMPARTED AND ABSORBED DOSE IN TERMS OF THE VECTORIAL ENERGY FLUENCE: CONNECTION TO RADIATION TRANSPORT THEORY

The mean energy imparted $\bar{\epsilon}$ to a body is determined by the net transport $\bar{R}_{in} - \bar{R}_{out}$ of radiant energy into the body corrected with the term $\bar{\Sigma Q}$, Eq. (13) (bars indicate that expectation values are considered). The net transport of radiant energy into the body is the sum of the net transport of radiant energy into it through all the surface elements dA of the area circumscribing the volume, Fig. 8.

The net transport of radiant energy through dA in the direction $-d\mathbf{A}$, i.e., into the body, is given by the scalar product $-\Psi \cdot d\mathbf{A}$. The mean energy imparted $\bar{\epsilon}$ is thus given by

$$\bar{\epsilon} = -\oint_{S(V)} \Psi \cdot d\mathbf{A} + \bar{\Sigma Q} \quad (34)$$

where $S(V)$ is the surface circumscribing the body of volume V .

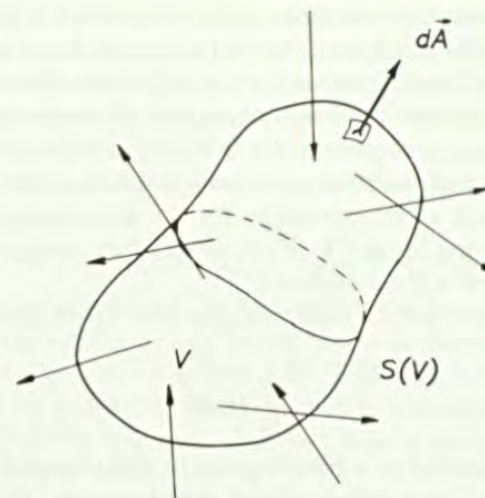


Fig. 8. A body of volume V is circumscribed by the area $S(V)$; $d\mathbf{A}$ is an area element of $S(V)$ with its area vector $d\mathbf{A}$ pointing out of the volume. Arrows indicate ionizing particles passing into and out of the body.

The surface integral in Eq. (34) can by application of Gauss's theorem (with the convention of $d\mathbf{A}$ pointing out of the volume)

$$\oint_{S(V)} \Psi \cdot d\mathbf{A} = \iiint_V \text{div } \Psi \, dV \quad (35)$$

be converted into a volume integral,

$$\bar{\epsilon} = -\iiint_V \text{div } \Psi \, dV + \bar{\Sigma Q} = \iiint_V \left(-\text{div } \Psi + \frac{d(\bar{\Sigma Q})}{dV} \right) dV \quad (36)$$

where in the last equality $\bar{\Sigma Q}$ is depicted as the volume integral of $d(\bar{\Sigma Q})/dV = \lim_{V \rightarrow 0} (\bar{\Sigma Q}/V)$.

According to the last equality in Eq. (36), the mean energy imparted gets a contribution at every point of the body. The formulation of $\bar{\epsilon}$ in Eq. (36) corresponds to the definition of ϵ in Eq. (11). The contribution to $\bar{\epsilon}$ at a point is the expectation value of the sum of the imparted energies in those basic processes occurring at that point (see Section III). It is of some interest to note that the assumptions that interactions occur at well-defined points, and that interaction forces are of zero range, are inherent in radiation transport theory (Roesch, 1968; Duderstadt and Martin, 1979).

When Eqs. (34) and (36) for $\bar{\epsilon}$ are applied in the expression for absorbed dose D , Eq. (14), one has

$$D = \frac{1}{\rho} \lim_{V \rightarrow 0} \left(\frac{\bar{\epsilon}}{V} \right) = -\frac{1}{\rho} \text{div } \Psi + \frac{1}{\rho} \frac{d(\bar{\Sigma Q})}{dV} \quad (37)$$

The expression for absorbed dose in Eq. (37) was derived by Rossi and Roesch (1962), Roesch (1968), and Spencer (1971) to exploit the connection between radiation dosimetry and radiation transport theory. This connection will be treated in more detail in Section VI. Local changes in the vectorial energy fluence Ψ as expressed by the scalar quantity $\text{div } \Psi$ result from interactions by ionizing particles and emission of such particles from radiation sources. The transport equation is formulated as an expression for $\text{div } \Phi_{r, a}$, which is easily transformed to an expression for $\text{div } \Psi$. First, some other aspects of the equations derived in this section will be explored.

Combining Eqs. (36) and (37) and changing to $m = \rho V$ as the variable of integration yields

$$\bar{\epsilon} = \int_m D dm \quad (38)$$

The mean energy imparted to a body equals the mass integral of the absorbed dose in it and has also been called integral absorbed dose. This relation can as well be derived from first principles by considering the fundamental connection between $\bar{\epsilon}$ and D expressed in the definition of D , Eq. (14).

At a point in a medium traversed by ionizing particles, the vectorial energy fluence Ψ may happen to be zero. For instance, one may think of two antiparallel beams of identical particles. It does not follow that $\text{div } \Psi$ equals zero. The following simplified, hypothetical example, Fig. 9, may clarify this point, which is easily misunderstood.

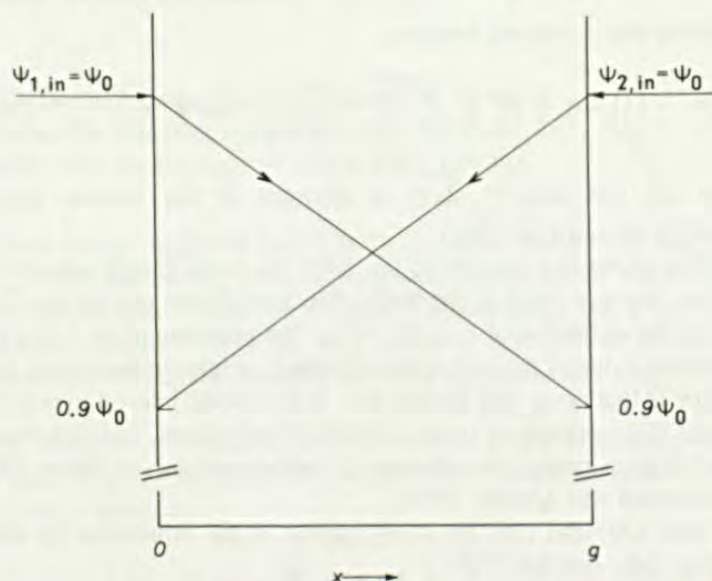


Fig. 9. Two broad antiparallel beams of charged ionizing particles are incident on a medium of thickness g . The incident energy fluences $\Psi_{1, in} = \Psi_{2, in} = \Psi_0$. The energy fluence in each beam decreases linearly in the medium.

Radiant energy is removed from the radiation field by means of interactions in the medium. It is assumed that after the interactions both the primary ionizing particles and the secondary radiations thereby created remain parallel to the incident beam.* Also assume that no nuclear or elementary particle transformations occur in the medium [$d(\sum Q)/dV = 0$].

The energy fluence in the two beams as a function of the depth x from the left boundary of the medium is

$$\Psi_1(x) = \Psi_0[1 - 0.1(x/g)], \quad \Psi_2(x) = \Psi_0[0.9 + 0.1(x/g)] \quad (39)$$

An expression for absorbed dose at any depth x can be derived by considering the decrease in the radiant energy of the beams in passing from x to $x + dx$ (beam 1) or from $x + dx$ to x (beam 2). This decrease, determined by the derivatives of $\Psi_1(x)$ and $\Psi_2(x)$, yields the energy imparted by the beams to the slab of thickness dx . In this case, absorbed dose is independent of depth and equals

$$D = 0.2\Psi_0/(g\rho) \quad (40)$$

with each beam contributing the same amount $0.1\Psi_0/(g\rho)$ to D . Although the energy fluence in each of these hypothetical beams decreases with depth, the absorbed dose from each single beam remains independent of depth. Physically, this situation approximately exists in the case of, e.g., protons. The energy fluence decreases as the energy of the particles decreases, but the fluence and stopping power may remain fairly constant over considerable depth intervals (cf. Section VI,C,4, Fig. 20).

The vectorial energy fluence $\Psi = \Psi_1 + \Psi_2$ can be written

$$\Psi = \Psi_x e_x + \Psi_y e_y + \Psi_z e_z \quad (41)$$

where Ψ_x , Ψ_y , and Ψ_z are the cartesian coordinates of the vector Ψ . The vectors Ψ_1 and Ψ_2 in Fig. 9 are parallel and antiparallel, respectively, to e_x and their coordinates are

$$\Psi_1 = \Psi_0[1 - 0.1(x/g), 0, 0] \quad (42)$$

$$\Psi_2 = -\Psi_0[0.9 + 0.1(x/g), 0, 0]$$

In coordinate form, the scalar quantity $\text{div } \Psi$ is defined from

$$\text{div } \Psi = \frac{\partial \Psi_x}{\partial x} + \frac{\partial \Psi_y}{\partial y} + \frac{\partial \Psi_z}{\partial z} \quad (43)$$

At $x = g/2$, $\Psi = \Psi_1 + \Psi_2$ is zero but, since $\text{div } \Psi = \text{div } \Psi_1 + \text{div } \Psi_2$, one has

$$D = -\frac{1}{\rho} \text{div } \Psi = 0.2 \frac{\Psi_0}{g\rho} \quad (44)$$

in accordance with the result in Eq. (40).

*The secondary radiations may alternatively be assumed to be in equilibrium. The energy fluence Ψ , Fig. 9, is then that of the primary particles.

F. EXAMPLES OF THE USEFULNESS OF THE VECTOR FORMALISM

Two illustrative examples will be given to demonstrate the power of the vector formalism in clarifying important quantities and problems in radiation dosimetry.

1. Determination of the Mean Energy Imparted to a Finite Body

In diagnostic radiology, it is of interest to determine the mean energy imparted to the patient during an examination for use as a crude basis for risk estimates. The mean energy imparted to the patient can be determined either as the mass integral of absorbed dose, Eq. (38), or as the net flow of radiant energy into the patient, Eq. (34), corrected with the term $\sum Q$ (taken here to be zero). In cases when photon fluence and absorbed dose vary rapidly with position in the body, the latter seems the simpler. However, one must be careful in exercising the idea, as shown in the following erroneous application from the literature, discussed by Carlsson (1963).

In the experiment to be discussed, the patient was simulated by a plane-parallel box of water. Ionization chambers were used to determine the exposure X at points all over the surfaces of the water box during irradiation.

Ionization chambers used to measure exposure X are, together with their walls, thin compared with the mean free paths of incident photons and respond, independently of their shape (Alm Carlsson and Carlsson, 1982), proportionally to the photon energy fluence Ψ at the measuring point. The relationship between X and Ψ is

$$\Psi = X \frac{\bar{W}}{e} \frac{1}{(\mu_{en}/\rho)_{air}} \quad (45)$$

Here, $\bar{W}$ is the mean energy imparted to air to form an ion pair, e the electronic charge, and $(\mu_{en}/\rho)_{air}$ a weighted mean of the mass energy absorption coefficient in air [derivations of weighted mean values of energy-dependent quantities for use in various applications are discussed by Greening (1981) and Alm Carlsson and Carlsson (1984); see also Section VII].

In the experiment, the mean energy imparted to the patient was calculated as the difference between the surface integral of Ψ_{in} of the incident photons and that of Ψ_{out} of escaping photons, Ψ_{in} and Ψ_{out} being derived from Eq. (45).

The incident photons are approximately parallel and perpendicular to the entrance surface such that $-\Psi_{in} \cdot d\mathbf{A} = \Psi_{in} dA$ ($d\mathbf{A}$ pointing out of the body following common convention). However, the photons scattered within, and escaping from, the body are multidirectional and $\Psi_{out} \cdot d\mathbf{A} < \Psi_{out} dA$ [cf. Eq. (33)]. Thus

$$\bar{\epsilon} = - \oint_{S(V)} \Psi_{in} \cdot d\mathbf{A} - \oint_{S(V)} \Psi_{out} \cdot d\mathbf{A} > \oint_{S(V)} \Psi_{in} dA - \oint_{S(V)} \Psi_{out} dA \quad (46)$$

1. THEORETICAL BASIS FOR DOSIMETRY

The escaping photons are nearly isotropic in the half space involved. The energy fluence differential with respect to the directions of motion $\Psi_{\Omega} = d\Psi/d\Omega$ takes on a constant value equal to $\Psi/2\pi$ in all directions of this half-space. Since $d\Omega = 2\pi \sin \theta d\theta$ defines the solid angle element occupied by directions Ω making an angle in the interval $d\theta$ around θ with $d\mathbf{A}$, one has [cf. Eq. (30)]

$$\Psi \cdot \frac{d\mathbf{A}}{dA} = \int_{\Omega=2\pi} d\Omega \Psi_{\Omega} \cos \theta = \Psi \int_0^{\pi/2} \sin \theta \cos \theta d\theta = \frac{\Psi}{2} \quad (47)$$

In this case with an isotropic energy fluence, the radiant energy passing through a fixed area element per unit area is only half of the energy fluence. Thus the mean energy imparted to the patient was seriously underestimated.

2. Variation of Fluence with Scattering

The fluence of particles incident on matter changes with depth due to absorption of the primary particles and creation of new secondary ionizing particles. Moreover, the fluence changes due to scattering of the particles, resulting in varying distributions of their directions of motion with depth. Here, we focus attention on the change of fluence due to the scattering of particles.

In Fig. 10, a broad uniform beam of particles is perpendicularly incident on a thin slab of a scattering medium. The incident particles are assumed neither to be absorbed nor to generate any new secondary ionizing particles, but just to undergo a change in direction and possibly energy. Also the scattering is assumed to occur in the forward direction only ($\theta < \pi/2$). The number of particles passing through equally large surface elements A at the upper and rear sides

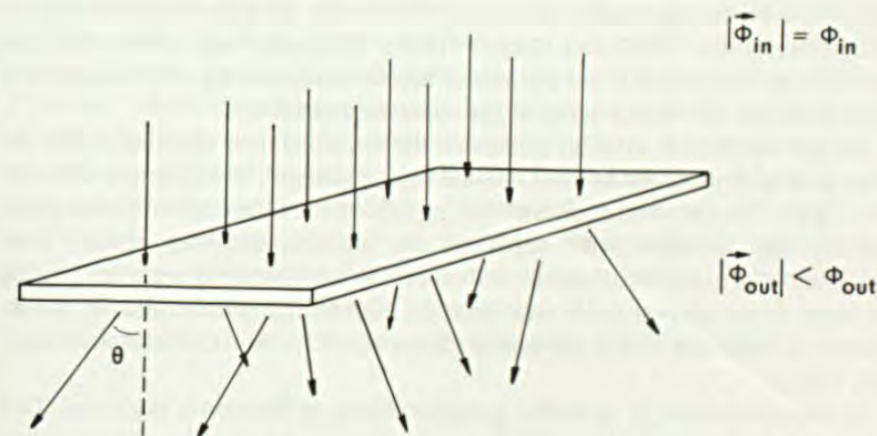


Fig. 10. A broad beam of particles is incident perpendicularly on a thin scattering slab; $|\vec{\Phi}_{in}| = \Phi_{in}$. The particles transmitted through the slab emerge from it distributed with respect to both azimuthal and polar angle, θ ; $|\vec{\Phi}_{out}| < \Phi_{out}$.

of the scattering slab is the same and since the emerging particles are multidirectional, one has

$$-\Phi_{in} \cdot \mathbf{A} = \Phi_{in} A = \Phi_{out} \cdot \mathbf{A} = |\Phi_{out}| A \cos \theta < \Phi_{out} A \quad (48)$$

Here, θ is the angle between the vectors Φ_{out} and $\mathbf{A}$ (θ should not be confused with the angle θ between the direction of motion of an individual particle and the normal to A in Fig. 10). The result in Eq. (48) derives from the basic Eq. (33).

When the incident particles are nonpolarized and the scattering medium is isotropic, a common situation, particles scattered at a given angle θ (Fig. 10) are uniformly distributed with respect to azimuthal angle. The vectorial fluence Φ_{out} is parallel to $\mathbf{A}$ and $\cos \theta = 1$. The last step in Eq. (48) follows by virtue of the fundamental relation, Eq. (26), independent of the value for $\cos \theta$.

If the directions of the particles in Fig. 10 were reversed, the fluence would be decreased by the resulting scattering. It can be generally concluded that the fluence is modified, either increased or decreased, by changes in the directions of motion of the particles. Another way to describe the increase in fluence in Fig. 10 is to consider the fluence as the total track length per unit volume of an infinitesimal volume element, independent of its shape. We choose the volume element as an infinitesimally thin slab at the rear of the scattering medium. Obviously the sum of the track lengths of particles passing in different directions through a plane slab exceeds the sum of the track lengths of those crossing it perpendicularly.

The increase in fluence with increasing depth in a scattering medium irradiated with a broad beam of perpendicularly incident electrons is demonstrated in Fig. 11, from calculations by Kessaris (1966), who used the moments method of Spencer (1955, 1959) and Roesch (1968). Electrons were assumed to lose their energy continuously and the Möller incoherent scattering cross section was used to derive the source term of the secondary electrons.

Berger and Seltzer (1982a) calculated the absorbed dose distribution for the same geometry using the Monte Carlo technique (Berger, 1963; Berger and Seltzer, 1969). The absorbed dose distribution reflects the fluence distribution modified by the variation with depth of the electron stopping power. Both distributions are, to a small extent, influenced by backscattered electrons passing the same depth interval more than once. In high-atomic-number media, the influence of backscattering is substantial (Spencer, 1959; Alm Carlsson and Carlsson, 1982).

In the calculations by Kessaris, bremsstrahlung production is neglected. Due to secondary electrons released by bremsstrahlung, as well as energy straggling by the primary electrons, the absorbed dose is not zero at $z/r_0 = 1$ in the calculations by Berger and Seltzer (Fig. 11).

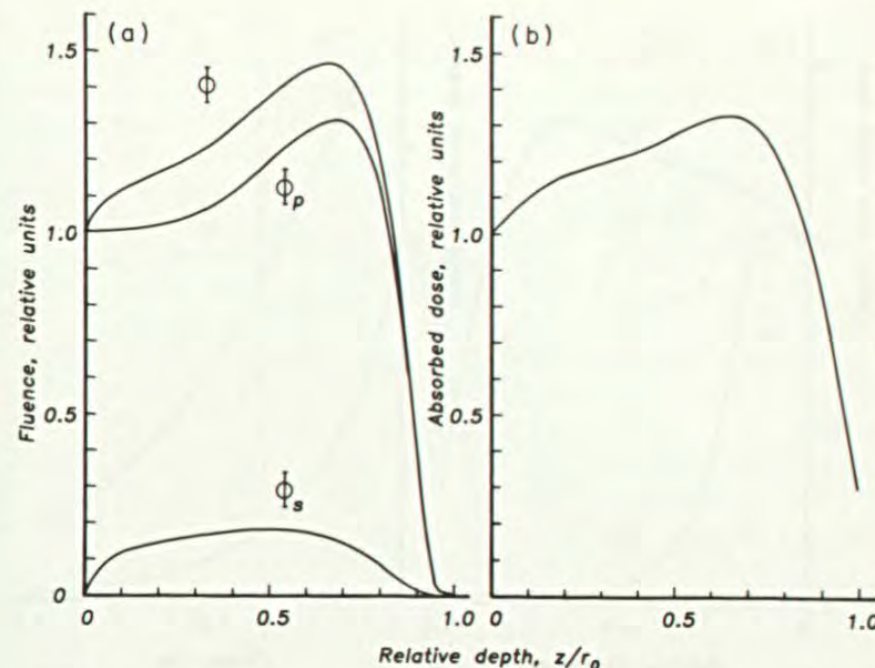


Fig. 11. The fluence of electrons (a) and the absorbed dose (b) as a function of depth in a semi-infinite water medium irradiated with a broad beam of perpendicularly incident 20-MeV electrons. The fluence Φ is the sum of the fluences of primary electrons Φ_p and secondary electrons Φ_s . The depth z is plotted as a fraction of the range r_0 calculated in the continuous slowing-down approximation (CSDA) of the incident electrons. [After Kessaris (1966) (a) and Berger and Seltzer (1982a) (b).]

An increase in fluence and absorbed dose with increasing depth can also be seen in media irradiated with broad beams of perpendicularly incident low-energy photons, see Fig. 12. The increase in fluence is due to incoherent (Compton) and coherent scattering, causing the photons to deviate from their initial directions of motion. (With photons of higher energies, the depth dose curves are influenced in a more complicated manner by the transport of both the photons and the secondary charged particles they set in motion.) In narrow-beam geometry, the scattering of particles away from the central axis is not compensated for by particles scattered in from more distant parts of the beam, and no corresponding increase in fluence occurs (Fig. 12).

The increase in fluence due to scattering (Fig. 10) is given by

$$\frac{\Phi_{out}}{\Phi_{in}} = \int_0^{\pi/2} \frac{N_\theta}{\cos \theta} d\theta / \int_0^{\pi/2} N_\theta d\theta \quad (49)$$

where $N_\theta d\theta$ is the number of particles emerging per unit area at an angle in

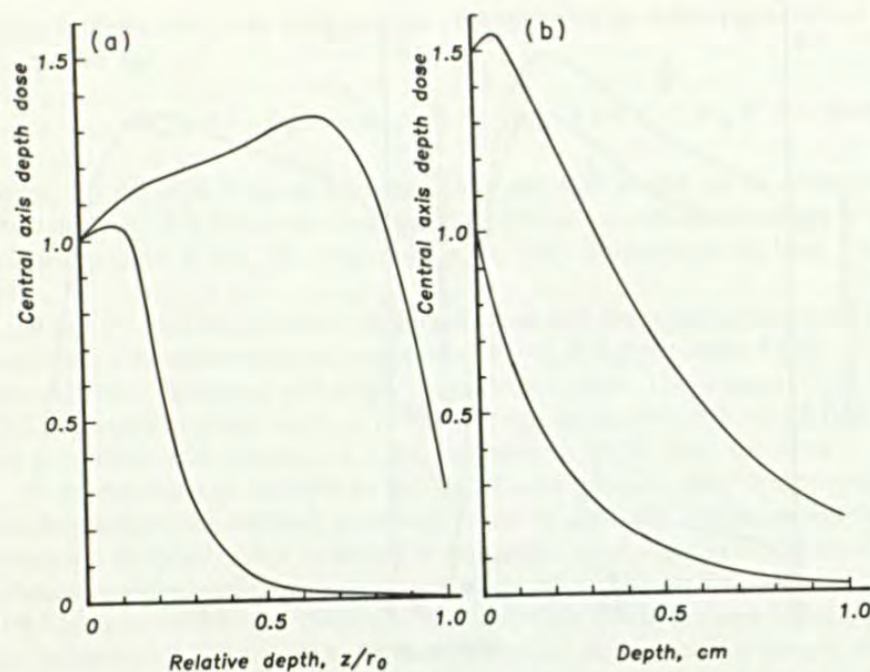


Fig. 12. Central axis depth doses with 20-MeV electrons (a) and x rays of HVT = 0.5 mm Cu (b) perpendicularly incident on semi-infinite water media. The lower curve was obtained in a narrow beam and the upper curve in a broad beam of incident radiation. [After Berger and Seltzer (1982a) and Carlsson (1963), respectively.]

the interval $d\theta$ around θ to the normal of the area. Each component $N_\theta d\theta$ of emerging particles yields a contribution to the fluence which is $N_\theta d\theta$ increased by the factor $1/\cos \theta$ (Whyte, 1959; Carlsson and Lidén, 1968; Brahme, 1982; Alm Carlsson and Carlsson, 1982).

Distributions N_θ of charged particles passing thin absorbers* are given in the multiple small-angle scattering theories [see, e.g., Zerby and Keller, (1967), for a review of various approaches to the treatment of multiple scattering problems] and have also been derived by means of Monte Carlo calculations, as illustrated in Fig. 13.

At large angles of emergence ($\theta > 45^\circ$) the gaussian scattering in the small-angle approximation (Eyges, 1948) yields unrealistic results mainly due to neglect of the random occurrence of large deflections in single scattering events. Also, N_θ does not decrease fast enough as θ approaches $\pi/2$ (Fig. 13), causing

*Distributions N_θ often refer to the number of particles escaping through the rear of a laterally infinite slab independent of position. Such distributions apply to the number of particles escaping per unit area when the slab is irradiated with a laterally infinite, homogeneous beam of incident particles.

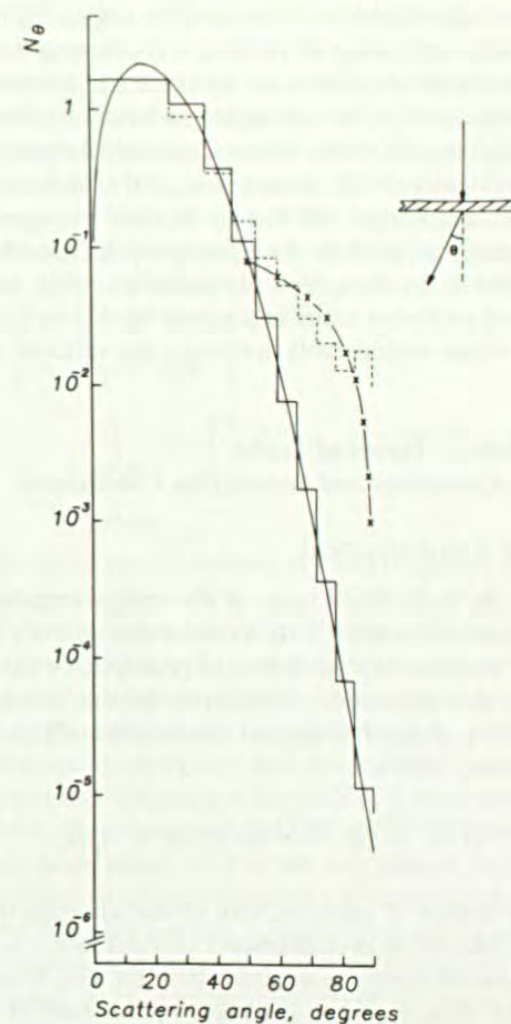


Fig. 13. The number of perpendicularly incident 2-MeV electrons transmitted through a 0.1-g/cm² foil of Al₂O₃ as a function of the angle θ of emergence. The solid line and histogram show analytical results obtained in the gaussian small-angle scattering approximation, and the dashed histogram results are from Monte Carlo calculations by Berger and Seltzer (1964). $\times$ — $\times$ is a function proportional to $\cos \theta$. More than 99% of the incident electrons are transmitted in both cases.

the integral for Φ_{out} , Eq. (49), to diverge. In reality, fluences of ionizing particles are finite, since particle ranges (or mean free paths) are finite and particles are absorbed.* For sufficiently large angles θ (a necessary condition being that

*Infinite fluences are not unknown but result from a mathematical abstraction of the real world. For instance, the fluence approaches infinity near a point source and at points in a pencil beam from a collimated point source emitting particles in one direction only.

$d/\cos \theta > R$; d = slab thickness, R = particle range), N_θ can be shown to decrease as $\cos \theta$ with increasing θ , yielding a converging integral for Φ_{out} .*

Statistical uncertainties inherent in the Monte Carlo calculations conceal to some extent the behavior of N_θ as approaching a function proportional to $\cos \theta$ as θ approaches $\pi/2$ (Fig. 13). Other Monte Carlo calculations of both backscattered (Mandour and Harder, 1975; Berger *et al.*, 1974) and transmitted particles (Mar, 1966; Seltzer and Berger, 1978) may be cited in support.

In Monte Carlo calculations of the fluence of particles penetrating a plane barrier, a cutoff value for θ , say, $\theta \leq 89.6^\circ$ (Leimdörfer, 1964), may be introduced to prevent scoring of excessive individual values for $N_\theta/\cos \theta$. The random occurrence of such values considerably increases the variance in the estimated value for Φ_{out} .

VI. Absorbed Dose in Terms of Scalar Radiometric Quantities and Interaction Coefficients

A. DEFINITION OF ABSORBED DOSE

Absorbed dose, the expectation value of the energy imparted per unit mass of an infinitesimal volume, is not strictly a measurable quantity (see Section III). On the other hand, its value at a point can, in principle, be calculated provided the radiation field and interaction coefficients of the ionizing particles are known. In terms of the scalar fluence and interaction coefficients the absorbed dose is (Alm Carlsson, 1981)

$$D = \int_{T_{cut}}^{\infty} \Phi_T \frac{\mu(T)}{\rho} \bar{\delta\epsilon}(T) dT + n_s \bar{\delta\epsilon}_s \quad (50)$$

where $\Phi_T dT$ is the fluence of particles with kinetic energies in the interval dT around T , $\mu(T)$ the interaction probability per unit distance, $\bar{\delta\epsilon}(T)$ the expectation value of the imparted energy in an interaction [Eq. (1)] of a particle of kinetic energy T , ρ the density of the medium, n_s the number of spontaneous nuclear transformations taking place per unit mass of the medium, and $\bar{\delta\epsilon}_s$ the expectation value of the imparted energy in such a process. The integration over kinetic energy extends down to the cutoff limit T_{cut} , below which the particle is no longer considered to be ionizing. The integration also includes a summation over all the types of ionizing particles present in the radiation field.

*The arguments to show this behavior of N_θ are similar to those inherent in the derivation of Lambert's cosine law dealing with the radiation (light) emitted from a thick medium containing a uniformly distributed isotropic source.

†Even charged particles are associated with a probability of interaction μ per unit length. Their mean free paths $1/\mu$, however, are generally so small that they are usually not considered.

The expression in Eq. (50) yields exactly the absorbed dose as defined in Eq. (14) and could serve as an alternative definition. It is the product of the expectation value of the number of basic processes taking place per unit mass and the expectation value of the imparted energy in each of these processes.

An expression for absorbed dose in terms of radiometric quantities and interaction coefficients can also be derived using transport theory, Eq. (37). The time-independent transport equation (valid when radiation sources are turned on for a finite time only and the radiation field is that integrated over this time interval) is formulated as an expression for the divergence of the differential vectorial fluence with respect to kinetic energy and direction of motion:

$$\begin{aligned} \text{div } \Phi_{T, \Omega} = & -\mu(T, \Omega) \Phi_{T, \Omega} \\ & + \int_{4\pi} d\Omega' \int_{T_{cut}}^{\infty} dT' \mu_{T, \Omega}(T', \Omega'; T, \Omega) \Phi_{T', \Omega'} \\ & + S_{T, \Omega} \end{aligned} \quad (51)$$

where $\mu(T, \Omega)$ is the interaction probability per unit length for a particle of kinetic energy T and direction of motion Ω , $\mu_{T, \Omega}(T', \Omega'; T, \Omega)$ is that of a particle of kinetic energy T' and direction of motion Ω' giving rise to a particle of kinetic energy T and direction of motion Ω per unit interval of these quantities, and S is the number of particles emitted per unit volume from radiation sources.

The divergence of the differential vectorial fluence $\Phi_{T, \Omega}$ yields the net transport of particles with kinetic energies T and directions of motion Ω out of an infinitesimal volume per unit volume and intervals of kinetic energy and direction of motion. The first term on the right-hand-side of Eq. (51) represents those particles in the incident beam which are lost per unit volume due to interactions in the infinitesimal volume. It is usually called the "outscattering" term since it yields the number of particles per unit volume which are "scattered" out of the intervals of kinetic energy and direction of motion considered (every particle is either energy degraded, or changes its direction of motion in an interaction, or both). The second term represents the particles which are created in the same volume in the interactions by other incident particles, and which add to those in the outgoing beam (particles created in an infinitesimal volume escape it without further interactions). This term is called the "inscattering" term, since it yields the number of particles per unit volume which are "scattered" into the intervals considered by particles which have their kinetic energies and directions of motion outside those intervals.

The lower integration limit for kinetic energy is usually taken to be equal to T in the inscattering term. Particles with kinetic energies $T' < T$ are not considered to be able to give rise to particles with kinetic energy T . Here, in Eq. (51), the lower value T_{cut} is used. In interactions in which rest-mass energy of nuclei

and elementary particles is released into radiant energy, it may well be that a particle appearing after the interaction has kinetic energy exceeding that of the interacting particle.

Since the divergence means a differentiation with respect to position only, multiplying the differential vectorial fluence by T gives [cf. Eq. (43)]

$$T \operatorname{div} \Phi_{T, \Omega} = \operatorname{div} T \Phi_{T, \Omega} = \operatorname{div} \Psi_{T, \Omega} \quad (52)$$

Finally, by integrating over T and Ω , $\operatorname{div} \Psi$ is obtained from

$$\begin{aligned} \operatorname{div} \Psi = & - \int_{4\pi} d\Omega \int_{T_{\text{cut}}}^{\infty} dT T \mu(T, \Omega) \Phi_{T, \Omega} \\ & + \int_{4\pi} d\Omega \int_{T_{\text{cut}}}^{\infty} dT T \int_{4\pi} d\Omega' \int_{T_{\text{cut}}}^{\infty} dT' \mu_{T, \Omega}(T', \Omega'; T, \Omega) \Phi_{T', \Omega'} \\ & + \int_{4\pi} d\Omega \int_{T_{\text{cut}}}^{\infty} dT T S_{T, \Omega} \end{aligned} \quad (53)$$

Apparently, $\operatorname{div} \Psi$ yields the net transport of radiant energy out of an infinitesimal volume per unit volume. In deriving an expression for $\operatorname{div} \Psi$ by means of the transport equation, the connection between dosimetry and radiation transport theory becomes quite clear. In addition, it is convincingly shown that $-\operatorname{div} \Psi$ is the scalar quantity which is related to absorbed dose, which was discussed to some extent in Section V.E.

Interaction probabilities depend on particle type (electrons, protons, photons, neutrons, etc.) as well as on kinetic energy. In nonisotropic media, they also depend on the direction of motion of the particles. In a mixed radiation field, a set of linked transport equations, one for each type of particle, has to be considered. The possibility of one type of particle creating a particle of another kind must be taken into account. The complexity in the set of equations increases rapidly in the presence of many types of particles (Rossi and Roesch, 1962).

By introducing the quantity $\delta\epsilon$ (the imparted energy in a basic process), it has been possible to reduce the complexity of the expression for absorbed dose. The values of $\delta\epsilon$ can be inferred from a knowledge of the differential interaction probabilities $\mu_{T, \Omega}(T, \Omega; T', \Omega')$ describing the possible outcomes of a certain type of interaction (including, when appropriate, subsequent deexcitations and neutralizations of the electronic structure). In the following, the simpler expression in Eq. (50) is preferred as a basis for introducing the basic dosimetric quantities.

Numerical evaluation of absorbed dose requires a detailed knowledge of the radiation field, the interaction cross sections, and the outcomes of the basic processes. Such detailed knowledge does not exist today, although, particularly within the field of microdosimetry, much effort is devoted to increasing the knowledge of the details of transport processes.

A first step in reducing the complexity of the expression for absorbed dose is to neglect the imparted energy $\delta\epsilon$ in (1) interactions of uncharged ionizing particles (e.g., photons and neutrons), (2) interactions involving nuclear or elementary particle reactions, and (c) spontaneous nuclear transformations. These three categories are neglected since their number is usually low compared to the number of interactions by charged particles. When, finally, the imparted energy in bremsstrahlung processes [bremsstrahlung may occur in the field of an atomic electron causing ionization, see, e.g., Seltzer and Berger (1982a)] is also neglected, the expression for absorbed dose reduces to

$$D = \int_{T_{\text{cut}}}^{\infty} \Phi_T \frac{S_{\text{col}}(T)}{\rho} k_{\text{col}}(T) dT \quad (54)$$

where S_{col} is the linear collision stopping power (ICRU, 1980) and k_{col} is the fraction of the kinetic energy lost by the interacting charged particle in a collision with atomic electrons that is converted into imparted energy in that collision (i.e., energy that does not reappear as radiant energy in the form of a δ particle, Auger or Coster-Kronig electrons, characteristic roentgen rays, or photons released in neutralization processes). The integration in Eq. (54) includes a summation over all kinds of *charged particles* (and charged particles only) present.*

Spencer (1971), in his fundamental work on cavity theory based on equations of transport theory, arrived at an expression for absorbed dose which mathematically takes quite another form but physically has the same significance as Eq. (54). His function $D(T)$ can be identified with $S_{\text{col}}(T)k_{\text{col}}(T)$.

B. PRACTICAL CALCULATIONS OF ABSORBED DOSE IN NONEQUILIBRIUM REQUIRING COMPLETE KNOWLEDGE OF THE FIELD OF CHARGED PARTICLES

Although calculations of absorbed dose can be reduced to evaluating Eq. (54), this is still a formidable task. When certain kinds of equilibrium prevail, it is possible to relax the requirement of complete knowledge of the field of charged particles. These will be treated in detail in the next section. For cases in which no kind of equilibrium exists, however, such as at locations close to interfaces between different media, the requirement of complete knowledge persists. In cavity theory, the Bragg-Gray detector represents an important case where no kind of equilibrium is established. A Bragg-Gray detector is, by definition, a detector so small that it does not appreciably disturb the fluence of charged particles in the surrounding medium. Ideally both it and the medium it replaces should be small compared to the ranges of all the charged particles present (Burch, 1955). The ideal Bragg-Gray detector approaches a "point" de-

*In cases where the primary particle loses so much energy that its residual kinetic energy falls below T_{cut} , S_{col} in Eq. (54) should be modified to include this residual kinetic energy.

tector (of infinitesimal size), in which the absorbed dose is calculated according to Eq. (54) with the fluence of charged particles identical to that in the surrounding medium.

Cavity theory has improved concurrently with improved information about the field of charged particles generated in slowing-down processes. Originally, slowing-down spectra or track-length distributions of electrons in photon-irradiated media were treated. Here, a brief outline of the improvements in evaluating true track-length distributions is given, aimed at elucidating Eq. (54).

Track-length distributions of charged particles can be regarded as solutions of the transport equation in an unbounded homogeneous medium containing a uniformly distributed source of charged particles (Roesch, 1968). The differential track length $y(T) dT$ is defined as the expectation value of the track length traversed by a charged particle (and its associated secondary charged particles) while having kinetic energy in the interval dT around T . It is proportional to the differential fluence with respect to kinetic energy in charged particle equilibrium (Section VI,C).

Calculated track-length distributions depend on the approximations employed in defining the transport process. A commonly employed approximation is the continuous slowing-down approximation (CSDA). The charged particles are assumed to lose their energy continuously at a rate given by the stopping power $S(T)$. In this model, no knock-on electrons (δ particles) or bremsstrahlung photons are created. The energy lost is immediately converted into imparted energy. The track-length distribution takes the simple form

$$y(T) dT = dT/S(T) \quad (55)$$

When fluences are calculated using the CSDA, absorbed dose is obtained from

$$D = \int_0^\infty \Phi_T \frac{S(T)}{\rho} dT \quad (56)$$

The continuous slowing-down approximation and the absorbed dose equation Eq. (56), are used, e.g., in the Bragg-Gray-Laurence cavity theory (Laurence, 1937; see also Burlin, 1968).

The continuous slowing-down approximation poorly describes electron transport. An electron may lose as much as half of its kinetic energy in a single atomic collision and the statistical fluctuations in the energy losses are appreciable. The generation and transport of high-energy secondary electrons (δ rays) was taken into account by Spencer and Fano (1954) using the Möller cross section, valid for collisions between free electrons. In this model, an energy limit Δ is chosen: energy losses above Δ are considered to result in the production of a δ particle, and those below Δ to represent imparted energy. The initial kinetic energy of the knock-on electron (δ ray) equals the energy lost by the primary electron; no

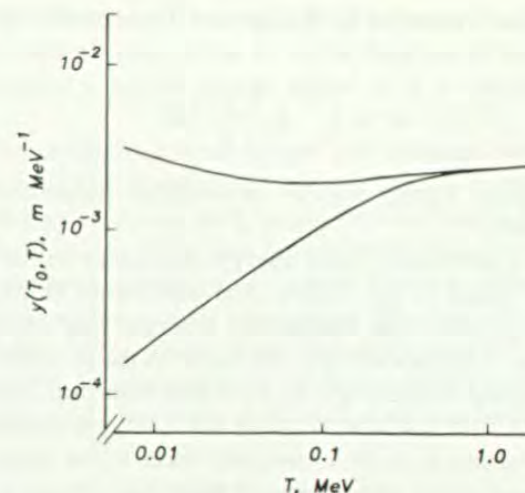


Fig. 14. Track-length distribution $y(T_0, T)$ in aluminum of an electron with initial kinetic energy $T_0 = 2$ MeV. Upper curve: total track-length distribution (primary and secondary electrons); lower curve: track-length distribution of primary electron only (approximately equal to that calculated in the continuous slowing-down approximation). [After Spencer and Fano (1954).]

energy is imparted to matter in the process. Figure 14 shows the track-length distribution, in aluminum, of a primary electron with initial kinetic energy $T_0 = 2$ MeV and its associated δ particles, calculated by Spencer and Fano.

In collisions between two electrons, the one emerging with the highest kinetic energy is, by convention, regarded as the primary electron. Consequently, a δ particle can acquire, at most, half of the kinetic energy of the primary electron. The resulting track-length distribution coincides with that of the primary electron alone for $T > T_0/2$ (Fig. 14). The contribution by δ particles to the complete track-length distribution is substantial, increasing with decreasing kinetic energy.

Figure 14 is interesting also in that it demonstrates the increase in knowledge about the true track-length distribution that is gained in going from the CSDA to the Spencer-Fano model. When bremsstrahlung losses are negligible as in Fig. 14, the track-length distribution of the primary electron (lower curve) comes close to that calculated in the CSDA. Bremsstrahlung causes energy losses to fluctuate widely and the track-length distribution of the primary electron to depart appreciably from the reciprocal stopping power, Eq. (55), when such losses are substantial. [Extensive tabulations of track-length distributions calculated in the Spencer-Fano model, covering a wide range of materials and initial kinetic energies of electrons from a few kilo-electron-volts to a few mega-electron-volts, have been published by McGinnies (1959).]

When fluences are calculated in the Spencer-Fano model, absorbed dose is obtained from

$$D = \int_{T_{\text{cut}}}^{\infty} \Phi_r \frac{L_{\Delta}(T)}{\rho} dT \quad (57)$$

where L_{Δ} is the linear energy transfer or restricted linear collision stopping power (ICRU, 1980).

In the Spencer-Fano model, more energy is assigned to the radiation field than in the model based on the CSDA. The equilibrium fluence of electrons (proportional to the track-length distribution) is larger, containing both primary electrons and δ rays. Correspondingly, the factor in the absorbed dose equation accounting for the imparted energy, L_{Δ}/ρ , is less than that, S/ρ , in the model based on the CSDA. It is essential to assure that energy is conserved, i.e., that the energy of interacting particles is assigned either to the radiation field or to imparted energy (and/or rest-mass energy of nuclei and elementary particles) but never to both. For instance, provided the electrons do not create any photons in slowing down, the imparted energy along the total track-length (including that of the δ rays when δ -ray production is considered) of an electron must equal the initial kinetic energy T_0 of this electron, independent of the model used in defining the transport process.

Track-length distributions calculated in the Spencer-Fano model have found important application in the extension of Bragg-Gray cavity theory by Spencer and Attix (1955). Due to the use of a more realistic track-length distribution, it yields results in much better agreement with experiments than the older Bragg-Gray-Laurence theory (see, e.g., Burlin, 1968).

In Spencer-Attix theory, the lower integration limit in Eq. (57) is Δ , the limit for energy losses resulting in δ -ray production, instead of the cutoff energy T_{cut} for ionizing particles. Since in the Spencer-Fano model the Möller cross section for collisions between free electrons is employed in calculating the generation of δ particles, accurate results can be expected only for values of Δ well above the binding energies of the majority of the atomic electrons. Thus, normally Δ exceeds T_{cut} . However, in the Spencer-Fano calculations no information about the track-length distribution of electrons with kinetic energies below Δ is given. The track-length distributions and correspondingly the fluences exhibit a cutoff at Δ such that the integrand in Eq. (57) vanishes below this value. Since energy losses below Δ are regarded as imparted energy, it is consistent to consider those electrons of the radiation field at trackends with kinetic energies below Δ to represent imparted energy as well. For reasons of energy conservation, this requires a change in the factor representing imparted energy. Spencer (1965) proved, by applying equations of radiation transport and the principle of energy conservation, that in changing the lower integration limit to Δ , the restricted linear collision stopping power for kinetic energies between Δ and 2Δ has to be

corrected to include the electron's residual kinetic energy for those collisions in which its kinetic energy drops below Δ . Only electrons in the energy interval Δ to 2Δ may acquire a kinetic energy below Δ in a collision with another electron.

Experimentally determined track-length distributions (McConnell *et al.*, 1965; Ritchie *et al.*, 1975; Tung *et al.*, 1977) reveal the occurrence of a large number of low-energy electrons, well below the cutoff limit Δ of the Spencer-Fano distributions (Fig. 15). Improved theoretical treatment of interaction cross sections for low-energy electrons (< 10 keV) is now developing, yielding cross sections that depend on the state of aggregation (Douthat, 1975; Ritchie *et al.*, 1975; Tung *et al.*, 1977, 1979; Tung and Ritchie, 1977; Ashley *et al.*, 1979a, b). In Fig. 15, experimental and theoretical track-length distributions in aluminum are compared. The low-energy end of the theoretical distribution is

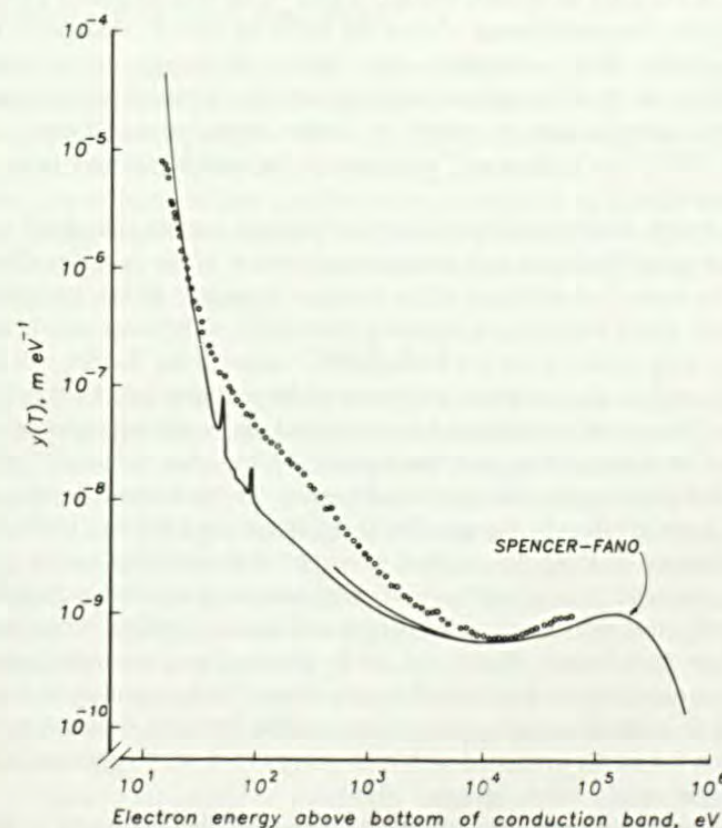


Fig. 15. Track-length distributions $y(T)$ of ^{198}Au beta rays in aluminum. Upper curve: experimental points; lower curves: Spencer-Fano theory for energies above 10 keV; below 10 keV, Spencer-Fano theory extended down to 0.5 keV and calculated values based on improved interaction cross sections for low-energy electrons. [After Tung and Ritchie (1977).]

based on improved cross sections which account for both the generation of electron-hole pairs and plasmon excitations of the valence electrons (being treated as an electron gas), as well as inner-shell excitations and ionizations including generation of Auger electrons in filling of inner-shell vacancies. Experimental and theoretical results agree well at high and very low energies. In the low- and intermediate-energy regions from 50 to 2000 eV the theoretical values are smaller than the experimental results by as much as a factor of four. Further studies are required before this discrepancy can be explained as a deficiency in either the experimental measurements or the theoretical calculations.

In going from the information given in Fig. 14 to that in Fig. 15, a large step has been taken toward complete knowledge about the radiation field. Ultimately, it might develop that absorbed dose can be evaluated from Eq. (54) where $S_{\text{col}} k_{\text{col}}$ evidently differs from L_{Δ} in Eq. (57). While energy losses below Δ may reappear in the form of radiant energy, a part of the energy losses above Δ is converted into imparted energy. Given the radiation source, changes in the assumed radiation field correspond (for reasons of energy conservation) to changes in the factor accounting for imparted energy. Detailed knowledge about low-energy electrons may be crucial for further improvement of cavity theory (Spencer, 1971) due to the rapid variation of the stopping power ratio at low electron energies.

Track-length distributions (equilibrium fluences) can be calculated without knowledge of the positions and directions of motion of the particles. They are obtained by numerical solutions of the transport equation. In bounded media or at interfaces where the particles penetrate into media of different atomic number or density, such solutions are not manageable. Values of the fluence will depend on position and on the directions of motion of the particles (see Section V,F,2). Here, the Monte Carlo technique has developed into a versatile instrument for calculation of radiation transport and absorbed dose. This technique applied to charged particle (in particular electron) transport has been worked out to a high degree of sophistication by Berger (1963) and Berger and Seltzer (1969). Extensive tabulations of three-dimensional absorbed dose distributions in a water phantom irradiated by a pencil beam of perpendicularly incident monoenergetic (1–60 MeV) electrons are given by Berger and Seltzer (1982a). Absorbed dose distributions from beams of any size can be obtained as a superposition of absorbed dose distributions from pencil beams. Monte Carlo calculations have also been used to evaluate stopping power ratios in the Spencer-Attix cavity theory formulation for media irradiated with high-energy (> 2 MeV) photons and electrons (Berger *et al.*, 1975; Nahum, 1978).

It is beyond the scope of this chapter to give a detailed treatment of methods for solving transport equations. The interested reader is referred to reviews by Fano *et al.* (1959), Roesch (1968), and O'Brien (1980).

C. RADIATION EQUILIBRIUM

Calculations of absorbed dose are considerably simplified if states of radiation equilibrium can be assumed. Such states may be established, e.g., within a radiation detector when this is equipped with a lining (wall) of an identical material. The requirement of a complete knowledge of the field of charged particles is then relaxed. The absorbed dose equations, Eqs. (56) and (57), which were derived as simplified expressions for absorbed dose, based on simplifying assumptions (although consistent with respect to energy conservation) about the transport process, come close to the truly valid equations when δ -particle equilibrium and partial δ -particle equilibrium, respectively, prevail. The concept of radiation equilibrium and expressions for absorbed dose, valid under various conditions of equilibrium, are treated in detail in this section.

1. Definition of Radiation Equilibrium

Although radiation equilibria in various forms such as charged particle equilibrium, δ -particle equilibrium, etc., are conditions which are often fulfilled at least approximately and utilized to facilitate calculations of absorbed dose, their general definition has been difficult to formulate.

Most authors have defined equilibrium as a condition of balance between the radiant energy of particles entering and leaving a specified volume (Roesch and Attix, 1968; Attix, 1979a, b). In cases when an infinitesimal volume (or a point) is considered, the definition is

$$\text{div } \Psi = 0 \quad (58)$$

Since absorbed dose is defined at a point (for an infinitesimal volume) and radiation equilibrium is utilized to facilitate determinations of absorbed dose, radiation equilibrium as defined for a point is of primary interest here.

The radiation field can be divided into different components, for instance, into the field of uncharged ionizing particles u , and charged ionizing particles c , respectively, such that $\Psi = \Psi_u + \Psi_c$. The condition defined in Eq. (58) is more specifically called complete radiation equilibrium. Charged particle equilibrium (CPE) exists when $\text{div } \Psi_c = 0$. Similarly, δ -particle and partial δ -particle equilibria exist when $\text{div } \Psi_{\delta} = 0$ and $\text{div } \Psi_{\delta, \Delta} = 0$, where Ψ_{δ} and $\Psi_{\delta, \Delta}$ are the vectorial energy fluences of δ particles (including Auger and Coster-Kronig electrons) and of δ particles generated with initial kinetic energies below Δ , respectively.

In particular, all kinds of equilibria exist at every point in an infinite homogeneous medium (homogeneous with respect to both atomic composition and density) containing a uniformly distributed radiation source. Then, the vectorial energy fluence Ψ and all its components are obviously constant in the

medium and, due to the constancy of the coordinates Ψ_x , Ψ_y , and Ψ_z , $\text{div } \Psi$, like $\text{div } \Psi_c$, equals zero [cf. Eq. (43)].

In the real world, media are finite and equilibrium conditions are rarely fulfilled completely. However, for one or more components of the radiation field, a state of equilibrium may be approximated to a high degree of accuracy. In a finite homogeneous medium, equilibrium prevails at a given point and for a particular kind of ionizing particle as soon as there exists a uniform production of such particles within all distances from the point up to the maximum range of the particles.* For instance, when a finite homogeneous medium is irradiated with an external beam of penetrating photons or neutrons, charged particle equilibrium is obtained approximately at points more than the maximum charged particle range from the boundaries of the medium. For a good approximation, the mean free paths of the photons or neutrons must be much larger than the maximum range of the charged particles they liberate.

Delta-particle equilibrium exists as soon as charged particle equilibrium is established. Delta-particle equilibrium also exists approximately at interior points of a finite homogeneous medium, externally irradiated with heavy charged particles having much longer ranges than the maximum range of the δ particles they liberate. Electrons liberate δ particles which may acquire up to half of their kinetic energy and δ -particle equilibrium is generally not attainable in media externally irradiated with electrons. A state of partial δ -particle equilibrium is possible in this case.

The condition $\text{div } \Psi = 0$ may occasionally be fulfilled at special points, along a line, or in a plane of media which are neither homogeneous nor exhibit a uniform liberation of particles. These cases are practically unimportant, and without real interest. Other definitions of radiation equilibrium have been given which implicitly exclude such points and restrict the concept of equilibrium to the energy balance existing in the particular case of a radiation source uniformly distributed within a homogeneous medium. Thus, ICRU (1980) states that "charged particle equilibrium exists if the energies, number and direction of the charged particles are constant throughout the volume of interest." Although not explicitly stated, the idea of a uniform liberation of charged particles within a homogeneous medium forms the basis of the definition. Only then can a volume be found such that the energies, number, and directions of the charged particles are constant throughout. Apparently with the same aim to restrict the concept of equilibrium to a particular case, Rossi and Roesch (1962) and Roesch (1968) defined charged particle equilibrium as $\text{div } \Psi_c = 0$, adding the requirement that $\text{div } \Psi_c$ vanishes throughout some nonzero volume. This requirement

*The requirement of a homogeneous density throughout the medium can, by virtue of Fano's theorem (Section VI,D), be relaxed if the liberation of particles is uniform with respect to the number of particles liberated per unit mass.

means that Ψ_c should be constant within this nonzero volume, i.e., coincides with the definition by ICRU (1980).

The ambiguities in defining radiation equilibrium at a point can be resolved by using Eq. (58). The particular case of radiation equilibrium existing when particles are uniformly liberated within a homogeneous medium would then be regarded as being the only one of any practical importance.

Although the uniform liberation of particles within a homogeneous medium represents an idealized situation, it is a useful exercise to derive expressions for absorbed dose that are valid in the ideal cases. It increases understanding of various dosimetric problems which can be analyzed in terms of equilibrium and departures from equilibrium. Besides derivations of expressions for absorbed dose in ideal cases of equilibrium, some practical implications will be discussed.

In the following sections, classification of ionizing particles as primary and secondary particles is essential. In a proper sense, primary particles are those emitted from radiation sources, e.g., radioactive nuclei and particle accelerators; secondary particles are created in the interactions by primary particles. Here, to facilitate the analysis, definitions of primary and secondary particles vary with the type of equilibrium (dosimetric problem) considered. The definitions are clearly stated in each section.

2. Complete Radiation Equilibrium

According to the proposed general definition of radiation equilibrium, $\text{div } \Psi = 0$, one has [from Eq. (37)]

$$D = \frac{1}{\rho} \frac{d(\sum Q)}{dV} \quad (59)$$

The condition of complete radiation equilibrium at a point in a medium requires the existence of a radiation source at that point. In vacuum, $\text{div } \Psi \equiv 0$ and complete radiation equilibrium exists but absorbed dose is zero. In a medium traversed by ionizing particles, interactions occur and absorbed dose takes on positive values. The $\text{div } \Psi$ cannot vanish, equilibrium cannot exist, unless $d(\sum Q)/dV > 0$. A case which immediately suggests itself, and on which attention is focused here, is that of a medium in which radioactive nuclei are distributed.

In the general case, both the distribution of radioactive nuclei and the transport of ionizing particles in the medium must be known to determine absorbed dose at a particular point. However, by assuming that $\text{div } \Psi = 0$ due to a uniform liberation of primary ionizing particles in a homogeneous medium, an expression for absorbed dose can be derived which does not require knowledge about the transport of the ionizing particles. This case is now treated, starting with a definition of primary and secondary particles.

Primary particles are ionizing particles emitted from radiation sources (radioactive and excited nuclei). Secondary particles are those created in interactions with atomic electrons and nuclei. If radioactive and excited nuclei are thereby formed, these emit primary particles upon their subsequent decay or de-excitation.

A primary particle emitted within an infinite medium is, together with all the generations of its secondary particles, completely stopped in it. The expectation value of the energy imparted to the medium, $\bar{\epsilon}_{T,p}$, from a primary particle with initial kinetic energy T and its associated secondary particles is, from Eq. (13) with $R_{in} = T$ and $R_{out} = 0$,

$$\bar{\epsilon}_{T,p} = T + \sum_n \bar{Q}_n \quad (60)$$

When the primary particles are emitted uniformly (i.e., with the same distributions of energies and directions of motion at all points) within a medium

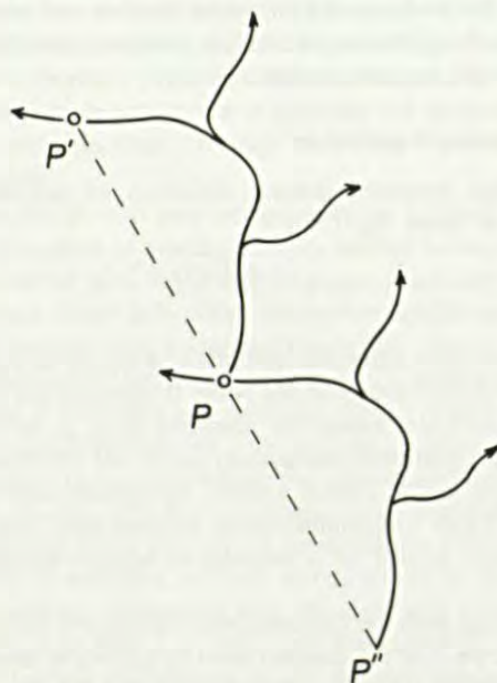


Fig. 16. Illustration of the reciprocity relation. Solid lines depict the tracks of primary particles and their secondary particles. For each primary particle emitted from P giving rise to a particle (primary or secondary) passing an infinitesimal volume at P' there exists on the average a corresponding particle emitted from P' giving rise to an identical particle traversal of an infinitesimal volume at P .

homogeneous with respect to both density and atomic composition a reciprocity relation, illustrated in Fig. 16, holds such that

$$d\epsilon(P \rightarrow P')/dV_{P'} = d\epsilon(P'' \rightarrow P)/dV_P \quad (61)$$

where $d\epsilon(P \rightarrow P')/dV$ is the expectation value of the energy imparted per unit volume at point P' per primary particle emitted from point P and P'' is a point at the same distance from P as P' but in the opposite direction. The reciprocity relation is true for nonisotropic emission of the primary particles as well as for nonisotropic media.

The expectation value of the energy imparted $d\epsilon$ to an infinitesimal volume element dV_P , around P , from interactions by primary and secondary particles, is then given by

$$\begin{aligned} d\epsilon &= \int_{P''} \frac{dS}{dV} dV_{P''} \frac{d\epsilon(P'' \rightarrow P)}{dV_P} dV_P \\ &= \int_{P'} \frac{dS}{dV} dV_P \frac{d\epsilon(P \rightarrow P')}{dV_{P'}} dV_{P'} = \frac{dS}{dV} dV_P \bar{\epsilon}_{T,p} \end{aligned} \quad (62)$$

Here, dS/dV is the number of primary particles emitted per unit volume, assumed to be constant throughout the homogeneous medium.

It is interesting to note that the reciprocity relation can also be used to demonstrate that

$$\Phi_T = (dS/dV)y(T) \quad (63)$$

i.e., in equilibrium the differential fluence (track length per unit volume) Φ_T is proportional to the differential track length $y(T)$ of the primary particles and the associated secondary particles. Equation (63) follows from Eqs. (61) and (62) when the total track length per unit volume of particles with kinetic energies in the interval dT around T is substituted for $d\epsilon/dV$ and $y(T) dT$ is substituted for $\bar{\epsilon}_{T,p}$.

Finally, the absorbed dose imparted in the interactions by primary and secondary particles is given by

$$D = (1/\rho)(dS/dV)\bar{\epsilon}_{T,p} \quad (64)$$

The double bar used with ϵ_T means that $\bar{\epsilon}_T$ in Eq. (62) is averaged with respect to the distribution of kinetic energies of the emitted primary particles.

In applications of Eq. (64), neutrinos emitted from β -decaying radioactive nuclei are not considered due to their weak interaction with matter.* Complete

*Note that neutrinos cannot, however, be disregarded in derivations of the energy imparted according to the definition, Eqs. (11) or (13). Since the kinetic energies of neutrinos are contained in the Q values of decaying nuclei, they must also be included in $\sum_i T_{a,i}$, Eq. (1), and R_{out} , Eq. (13). If not, kinetic energies of neutrinos would form part of the energy imparted.

radiation equilibrium (excluding neutrinos) is obtained approximately at interior points of a homogeneous medium containing a uniformly distributed radioactive source within distances from the points equal to at least a few mean free paths of the most penetrating photons (Roesch and Attix, 1968).

The absorbed dose equation, Eq. (64), is basic to all the kinds of equilibria treated in the following sections. In equilibrium, absorbed dose imparted in the interactions by primary and secondary particles equals the expectation value of the number of primary particles emitted per unit mass times the expectation value of the imparted energy in the interactions by a primary particle and all its secondary particles during complete slowing down. The explicit form of the absorbed dose equations derived will vary, however, since the classification of primary and secondary particles changes with the type of equilibrium considered. The imparted energy in those basic processes (e.g., decays and deexcitations of nuclei) in which the primary particles are emitted is neglected in Eq. (64).

3. Charged Particle Equilibrium; Kerma and Collision Kerma

Charged particle equilibrium (CPE) is the state of equilibrium which has gained most attention in the past. It is of particular importance when finite homogeneous media are irradiated externally with uncharged ionizing particles such as photons and neutrons. If the mean free paths ($1/\mu$) of the uncharged particles are much larger than the maximum ranges R of the charged particles they liberate ($\mu R \ll 1$), approximate CPE is obtained for points at depths exceeding the maximum ranges of the charged particles. In judging the degree of CPE obtained, those secondary uncharged particles created in interactions in the medium must be taken into account.

In CPE, interest is focused on the charged ionizing particles. Interactions by uncharged ionizing particles are considered as sources of primary charged particles. Like the condition of complete radiation equilibrium, CPE cannot be established (except in vacuum) at a point without the existence of a source of primary charged particles at that point. Charged particle equilibrium cannot exist in a medium irradiated with an external beam of charged particles.

Charged ionizing particles are classified as primary and secondary charged particles in the following way: * *primary charged particles* are charged ionizing particles emitted in spontaneous nuclear transformations or created in interactions of uncharged particles, and *secondary charged particles* are those created in interactions of charged particles, Fig. 17.

*Since CPE cannot exist with an external beam of charged particles, accelerator-produced charged particles are not considered in the definition.

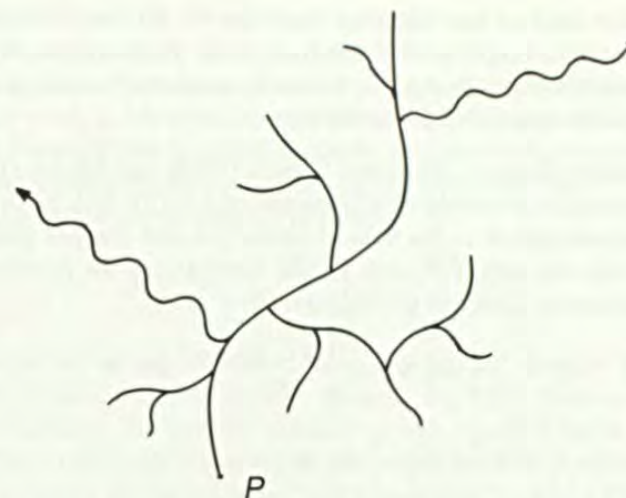


Fig. 17. A primary charged particle emitted at P slows down in the medium, generating new ionizing particles, both secondary charged particles and uncharged particles. The tracks of the primary charged particle and its associated secondary charged particles are illustrated as a cascade of solid lines; uncharged particles are shown as waves. Uncharged ionizing particles create new primary charged particles on their subsequent interactions, usually far away from the charged particle cascade. [After Alm Carlsson (1981).]

The expectation value of the imparted energy $\bar{\epsilon}_{T,c} = (\sum \delta\epsilon)_c$ in the interactions of a primary charged particle with initial kinetic energy T and its associated secondary charged particles (along the cascade of solid lines in Fig. 17) is

$$\bar{\epsilon}_{T,c} = \left(T - \sum_j T_{u,j} + \sum_j Q_j \right)_c \quad (65)$$

where $\sum_j T_{u,j}$ is the sum of the kinetic or quantum energies of all those uncharged ionizing particles created in the interactions by the primary charged particle and its associated secondary charged particles and $\sum_j Q_j$ is the total release of rest-mass energy of nuclei and elementary particles in the same interactions.

When CPE exists due to a uniform liberation of primary charged particles in a homogeneous medium, absorbed dose imparted in interactions of the charged particles is, from Eq. (64)

$$D = n_c \bar{\epsilon}_{T,c} \quad (66)$$

where n_c is the number (expectation value) of primary charged particles emitted per unit mass in the medium and $\bar{\epsilon}_{T,c}$ is given in Eq. (65). The imparted energy resulting from interactions by uncharged ionizing particles and spontaneous nuclear transformations is neglected.

Equation (66) implies that absorbed dose can be derived without detailed knowledge about the charged particle transport in the medium. Some knowledge about the properties of the medium is, however, needed since the value of $\bar{\epsilon}_{T,c}$, Eq. (65), depends on its atomic composition.

a. The Quantity Kerma. Rossi and Roesch (1962) and Roesch (1968) derived an expression for absorbed dose by means of Eq. (37) and the two-particle transport equation applied to the field of uncharged and charged particles. By separating Ψ into the sum of Ψ_u and Ψ_c and disregarding the presence or formation of radioactive nuclei in the medium, they get

$$\begin{aligned} D &= -\frac{1}{\rho} \operatorname{div} \Psi_u + \frac{1}{\rho} \frac{d(\sum Q)}{dV} - \frac{1}{\rho} \operatorname{div} \Psi_c \\ &= K - B - \frac{1}{\rho} \operatorname{div} \Psi_c \end{aligned} \quad (67)$$

Since the presence and formation of radioactive nuclei are neglected, only interactions by ionizing particles contribute to the term $(1/\rho)(d(\sum Q)/dV)$ in Eq. (67). The quantities K and $-B$ contain the contributions from uncharged and charged particle interactions, respectively, to $(-\operatorname{div} \Psi_u + d(\sum Q)/dV)$ in the intermediate equality.

The quantity K is called kerma. Kerma has been defined (ICRU, 1980) as (and is an acronym for) the sum of the initial kinetic energies of all those primary charged particles released by uncharged particles per unit mass. Strictly, K in Eq. (67) also contains the sum of the imparted energies resulting from the interactions by the uncharged particles per unit mass.

The quantity B , which has not been given any particular name, is the sum of the kinetic energies of all those uncharged ionizing particles which are generated per unit mass in the interactions by charged particles [contributing to $(1/\rho) \operatorname{div} \Psi_u$, cf. the second "inscattering" term of Eq. (53)] minus the net release per unit mass of rest-mass energy of nuclei and elementary particles in the same interactions.

In CPE, $\operatorname{div} \Psi_c = 0$ and $D = K - B$. This relation is valid without presupposing a uniform liberation of charged particles in a homogeneous medium.

In terms of radiometric quantities and interaction coefficients, kerma can be written

$$K = \int_{T_{\min}}^{\infty} \frac{\mu_{tr}}{\rho} \Psi_{T,u} dT \quad (68)$$

where μ_{tr}/ρ is the mass energy transfer coefficient (ICRU, 1980) and $\Psi_{T,u} dT$ is the energy fluence of the uncharged ionizing particles with kinetic or quantum energies in the interval dT around T .

The "kerma factor" k_f , commonly employed in neutron physics (Caswell *et al.*, 1982), is defined as the product of T and μ_{tr}/ρ . The integrand in Eq. (68) is then the product of the kerma factor and the differential fluence.

In CPE as existing when the primary charged particles are uniformly liberated within a homogeneous medium, B can be identified with the second term in the following expression derived from Eqs. (65) and (66) with $K = n_c \bar{T}_c$ (disregarding the formation of radioactive nuclei):

$$D = K - n_c \left(\sum_j \bar{T}_{u,j} - \sum_j \bar{Q}_j \right) \quad (69)$$

b. Collision Kerma or Absorbed Dose in Charged Particle Equilibrium. When CPE prevails, absorbed dose is given by Eq. (69). Even in cases when CPE is not fulfilled, the quantity defined by this equation exists and takes a definite value at each point traversed by uncharged ionizing particles. It may be convenient to assign the special name "collision kerma" (Attix, 1979a, b) to this quantity although the word "collision" is somewhat ambiguous since all interactions can be characterized as "collisions." Collision kerma has also been called "absorbed dose under conditions of CPE, D_{eq} " (Alm Carlsson, 1981).

By defining the mass energy absorption coefficient μ_{en}/ρ as [from Eqs. (65) and (66)]

$$\frac{\mu_{en}}{\rho} = \frac{\mu_{tr}}{\rho} \frac{\bar{\epsilon}_{T,c}}{\bar{T}_c} = \frac{\mu_{tr}}{\rho} \left[1 - \frac{(\sum_j \bar{T}_{u,j} - \sum_j \bar{Q}_j)_c}{\bar{T}_c} \right] \quad (70)$$

collision kerma K_c can be expressed in a way analogous to that of kerma in Eq. (68):

$$K_c = \int_{T_{\min}}^{\infty} \frac{\mu_{en}}{\rho} \Psi_{T,u} dT \quad (71)$$

The clear definition of the mass energy absorption coefficient has not been without problem. ICRU (1980) defines it as the product of μ_{tr}/ρ and $(1 - g)$, where g is that part of the initial kinetic energies of the liberated charged particles which reappears in the form of bremsstrahlung only. The significance of the mass energy absorption coefficient as being related to collision kerma is not discussed and may not have been considered as the basis for the definition. In tabulated values of mass energy absorption coefficients for photons (Hubbell, 1969, 1977, 1982), positron annihilation in flight is also considered* as contributing to the difference between μ_{en}/ρ and μ_{tr}/ρ . Annihilation in flight

*Annihilations at rest do not influence the value of μ_{en}/ρ since the positron loses no kinetic energy to the annihilation photons. In Eq. (70), the energies $= 2m_0c^2$ of the annihilation photons, contained in $\sum_j \bar{T}_{u,j}$, are in this case canceled by the release $= 2m_0c^2$ of rest-mass energy when a positron annihilates with a negatron, contained in $\sum_j \bar{Q}_j$.

reduces the value of μ_{en}/ρ by 2 to 3% at photon energies of 10 MeV in heavy elements. Its significance increases with increasing photon energy due to the increased relative probability of pair production at higher photon energies. The energies of characteristic x rays are not excluded from μ_{en}/ρ , however. This is a quantitative approximation rather than a matter of definition. The definition was implicitly stated by Hubbell (1969, p. 15) in the form of a flow diagram describing all pathways of flow of energy back to the photon radiation field.

Some bremsstrahlung photons and characteristic x rays generated by a charged particle have such low energies that their mean free paths are small compared to the range of the charged particle initially emitted. When CPE exists, these low-energy photons are also in a state of equilibrium such that they can be disregarded, i.e., included in the μ_{en}/ρ values. However, for reasons of energy conservation, they must be disregarded as part of the photon radiation field when K_c is determined from Eq. (71). It is consistent with this approach to include in the kerma the energies of low-energy characteristic x rays emitted upon photon interactions.

Equation (69) for absorbed dose in CPE is strictly valid when the imparted energies in the interactions by uncharged ionizing particles can be neglected. This restriction can easily be avoided by including the $\bar{\delta\epsilon}$ of the uncharged particle interaction itself into the kerma.* In fact, in using tabulated μ_{tr}/ρ values for calculations of kerma in photon fields, this is usually done. The energy transferred to kinetic energy of charged particles is derived as the difference between the photon energy and the expectation value of the energies of all secondary photons created in (or subsequent to) the photon interaction [including where appropriate the Q value, Eq. (1), of the photon interaction]. This treatment is in accordance with definitions of μ_{tr}/ρ given by ICRU (1980). It is also consistent with the quantity K as strictly defined from Eq. (67).

c. Relation between Collision Kerma and Absorbed Dose in Transient Equilibrium. Since neutrons liberate heavy charged particles but photons generate electrons with long path lengths, CPE ($\mu R \ll 1$) is more readily fulfilled for neutrons than for photons. The product μR increases with increasing energy; at 10 MeV in water it is about 0.01 for neutrons and about 0.1 for photons. The difference between absorbed dose D and collision kerma K_c is there-

*Including $\bar{\delta\epsilon}$ of the photon interaction into collision kerma means that μ_{en}/ρ , Eq. (70), must be redefined:

$$\frac{\mu_{en}}{\rho} = \frac{\mu}{\rho} \frac{\bar{\delta\epsilon}}{h\nu} + \frac{\mu}{\rho} \frac{\sum_i T_{e,i}}{h\nu} \frac{\bar{\epsilon}_{T,c}}{T_c}$$

where μ/ρ is the total mass attenuation coefficient, and $\sum_i T_{e,i}$ the sum of the initial kinetic energies of all ionizing charged particles liberated by the photon's interaction. For photon energies near an absorption edge, $(\mu/\rho)(\bar{\delta\epsilon}/h\nu)$ may contribute significantly to μ_{en}/ρ (Charlton *et al.*, 1983).

fore of particular concern for high-energy photon beams, and is treated in the following. Many of the arguments are also applicable to neutron beams.

Roesch (1958) was the first to work out a method for calculating the quotient between absorbed dose and kerma based on calculations of the transport of secondary electrons. In fact, the concept of kerma has its origin in this work. Bremsstrahlung losses were, however, neglected. When such losses are considered, the same type of calculations yield basically the quotient between absorbed dose and collision kerma (Attix, 1979a, b).

Consider a semi-infinite homogeneous object irradiated with a narrow beam of perpendicularly incident monoenergetic photons. Buildup of secondary photons along the central axis is neglected. On the other hand, it is assumed that the beam is large enough to establish a balance laterally between in- and outscattering of electrons at this axis.

At depth z on the central axis, absorbed dose is imparted by primary electrons (and their associated secondary electrons) liberated at varying depths z' such that $z - R < z' < z + R$; R being the maximum range of the electrons. Furthermore, one primary electron liberated at depth z' imparts on the average (together with its associated secondary electrons) the energy $d\epsilon(z - z')$ in a layer of thickness $d(z - z')$ at the distance $(z - z')$ from z' . The integral over the distribution $d\epsilon(z - z')/d(z - z')$ equals

$$\int_{-R}^R \frac{d\epsilon(z - z')}{d(z - z')} d(z - z') = \bar{\epsilon}_{T,c} \quad (72)$$

In the geometry considered and when $n_c(z')$ primary electrons are liberated per unit mass at depth z' , absorbed dose $D(z)$ at depth z ($z \geq R$) on the central axis is given by

$$D(z) = \int_{z-R}^{z+R} n_c(z') \frac{d\epsilon(z - z')}{d(z - z')} dz' = n_c(z) \int_{-R}^R e^{\mu z''} \frac{d\epsilon(z'')}{dz''} dz'' \quad (73)$$

where $z'' = z - z'$.

The last equality in Eq. (73) derives from assuming an exponential decrease with increasing depth of the number of liberated primary electrons per unit mass. This is the case when buildup of secondary photons along the central axis can be neglected.

Collision kerma $K_c(z)$ at depth z is the product of $n_c(z)$ and $\bar{\epsilon}_{T,c}$. Thus, the quotient between absorbed dose and collision kerma takes a value which is independent of depth and *transient equilibrium* is said to exist.

Provided $\mu R \ll 1$, the factor $e^{\mu z''}$ in the integrand of Eq. (73) can be approximated by $e^{\mu z''} \approx 1 + \mu z''$ and $D(z)$ reduces to

$$D(z) = K_c(z)[1 + \mu x], \quad z \geq R \quad (74)$$

where x is the first moment of the frequency function $(d\epsilon(z'')/dz'')/\bar{\epsilon}_{T,c}$:

$$x = \frac{1}{\bar{\epsilon}_{T,c}} \int_{-R}^R z'' \frac{d\epsilon(z'')}{dz''} dz'' \quad (75)$$

High-energy photons liberate electrons which are directed mainly in the direction of the incident photon such that $d\epsilon(z'')/dz''$ is large for positive values of z'' and x becomes positive. Values of absorbed dose exceed values of collision kerma. With isotropic emission of the primary electrons, x equals zero and absorbed dose equals collision kerma. In high-atomic-number media, the secondary electrons are highly scattered (Rutherford scattering) resulting in a distribution $d\epsilon(z'')/dz''$ that is more symmetric than that in low-atomic-number

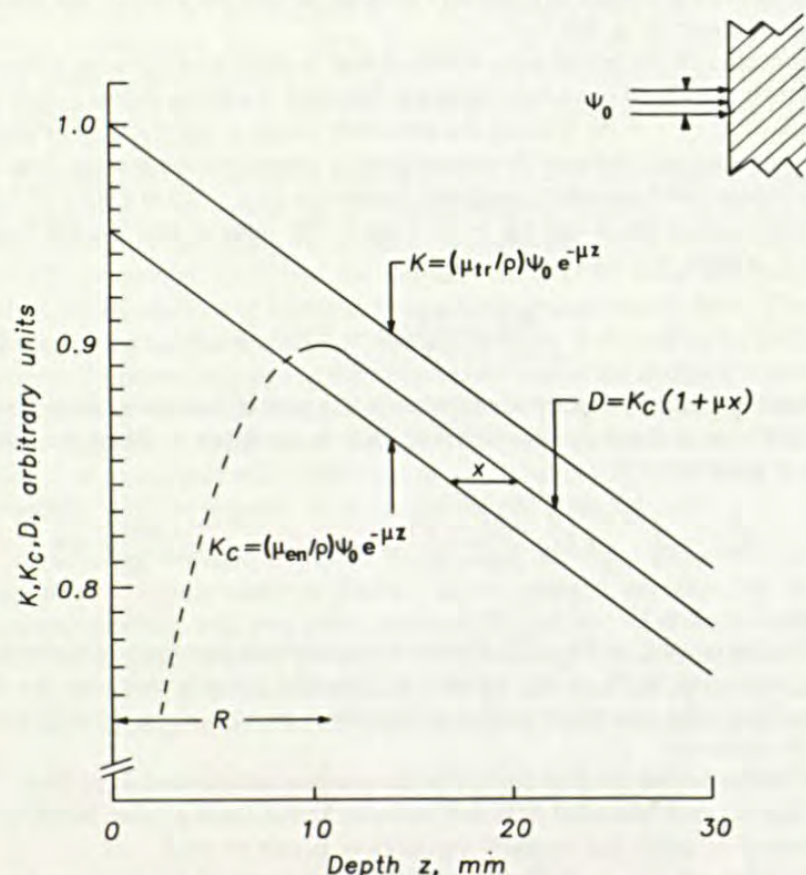


Fig. 18. Kerma K , collision kerma K_c , and absorbed dose D as a function of depth along the central axis in a slab of aluminum irradiated with perpendicularly incident 6-MeV photons. The incident beam diameter is twice the maximum range R of the charged particles, so that transient equilibrium [Eq. (74)] is established at depths $z > R$. Buildup of secondary photons along the central axis is neglected.

media irradiated with the same photons. The degree to which CPE is obtained in a particular case depends not only on the relation between $1/\mu$ and R but also depends on the directions of emission of the primary charged particles and the scattering properties of the medium.

In Fig. 18, kerma, collision kerma, and absorbed dose are given for 6-MeV photons perpendicularly incident on a semi-infinite aluminum absorber; narrow-beam geometry is assumed as defined in deriving Eq. (74). In transient equilibrium ($z \geq R$), $D(z) = K_c(z - x)$. The value of absorbed dose can be interpreted as imparted by the electrons liberated a distance x "upstream." The ratio D/K has (in cases when bremsstrahlung is negligible) been called a "correction factor for the mean origin of electrons" (Boutillon and Niatel, 1973).

When bremsstrahlung losses are so large that $\mu_{tr}/\mu_{en} > 1 + \mu x$, the D curve will not rise above the K curve (Attix, 1968, 1979a, b). It is not generally valid that absorbed dose exceeds kerma in the region of transient equilibrium. The latter result is obtained only if bremsstrahlung is neglected as in the original treatment by Roesch (1958). This treatment is applicable to the case of neutrons, where the liberated heavy charged particles lose a negligible fraction of their kinetic energies to bremsstrahlung.

Calculated values of D/K_c in transient equilibrium depend on the model used for the charged particle transport which determines the distribution $d\epsilon(z'')/dz''$. The values in Fig. 18 were obtained using $x = 3.4$ mm as determined by Roesch (1958), who followed a method described by Lewis for calculating first moments of point-collimated electron sources (see also Roesch, 1968); values of μ , μ_{en} , and μ_{tr} were taken from Hubbell (1969). Jun and Loevinger (1980) used energy dissipation functions tabulated by Spencer (1959) for plane-perpendicular electron sources, following a method by Allisy (1967). They calculated values of D/K_c in polystyrene, carbon, air, and aluminum for photon energies between 0.4 and 2 MeV. Results for aluminum and air are shown in Fig. 19.

The departure from CPE increases with increasing photon energy; however, it is less pronounced in aluminum than in air, Fig. 19, even though the product μR (R calculated in the continuous slowing-down approximation) is somewhat larger in aluminum than in air (Hubbell, 1969). This results from the stronger Rutherford scattering in aluminum. Since Compton scattering predominates in both aluminum and air, the initial directions of motion of the liberated electrons are the same in both media.

With a broad beam of incident monoenergetic photons, the buildup of secondary photons along the central axis can no longer be neglected. The degree of CPE obtained depends on the energy and angular distributions of the total photon fluence consisting of primary as well as of secondary photons. Attix (1979a, b) estimates the quotient of D and K_c to be $1 + \mu'x$, where μ' is an "effective," depth-dependent attenuation coefficient for the broad beam attainable from calculated buildup factors reported in the literature; x is the first moment, Eq. (75),

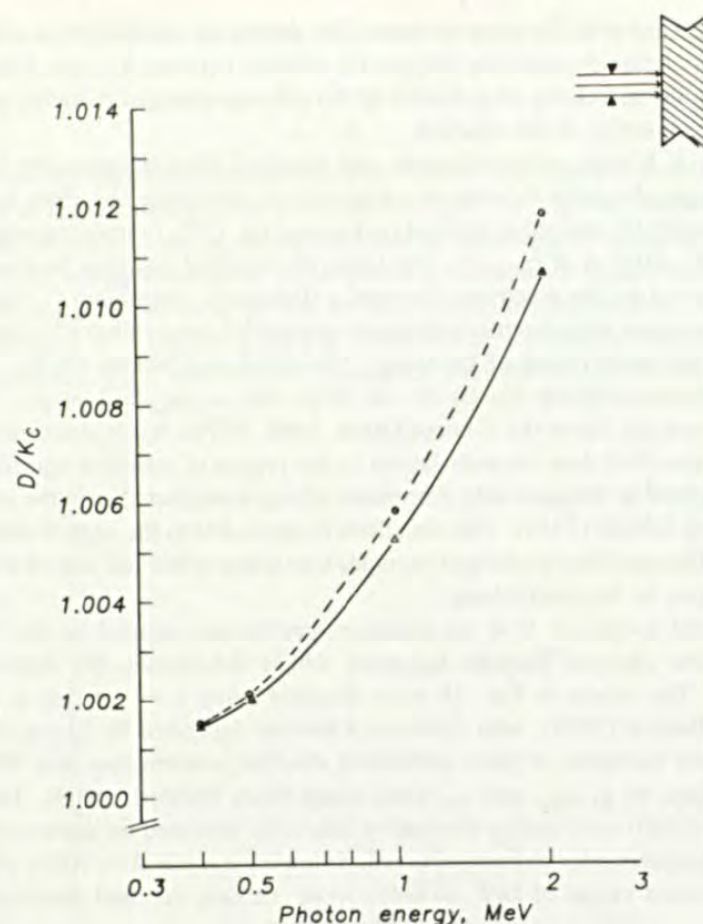


Fig. 19. The quotient in transient equilibrium between absorbed dose D and collision kerma K_c in aluminum (Δ — Δ) and air ($\circ$ — $\circ$) for photon energies from 0.4 to 2 MeV. [After Jun and Loevinger (1980).]

considering the complete energy distribution of the photon fluence at the point of interest. For each direction of motion of the photons, the z axis is along this direction and knowledge about the angular distribution is not required for calculating x .

To obtain an upper limit for the case with both primary and secondary photons present, the value of x may be taken as that valid for the primary photons only (Attix, 1979a, b). Then for 6-MeV photons incident on aluminum (broad beam), D/K_c equals, in transient equilibrium, 1.015 over depths up to 5 cm compared with its value of 1.025 (Fig. 18) in a narrow beam ($\mu' = 0.044 \pm 0.001 \text{ cm}^{-1}$, $\mu = 0.0718 \text{ cm}^{-1}$). More accurate derivations of this quotient for broad-beam conditions could be performed using, e.g., cou-

pled photon-electron Monte Carlo transport calculations (Berger and Seltzer, 1969; Nahum, 1978; Nelson, 1980).

d. The Quantity Exposure. The exposure X at a point is defined only for photons and, contrary to kerma, is restricted to one reference material, viz., air. In terms of radiometric quantities and interaction coefficients it is defined as (ICRU, 1980)

$$X = \frac{e}{W} \int_{h\nu_{\text{cut}}}^{\infty} \left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{air}} \Psi_{h\nu} d(h\nu) \quad (76)$$

The integral is the air collision kerma. Exposure is the ionization equivalent of this quantity.

Primary standards for radiation measurements in photon fields are mostly based on the absolute determination of exposure. Initially, the free-air chamber was constructed and used for ionization measurements under conditions of CPE in air (see, e.g., Roesch and Attix, 1968; Wyckoff, 1980). To the extent that W is accurately known, these measurements also mean absolute determinations of air collision kerma. With increasing photon energy, the ranges of the secondary electrons increase and measurements with the free-air chamber become increasingly difficult. Absolute determinations of exposure with 1- to 2-MeV photons are preferably made using a small but well-determined air volume enclosed within walls of graphite just thick enough to establish CPE at the site of the air volume. This is treated as a Bragg-Gray cavity within graphite, and cavity theory is employed to convert the ionization measured into exposure. Among corrections which have to be made, one considers the degree to which CPE is actually established in the graphite at the site of the air volume (Boutillon and Niatel, 1973; Loftus and Weaver, 1974). By improving knowledge about the ratio between D and K_c , cf. Fig. 19, for higher photon energies, accurate determinations of exposure can in principle be extended to photon energies > 3 MeV (Attix, 1979a, b), commonly considered to be the upper limit of photon energy for practical exposure measurements.

The term "exposure" is, among nonspecialists, often used to describe irradiation conditions in a sense that approaches incident radiant energy. Use of the concept air collision kerma instead of exposure could help eliminate confusion. In practical dosimetry, there exists a trend to change to this usage, encouraged by the introduction of SI units. Formerly, the conversion factor relating exposure (unit R) to water collision kerma (unit rad) was close to unity. With SI units, this conversion factor for exposure (unit C/kg) to water collision kerma (unit Gy) is approximately 30. By considering instead air collision kerma, the conversion factor $(\mu_{\text{en}}/\rho)_{\text{water}}/(\mu_{\text{en}}/\rho)_{\text{air}}$ remains close to unity.

4. Delta-Particle Equilibrium

Delta-particle equilibrium is approximately established within a homogeneous medium externally irradiated with heavy charged particles. The conditions are that both the range and the mean free path for nuclear reaction of the heavy charged particles are much larger than the maximum range of the δ particles they liberate. These conditions are fulfilled, e.g., for protons of kinetic energies up to at least 1 GeV in a low-Z material. Figure 20 may serve as an illustration of the conditions met in irradiations with energetic heavy charged particles.

The maximum energy that can be transferred to an atomic electron by 185-MeV protons is 450 keV (Evans, 1955; Janni, 1966) having a mean track length (CSDA) in $(C_2F_4)_n$ of 0.8 mm (Berger and Seltzer, 1982b). Most of the δ particles have much lower initial kinetic energies and the effective δ -particle buildup region extends only to about 0.4 mm, Fig. 20, corresponding to a depth of about 0.8 mm in tissue. Secondary protons are created in cascade and evaporation processes and cause the slower portion of the buildup region (Zerby and

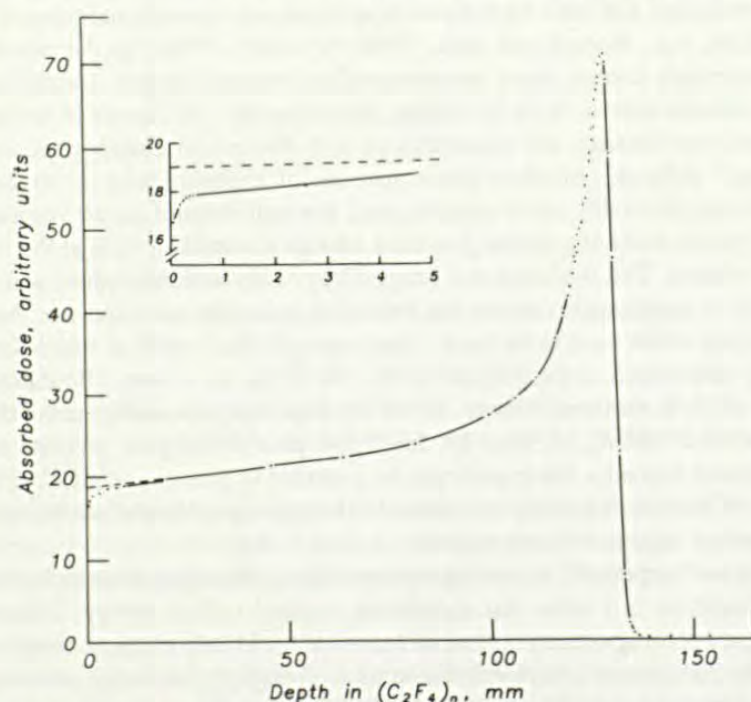


Fig. 20. Absorbed dose as a function of depth in Teflon $(C_2F_4)_n$ for a broad beam of perpendicularly incident 187-MeV protons as measured with 0.1-mm-thick, disk-shaped LiF dosimeters. The inserted enlargement of the buildup region at the surface demonstrates a rapid buildup over the first 0.4 mm due to δ particles followed by a slower buildup from secondary protons released in nuclear reactions by the primary protons. [After Carlsson and Alm Carlsson (1977).]

Kinney, 1965) extending to a depth of about 20 mm in $(C_2F_4)_n$. Secondary neutrons and gamma rays created in the nuclear reactions cause a low tail of absorbed dose beyond the Bragg peak of the proton absorbed dose.

At depths beyond the δ -particle buildup region, δ -particle equilibrium is a good approximation for protons of all energies; both primary and secondary protons are in equilibrium with their δ rays. In Monte Carlo calculations of proton transport and absorbed dose in soft tissue (Turner *et al.*, 1964; Zerby and Kinney, 1965; Neufeld *et al.*, 1966; Wright *et al.*, 1969), δ -particle equilibrium is usually presupposed for protons of all generations. The contribution to absorbed dose by secondary gamma rays liberated in the nuclear reactions is mostly neglected.

In the following derivation of an expression for absorbed dose in δ -particle equilibrium, primary and secondary charged particles are defined as follows: a primary charged particle is one emitted from an accelerator, emitted in a nuclear decay or deexcitation, liberated in the interactions of uncharged ionizing particles, or created in nuclear or elementary particle reactions; a secondary charged particle is one created by a primary charged particle in electronic collisions and is a δ particle or an Auger or Coster-Kronig electron (for simplicity all are classified as δ -particles in the following). It is assumed that such a secondary electron is not energetic enough to cause nuclear reactions, i.e., cannot give rise to primary charged particles.

When the δ particles are uniformly liberated at all distances from a point up to their maximum range in a homogeneous medium, the absorbed dose imparted in interactions of charged particles is given by

$$D = n_{\delta} \left(\bar{T} - \sum_j T_{u,j} \right)_{\delta} + n_{i,c,p} \bar{\delta\epsilon}_{i,c,p} \quad (77)$$

The first term results from reciprocity [Eq. (64)]* and is the absorbed dose imparted in the interactions of δ particles: n_{δ} is the number of δ particles liberated per unit mass by primary charged particles, $\bar{T}$ their mean kinetic energy, and $\sum_j T_{u,j}$ the sum of the energies of all uncharged ionizing particles generated in the slowing down of a primary δ particle and its associated δ particles. Contrary to the case in CPE, the energies imparted by particles in the interactions that liberate those particles with which equilibrium is established cannot be neglected: $n_{i,c,p}$ is the number of interactions by primary charged particles per unit mass, and $\bar{\delta\epsilon}_{i,c,p}$ is the expectation value of the imparted energy in such an interaction.

*The primary particles, Eq. (64), considered to be in a state of equilibrium here are the *primary δ particles*, viz., those liberated in the interactions by primary charged particles. The primary δ particles liberate *secondary δ particles* in slowing down.

By neglecting the imparted energies in those interactions by the primary charged particles in which bremsstrahlung or nuclear reactions occur, Eq. (77) can, by introducing the expression for $\delta\epsilon$ in Eq. (2) for electronic Coulomb collisions, be reduced to

$$D = \int_{T_{\text{cut}}}^{\infty} \Phi_{T,p} \left(\frac{S_{\text{col}}}{\rho} \right) [1 - p_{\text{col}}(T)] dT \quad (78)$$

where Φ_p is the fluence of primary charged particles and p_{col} is that fraction of the kinetic energy lost by a primary particle in an electronic Coulomb collision which either reappears as characteristic x rays subsequent to this collision, or else reappears as photon energy in the slowing down of the released δ particles.

In cases when p_{col} can be neglected as in low-atomic-number media, the integrand in the expression for absorbed dose reduces to a more familiar one [cf. Eq. (56)] containing the product of fluence and mass collision stopping power only. In high-atomic-number media, neglect of p_{col} might not be a good approximation due to high probability of emission of characteristic x rays in filling inner-shell vacancies and, possibly, the generation of bremsstrahlung in the slowing down of δ particles. The electrons generated when the photons subsequently interact, including electrons liberated by gamma rays from nuclear reactions, are in the formalism here regarded as primary charged particles. If CPE exists for the electrons generated by photons, δ -particle equilibrium also exists and Eq. (78) is still applicable, provided the integration is extended to include a summation over primary electrons. Otherwise, absorbed dose from electrons liberated by photons must be accounted for separately.

Rossi and Roesch (1962) derived an expression for absorbed dose in δ -particle equilibrium by applying the two-particle field approximation to the field of primary charged particles and their δ particles. They restrict interactions by charged particles to collisions with $Q = 0$ [Eq. (2)]. Uncharged ionizing particles are regarded merely as sources of charged particles. In δ -particle equilibrium, $\text{div } \Psi_{\delta} = 0$, their expression for absorbed dose [Eq. (33) in the cited reference] reduces to Eq. (78) with $p_{\text{col}} = 0$ and the total stopping power substituted for the collision stopping power. This is an artifact of the two-particle field approximation, the nature of which can be disclosed by returning to Eq. (67). The $\text{div } \Psi_e$ can be divided into the sum of $\text{div } \Psi_{e,p}$ (the vectorial energy fluence of primary charged particles) and $\text{div } \Psi_{\delta}$. In the two-particle field approximation, K is included in the source term of $\text{div } \Psi_{e,p}$ [see last term of $\text{div } \Psi$, Eq. (53)]. However, a primary charged particle can, in an interaction, give rise only to an energy-degraded primary charged particle or a δ particle, but not to a third type of particle, e.g., a photon. The term B in Eq. (67) cannot be accounted for. While photons or neutrons incident on an infinitesimal volume element are potential sources of charged particles emitted from that volume,

photons (bremsstrahlung, characteristic x rays) liberated by charged particles within it and escaping it are not such sources. If the absorbed dose equation derived in the two-particle field approximation is corrected to take account of B , one gets Eq. (78) presupposing that equilibrium is established due to a uniform liberation of δ particles in a homogeneous medium.

5. Partial δ -Particle Equilibrium

Since electrons liberate δ particles which may acquire half of the kinetic energy of the interacting electron, δ -particle equilibrium cannot be established in media externally irradiated with electrons. Only in cases when CPE exists does δ -particle equilibrium also exist for electrons. On the other hand, for a limited group of low-energy δ particles, a state of equilibrium (partial δ -particle equilibrium) may be reached as indicated in, e.g., Monte Carlo calculations by Berger and Seltzer (1969) of the penetration and diffusion into a water phantom of high-energy (> 10 MeV) electrons. The fluence of secondary electrons with kinetic energies less than about 100 keV remains fairly constant with depth when normalized to the value of absorbed dose.

When equilibrium is restricted to δ particles with initial kinetic energies below a limit Δ , the first (equilibrium) term in Eq. (77) is restricted to take into account the generation of these low-energy δ particles only. Absorbed dose imparted in interactions by δ particles with initial kinetic energies above Δ must instead be considered in the second term and these δ particles are now included among the primary electrons. Then, in partial δ -particle equilibrium

$$D = \int_{T_{\text{cut}}}^{\infty} \Phi_{T,p} \left(\frac{S_{\text{col}}}{\rho} \right) (1 - p_{\text{col},\Delta}) dT \quad (79)$$

where p_{col} , Eq. (78), is increased to $p_{\text{col},\Delta}$ by excluding also that fraction of the kinetic energy lost by the interacting particle which reappears as a δ particle with initial kinetic energy above Δ .

In practical applications, the choice of a value for Δ is somewhat arbitrary and depends on the energies of the primary electrons (a value of $\Delta \approx 100$ keV is indicated above for high-energy electron beams). Moreover, absorbed dose is generally calculated according to the Spencer-Attix formulation, Eq. (57), with T_{cut} replaced by Δ as a lower integration limit. In terms of concepts of equilibrium, the corresponding modification of the restricted mass collision stopping power for $\Delta < T \leq 2\Delta$ (see Section VI,B) implies a state of equilibrium for the trackends of primary and high-energy δ particles. In CPE, trackend and partial δ -particle equilibria are complete independent of the value chosen for Δ .

The Spencer-Attix formulation of absorbed dose originates from the Spencer-Attix cavity theory, dealing with a bare Bragg-Gray cavity (detector) within a medium of differing atomic composition. It represents a case in which no kind of equilibrium exists. With respect to the cavity, the Spencer-Attix formulation yields a value of the spatial mean of absorbed dose in it, and not a value of absorbed dose at a particular point. Electrons entering from the medium generate δ particles in the cavity, the transport of which causes absorbed dose to vary from point to point in it. The appropriate value for Δ depends on cavity geometry and was originally taken to be equal to the kinetic energy of an electron that can just traverse it. In a later work, Spencer (1971) suggested a more sophisticated method, based on equations of transport theory, for determining spatial averages of absorbed dose in the cavity. Janssens (1981) performed calculations on the basis of this method and found that in most cases the original formulation, Eq. (57), is satisfactory provided Δ is instead chosen to be equal to the kinetic energy of an electron having its range equal to half the average chord length of the cavity.

A situation quite different in principle arises when the cavity is surrounded by a thin lining of an identical material, by means of which a state of partial δ -particle equilibrium is established at all points in it. Absorbed dose does not vary within the cavity and can be derived using Eq. (57), with the important distinction, however, that Δ is now determined by the thickness of the equilibrium-generating lining, and not by the dimensions of the cavity itself. The two different cases of cavity geometry and their influence on absorbed dose are demonstrated in Fig. 21.

In the bare cavity, the spatial average of absorbed dose varies with thickness indicating conditions of nonequilibrium (Fig. 21). The Spencer-Attix cavity theory treats this variation, taking it into account through the parameter Δ , which depends on cavity dimension. When lined with 0.25-mm graphited paper, the spatial average of absorbed dose is essentially independent of air thickness, indicating equilibrium conditions, i.e., a constant value of absorbed dose. With increasing thickness of the bare cavity, partial δ -particle equilibrium begins to be established at interior points. At a thickness of 1.4 mg/cm² (corresponding to the range of a 25-keV electron in the continuous slowing-down approximation) the volume within which equilibrium prevails is large enough to conceal the variations of absorbed dose at those points lacking equilibrium. This situation is achieved at even smaller thicknesses (>0.5 mg/cm²) when the cavity is lined with 0.25-mm graphited paper, the departure from equilibrium at points with nonequilibrium being smaller the smaller the difference in atomic composition of the adjacent materials.

a. Nonequilibrium Limit. To complete the exposition of various kinds of equilibrium, proceeding from complete equilibrium down to states of equilibri-

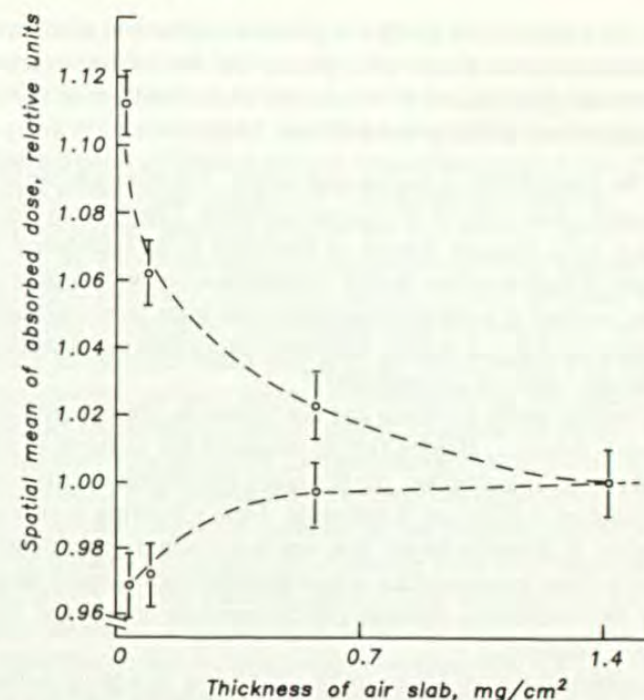


Fig. 21. Spatial mean of absorbed dose in a bare slab of air within lead (upper curve) and in the same slab when lined with 0.24-mm graphited paper (lower curve) as a function of air thickness (being varied by varying the pressure in a 12-mm air gap). Kinetic energy of incident electrons $T_0 = 20$ MeV. [After ICRU (1984a).]

um for successively smaller groups of ionizing particles, it is of some interest to go to the limit of nonequilibrium. When equilibrium can no longer be assumed, the first (equilibrium) term in Eq. (77) vanishes and absorbed dose is given solely by the second term, where now all charged ionizing particles are considered as primary particles. At this final stage, we are, as should be the case, faced with an expression, the sense of which is identical to that in Eq. (54) at the beginning of the section.

D. FANO'S THEOREM

Equilibrium may also be established in media which are not completely homogeneous but which exhibit variations in density. This is stated in Fano's theorem and has been of great practical importance in the construction of ionization chambers for use in dosimetry as well as in experimental work of microdosimetry. It was originally formulated for the case with an infinite medium exposed to a uniform fluence of uncharged ionizing particles generating a distributed source of charged particles in the medium.

Theorem: In a medium of given composition exposed to a uniform flux of primary radiation (such as x rays or neutrons) the flux of secondary radiation is also uniform and *independent of the density* of the medium as well as of the density variations from point to point (Fano, 1954).

The proof by Fano (1954) is elegant and simple. The theorem can be generalized to cases with other sources of ionizing particles. The general requirements for equilibrium (or a constant fluence of particles) to be established within an infinite medium of homogeneous atomic composition but with density variations are (1) that the number of particles emitted per unit mass from radiation sources remains constant throughout and (2) that basic interaction cross sections do not depend on density (state of aggregation).*

In the real world, media are finite and not infinite as presupposed in the derivation of Fano's theorem. Harder (1974) discusses the theorem in connection with bounded media, in particular, the boundary conditions required for obtaining a unique solution in this case. Apparently, Fano's theorem is only applicable at interior points, at distances larger than one maximum particle range from the boundaries, of a finite medium. This is instructively demonstrated in an alternative proof of the theorem by Spencer (1975) based on an integral formulation of the transport equation.

The requirement for equilibrium to be established at a point within a finite medium with density variations could be formulated: equilibrium exists at a point in a finite medium as soon as there is a uniform liberation of particles per unit mass at all distances from the point up to the maximum range of the particles (a condition that will, of course, never be completely fulfilled in actual cases). A reciprocity relation such as that in Eq. (61), illustrated in Fig. 16, cannot, however, be assumed. It is well known from microdosimetric measurements of single particle traversals through low-density cavities inside solid media (Rossi, 1968) that the particle tracks are thereby distorted compared to those in the homogeneous solid medium. However, the total number of particle traversals is also changed such that the expectation value of the total track length of the particles per unit volume, the particle fluence, remains unaffected (Berger, 1974).

The application of Fano's theorem to the construction of walled ionization chambers for use in a field of uncharged ionizing particles has been extensively discussed by Failla (1956). A state of CPE is attainable at all points within a gaseous volume surrounded by a sufficiently thick (equal to the maximum range of the charged particles) solid wall of identical atomic composition. The mean free paths of the uncharged particles must be large compared with the maximum range of the charged particles, and the beam of uncharged particles must remain uniform throughout the chamber wall and gas volume. The latter requirement may be violated in, e.g., a strongly divergent beam or, possibly, when a cylin-

*The interaction cross section *per unit mass* should remain independent of density.

drical or spherical ionization chamber is introduced into a solid medium. The displacement of a medium volume by the chamber may cause appreciable changes in the field of uncharged ionizing particles (showing up as a shift of the "effective point of measurement" from the center of the chamber) in such a way that the different parts of the chamber are nonuniformly irradiated. This problem does not arise to the same extent with a plane-parallel chamber.

When the uncharged particles are distributed in energy, the wall thickness is determined by the maximum range of all charged particles liberated by the uncharged particles. However, CPE does not require that the mean free paths of all the uncharged particles be much larger than this maximum range but only that, for each energy, the mean free path be much larger than the maximum range of the charged particles liberated by particles of that energy. Also, for each energy, the requirement of beam uniformity applies to the gas volume plus that part of the wall from which the liberated charged particles can reach the gas. Absorbed dose in CPE is determined by the spectral distribution with respect to kinetic energy of the uncharged particles at that point at which CPE prevails, Eq. (71). The incident uncharged particles are, when distributed in energy, attenuated differently by the chamber wall, which may have to be considered, e.g., in applications of cavity theory (cf. Section VII).

Fano's theorem can also be referred to in assuring that, e.g., partial δ -particle equilibrium can be reached in a gas volume surrounded by a solid lining. The achievement of a state of equilibrium means that the dimensions of the cavity (gas volume) are not crucial, cf. Fig. 21.

An important aspect of Fano's theorem is the requirement that the interaction cross sections should be independent of density. This requirement is never fulfilled completely and deserves attention. At high energies, the so-called density effect due to polarizability of the medium causes the mass stopping power of charged particles to be lower in a condensed medium than in a gas of identical atomic composition. At low energies, values of the mass stopping power are known to be dependent to some extent on chemical binding and the state of aggregation (see, e.g., Seltzer and Berger, 1982b; ICRU, 1984b). The influence of these departures from density-independent cross sections on the accuracy to which equilibrium can be obtained is not well known.

VII. Cavity Theory

Cavity theory deals with the determination of absorbed dose at a specified point in a medium from a measurement with a radiation detector, which is a most important task of radiation dosimetry. To bring problems of cavity theory into the context of the present review of basic quantities and concepts, the particular problem of formally defining corrections for the perturbing effect of, above all, an air cavity in a condensed medium is discussed here.

When a bare detector is introduced into a medium of differing atomic composition, it will inevitably be inhomogeneously irradiated due to differences in the generation and slowing down of secondary radiations. Only in the unrealistic, limiting case of a point detector is the response solely due to radiations generated and slowed down in the surrounding medium. A possible way to avoid inhomogeneous irradiations is to surround the detector with a wall of identical atomic composition to generate a state of equilibrium for the secondary radiations at all points in it. This possibility is limited by the relative ranges (mean free paths) of the primary and secondary radiations. In the general case, we will have to deal with spatial averages of absorbed dose $\bar{D}_{\text{det}}$ in the detector.

The signal M from the detector is (when statistical fluctuations in the energy imparted can be neglected) related to the spatial average of absorbed dose in it, Eq. (16):

$$M = \bar{\eta}\bar{\epsilon}_{\text{det}} = \bar{\eta}m(\bar{\epsilon}_{\text{det}}/m) = (1/c)\bar{D}_{\text{det}} \quad (80)$$

where m is the mass of the detector and $\bar{\eta}$ is its efficiency. The relation of M to the absorbed dose at a point in the medium D_{med} in the absence of the detector is given by the identity

$$D_{\text{med}} = [M/(M/\bar{D}_{\text{det}})](D_{\text{med}}/\bar{D}_{\text{det}}) \quad (81)$$

Determination of the conversion factor $\bar{D}_{\text{det}}/D_{\text{med}}$, depending only on the radiation absorption properties of the medium and the detector, is the aim of cavity theory. The conversion factor is well established for some ideal cases in which certain properties of the detector can be assumed, e.g., for a detector with dimensions so small that it does not disturb the fluence of charged particles in the medium at that point at which it is inserted. Alternatively, the detector may be one in which secondary particle equilibrium is established at all points in it and the fluences of the primary radiations at these points coincide with that in the undisturbed medium at the point of interest. For these ideal cases, calculations of the conversion factor can be performed using equations derived in the preceding Section VI.

Ideal cases are rarely met in practice, but often a detector used in a given situation can be thought of as nearly ideal. Deviations from the ideal cases are accounted for by so-called "perturbation corrections." The perturbation correction factor is formally a quotient between values of absorbed dose, defined by the identity

$$\bar{D}_{\text{det}}/D_{\text{med}} = (\bar{D}_{\text{det}}/D_{\text{med}})_{\text{theory}}(\bar{D}_{\text{det}}/\bar{D}_{\text{det, theory}}) \quad (82)$$

where $\bar{D}_{\text{det}}$ is the actual value of the spatial average of absorbed dose in the detector and $\bar{D}_{\text{det, theory}}$ is that presupposing a detector with ideal properties. The perturbation correction is the quotient between the values of $\bar{D}_{\text{det}}$ and $\bar{D}_{\text{det, theory}}$.

As an example, consider a case where the detector is supposed to meet the requirements of a Bragg-Gray detector in a radiation field of electrons which may (but not necessarily) have been generated by photons. Then in the Spencer-Attix formulation

$$\left[\frac{\bar{D}_{\text{det}}}{D_{\text{med}}}\right]_{\text{theory}} = \frac{\int_{\Delta}^{\infty} \Phi_T(L_{\Delta}/\rho)_{\text{det}} dT}{\int_{\Delta}^{\infty} \Phi_T(L_{\Delta}/\rho)_{\text{med}} dT} = \frac{(\bar{L}_{\Delta}/\rho)_{\text{det}, \Phi_T}}{(\bar{L}_{\Delta}/\rho)_{\text{med}, \Phi_T}} \quad (83)$$

where Φ_T is the differential fluence of primary electrons ($T > \Delta$) at the point of interest in the medium and Δ is determined by the dimensions of the detector (see Section VI, C, 5)* The point of interest is usually the point in the medium corresponding to the geometrical center of the detector. The weighted averages of the restricted mass collision stopping powers are given by $\bar{L}_{\Delta}/\rho$; index Φ_T indicates the fluence distribution used for averaging.

In particular, when the detector consists of a gas volume in a condensed medium and the requirements of the Fano theorem are not fulfilled, the fluence of primary electrons is likely to vary within it even in cases when the fluence was constant throughout the displaced medium volume before introducing the detector. There exists a nonbalance of in and outscattering of electrons with respect to the gas volume (Harder, 1968). Fluence variations within the displaced medium volume contribute, when present, additionally to fluence variations within the detector. The perturbation correction factor $\bar{D}_{\text{det}}/\bar{D}_{\text{det, theory}}$ [Eq. (82)] can be written†

$$\begin{aligned} \frac{\bar{D}_{\text{det}}}{\bar{D}_{\text{det, theory}}} &= \frac{\int_{\Delta}^{\infty} \bar{\Phi}_T(L_{\Delta}/\rho)_{\text{det}} dT}{\int_{\Delta}^{\infty} \Phi_T(L_{\Delta}/\rho)_{\text{det}} dT} \\ &= \frac{(\bar{L}_{\Delta}/\rho)_{\text{det}, \bar{\Phi}_T}}{(\bar{L}_{\Delta}/\rho)_{\text{det}, \Phi_T}} \quad (84) \end{aligned}$$

where $\bar{\Phi}_{\text{det}}$ is the spatial average of the fluence of primary electrons in the detector and Φ is the same fluence at the point of interest in the medium. Due to differences in the distributions with respect to kinetic energies of these fluences, the weighted averages of the restricted mass collision stopping powers in the numerator and denominator in Eq. (84) may not coincide.

*When D_{med} is calculated according to the Spencer-Attix formulation, partial δ -particle and track-end equilibria are presupposed in the undisturbed medium. The choice of a value for Δ is somewhat arbitrary and is, for simplicity, usually taken as that used in calculating $\bar{D}_{\text{det}}$.

†Use of the Spencer-Attix formulation to calculate $\bar{D}_{\text{det}}$ is justified as soon as there are δ particles generated within the detector which tend to impart most of their energies outside it. The procedure to choose a value for Δ , in cases when the primary electron fluence varies inside the detector, may have to be modified compared to that in conventional Bragg-Gray dosimetry, where this fluence is considered to be constant (identical to that in the surrounding medium).

Svensson and Brahme (1979) and Loevinger (1981) suggest that perturbation correction factors be defined as the quotient between $\bar{\Phi}_{\text{det}}$ and Φ . This is justified in cases when the perturbing effect is so small that the quotient of the weighted averages of the restricted mass collision stopping powers, Eq. (84), is likely to deviate considerably less from unity than the quotient between $\bar{\Phi}_{\text{det}}$ and Φ .

In cases when the fluence in the detector varies due to a varying fluence in the displaced medium volume, the perturbation* can instead conveniently be expressed as a shift of the "effective point of measurement" away from the geometrical center P of the detector to a point P_{eff} (Dutreix and Dutreix, 1966; Hettinger *et al.*, 1967; Svensson and Brahme, 1979). The correction $\bar{D}_{\text{det}}/\bar{D}_{\text{det, theory}}$, Eq. (84), becomes equal to one if a point P_{eff} in the medium is chosen such that

$$(\bar{L}_{\Delta}/\rho)_{\text{det}, \bar{\Phi}_{T, \text{det}}} \bar{\Phi}_{\text{det}} = (\bar{L}_{\Delta}/\rho)_{\text{det}, \bar{\Phi}_{T, P_{\text{eff}}}} \Phi_{P_{\text{eff}}} \quad (85)$$

i.e., P_{eff} is that point in the medium at which the absorbed dose to an ideal detector at that point $\bar{D}_{\text{det, theory}}(P_{\text{eff}})$, equals the actual value $\bar{D}_{\text{det}}$. In determining $D_{\text{med}}(P_{\text{eff}})$ instead of $D_{\text{med}}(P)$, the conversion factor, Eq. (82), remains identical to that, Eq. (83), of an ideal Bragg-Gray detector at point P_{eff} . If the perturbing effect of the detector is small, values of $(\bar{L}_{\Delta}/\rho)$, Eq. (85), are likely to be insensitive to whether $\bar{\Phi}_{T, P_{\text{eff}}}$ or $\bar{\Phi}_{T, \text{det}}$ is used in averaging and P_{eff} can be identified as that point in the medium at which $\Phi_{P_{\text{eff}}} = \bar{\Phi}_{\text{det}}$.

Modern solid-state detectors are convenient for measurements in condensed or liquid media. They can be used without displacing large volumes of the medium. However, compared with the ranges of the charged particles present in the medium or generated by uncharged ionizing particles within them, they are neither thin nor thick. The conversion factor $\bar{D}_{\text{det}}/D_{\text{med}}$ is usually not derived as a perturbation of an ideal case. Burlin (see, e.g., Burlin, 1968) derived an expression for the conversion factor of a bare detector in photon-irradiated media considering explicitly the dose gradients within the detector caused by electrons entering from the medium, as well as from those generated by photons in the detector. Derivations of corresponding conversion factors for use of the same bare detectors within electron-irradiated media have been performed, but the results are still confusing, calling for increased attention to the details of electron transport at the interface between media of differing atomic numbers and density.

*Using a completely medium-equivalent detector, the fluence of electrons in the medium is not disturbed by the detector. The correction factor, Eq. (84), is not truly a perturbation correction but exists, being equal to the quotient between the spatial mean of absorbed dose in a finite medium volume (the detector) and the absorbed dose at the point of interest. The concept of "effective point of measurement" is applicable when this quotient deviates from unity.

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2

Fundamentals of Microdosimetry*ALBRECHT M. KELLERER*

INSTITUT FÜR MEDIZINISCHE STRAHLENKUNDE
 JULIUS-MAXIMILIANS-UNIVERSITÄT WÜRZBURG
 WÜRZBURG, FEDERAL REPUBLIC OF GERMANY

I. Introduction	78
II. General Concepts and Basic Quantities	79
A. Nature of the Problem	79
B. Conventional Parameters	81
C. Two Aspects of Microdosimetry	87
D. The Inchoate Distribution of Energy Transfers	88
E. The Stochastic Quantities and Their Distributions	90
F. The Single-Event Distribution	93
III. The Compound Poisson Process in Microdosimetry	97
A. The Double Role of the Compound Poisson Process	97
B. The Basic Equation	99
C. An Additivity Relation and the Resulting Solution	105
D. Relations for the Moments	107
IV. Determination and Utilization of the Microdosimetric Parameters	113
A. Aspects of Dosimetric and Microdosimetric Measurements	113
B. Application of Microdosimetric Data in Radiation Protection	117
C. Biophysical Implications of Microdosimetric Data	120
V. The Straggling Problem and the Single-Event Spectrum	125
A. Energy-Loss Straggling	126
B. Comparison of the Various Random Factors	132
VI. Geometric Aspects of the Inchoate Distribution	137
A. Uniform and Weighted Random Sampling	137
B. Formulas for the Associated Volume and for the Unweighted Averages	141
C. The Random Intercept of Two Geometric Objects	143
D. Application to Microdosimetry	146
E. A Biophysical Model	152
Appendix: Algorithm for the Compound Poisson Process	154
A. Solution by Successive Convolutions	154
B. Simulation of Specific Energy Distributions	157
References	158

I. Introduction

The remarkable feature of ionizing radiations is their discontinuous interaction with matter. However, absorbed dose and absorbed dose rate are defined as statistical averages that disregard the resulting random fluctuations [1]. For most radiobiological considerations these quantities therefore lose all direct meaning. The knowledge of absorbed dose may permit no statement on energy actually imparted to individual cells or to subcellular structures. The deviations are most substantial for small volumes, for small doses, or for densely ionizing radiations; in many cases the energy imparted at a given absorbed dose can be zero or it can exceed the expectation value by orders of magnitude. Additional concepts and quantities are therefore required to specify the energy concentrations in microscopic regions. This is the subject matter of microdosimetry.

The random nature of energy deposition attracted considerable attention early in the history of radiation biology; this led to the *target theory* and various related approaches. These approaches were aimed at an explanation of the shape of dose-effect relations in terms of the statistics of energy deposits that were termed *hits* and that were alternatively thought to be single ionizations [2], or events of *point heat* [3]. As a heuristic principle this interpretation was useful. The first monograph on the subject [4] still deserves attention, and some principles of target theory will be considered in Section IV. In many applications, however, the approach failed because it made only limited use of physics data or the general tools of probability theory. Absorbed dose remained the sole reference quantity. The parameters of actual interest were derived from simple formulae based on the Poisson statistics for assumed identical and statistically independent events of energy deposition.

A far more realistic treatment emerged in the work of Lea, who attempted a more detailed description of the random configurations of energy deposition in the tracks of charged particles. His monograph "Actions of Radiations on Living Cells" [5] is still a classic; the premature death of Lea (1947) interrupted a promising development.

Microdosimetry in its present sense was, however, not the result of a gradual development. Instead it is founded entirely on an original approach introduced by H. H. Rossi when he recognized the fundamental difference between absorbed dose and the corresponding random variables that need to be described by probability distributions. He and his colleagues proceeded then to develop techniques for measuring the random fluctuations of energy deposition, to construct a novel conceptional and mathematical framework, and to apply the new concepts and methods to radiobiology (see, e.g., [6-9]).

The theoretical basis of microdosimetry is treated in this chapter. Experimental techniques, numerical data, and applications are referred to for illustration of essential concepts and results. A recent report of the ICRU [10] can be consulted for additional details, for technical aspects, and for a compilation of nu-

merical data. The chapter by Rossi in the earlier, related monograph [8] retains special value as an overall introduction and as a synopsis of the essentials of the field.

II. General Concepts and Basic Quantities

A. NATURE OF THE PROBLEM

The theory of radiation quality has a twofold aspect. The physicist asks for a precise description of a radiation field; the biophysicist requires information on the energy distributions in microscopic structures, or on the local concentrations of the radiation-induced ionizations and subsequent radiation products.

For a description of a radiation field one can employ *radiometric quantities*, such as the fluences of the different particles and their spectral distributions in energy and direction. The computation or the experimental determination of these quantities will usually be complex, and even if they are fully known a complete description of the radiation field is not attained. Such a description would require additional information on the spatial and temporal correlation of the fluence of different secondary particles that always occurs when a primary ionizing particle interacts with matter.

To obtain information on the energy distribution in the exposed medium one requires *material constants*, such as cross sections and stopping powers, in addition to the radiometric quantities. The necessary computations may be difficult, but—provided the input information is sufficient—they are always possible, for example, by Monte Carlo simulation. A far more fundamental problem is the inherent complexity of the resultant information. The microscopic patterns of energy deposition differ so widely with different types of radiation and on different levels of spatial resolution that even the most detailed description or simulation may contribute little toward the recognition of those features and parameters that are critical for the biological effect. The purpose of microdosimetry is therefore not the unlimited generation of data, but their deliberate reduction to the most essential parameters.

The radiobiologist has always used highly simplified parameters to elucidate the role of the microscopic distribution of energy and the effectiveness of different types of ionizing radiations. The treatment in terms of linear energy transfer (LET) is an extreme simplification, but is very widely utilized and, in view of its simplicity, it has been remarkably successful. The treatment in terms of microdosimetry is more complex and far more realistic. However, it too rests on various approximations. Such approximations are acceptable when they bring out those aspects that differ for the different types of ionizing radiation; they need not take full account of features that are common to different radiations. For this reason quantities can be employed that are linked to the energy imparted without specific regard to the nature of the electronic alterations; the latter can

be assumed to be largely similar for different types of ionizing radiations. For example, there is reason to assume that excitations contribute little to the action of ionizing radiations on the cell, but this need not be reflected in the definition of microdosimetric quantities. The definitions refer only to total energy imparted. Nevertheless, the quantities remain valid because the ratio of excitations to ionizations and their spatial interrelation are similar for different radiations. Furthermore, one may note that, in contrast to their definitions, the microdosimetric quantities are experimentally determined by ionization measurements in gas. This leads to inaccuracies that are usually not critical. The spatial distribution of ionizations produced by a charged particle in tissue-equivalent gas is largely representative also of the distribution of all energy transfers along the particle's track in tissue.

A further simplification in microdosimetry results from the fact that only the tracks of *charged* particles need to be considered. All ionizing radiations work ultimately through charged particles. Electromagnetic radiations release in matter electrons and positrons, and neutrons release nuclear recoils or nuclear fragments. These charged particles produce the majority of electronic excitations and ionizations that are responsible for the biological effect. It is, therefore, sufficient to treat only the *directly ionizing*, charged particle radiation field. Because the mean free path of photons or neutrons is larger than the structures of interest in microdosimetry, it is possible to disregard any spatial correlation of charged particles even if they are produced by the same uncharged particle. An important exception is the multiplicity of electrons released in an Auger cascade. Other exceptions are α - or β -emitting hot spots. These exceptional cases (see, e.g., [11-13]) are not treated in this chapter, but they can be of considerable radiobiological interest.

Figure 1 serves as a rough illustration of the configuration of charged particle tracks. The dots symbolize individual ionizations in tracks that are computed on the basis of random numbers. The diagrams are somewhat simplified. In particular, the track of the densely ionizing 500-keV proton is actually considerably more narrow than indicated in the diagram, where an attempt has been made to resolve the individual ionizations. Furthermore, it will be noted that the diagram fails to indicate the complex spatial orientation of tracks that occur randomly in the exposed medium. An essential point, however, is the substantial increase of the ionization density with decreasing velocity of the particle. This increase leads to considerable differences in the concentration of the energy transferred to the cell and it has evident implications for the biological effectiveness of ionizing radiations. The local fluctuations of energy deposition and the resultant complexities of energy distribution are equally important and they are the objective of the microdosimetric analysis.

There is, at present, no satisfactory method for obtaining images of tracks experimentally. Nuclear emulsions have been employed for studies of track struc-

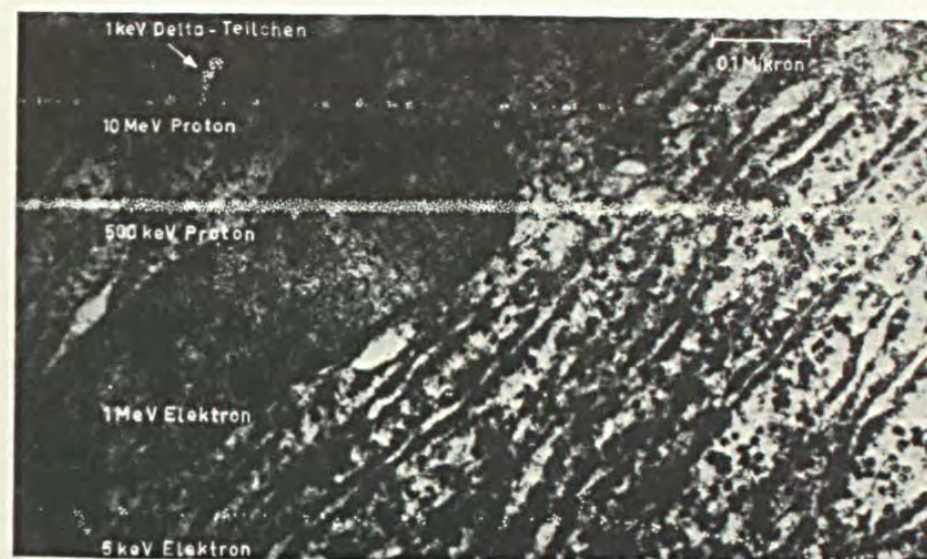


Fig. 1. Diagram of track segments in tissue of electrons and protons with various energies. The dots represent ionizations. The lateral extension of the track core is somewhat enlarged to permit the resolution of individual energy transfers. The length of the track segments is 1 μm .

ture [14, 15], but they do not have sufficient resolution to represent individual ionizations. Somewhat better resolution has been obtained with the more sensitive cloud chamber, which permits a fairly accurate visualization of the tracks, at least of sparsely ionizing particles [16]. However, the only common and practicable method for obtaining geometric representations of charged particle tracks is Monte Carlo simulation on the basis of known and interpolated collision cross sections. Such studies have provided the basis for a quantitative evaluation of particle tracks. In view of the available computer codes and their descriptions (see, e.g., [17-19]) it will not be necessary to give details here. The simulated charged particle tracks can be considered as available input information.

B. CONVENTIONAL PARAMETERS

Certain basic features of the microdistribution of energy deposition by ionizing radiations can be described in terms of conventional parameters; this facilitates the subsequent introduction of the microdosimetric concepts and their interpretation.

1. Fluence

The most important radiometric quantity is the fluence ϕ of a specified type of particle. It is defined as the mean number (expectation value) of particles entering a sphere of unit cross section (diameter: $d = 2/\sqrt{\pi}$). The definition is equivalent to the statement that, on the average, $\phi/2$ particles traverse a unit

plane surface area randomly oriented, or, in a unidirectional field, ϕ particles traverse a unit surface element orthogonal to the direction of the field. Another equivalent statement is that the fluence is equal to the mean total length of particle trajectories per unit volume.

Figure 2 gives the fluence per unit absorbed dose in water for monoenergetic photons, neutrons, and charged particles. The diagram shows that the fluence of uncharged particles, necessary to deliver a given dose, exceeds that of charged particles very substantially. However, a cell or a subcellular structure can be traversed by many uncharged particles without any interaction, and it is therefore the fluence of charged particles that is directly relevant to microdosimetry.

2. Mean Free Path and Range

Figure 3 gives the range of charged particles and the mean free path of uncharged particles in water. The mean free path of photons and neutrons is large compared to cells and cellular structures, i.e., to sites of interest in microdosimetry. The spatial correlation of charged particles set in motion by the same un-

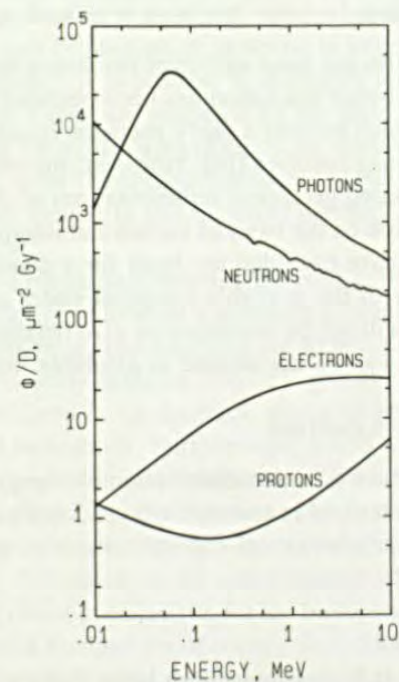


Fig. 2. Fluence per unit absorbed dose in water for charged and uncharged particles with specified initial energies. The fluence of the uncharged particles is always considerably larger than the fluence of the charged secondaries, for equal doses. Charged particle equilibrium is assumed to exist for the uncharged particle fields.

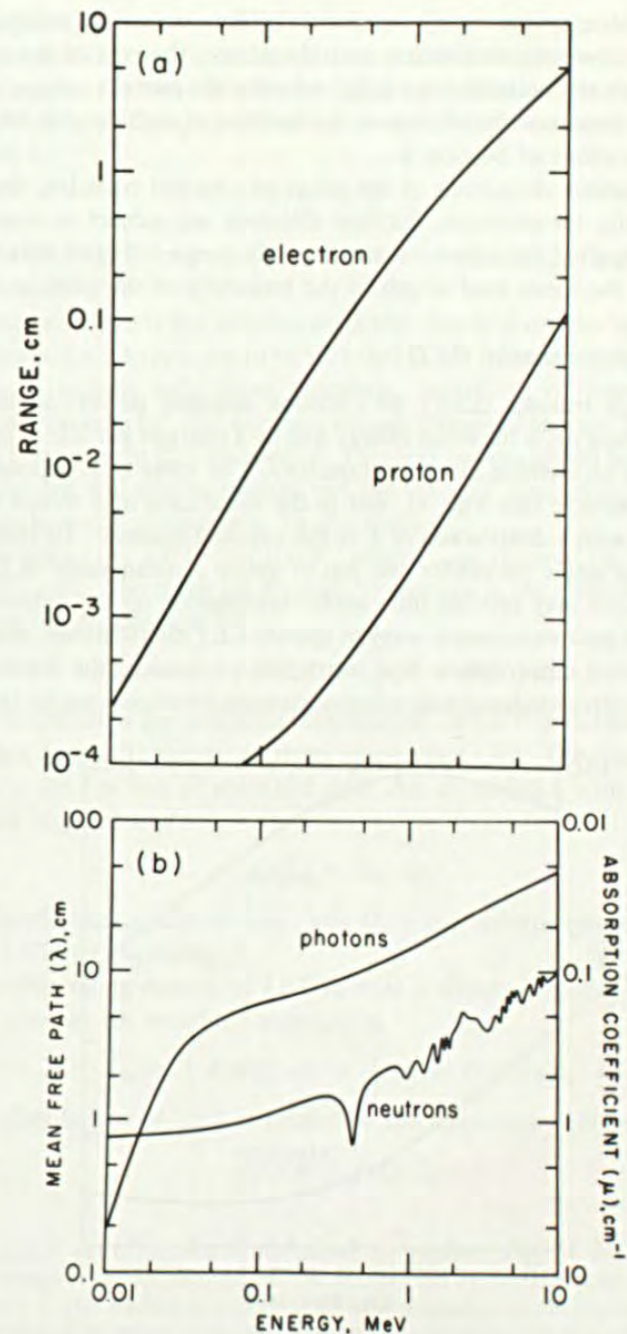


Fig. 3. (a) Range of electrons and protons and (b) mean free path of photons and neutrons of specified energy in water. (From Kellerer and Rossi [41].)

charged particle can therefore, as stated in the preceding paragraph, be disregarded in most microdosimetric considerations. The plot of the ranges of charged particles can be utilized to judge whether the particle ranges are large compared with the structures of interest, a condition of considerable importance for the considerations of Section V.

There are various definitions of the range of charged particles; they differ most substantially for electrons, because electrons are subject to considerable energy-loss straggling and angular scattering. The range in Fig. 3 is an integrated range, i.e., the mean total length of the trajectory of the particle.

3. Linear Energy Transfer (LET)

Linear energy transfer (LET), or collision stopping power, is defined as $L = dE/dx$, where dE is the mean energy lost by a charged particle in electronic collisions along an element dx of its trajectory. The value of L depends on the energy of the particle (see Fig. 4), and in the usual case of a mixed radiation field one deals with a distribution of L in the exposed material. To characterize a radiation by a single parameter one has to utilize a mean value of LET, but such a mean value may provide little useful information on the radiation field. Moreover there are two common ways to specify LET distributions, and accordingly two different mean values. It is instructive to consider the distinction because it has its correspondence with microdosimetric functions (see Section II, F).

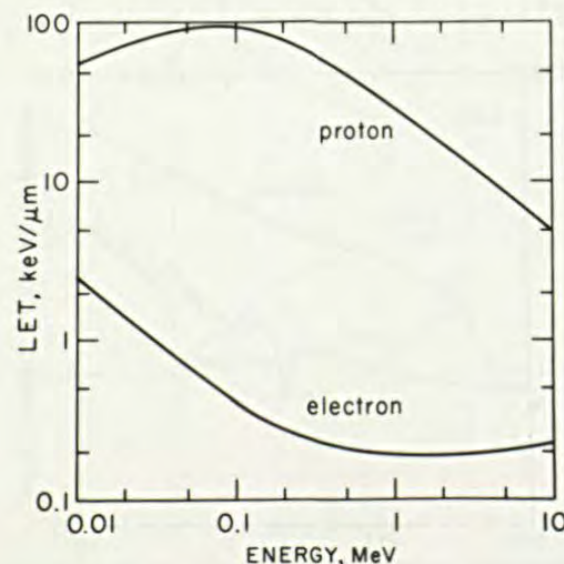


Fig. 4. Linear energy transfer (LET) for electrons and protons in water as a function of their energy. (From Kellerer and Rossi [41].)

The *frequency distribution* of LET is defined in terms of total track length of the charged particles or, equivalently, in terms of the particle fluence. The *distribution function* (or *sum distribution*) $F(L)$ is the fraction of total track length, i.e., the fraction of fluence, that is associated with linear energy transfer not larger than L :

$$F(L) = \phi_L / \phi \quad (1)$$

where ϕ is the total fluence and ϕ_L is the fluence of particles with LET not exceeding L .

For electrons there is the obvious difficulty that it is unclear whether secondary electrons, i.e., δ rays, are to be included in the definitions. A possible convention is to include only those electrons, regardless of whether they are primaries or secondaries, that exceed a certain threshold energy. A suitably restricted LET value must then be applied. This will be considered later in this subsection, where it will also be pointed out that the LET concept is, for additional reasons, particularly problematic for electrons.

The density of LET in track length or fluence is denoted by $f(L) = dF(L)/dL$. The *track* (or *frequency*) *average* is the mean value that corresponds to the distribution:*

$$\bar{L}_F = \int L f(L) dL = \int [1 - F(L)] dL \quad (2)$$

The *dose distribution* (or *weighted distribution*) of LET is defined in terms of the absorbed dose delivered by particles of specified LET. The distribution function $D(L)$ is the fraction of absorbed dose due to particles with linear energy transfer not larger than L :

$$D(L) = D_L / D \quad (3)$$

where D is the total absorbed dose, and D_L is the absorbed dose due to particles with LET not exceeding L .

The corresponding density of LET in dose is denoted by $d(L) = dD(L)/dL$. The *dose average* (or *weighted average*) is

$$\bar{L}_D = \int L d(L) dL = \int (1 - D(L)) dL \quad (4)$$

The dose distribution of LET is related to the frequency distribution:

$$d(L) = L f(L) / \bar{L}_F \quad (5)$$

*The indices F and D are used in this chapter for *frequency averages* and *dose* (or *energy*) *-weighted averages* of various distributions. The densities and weighted densities are designated by $f(\cdot)$ and $d(\cdot)$. This notation is a compromise with common usage. In the more mathematical treatment of Section III, D and in the discussion of the straggling problem a different notation is used; for example, the frequency mean and the weighted mean of the collision spectrum are denoted by δ_1 and δ_2 rather than by $\bar{\delta}_F$ and $\bar{\delta}_D$.

Accordingly one can express the dose average LET in terms of the first two moments of $f(L)$:

$$\bar{L}_D = \bar{L}_F^2 / \bar{L}_F \quad (6)$$

For any distribution the variance equals the second moment minus the square of the first moment. Thus one obtains for the variance and for the relative variance:

$$\sigma_L^2 = \bar{L}_F^2 - \bar{L}_F^2 \quad (7)$$

$$V_L = \sigma_L^2 / \bar{L}_F^2 = (\bar{L}_D / \bar{L}_F) - 1 \quad (8)$$

It follows that the dose average is always larger than the frequency average, the difference being proportional to the variance of $f(L)$:

$$\bar{L}_D - \bar{L}_F = (\bar{L}_F^2 - \bar{L}_F^2) / \bar{L}_F = \sigma_L^2 / \bar{L}_F \quad (9)$$

In spite of its complexities the LET concept can, at best, provide a crude characterization of the charged particle tracks that occur in the exposed medium. Three features are essential in describing charged particle tracks:

- (1) The finite range of the particles and the change of LET along the track.
- (2) The lateral extension of the particle tracks due to the finite range of δ rays.
- (3) The statistical fluctuation of energy loss along the particle track, often termed *energy-loss straggling*.

All three features, and others, e.g., the angular scattering, are disregarded in the LET concept. Only the second point is sometimes taken into account by consideration of a restricted LET, i.e., a stopping power that includes only collisions with energy transfer below a specified cutoff [21]. However, this modification is of doubtful value; one requires a set of LET distributions that belong to different cutoff values but one gains little information about the actual structure of particle tracks.

If a charged particle traverses a spherical reference volume of diameter d , the mean chord length is $2d/3$ and one would, therefore, expect the energy loss $\frac{2}{3}Ld$. However, the actual energy losses can deviate substantially from this value. A quantitative evaluation [22] shows that the LET concept is never adequate for electrons; there are no sites sufficiently small to disregard the finite range of the electrons and at the same time sufficiently large to discount the energy-loss straggling and lateral escape of δ rays. For heavy ions, on the other hand, there are site sizes and particle energies for which the LET concept predicts adequately the energy deposition. But even in this case the LET is of limited value since it permits no statement on the energy distribution within the sites, although this distribution can differ substantially for particles that have the same LET but different velocities.

These limitations of the LET concept illustrate the need for a treatment that is based on random variables rather than statistical expectation values. This has been the historical root of microdosimetry. The spherical proportional counters that are now the main tool of microdosimetry were first utilized in the attempt to measure LET spectra in various radiation fields. In the course of these experiments Rossi realized the inherent impossibility of determining exact LET distributions. Microdosimetry originated when he recognized that the seemingly inadequate response of the detectors was, in fact, superior to the information originally sought. Not the values of LET, but the actual energy concentrations determine the biological effect. These energy concentrations need therefore to be analyzed.

C. TWO ASPECTS OF MICRODOSIMETRY

The spatial patterns of energy deposition in the tracks of charged particles and the resulting biological effectiveness of radiations can be regarded from two different points of view. This needs to be explained before the definitions of basic quantities are considered.

The concept that has been utilized first and that is still most familiar involves certain sensitive structures (*sites*) in the cell, and postulates that the biological effect is determined by the amount of energy deposited in these structures. In a first approximation the spatial distribution of energy within the site is often disregarded, although it must evidently codetermine the effectiveness of the energy imparted. For any considerations in terms of the site concept one requires the probability distribution, and certain expectation values, of energy imparted in the specified structures (see Section II, E). Of equal importance is the probability distribution of energy imparted if exactly one particle and/or its associated particles affect the site (see Section II, F). The probability distributions of energy imparted depend on size and shape of the structure and on the type of radiation.

Energy imparted and its probability distributions can be determined, for any radiation field, with walled or wall-less proportional counters developed for this purpose. Such detectors, termed *Rossi counters*, simulate spherical or cylindrical tissue regions with linear dimensions not less than fractions of a micrometer (see, e.g., [10, 23]). It is essential that microdosimetry has been developed in terms of quantities that are readily measurable, even for an unknown radiation field. These quantities have remained the conventional basis of microdosimetry. In the present chapter, too, most definitions and a major part of the theoretical considerations relate to functions and parameters linked to the *site concept*.

Microdosimetry and microdosimetric quantities are, however, not limited to the site concept or to the notion of energy imparted to certain structures. With the advancement of computational methods and the availability of Monte Carlo simulations of charged particle tracks a more general view has been adopted and

new methods were sought to quantitate or parametrize the microgeometric patterns of energy deposition. While there is, of course, an unlimited variety of possible approaches, one can make the rough distinction between methods that relate the random pattern of energy deposits to a site, i.e., a spatial "probe," and methods that characterize the pattern in terms of spatial interrelations that do not refer to a reference geometry but only to the spatial structure of the track itself. Parameters obtained by this latter method remain meaningful even in a uniform extended medium.

The site concept and the measurements with microdosimetric detectors correspond to the first class of approaches. The LET concept is perhaps the simplest example of the second method. More sophisticated descriptions in terms of the *proximity concept* utilize the notion of the distribution of distances between energy transfers. It is evident that the spatial proximity of energy transfers governs the probabilities of interactions, according to the lifetime and the mobility of radiation products such as free radicals or, on a more complex scale, of macromolecular lesions. Concepts, such as the *proximity function* [10, 24], have therefore extended the application of microdosimetry to radiation chemistry [25] and radiation biology [26, 27]. The novel approaches cannot be adequately treated within this chapter—and some of the results still lack a satisfactory mathematical basis. Some essentials are, however, explained in Section VI. The same section gives the fundamental interrelation that links essential quantities for the site and the proximity concepts.

D. THE INCHOATE DISTRIBUTION OF ENERGY TRANSFERS

For all geometric considerations in microdosimetry the notion of the *inchoate distribution of energy transfers* is essential. It can also help to clarify the definition of the conventional microdosimetric quantities, and it is therefore introduced here.

A convention on terminology is required first. The term *ionizing particle* is subsequently used for a particle with kinetic energy exceeding a specified threshold. The threshold energy will depend on the type of particle and the exposed material, but the choice of the numerical value is not of concern for the present considerations. It is sufficient that appropriate values can be selected for specific circumstances, and one may note that the selected value need not strictly conform to the actual threshold for ionizations in the material.

Ionizing charged particles undergo interactions and lose energy at certain points in the exposed medium. The point of interaction is termed a *transfer point*. The particle of kinetic energy E_1 can be stopped at the transfer point or it can emerge with reduced kinetic energy E_2 . It will be treated as an emerging particle only if its kinetic energy still exceeds the specified cutoff. The interaction may also produce one or more secondary ionizing particles with kinetic

energies E_3 , E_4 , etc. If no ionizing particle emerges from the point of interaction, the point is the end of a charged particle track or of one of its branches. The various possibilities are exemplified in the diagram of Fig. 5.

At any transfer point T_i , the *energy transfer* ϵ_i is the energy that has left the field of ionizing radiations, i.e., it equals the kinetic energy of the incoming particle minus the kinetic energy of all emerging ionizing particles.*

In an exposed medium a random configuration of transfer points occurs. The term *particle track* is used for the configuration of transfer points and of associated charged particles. The entire constellation of transfer points in the exposed medium has been called the *inchoate distribution* (of energy transfers) [24]. The term *inchoate* refers to the fact that this is the incipient distribution before the subsequent processes of further energy degradation.

The notion of the inchoate distribution of energy transfers is particularly pertinent to microdosimetric computations, because any microdosimetric variable and its probability distribution can be determined from repeated random realizations of inchoate distributions or from multiple sampling of a sufficiently extended inchoate distribution. This will be dealt with in Section VI.

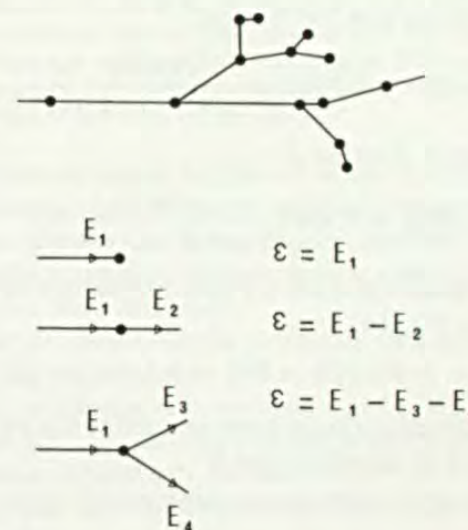


Fig. 5. Schematic diagram of a segment of a charged particle track with indication (•) of the transfer points and the corresponding energy transfers (ϵ_i).

*A rigorous definition has to account also for possible changes of rest mass. In the interest of brevity it is here omitted; it is sufficient to note that it is analogous to the formulation in the definition of absorbed dose [1, 28].

E. THE STOCHASTIC QUANTITIES AND THEIR DISTRIBUTIONS

Energy imparted ϵ is the sum of all energy transfers within a specified site S . It is a random variable and its relative fluctuations are greatest for small sites, for densely ionizing radiations, and for small doses.

$$\epsilon = \sum \epsilon_i \quad (\epsilon_i \text{ in } S) \quad (10)$$

Energy imparted has a uniquely defined value in a specified region after an exposure has taken place. The values vary with repeated irradiations, and predictions can be made only on the basis of probability distributions.

A closely related quantity is the *specific energy*. The specific energy z is defined as the energy imparted divided by the mass m of the specified region:

$$z = \epsilon/m \quad (11)$$

Although z and ϵ are closely related quantities, it is often more convenient to use z because it is the random analog of absorbed dose D . In many radiobiological applications one utilizes the energy unit keV and expresses the mass as the product of the volume V in cubic micrometers and the density ρ in grams per cubic centimeter. One has then the relation

$$z \text{ (Gy)} = 0.1602 \frac{\epsilon \text{ (keV)}}{V \text{ (}\mu\text{m}^3) \cdot \rho \text{ (g/cm}^3\text{)}} \quad (12)$$

or for a spherical site of diameter d

$$z \text{ (Gy)} = 0.306 \frac{\epsilon \text{ (keV)}}{[d \text{ (}\mu\text{m)}]^3 \cdot \rho \text{ (g/cm}^3\text{)}} \quad (13)$$

The probability *distribution function* (or *sum distribution*) of z at an absorbed dose D is denoted by $F(z; D)$:

$$F(z; D) = P(\underline{z} \leq z | D) \quad (14)$$

i.e., the distribution function is equal to the probability that the random variable $\underline{z}$ does not exceed z at an absorbed dose D .

The probability density (or differential distribution) $f(z; D)$ of z is the derivative of $F(z; D)$:

$$f(z; D) = dF(z; D)/dz \quad (15)$$

The definitions of the distributions for energy imparted are entirely analogous and therefore need not be stated. The same applies to other subsequent considerations that are formulated in terms of specific energy but can equally be given for energy imparted.

Sum distribution and probability densities of specific energy are illustrated by a numerical example in Figs. 6 and 7. The distributions are computed from the

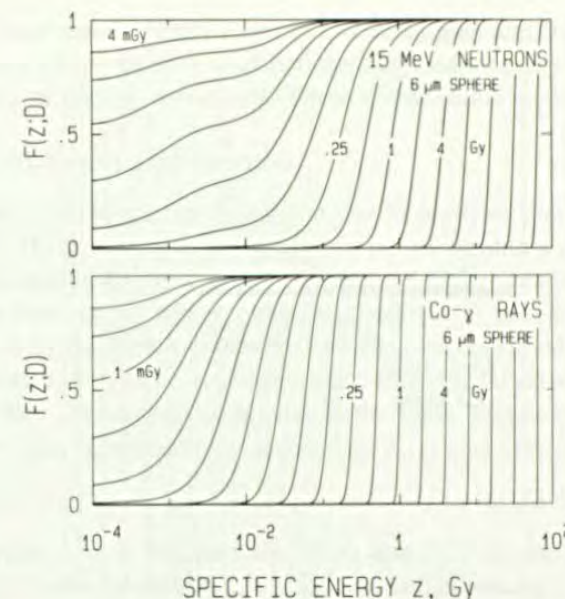


Fig. 6. Sum distributions $F(z; D)$ of specific energy in a tissue sphere of $6\text{-}\mu\text{m}$ diameter and unit density exposed to different doses of ^{60}Co γ rays and to 15-MeV neutrons. This and subsequent figures are based on data by Kliuga and Dvorak [66] for ^{60}Co γ rays and on data by Booz and Coppola [61] for neutrons. The distributions are calculated by the algorithm of successive convolutions that is explained in Section III and the Appendix.

single-event distributions that are introduced in the next subsection. The computational procedures and their theoretical basis are considered in Section III and the Appendix. One may note that dose-dependent distributions $f(z; D)$ were obtained experimentally in the early microdosimetric studies [29] before the computational procedures were developed.

The function $f(z; D)$ determines the probability for a specified value z of the specific energy at the absorbed dose D , i.e., $f(z; D) dz$ is the probability that the specific energy assumes a value between z and $z + dz$. At low doses the relative fluctuations are large, and it is then impractical to represent the distributions on a linear scale. Accordingly one uses, as in Figs. 6 and 7, a logarithmic scale. On a logarithmic scale the densities are transformed; one must plot the dimensionless quantity

$$dF(z; D)/d \ln z = z \cdot f(z; D) \quad (16)$$

This transformed density is properly normalized with respect to the natural logarithm of z . Normalization relative to the base-10 logarithm would require an additional numerical factor $\ln(10) \sim 2.30$.

There is always a finite probability $F(0; D)$ for $z = 0$. Whenever this probability cannot be disregarded, a δ -function $F(0; D) \cdot \delta(z)$ has to be included in $f(z; D)$. When the densities are given on a logarithmic scale of z , this discrete com-

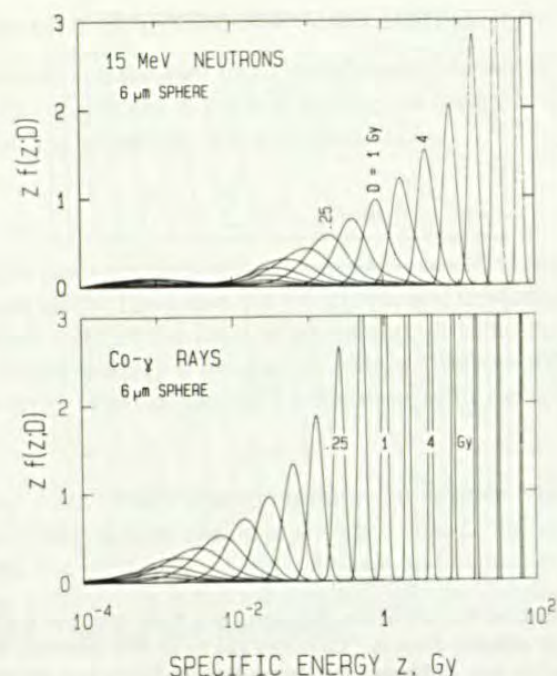


Fig. 7. Densities of specific energy that correspond to the sum distributions in Fig. 6.

ponent at $z = 0$ cannot be represented. However, at low doses the area under the curves is less than unity, and, as in Fig. 7, the defect of the area indicates the magnitude of the δ function.

The average (expectation value) specific energy in a site

$$\bar{z} = \int_0^{\infty} z f(z; D) dz = \int_0^{\infty} [1 - F(z; D)] dz \quad (17)$$

is equal to the absorbed dose D when the site is uniform and is exposed to a uniform radiation field. Otherwise $\bar{z}$ equals the average absorbed dose in the site. Under nonuniform conditions a rigorous definition of absorbed dose must be given in terms of the limit value:

$$D = \lim_{m \rightarrow 0} \bar{z} \quad (18)$$

This relation illustrates the fact that the random variable z and its probability distribution are more fundamental than the absorbed dose.

For a specified reference site and a specified radiation one deals with functions, $F(z; D)$ or $f(z; D)$, of two variables. In practice, it is rarely necessary to utilize the explicit functions; essential biophysical and radiobiological arguments can instead be based on a few parameters of these distributions (see Sec-

tion IV). In those cases where the explicit distributions are required, they can be derived from the probability distributions of specific energy produced in individual events of energy deposition. These distributions are considered next.

F. THE SINGLE-EVENT DISTRIBUTION

An *event* in a site is energy deposition due to particles that are statistically correlated [1]. If, for example, an α -particle passes outside a reference region and injects a number of δ rays, these δ rays belong to the same event in the site. Similarly electrons of an Auger cascade contribute to the same event. Two recoils released by the same neutron, or two electrons liberated by one photon, are also correlated, but, as pointed out earlier, their spatial separation is usually so large that they are unlikely to appear in the same microscopic site.

The *single-event distribution* is denoted by $F_1(z)$ and $f_1(z)$:

$$F_1(z) = P(\bar{z} < z \mid \nu = 1) \quad \text{and} \quad f_1(z) = dF_1(z)/dz \quad (19)$$

The specification $\nu = 1$ indicates that $F_1(z)$ and $f_1(z)$ are the distributions of specific energy under the condition that exactly one event has taken place in the site. It must be noted that the single-event distributions do not contain a discrete component at $z = 0$. By definition, an event requires energy deposition; the mere passage of a charged particle without energy transfer to the site is therefore not counted as an event.

The single-event distributions for energy imparted are defined in an analogous way. There is, furthermore, a related variable that has, originally in somewhat different form, been introduced by Rossi and colleagues as the random analog to LET. The *lineal energy* y is defined as the energy imparted in one event divided by the *mean chord length* $\bar{l}$ that results from the random interception of the site by a straight line:*

$$y = \epsilon / \bar{l} \quad (20)$$

The mean chord length is equal to $4V/S$ for a convex site of volume V and surface S (see, e.g., [30-32])†

It is customary to give y in kilo-electron-volts per micrometer. For the density $\rho = 1 \text{ g/cm}^3$, as often assumed in radiobiological applications, one has the re-

*The utilization of the mean chord length $\bar{l}$ in the definition of y is somewhat arbitrary, because $\bar{l}$ is the mean value for one special type of randomness, i.e., uniform isotropic randomness. Other types of randomness exist and are associated with other mean chord lengths (see Section VI).

†For a spherical site of diameter d one has $\bar{l} = 2d/3$; for a circular cylinder of diameter d and height h the mean chord length is $\bar{l} = 2dh/(d + 2h)$; for a spheroid with two axes d and one smaller axis ($e \cdot d$) one has, with $\epsilon = \sqrt{1 - e^2}$, the mean chord length $\bar{l} = d/[1/2e + \ln(1/e + \epsilon^2)/2\epsilon]$ (see also Section V, B, Fig. 24).

lation between the specific energy and the lineal energy in a sphere of diameter d :

$$z \text{ (Gy)} = 0.204 \frac{y \text{ (keV}/\mu\text{m})}{[d \text{ (}\mu\text{m)}]^2} \quad (21)$$

The definition of y is restricted to energy imparted in one event; this conventional restriction appears reasonable because y is the random analog of LET.

The single-event distributions of lineal energy are largely equivalent to the single-event distributions of specific energy. The values of the sum distributions are equal for corresponding values of z and y :

$$F(y) = F_1(z) \quad (22)$$

The relation is somewhat more complex for the corresponding densities. For a site of density $\rho = 1 \text{ g/cm}^3$ with volume V and surface area S one has

$$f(y) = (V/\bar{l})f_1(z) = (S/4)f_1(z) \quad (23)$$

where the last equality applies to convex sites.

In the convenient logarithmic representation, considered in the preceding subsection, the densities of specific energy and lineal energy are equal and can be represented in the same plot. Characteristic single-event spectra are represented in Fig. 8. They are given in terms of the weighted distributions that are introduced later in this section [see Eq. (27)]. One of the striking features of these distributions is the wide range of values assumed by the random variable. Narrow distributions result only if low-energy monoenergetic photons release photoelectrons in a site with linear dimensions substantially larger than the electron ranges. In the typical cases of very broad single-event distributions it is always appropriate to utilize a logarithmic scale of ϵ , z , or y , and the density must then be transformed, as explained in the preceding section.

The single-event distributions are of far greater pragmatic importance than the dose-dependent distributions. Because individual events are statistically independent it is sufficient to derive, experimentally or computationally, the single-event distributions. The dose-dependent distributions can then be computed, as will be explained in subsequent sections. These sections will also deal in some detail with the *moments* of the microdosimetric distributions. However, the most essential results will first be given without derivation, so that they are accessible without the need to penetrate the mathematical treatment.

The average specific energy produced by an event in the site is

$$\bar{z}_F = \int_0^\infty z f_1(z) dz = \int_0^\infty (1 - F_1(z)) dz \quad (24)$$

*The index 1 is not required with the distribution of y , because y relates by definition only to a single event.

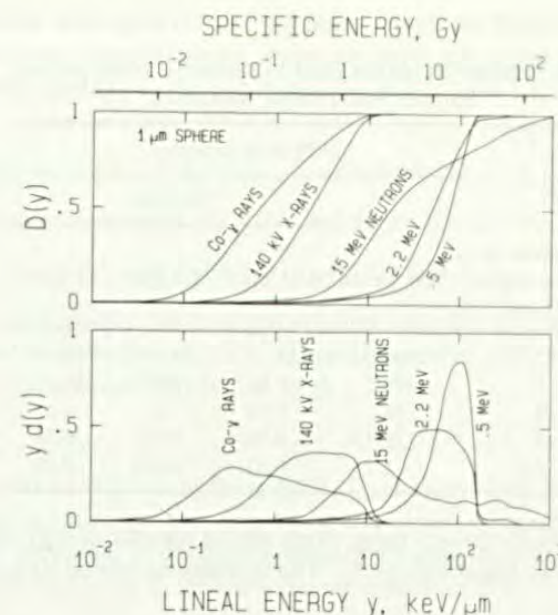


Fig. 8. Distributions of lineal energy in spherical tissue regional of 1- μm diameter exposed to various radiations. In the lower panel the distributions are represented as dose-weighted densities $y d(y)$ relative to a logarithmic scale of lineal energy y . These spectra determine the fraction of absorbed dose delivered per unit logarithmic interval of lineal energy. In the upper panel the corresponding sum distributions $D(y)$ are given, and they specify the fraction of events up to a lineal energy y . On top of the upper panel an additional abscissa is given for the specific energy z . Relative to this scale the curves in the lower panel are the weighted densities $z d_1(z)$ of specific energy in single events; the curves in the upper panel are the sum distributions $D_1(z)$ of specific energy in single events.

The index F is used to distinguish this *frequency* average from the *weighted* average that plays a considerable role in many applications of microdosimetry and that will be considered later in this section.

The average specific energy at the absorbed dose D is the product of the mean event size $\bar{z}_F$ and the mean number $\bar{\nu}$ of events. But it is also equal to D :

$$\bar{z}_F \bar{\nu} = \bar{z}(D) = D \quad (25)$$

It follows that the average number $\bar{\nu}$ of events is equal to $D/\bar{z}_F$. In particular, one concludes that the event frequency per unit absorbed dose is

$$\phi(0) = 1/\bar{z}_F \quad (26)$$

Knowledge of the event frequencies permits general and important conclusions in radiobiological applications (see Section IV, B). Table I gives a synopsis of event frequencies for several site diameters and different types of ionizing radiations.

TABLE I
EVENT FREQUENCIES (ϕ) PER GRAY IN SPHERICAL TISSUE REGIONS
EXPOSED TO DIFFERENT RADIATIONS

Diameter of critical region d (μm)	Type of radiation			
	Neutrons			
	^{60}Co γ rays	0.43 MeV	5.7 MeV	15 MeV
12	2000	55	51	61
5	360	4.2	8.6	11
2	58	0.39	1.2	1.6
1	12	0.08	0.32	0.38
0.5	1.7	0.02	0.073	0.09

In analogy to the frequency mean event size of specific energy one can define the frequency mean lineal energy $\bar{y}_F$. The quantity is related to $\bar{z}_F$ according to Eq. (21).

The frequency mean lineal energy is largely analogous to the frequency mean LET introduced in Section II, B. When energy-loss straggling and the lateral escape of δ rays out of the reference region play no role and when the range of the ionizing particle is sufficiently large, the two mean values $\bar{L}_F$ and $\bar{y}_F$ are approximately equal.

In analogy to the definitions of the LET distributions one can consider weighted distributions of lineal energy or of specific energy:

$$d(y) = y f(y)/\bar{y}_F \quad d_1(z) = z f_1(z)/\bar{z}_F \quad (27)$$

The weighted distributions are also termed *dose distributions*. They determine the fractions of absorbed dose or of energy imparted that are associated with certain values of y or z , and they are therefore relevant to all considerations of the effectiveness of radiation as a function of local energy concentration in microscopic sites.

The mean values of the weighted distributions are

$$\begin{aligned} \bar{y}_D &= \int y d(y) dy = \int y^2 f(y) dy / \bar{y}_F = \bar{y}_F^2 / \bar{y}_F \\ \bar{z}_D &= \int z d_1(z) dz = \int z^2 f_1(z) dz / \bar{z}_F = \bar{z}_F^2 / \bar{z}_F \end{aligned} \quad (28)$$

The variance of any distribution is equal to the second moment minus the square of the mean,

$$\sigma_F^2(y) = \bar{y}_D^2 - \bar{y}_F^2 \quad (29)$$

From this and the analogous relation for z one concludes that the dose averages $\bar{y}_D$ and $\bar{z}_D$ are always larger than the frequency averages $\bar{y}_F$ and $\bar{z}_F$. Furthermore one has, again in analogy to the corresponding relation for LET (see Section II, B),

$$\sigma_F^2(y) = (\bar{y}_D - \bar{y}_F) \cdot \bar{y}_F, \quad \sigma_F^2(z) = (\bar{z}_D - \bar{z}_F) \cdot \bar{z}_F \quad (30)$$

The relative variance is equal for $f(y)$ and $f_1(z)$:

$$V_1 = (\bar{y}_D / \bar{y}_F) - 1 = (\bar{z}_D / \bar{z}_F) - 1 \quad (31)$$

A less evident but equally fundamental relation links the second moment of the dose-dependent distributions $f(z; D)$ of specific energy with the two first moments of the single-event distribution $f_1(z)$:

$$\bar{z}_D^2 = (\bar{z}_D + D) \bar{z}_F \quad (32)$$

This relation will be derived in Section III, D; it is essential to the analysis of radiobiological mechanisms that depend on the square of specific energy. The relation is also important in dosimetry because it yields the relation for the variance of specific energy at a certain value of absorbed dose:

$$\sigma_z^2 = \bar{z}_D^2 - \bar{z}_F^2 = \bar{z}_D \cdot D \quad (33)$$

The relative standard deviation of z is therefore

$$\sigma_z / D = \sqrt{\bar{z}_D / D}, \quad \text{since } \bar{z}_F = D. \quad (34)$$

Applications of these relations are discussed in Section IV.

III. The Compound Poisson Process in Microdosimetry

A. THE DOUBLE ROLE OF THE COMPOUND POISSON PROCESS

When a microscopic site is exposed to a radiation field, events of energy deposition occur in a random sequence. According to ICRU definition [1] the events are statistically independent: each individual event can be due to the appearance of one or several correlated charged particles in the site with consequent energy deposition. Frequently one deals with the simple case where independent charged particles traverse the site in straight random paths, and this situation is schematically represented in the left-hand panel of Fig. 9. During the time interval that corresponds to a specified absorbed dose D events occur randomly, and if the mean specific energy per event is $\bar{z}_F$ the expected number of events is, as pointed out in the preceding subsection, $\bar{\nu} = D/\bar{z}_F$. The actual number ν of events is subject to statistical fluctuations and, because the events are independent, it follows a Poisson distribution. Accordingly, if the event frequency per unit absorbed dose $1/\bar{z}_F$ is known and if one is interested merely in

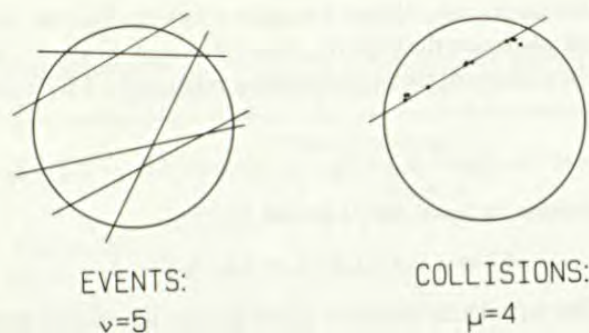


Fig. 9. Schematic diagram that indicates the double role of the compound Poisson process for the energy deposition in a site at specified dose. In the left panel ν events are represented that correspond to statistically independent traversals of charged particles. In the right panel one of the events is selected and is represented as a sequence of μ statistically independent collision processes along the particle track.

the number of events, one deals with a simple Poisson process. The probabilities for 0 events, for 1 event, or for any specified number of events can then be readily calculated.

The assessment of energy imparted is, however, far more complex because the energy imparted per event varies widely. As is apparent from the examples in Fig. 8, typical single-event distributions span several orders of magnitude of the random variable. The statistical fluctuations of energy imparted to a site are, in fact, predominantly determined by the varying amount of energy imparted per event. As will be shown in Section V, B, the fluctuations of the number of events, although they are always present, are far less consequential. It is therefore the essential feature of energy imparted that it results from a *mixed* (or *compound*) *Poisson process*, i.e., a process of independent events of varying magnitude. Formally this can be expressed by the relation

$$\epsilon = \sum_{i=1}^{\nu} \epsilon_i \quad (35)$$

The ϵ_i are the energies imparted in individual events. ν is the number of events, which follows the Poisson distribution. The subsequent section deals with the mathematical and numerical essentials of the compound Poisson process. Basic parameters of the dose-dependent distributions of energy imparted or specific energy in a site will be expressed in terms of the corresponding parameters of the single-event distributions. Furthermore, the explicit relation between the dose-dependent distributions and the single-event distributions will be treated, and the computational procedure will be described that can be utilized to derive the dose-dependent distributions from the single-event distributions.

The compound Poisson process is treated with emphasis on the mathematical relations rather than on the connection between single-event distributions and dose-dependent distributions. This is done because the compound Poisson

process plays a double role in microdosimetry. It applies equally to another step in the chain of random events, namely, the statistical sequence of energy losses of a charged particle traversing the site. This process, commonly termed *energy-loss straggling*, is treated in Section V, and it will be seen that the same mathematical relations and the same numerical procedures link, on the one hand, $f(z; D)$ with $f_1(z)$ and, on the other hand, the distribution of energy lost by a particle along a track segment with the distribution of energy losses in individual collisions.

The left panel of Fig. 9 indicates the events, i.e., the passages of charged particles, merely as line segments. On the right panel one such event is selected and is represented as a succession of *collision events*, i.e., energy losses by the charged particle. The collisions may result in excitations, individual ionizations, or δ rays.

If the track segment within the site is much shorter than the range of the particle, any variations of LET of the particle within the site can be disregarded. As an important consequence the collisions along the track segment can be treated as independent. The number μ of collisions is then again subject to the Poisson distribution. Its expectation value $\bar{\mu}$ is proportional to the length of the segment and to the stopping power of the particle and is inversely proportional to the average energy imparted to the site in a collision. The Poisson fluctuations of the number μ of collisions are always present. But, as in the analogous case of $f_1(z)$ and $f(z; D)$, their influence is far smaller than the influence of the variations of energy lost by the particle, or energy imparted to the site, in individual collisions.

In summary, one can state that there is remarkable similarity on the two levels of the hierarchy of random events. The random variables ϵ_i in Eq. (1) are themselves the result of a compound Poisson process:

$$\epsilon_i = \sum_{j=1}^{\mu_i} \epsilon_j \quad (36)$$

Accordingly the energy imparted to the site is

$$\epsilon = \sum_{i=1}^{\nu} \left(\sum_{j=1}^{\mu_i} \epsilon_j \right) \quad (37)$$

where the inner summation stands for the Poisson process on individual track segments (the energy-loss straggling) while the outer summation represents the Poisson process of charged particles traversing the site (the random sequence of events).

B. THE BASIC EQUATION

At a specified absorbed dose, the energy imparted to the site and the related variable specific energy are, as stated in the preceding subsection, the result of

a *compound* (or *mixed*) *Poisson process*. The term *Poisson process* refers to the independence of events; the term *compound* refers to the fact that the size of the individual events is variable. The *spectrum* of the Poisson process is the single-event distribution $f_1(z)$. The *solutions* of the compound Poisson process are the dose-dependent distributions $f(z; D)$. For brevity the term *event fluctuation* is utilized to refer to this process and its mathematical treatment.

Although this and the following subsection refer to the distributions $f_1(z)$ and $f(z; D)$, the mathematical treatment applies equally to the *energy-straggling* problem, i.e., to the random energy loss of a charged particle along a specified track segment. In all subsequent relations one can, accordingly, substitute $f_1(z)$ by $c_1(\epsilon)$, the probability density of energy ϵ lost by the charged particle in individual collisions. The solutions are then $f(\epsilon; \Delta)$, the probability densities of total energy lost along specified track segments, with expected energy loss $\bar{\epsilon} = \Delta$. These probability densities are termed *straggling distributions*.

The average specific energy produced by a single event is the mean value of $f_1(z)$. This mean value $\bar{z}_F$, which was introduced in Section II, F, is a fundamental parameter because it determines the event frequency $\phi(0) = 1/\bar{z}_F$ per unit absorbed dose [see Eq. (26) and Table I in Section II, F]. Since events are by definition statistically independent, their number ν in a specified site at a specified absorbed dose follows a Poisson distribution:

$$p(\nu) = \exp(-n)n^\nu/\nu!, \quad \text{with } n = \bar{\nu} = D/\bar{z}_F \quad (38)$$

Even if the number ν of events is fixed, the specific energy in the site can vary widely. Its distribution is then the ν -fold convolution of the single-event distribution. This convolution is denoted by $f_\nu(z)$, and it can be defined by the recurrence formula:

$$f_\nu(z) = \int_0^z f_1(x)f_{\nu-1}(z-x)dx \quad (\nu = 2, 3, \dots) \quad (39)$$

$f_\nu(z) dz$ is the probability that the specific energy has a value between z and $z + dz$, if exactly ν events have taken place in the site.

Accordingly one obtains the relation for the dose-dependent distributions of specific energy:

$$f(z; D) = \sum_{\nu=0}^{\infty} e^{-n} \frac{n^\nu}{\nu!} f_\nu(z), \quad \text{with } n = \frac{D}{\bar{z}_F} \quad (40)$$

$f_0(z)$ equals $\delta(z)$, i.e., the delta function at $z = 0$. Accordingly $f(z; D)$ contains always a discrete probability e^{-n} for no event, i.e., for $z = 0$.

The essence of the compound Poisson process is illustrated by the diagrams of Fig. 10. Individual Monte Carlo realizations of the process are represented for 15-MeV neutrons in the plane of the two variables D and z . Any combination of values D and z corresponds to a point; $F(z; D)$ is the probability that the random path runs below the point. Those lines that run below the point pass it on

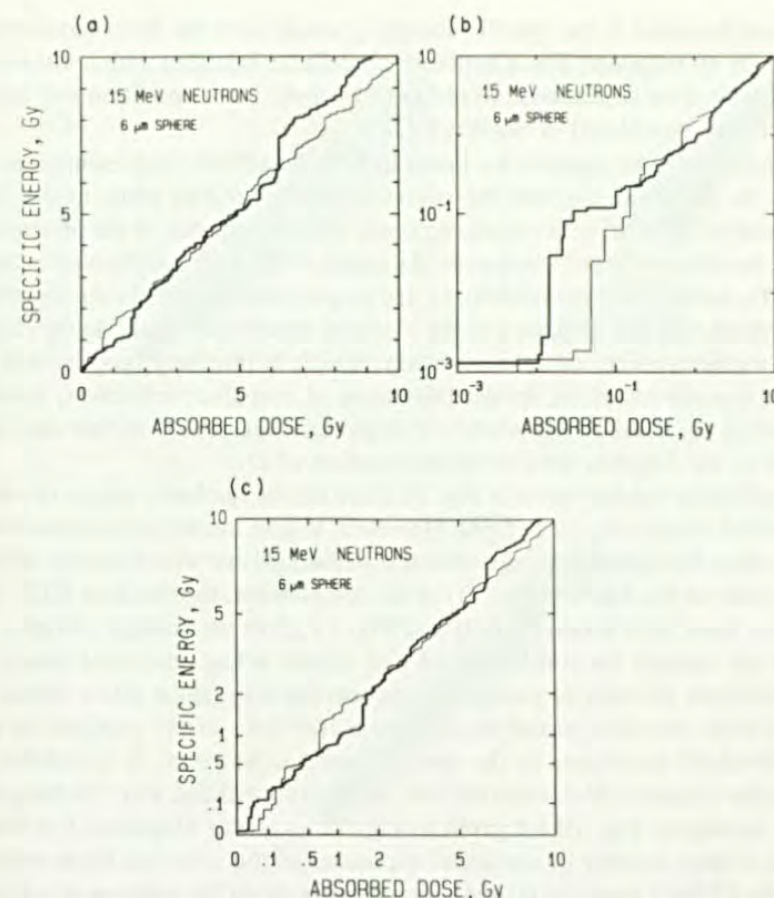


Fig. 10. Two random paths that represent the stochastic sequence of events of energy deposition in a 6- μ m tissue sphere exposed to 15-MeV neutrons. The two random sequences are represented (a) on a linear scale of dose and specific energy, (b) on a logarithmic scale, and (c) on a square-root scale. The absolute deviations of specific energy from absorbed dose increase with absorbed dose, while the relative deviations decrease. On the graph with the square-root scale the magnitude of the deviations remains on the average constant as the dose increases.

the right; i.e., they reach the value z at a dose exceeding D . The conclusion is that $F(z; D)$ is a sum distribution both with reference to z as random variable and with reference to D as random variable:

$$F(z; D) = \text{Prob}\{z \leq z \mid D\} \quad (41)$$

$$G(D; z) = 1 - F(z; D) = \text{Prob}\{D \leq D \mid z\} \quad (42)$$

One must note that the densities of the two sum distributions are not closely linked. The function $G(D; z)$ can be invoked whenever one considers a response with sharp threshold of energy imparted or specific energy. An instrument with

a response threshold at the specific energy z_c would have the dose dependence $F(z_c; D)$ for no response. For a hypothetical cellular structure with a threshold of z_c the same dose dependence would have to apply. This condition and related matters are considered in Section IV, C.

It is informative to compare the linear and the logarithmic representations in Fig. 10. In the linear diagram the distances of the random paths to the line $z = D$ tend to increase with increasing dose; this corresponds to the increasing standard deviation of z as D increases. As stated in Eq. (33) and derived in Section III, D, the standard deviations of z are proportional to $\sqrt{D}$. In the logarithmic representation the distances to the diagonal tend to decrease, as they correspond to the *relative* standard deviation, which is inversely proportional to $\sqrt{D}$. The dependence of the standard deviation of z on absorbed dose is further illustrated in the third panel, where $\sqrt{z}$ is plotted versus $\sqrt{D}$. In this case the distances to the diagonal tend to be independent of D .

The individual random paths in Fig. 10 illustrate the stochastic nature of energy deposition in microscopic regions. However, to give the full information contained in dose-dependent microdosimetric distributions one would have to utilize suitable plots of the function $F(z; D)$ or its complement, the function $G(D; z)$. Such plots have been produced [33] and Fig. 11 gives an example. Graphs of this type are suitable for considerations that require actual numerical values of the probabilities to reach or exceed certain specific energies at given values of absorbed dose; they also permit the construction of dose-effect relations for assumed threshold reactions. In the present context, however, it is helpful to visualize the character of the distributions in a less formalized way. To this purpose the analogs to Fig. 10 are given in Fig. 12 as scatter diagrams. For these diagrams a large number of simulated exposures of the spherical tissue region of $6 \mu\text{m}$ by 15-MeV neutrons is used. Each dot represents the outcome of a simulated exposure. For a specified absorbed dose D a random value z of specific

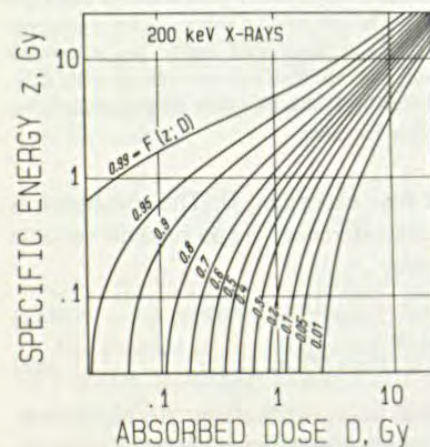


Fig. 11. A representation of the dose-dependent distributions of specific energy in terms of lines of equal values of the function, $F(z; D)$. The parameter on the curves gives the value $F(z; D)$ or its complement $1 - F(z; D)$. (Redrawn from Kellerer [33].)

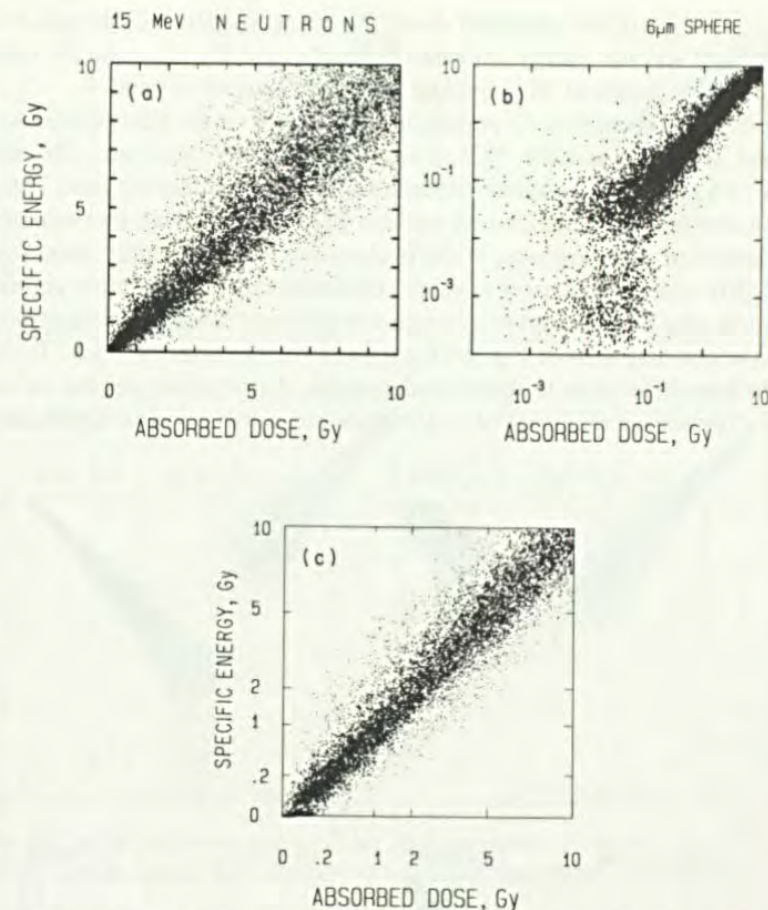


Fig. 12. Scatter diagram of the distribution of specific energy at specified absorbed doses in spherical tissue regions of $6\text{-}\mu\text{m}$ diameter exposed to 15-MeV neutrons. In analogy to Fig. 10, (a) linear scales, (b) logarithmic scales, and (c) square-root scales of absorbed dose and specific energy are used. In each diagram a large number of dose values are uniformly distributed on the scale that is being used. Each dot represents the value of specific energy from a random simulation of the exposure with the specified absorbed dose. The reduction of the number of points at low doses reflects the increasing probability for zero events that are not visible in the graph. This and subsequent scatter diagrams are obtained by the algorithm described in the Appendix.

energy is computed and is represented by the corresponding point in the D - z plane. Dose values are randomly selected in such a way that they are uniformly distributed along the abscissa that is used in the representation. The scatter diagrams permit the visualization of the densities of specific energy as function of absorbed dose. Since the zero events are not represented, fewer points appear on the left-hand side of the graphs, where the event probabilities are substantially less than unity. The essential point comes out most clearly in the logarithmic

plot: at sufficiently low absorbed doses the event frequencies decrease but not the values of specific energy. At small doses they merely represent the distribution of values produced in individual energy deposition events.

The four panels of Fig. 13 permit the comparison of the distributions for two different site sizes and for ^{60}Co γ rays and 15-MeV neutrons. The site diameter of $6\text{ }\mu\text{m}$ is chosen to approximate the size of a cellular nucleus. The average volume of a mammalian cell nucleus exceeds somewhat the volume of a $6\text{-}\mu\text{m}$ spherical site; however, if the nucleus is a spheroid rather than a sphere the slightly reduced diameter is more representative for the actual geometry.

A somewhat more complete synopsis for different radiation qualities is given in the various diagrams of Fig. 14 for a fixed site diameter of $1\text{ }\mu\text{m}$. To recognize the fine differences in the microdosimetric distributions one has to consult Fig. 8. The scatter diagrams of z , D -values are suitable for an appreciation of

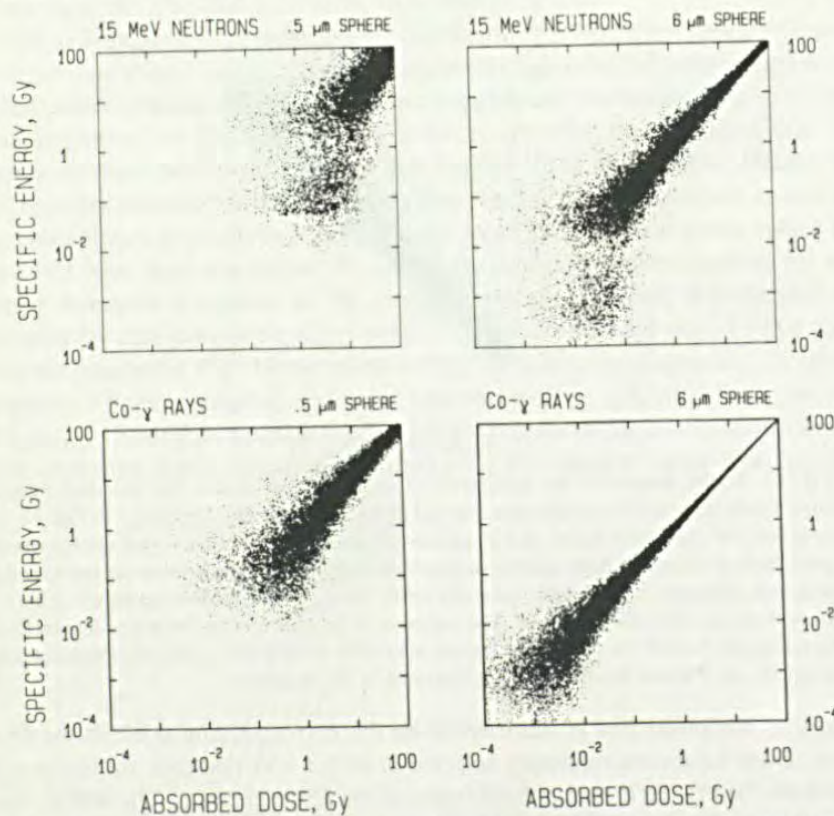


Fig. 13. Scatter diagrams for a comparison of z distributions in small sites and in sites that correspond roughly to the diameter of the nucleus of a cell ($6\text{ }\mu\text{m}$). Results are given for ^{60}Co γ rays and 15-MeV neutrons. Here and in Fig. 14 each panel contains 4000 values per decade of D , i.e., 24,000 simulations are utilized per graph. The actual number of points is considerably less at low doses because the events with $z = 0$ are not visible.

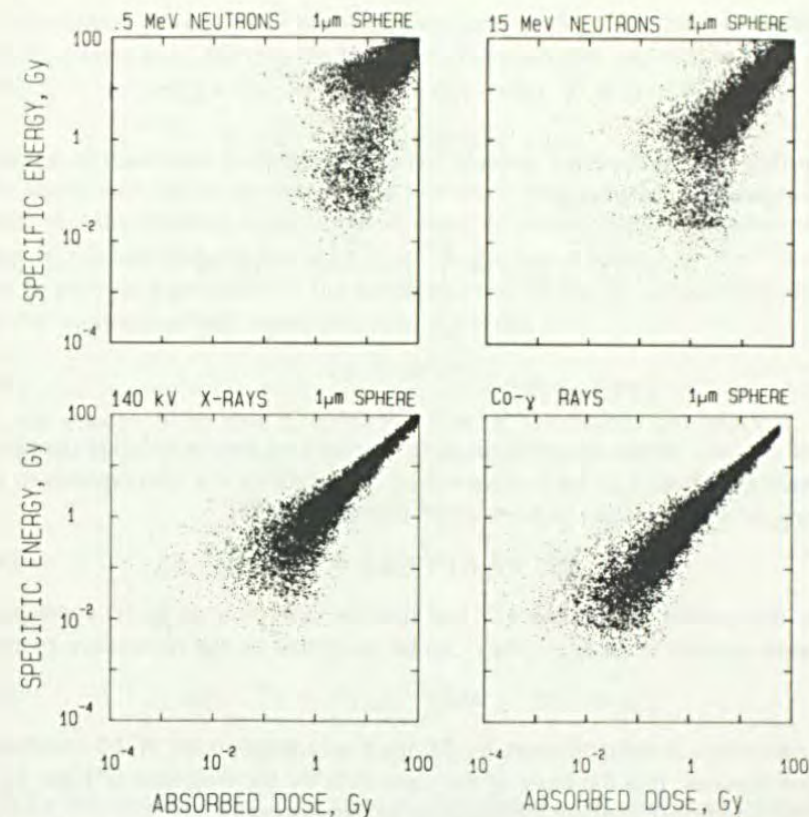


Fig. 14. Scatter diagrams as in Fig. 13 but for a comparison of 140-kV x rays, ^{60}Co γ rays, 0.55-MeV neutrons, and 15-MeV neutrons for a spherical tissue site of $1\text{-}\mu\text{m}$ diameter.

the general features of the distributions and of their similarities for sparsely ionizing radiations on the one hand, and densely ionizing radiations on the other.

In the early microdosimetric studies Eq. (40) was utilized [29] to compute the dose-dependent distributions. However, this approach is inconvenient because a large number of convolutions $f_n(z)$ is required. It is therefore more efficient to base numerical evaluations on another relation, fundamental to microdosimetry, which will be discussed in the next section.

C. AN ADDITIVITY RELATION AND THE RESULTING SOLUTION

Because the convolution operation is fundamental in probability theory, and because it has all the characteristics of multiplication, it is convenient to abbreviate the integral. One writes

$$f_n(z) = \int_0^z f_1(x) \cdot f_{n-1}(z-x) dx = f_1(z) * f_{n-1}(z) \quad (43)$$

or generally

$$f_{\nu+\mu}(z) = \int_0^z f_\nu(x) \cdot f_\mu(z-x) dx = f_\nu(z) * f_\mu(z) \quad (44)$$

In a further step the repeated convolution of a distribution with itself is denoted by an exponent. For example,

$$f_1^{*2}(z) = f_2(z) = \int_0^\infty f_1(z-x) \cdot f_1(x) dx \quad (45)$$

or

$$f_1^{*\nu}(z) = f_\nu(z) \quad (46)$$

One can also utilize the multiplicative character of the convolution operation to obtain $f_\nu(z)$ from $f_1(z)$ by a sequence of convolutions that corresponds to the splitting of ν into integer powers of 2. For example,

$$f_{84}(z) = f_4(z) * f_{16}(z) * f_{64}(z) = f_{|2|} * f_{|4|} * f_{|6|} \quad (47)$$

where the symbol $| \nu |$ indicates 2^ν and thus the convolutions $f_{|\nu|}(z)$ correspond to integer powers 2^ν of $f_1(z)$. They can be computed by the recurrence relation

$$f_{|0|} = f_1(z) \quad \text{and} \quad f_{|\nu|}(z) = f_{|\nu|-1}^{*2}(z) \quad (48)$$

This procedure is very efficient for Monte Carlo simulations of the compound Poisson process. It is the basis of the algorithm for the diagrams of Figs. 12 to 14. The numerical method is explained in the Appendix.

The computation of the distribution $f(z; D)$ for a specified dose requires a somewhat modified algorithm. It can, however, be based on the same principle. The convolution relation applies not only to distributions of z for specified numbers ν of events. It holds equally for the dose-dependent distributions. In the Poisson process the number and magnitude of events during two time periods, or due to two absorbed doses, are independent. The distribution of the sum of the two random variables equals the convolution of their probability distributions. Hence the specific energy at absorbed dose $D_1 + D_2$ has the distribution

$$f(z; D_1 + D_2) = \int_0^z f(z-x; D_1) f(x; D_2) dx = f(z; D_1) * f(z; D_2) \quad (49)$$

and specifically

$$f(z; D) = f(z; D/2) * f(z; D/2) \quad (50)$$

Starting from an approximation of $f(z; D)$ that is valid at low doses, and gaining a factor of 2 in D with each convolution, one can then derive the distributions for arbitrary doses. The approximation of the z distribution is simple at doses

that correspond to very small event frequencies. At the small dose $\eta = \epsilon \bar{z}_p$ the event frequency (i.e., the expected number of events) is ϵ , and with $\epsilon \ll 1$ one has

$$f(z; \eta) = (1 - \epsilon)\delta(z) + \epsilon f_1(z) \quad (51)$$

if the terms with higher powers of ϵ are omitted. In the next subsection it will be seen that the resulting error of the standard deviation of the computed distribution of z is less than the factor $(1 - \epsilon)$. In practice a value $\epsilon < 10^{-2}$ is adequate to provide a precision of the numerical results that is considerably better than the accuracy of any input data $f_1(z)$. One can set

$$\eta = D/2^N \quad (52)$$

and can choose N so that $\epsilon < 10^{-2}$. With N successive convolutions one reaches the desired distribution for the absorbed dose D :

$$\begin{aligned} f(z; 2\eta) &= f(z; \eta) * f(z; \eta) \\ f(z; 4\eta) &= f(z; 2\eta) * f(z; 2\eta) \end{aligned} \quad (53)$$

$$\begin{aligned} &\cdot \quad \cdot \quad \cdot \\ &\cdot \quad \cdot \quad \cdot \\ &\cdot \quad \cdot \quad \cdot \end{aligned}$$

$$f(z; D) = f(z; D/2) * f(z; D/2)$$

Formally this procedure of N successive convolutions can also be expressed as

$$f(z; D) = f(z; h) *^{2^N} \quad (54)$$

The process requires relatively few convolutions. For example, a total of 14 convolutions are required to reach distributions that correspond to average event numbers around $n = 100$. The distributions in Figs. 6 and 7 exemplify the procedure.

The Appendix contains the computer algorithm for the solution of the Poisson process in terms of successive convolutions. Because the spectrum of the process [i.e., the distribution $f_1(z)$] can span several orders of magnitude of the random variable, the convolution has to be executed on a suitable scale; a logarithmic scale of z is chosen for the purpose.

D. RELATIONS FOR THE MOMENTS

Frequently the explicit dose-dependent distributions of specific energy are not required. The moments of the distributions or related parameters are of far greater pragmatic importance. They can be expressed in terms of the moments of the single-event distribution. The second moment and the variance of z play

the greatest role in various applications; they are therefore derived first. The more complicated relations for the higher moments are given subsequently. They are less relevant to radiobiological applications of microdosimetry, but they can be useful tools in any quantitative assessment of experimental or computed microdosimetric data and of their interrelations.

1. Second Moment and Variance

The expectation of specific energy is, as pointed out in Sections II, E and II, F, equal to absorbed dose:*

$$\bar{z} = n\bar{z}_1 = D \quad (55)$$

The variance $\sigma_z^2(D)$ of the specific energy z at absorbed dose D is readily obtained. One utilizes the fact that the variance of the sum of two independent random variables is equal to the sum of their variances, i.e., that the variances are additive in the convolution of two distributions. It follows that the variance of z at dose $D_1 + D_2$ is equal to the sum of the variances at dose D_1 and at dose D_2 . The variance must, accordingly, be proportional to absorbed dose:

$$\sigma_z^2 = cD \quad (56)$$

As a next step one can derive the constant c . The variance of a random variable is equal to the second moment minus the square of the expectation value:

$$\sigma_z^2 = (\overline{z^2} - \bar{z})^2 = \overline{z^2} - \bar{z}^2 = \overline{z^2} - D^2 \quad (57)$$

The second moment can be expressed in terms of Eq. (40):

$$\overline{z^2} = \sum_{\nu=0}^{\infty} e^{-n} \frac{n^{\nu}}{\nu!} \int_0^{\infty} z^2 f_{\nu}(z) dz = e^{-n} \sum_{\nu=1}^{\infty} \frac{n^{\nu}}{\nu!} \overline{z_{\nu}^2} \quad (58)$$

where $\overline{z_{\nu}^2}$ is the second moment of the ν -event distribution $f_{\nu}(z)$; in contrast to $\overline{z^2}$, the values $\overline{z_{\nu}^2}$ are not dependent of dose.

The power expansion of Eq. (58) is

$$\begin{aligned} \overline{z^2} &= (1 - n + \frac{1}{2}n^2 - \dots) \cdot (n\overline{z_1^2} + \frac{1}{2}n^2\overline{z_2^2} + \dots) \\ &= \overline{z_1^2}n + (\overline{z_2^2} - \overline{z_1^2})n^2 + \dots \\ &= (\overline{z_1^2}/z_1)D + [(\overline{z_2^2} - \overline{z_1^2})/\bar{z}_1^2] \cdot D^2 + \dots \end{aligned} \quad (59)$$

*In the context of this and the following subsection it is practical to utilize the notation $\bar{z}_1$ and $\bar{z}_1^2$, rather than the notation $\bar{z}_F$ and $\bar{z}_F^2$, which is employed whenever the discrimination of the frequency distributions (index F) from the dose-weighted distributions (index D) is essential. For the expectation values at a specified absorbed dose no index is used and the argument D is omitted whenever the meaning is clear from the context. Thus $\bar{z}$, $\bar{z}^2$, and σ_z^2 stand for $\bar{z}(D)$, $\bar{z}^2(D)$, and $\sigma_z^2(D)$, respectively.

Accordingly, one has

$$c = \sigma_z^2/D = \overline{z_1^2}/\bar{z}_1 + [(\overline{z_2^2} - \bar{z}_1^2)/\bar{z}_1^2] \cdot D + \dots \quad (60)$$

Since c is a constant, one can obtain its value from the limit $D \rightarrow 0$:

$$c = \lim_{D \rightarrow 0} c = \frac{\overline{z_1^2}}{\bar{z}_1} \quad (61)$$

Thus one obtains the essential relations

$$\sigma_z^2 = (\overline{z_1^2}/\bar{z}_1) \cdot D \quad \text{and} \quad \bar{z}^2 = (\overline{z_1^2}/\bar{z}_1) \cdot D + D^2 \quad (62)$$

that were cited without derivation in Section II. F.

In view of important radiobiological applications (see Section IV. C), the term $\overline{z_1^2}/\bar{z}_1$ has been given a special symbol:

$$\zeta = \overline{z_1^2}/\bar{z}_1 \quad (63)$$

As pointed out in Section II, F, this is the mean value of the dose-weighted single-event distribution $d_1(z)$.

2. Utilization of the Relation for the Variance

Section IV deals with various applications of microdosimetry and Eq. (62) plays a prominent role in these applications. Two specific applications are of interest already in the immediate context of the present section.

Equation (62) can be utilized to assess the error that is caused by the omission of the multiple-event terms in the low-dose approximation in Eq. (51). The mean value of this approximation is correct:

$$\bar{z} = \epsilon \int z f_1(z) dz = \epsilon \bar{z}_1 = \eta \quad (64)$$

However, the second moment is somewhat smaller than the exact value $\zeta\eta + \eta^2$. It is instead

$$\bar{z}^2 = \epsilon \int z^2 f_1(z) dz = \zeta\eta \quad (65)$$

For σ_z^2 one has therefore $\zeta\eta - \eta^2$ instead of the correct value $\zeta\eta$. The standard deviation of the approximation used in Eq. (51) is, accordingly, too small by the factor

$$f = (1 + \epsilon\bar{z}_1/\zeta)^{1/2} \quad (66)$$

The dose average ζ of the single-event distribution is always larger than the frequency average $\bar{z}_1$, and the error factor is therefore substantially closer to unity than $(1 - \epsilon/2)$. Hence the condition $\epsilon < 10^{-2}$ ensures adequate precision of the iterative convolution algorithm described in the preceding subsection.

The second consideration relates to the relative role of the fluctuations of the event number and the event size in the compound Poisson process. In Section III, A the statement has been made that the variations of event size are the dominant factor. To quantify this statement it is practical to consider the *relative* variance $V_z = \sigma_z^2(D)/D^2$ of the distribution $f(z; D)$ and express it in terms of the relative variance $V_1 = \sigma_1^2/\bar{z}_1^2$ of the single-event spectrum:

$$V_z = (V_1 + 1)/n \quad (67)$$

Let $n = D/\bar{z}_1$ be the mean event number at dose D ; for simplicity D may be assumed to be an integer multiple of $\bar{z}_1$. Without fluctuations of the event number one would obtain the distribution $f_1^{*n}(z)$ instead of $f(z; D)$. The variance of this distribution is

$$\sigma_z^2(n) = n\sigma_1^2 \quad (68)$$

The relative variance is

$$V_z' = \sigma_z^2(n)/D^2 = V_1/n \quad (69)$$

where $1/n$ is the relative variance V_p of the number of events. Therefore, V_z is the sum of the term V_1/n , due to the fluctuations of event size, and the term $1/n$, due to the variations of event number. For the single-event spectra V_1 is commonly considerably larger than one; the fluctuations of event size are, therefore, more important than the variations of event number. The same statement applies to the energy-loss straggling problem that is treated in Section V.

3. General Relations for the Moments

The remainder of this subsection has a somewhat more mathematical character. It deals with the higher moments for the compound Poisson process. The results—although applicable to a variety of problems—are not required in the subsequent sections.

The higher moments could be obtained by a method largely analogous to the considerations in the preceding section. The subsequent, less elementary treatment has, however, the advantage that it uses concepts and relations that are also of interest and utility in themselves. There is particular relevance to the energy-loss straggling problem that is considered in Section V.

The first step in the derivation is the introduction of certain combinations of the moments that are termed *semi-invariants* and that are additive in convolutions. Let $\phi(t)$ be the characteristic function, i.e., the Fourier transform, of the probability density $f(z)$:

$$\phi(t) = \int_0^\infty e^{itz} f(z) dz \quad (70)$$

2. FUNDAMENTALS OF MICRODOSIMETRY

The power expansion of $\phi(t)$ is

$$\begin{aligned} \phi(t) &= \sum_{\lambda=0}^{\infty} \int_0^\infty \frac{(it)^\lambda}{\lambda!} z^\lambda f(z) dz \\ &= \sum_{\lambda=0}^{\infty} m_\lambda \frac{(it)^\lambda}{\lambda!} \end{aligned} \quad (71)$$

where m_λ are the moments of the distribution $f(z)$.

A different representation of $\phi(t)$ can be obtained by expanding the logarithm of $\phi(t)$ into a power series:

$$\ln \phi(t) = \sum_{\lambda=1}^{\infty} k_\lambda \frac{(it)^\lambda}{\lambda!} \quad (72)$$

The resulting coefficients k_λ are termed semi-invariants. In the convolution of two distributions the logarithms of the characteristic functions are additive, and so are the semi-invariants.

To express the semi-invariants in terms of the moments m_λ one can juxtapose the expressions from Eqs. (71) and (72):

$$\begin{aligned} \sum_{\lambda=0}^{\infty} m_\lambda \frac{(it)^\lambda}{\lambda!} &= \exp \sum_{j=1}^{\infty} k_j \frac{(it)^j}{j!} \\ &= \sum_{\mu=0}^{\infty} \left(\sum_{j=1}^{\infty} k_j \frac{(it)^j}{j!} \right)^\mu / \mu! \\ &= 1 + k_1(it) + (k_2 + k_1^2) \frac{(it)^2}{2!} \\ &\quad + (k_3 + 3k_1k_2 + k_1^3) \frac{(it)^3}{3!} + \dots \end{aligned} \quad (73)$$

The comparison of the coefficients and resolution for the k_i yields

$$\begin{aligned} k_1 &= m_1 \\ k_2 &= m_2 - m_1^2 = \sigma_2 \\ k_3 &= m_3 - 3m_2m_1 + 2m_1^3 = \sigma_3 \\ k_4 &= m_4 - 4m_3m_1 - 3m_2^2 + 12m_2m_1^2 - 6m_1^4 = \sigma_4 - 3\sigma_2^2 \end{aligned} \quad (74)$$

where σ_λ are the central moments:

$$\sigma_\lambda = \int_0^\infty (x - \bar{x})^\lambda f(x) dx \quad (75)$$

The relations between the central moments σ_λ and the noncentral moments m_λ that are used in Eq. (74) are readily verified. Equation (74) shows that, next to the mean and the variance, the third central moment is additive in a convolution and generally in a Poisson process.

It remains to express the semi-invariants in terms of the moments of the single-event spectrum. The relations for k_1 and k_2 have already been obtained in the preceding subsection:

$$\begin{aligned} k_1 &= n\bar{z}_1 = D \\ k_2 &= n\bar{z}_1^2 = (\bar{z}_1^2/\bar{z}_1)D \end{aligned} \quad (76)$$

The general relation is of the same form:

$$k_\lambda = n\bar{z}_1^\lambda = (\bar{z}_1^\lambda/\bar{z}_1)D \quad (77)$$

This result could be derived in analogy to Eqs. (58)–(61); i.e., one could utilize the limit of $f(z; D)$ as D goes to zero. Instead, a derivation will be used that is instructive because it demonstrates also the solution of the compound Poisson process in terms of the Fourier transform.

For the Fourier transforms the convolution reduces to a multiplication. Accordingly one can write the equation for the compound Poisson process in terms of the transforms:

$$\phi(t; D) = \sum_{n=0}^{\infty} e^{-n} \frac{n^n}{n!} \phi_1^n(t) = e^{n[\phi_1(t) - 1]} \quad (78)$$

or

$$\ln \phi(t; D) = n[\phi_1(t) - 1] \quad (79)$$

where $\phi(t; D)$ and $\phi_1(t)$ are the Fourier transforms of $f(z; D)$ and $f_1(z)$.

Inserting the relation from Eq. (71)

$$\phi_1(t) = \sum_{\lambda=0}^{\infty} \bar{z}_1^\lambda \frac{(it)^\lambda}{\lambda!} = 1 + \sum_{\lambda=1}^{\infty} \bar{z}_1^\lambda \frac{(it)^\lambda}{\lambda} \quad (80)$$

into Eq. (79) one has

$$\ln \phi(t; D) = \sum_{\lambda=1}^{\infty} n\bar{z}_1^\lambda \frac{(it)^\lambda}{\lambda!} \quad (81)$$

From the identity of the coefficients one obtains the result of Eq. (77). The semi-invariants of the compound Poisson process are proportional to the noncentral moments of the spectrum $f_1(z)$, of the Poisson process.

Equation (78) is of interest in itself, since it gives the solution of the Poisson process in terms of the Fourier transforms. The use of this solution and the fast

Fourier-transform algorithm might appear to be the most efficient method of computing the solutions of the compound Poisson process. However, the microdosimetric distributions $f_1(z)$ —and the same applies to the collision spectra $f_c(\epsilon)$, which will be considered in Section V—cover such a broad range of the random variable that the Fourier transform requires impractically large arrays or, alternatively, necessitates a splitting of the spectrum into sections that are separately processed. The method of the Fourier transform is therefore not superior to the direct convolution that can be executed on a suitable logarithmic scale with arrays of moderate size (see Appendix).

IV. Determination and Utilization of the Microdosimetric Parameters

Most microdosimetric measurements are performed with proportional counters filled with tissue-equivalent gas. The instruments are either wall-less or have tissue-equivalent walls. These *Rossi counters* can be utilized under a wide variety of experimental conditions and for a multiplicity of purposes. It is not the objective of this chapter to deal with the experimental methods and with the scientific or practical applications of the microdosimetric data. A brief consideration of certain aspects of the measurements and of some principles of utilization of the results will, however, facilitate the comprehension of the fundamentals of microdosimetry.

The microdosimetric spectra, and particularly the important single-event spectra, cover broad ranges of values of lineal energy or specific energy. It is evident that such spectra cannot, without loss of information, be characterized by their first two moments only. Nevertheless, the majority of the applications of microdosimetry utilize merely these first moments, or the equivalent parameters *frequency average* and *dose average* of event size. There are, of course, radiations—and the most important and interesting case is that of heavy ions, which are not treated in this chapter—where the pairs of values $\bar{y}_F$ and $\bar{y}_D$ or $\bar{z}_F$ and $\bar{z}_D$ ($=\bar{z}$) are grossly inadequate. But these are exceptions and if one does not deal with particles of extremely high LET, such as heavy ions, or with particles of very short ranges, such as the low-energy electrons released by ultrasoft x rays, the two mean values of the single-event spectra are of fundamental importance. This is the justification for the detailed treatment given to the moments of the microdosimetric spectra in the preceding section. For the same reason these parameters will be considered further from a more practical point of view.

A. ASPECTS OF DOSIMETRIC AND MICRODOSIMETRIC MEASUREMENTS

1. Inherent Imprecision of Dose Determinations

The statistical fluctuations of energy deposition set an absolute limit to the precision of dose determinations. The physical size of the sensitive volume of

an instrument determines the expected number of events and its standard deviation. It also determines the expectation value of the energy imparted or specific energy and the standard deviation of these quantities. Even an ideal instrument cannot attain a precision beyond that determined by these standard deviations.

Consider a spherical, gas-filled proportional counter. For simplicity it is assumed that it is tissue equivalent. Let ρ be the density of the counting gas in grams per cubic centimeter. The value ρ then is also the ratio of the density of the counting gas to the density of tissue, which is assumed to be 1 g/cm³. If the counter has the diameter d^* , the diameter of the equivalent tissue region is $d = \rho d^*$. Neglecting density effects and possible wall effects in the measurement, the single-event spectrum in the counter equals the spectrum in the corresponding tissue site. The mass of the counting gas exceeds the mass of the corresponding tissue site by the factor $1/\rho^2$. The event frequency is higher, by the same factor, in the gas cavity than in the tissue site. The expected number of events in the gas is

$$n = \phi \cdot D/\rho^2 \quad (82)$$

where ϕ is the event frequency per unit absorbed dose in the tissue region of diameter d .

The actual number of events follows the Poisson distribution. The relative standard deviation of the number of events is therefore

$$\sigma_n/n = 1/\sqrt{n} = \rho/\sqrt{\phi D} \quad (83)$$

A numerical example can illustrate the relation. Consider a spherical detector with a diameter of 1 cm that is filled with tissue-equivalent gas at atmospheric pressure ($\rho \approx 10^{-3}$ g/cm³) and is exposed to a field of 15-MeV neutrons. The corresponding tissue site has a diameter of 10 μ m and the event frequency for 15-MeV neutrons is roughly 50/Gy (see Table I, Section II). The relative standard deviation in the determination of 1 mGy is then

$$\sigma_n/n = 10^{-3}/\sqrt{50 \cdot 10^{-3}} = 0.0045 \quad (84)$$

This means that even an ideal instrument with the specified size and gas pressure cannot reach a precision of better than 0.5% in determining a dose of 1 mGy of 15-MeV neutrons. Utilization of the detector in the counting mode is, of course, a procedure that requires previous knowledge of the event frequency for the radiation in question. Event frequencies are sensitive to minor modifications of radiation quality. To measure energy imparted is therefore, in general, more appropriate than counting pulses. The relation for the inherent imprecision of a measurement of energy imparted, or of specific energy, is accordingly of main pragmatic interest. It will be considered next.

According to the relation obtained in the preceding section [see Eqs. (62) or the earlier statement in Eq. (33)], the standard deviation of specific energy in

the tissue region is

$$\sigma_z^2 = \zeta D \quad (85)$$

where ζ is the dose-weighted average [see Eq. (28)] of the single-event spectrum. In the gas cavity the number of events is increased by the factor $1/\rho^2$. The variance of the specific energies is proportional to the number of events and is therefore equal to

$$\sigma_z^2 = \zeta D/(1/\rho^2) \quad (86)$$

Hence one obtains the relative standard deviation of the specific energy:

$$\sigma_z/D = \sqrt{\zeta/D}(\rho) \quad (87)$$

In the above example $\zeta = 0.15$ Gy (see Fig. 15). Accordingly, at an absorbed dose of 1 mGy the inherent relative standard error is

$$\sigma_z/D = \sqrt{0.15/10^{-3}}(10^{-3}) = 0.012 \quad (88)$$

The specified instrument can therefore, ideally, attain a precision of 1.2% in a measurement of specific energy or energy imparted at an absorbed dose of 1 mGy of 15-MeV neutrons. This inherent imprecision may appear to be insignificantly low. However, in radiation protection applications, specifically in area monitoring, one commonly deals with very low dose rates. The dose determinations are then associated with correspondingly larger relative errors that may necessitate the utilization of considerably larger detectors.

2. The Variance Method

The weighted mean values $\bar{y}_D$ and ζ of single events are fundamental parameters of radiation quality. It is therefore of particular interest that they can

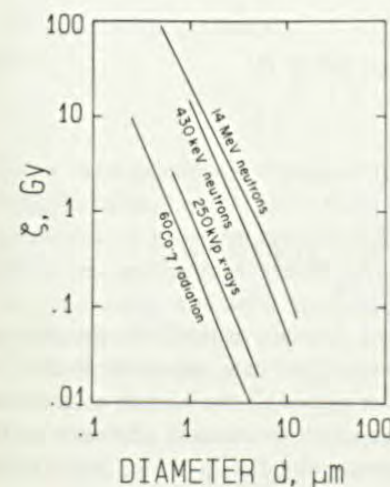


Fig. 15. Weighted mean ζ of the specific energy produced in individual events by different radiations in spherical tissues sites of diameter d . (From Kellerer and Rossi [41].)

be obtained directly without determination of the explicit spectra. The direct measurement is important also because it is applicable to site sizes that are smaller than those for which the spectra themselves can be measured with present experimental techniques.

A Rossi counter measures the values ϵ_k of energies imparted due to successive single events. The average of a sufficient number of the squares of observed values is the second moment $\bar{\epsilon}_D^2$; the average of the observed values is the mean $\bar{\epsilon}_F$. Accordingly one can determine both mean values $\bar{\epsilon}_D$ and $\bar{\epsilon}_F$ from averages over a sufficient number of measured pulses:

$$\bar{\epsilon}_D = \sum_1^K \epsilon_k^2 / \sum_1^K \epsilon_k \quad \text{and} \quad \bar{\epsilon}_F = \frac{1}{K} \sum_1^K \epsilon_k \quad (89)$$

Related quantities are $\bar{z}_D = \bar{\epsilon}_D/m$ and $\bar{y}_D = \bar{\epsilon}_D/\bar{l}$.

Proportional counters cannot simulate sites with diameters less than approximately 0.3 μm . If the gas pressure in the instruments is too low, one cannot achieve sufficient multiplication with adequate uniformity of response throughout the sensitive region. In view of this limitation, Bengtsson [35] and Forsberg and Lindborg [36] have introduced an alternative method that is linked to the fundamental relation for the variance of specific energy. They utilize the fact that one can perform measurements with little or no multiplication, if the detector is set to integrate the response over finite dose increments, D . Let the observed values of energy imparted in repeated measurements with absorbed dose D be denoted by the symbol e_k ($k = 1, \dots, K$). With a sufficient number of measurements the moments of the distribution $f(z; D)$ are obtained as

$$\bar{z}^2 = \frac{1}{K} \sum_1^K \left(\frac{e_k}{m} \right)^2, \quad \bar{z} = D = \frac{1}{K} \sum_1^K \frac{e_k}{m} \quad (90)$$

From the relation [see Eq. (62)]

$$\bar{z}^2 = \sigma_z^2(D) + \bar{z}^2 = \zeta D + D^2 \quad (91)$$

one obtains

$$\begin{aligned} \zeta &= \frac{\sigma_z^2}{D} = \frac{\bar{z}^2}{D} - D \\ &= \frac{\sigma_z^2/\bar{e}}{m} = \frac{\sum_1^K e_k^2 / \sum_1^K e_k - (1/K) \sum_1^K e_k}{m} \end{aligned} \quad (92)$$

This determination of ζ , the *variance method*, depends critically on the absence of experimental fluctuations that contribute to σ_z^2 and can, therefore, lead to erroneously large values of ζ . With variable or pulsed sources, such as accelerators, a modified method is accordingly required. A suitable approach is the utilization of two detectors that work independently, but in phase, in the radia-

tion field. With the twin detectors the parameter ζ is obtained from the variance of specific energy in one of the detectors minus the covariance of specific energy in the two detectors [37]. The *variance-covariance method* exemplifies the fundamental role of the relations between the moments of the microdosimetric distributions. It makes it possible to perform microdosimetric measurements in accelerator-produced radiation fields that are often too intense during pulses to permit the resolution of individual events in microdosimetric detectors.

B. APPLICATION OF MICRODOSIMETRIC DATA IN RADIATION PROTECTION

There are a variety of uses for microdosimetric data. A brief survey exemplifies first some applications to radiation protection. Biophysical implications will be considered subsequently.

Perhaps the most important application of microdosimetric techniques and data is the exploration of unknown or inadequately known radiation fields, or the monitoring for changes in radiation quality. The need for such procedures arises in radiation protection but also in applications of ionizing radiation, such as radiation therapy. For most fields of high-energy photons or neutrons, conventional, walled counters can be utilized to obtain the microdosimetric spectra or their mean values. Only in exceptional situations, for example, with high-energy charged particles, must wall-less counters be used.

For purposes of radiation protection one can derive adequately precise values of the quality factor by substituting spectra of lineal energy for the unknown and often unmeasurable LET spectra, which enter the strict definition of the quality factor. Instead of the relation ([38]; see Section II. B for the notation)

$$\bar{Q} = \int Q(L) d(L) dL = \int Q(L) L t(L) dL / \bar{L}_F \quad (93)$$

one utilizes the equation

$$\bar{Q} = \int Q(y) d(y) dy = \int Q(y) y f(y) dy / \bar{y}_F \quad (94)$$

where the dependence of the quality factor on LET and on lineal energy is taken to be the same, i.e., $Q(y) = Q(L)$ for $y = L$. The numerical differences caused by the use of y spectra instead of LET spectra are of little concern in radiation protection applications. It is therefore feasible to couple a Rossi counter to suitable electronics and to a microprocessor, and to use the device as a dose-equivalent-rate meter.

The deficiencies of the LET concept have been considered in Section II, and it has been noted that LET is less relevant to the biological effectiveness of radiations than the actual energy concentrations that determine the y spectra or z spectra. If one thinks merely in terms of LET, it remains unclear why the quality

factor, according to its present definition [1, 38], should be constant up to 3.5 keV/ μm before it increases with LET. In terms of lineal energy, however, this feature of the definition is readily understood. For densely ionizing radiations the values of lineal energy tend to be proportional to LET. For very sparsely ionizing radiations the values of lineal energy do not decrease proportionally, but tend to be larger than LET. The reason is the energy-loss straggling, i.e., the clustered deposition of energy in the volume by δ rays or groups of δ rays. In the same context one may note that a definition in terms of lineal energy could adequately take into account any increase of biological effectiveness due to Auger cascades or other correlated ionizing particles. The increase cannot be accounted for in terms of the LET concept.

If the quality factor were defined in terms of lineal energy in sites of diameter 0.3 μm [39], the quality factor could be set to be proportional to lineal energy for sparsely ionizing and moderately densely ionizing radiations, and almost the same numerical values would be obtained as with the present definition of Q in terms of LET (see also [10]). Proportionality between the quality factor and y would have the practical advantage that the mean quality factor $\bar{Q}$ for a radiation could be obtained directly from $\bar{y}_D$ without knowledge of the explicit y spectrum. With $Q(y) = qy$ one obtains [see Eq. (94)]

$$Q = \int qy^2 f(y) dy / \bar{y}^2 = q\bar{y}_D = Q(\bar{y}_D) \quad (95)$$

With the definition in terms of LET one cannot, strictly, set $\bar{Q} = Q(\bar{L}_D)$, although this is often done. The preceding considerations apply, of course, only to sparsely ionizing radiations. For very densely ionizing radiations ($L > 100$ keV/ μm or $y > 100$ keV/ μm) nonlinearities enter because of the saturation effect.

Rossi has reexamined the definition of the quality factor [40] and has proposed a change that goes beyond a mere substitution of LET by lineal energy. His modified definition of the quality factors in terms of lineal energy would be in better agreement with radiobiological data than the present quality factors, which do not sufficiently account for the different effectiveness of densely ionizing and sparsely ionizing radiations at low doses.

Beyond its practical uses, microdosimetry has fundamental implications for radiation protection. In radiation protection one discriminates between *stochastic* effects, such as hereditary damage and radiation carcinogenesis, and *non-stochastic* effects, such as skin damage, impairment of fertility, lens opacification, and prenatal malformations. It is an objective of radiation protection to avoid nonstochastic effects entirely. One assumes that this objective can be reached because nonstochastic effects depend on damage to a multiplicity of cells in a tissue, with resultant curvilinear, or even thresholdlike, dose dependence. Stochastic effects, on the other hand, cannot be avoided entirely. One assumes that they result from damage to individual cells and that they exhibit linear

dose dependences at small doses. The aim of radiation protection can therefore be merely to minimize radiation exposures and to reduce thereby the risk of stochastic effects to a level comparable to or substantially below other risks associated with human activities.

The assumption of linear dependences for the stochastic effects of low doses is crucial and specific to the risk considerations for ionizing radiations. Similar assumptions are not common in the consideration of chemical carcinogens. Microdosimetric principles clarify this fundamental difference. The effects of ionizing radiations and their dose dependences reflect the stochastic distribution of energy in individual cells. Chemicals, on the other hand, affect individual cells according to their concentration without substantial random fluctuations.

Knowledge of event frequencies in the nuclei of mammalian cells permits statements where dose dependences for effects on individual cells must be linear. If the dose is smaller than $\bar{z}_F$ the event frequency $D/\bar{z}_F$ is less than 1. Only a few cell nuclei then receive multiple events of energy deposition, and the specific energy is predominantly determined by the single-event spectrum; i.e., it depends on radiation quality and not on absorbed dose. At sufficiently low doses most cells receive no energy deposition; the number of affected cells is proportional to absorbed dose. Rossi has introduced the notion of radiation effects on *autonomous* cells [40], and implies by this term that cells are not influenced by radiation effects on neighboring cells or by radiation effects on the tissue level. For effects on autonomous cells—and the mutations that cause hereditary effects are the main example—linear dose-effect relations must be assumed at sufficiently low doses. Figure 16 indicates for different types of ionizing radiations those combinations of doses and site diameters that correspond to event frequencies less than 1.

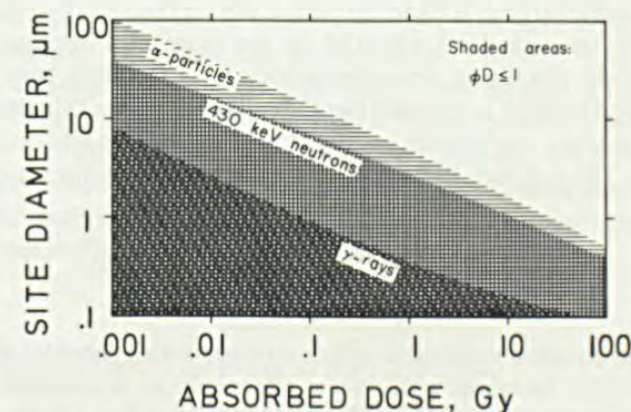


Fig. 16. Diagram of site diameters and of absorbed doses that correspond to mean event frequencies ϕD less than one. The regions with $\phi D \leq 1$ are indicated for three different radiations.

However, it must be noted that linear dose-effect relations cannot be postulated for radiation carcinogenesis, which may be codetermined by radiation-induced alterations on the tissue level. Nonlinear dose-effect relations have been found for tumors induced by 400 keV neutrons. Remarkably, these dependences correspond to dose exponents less than 1 at neutron doses where only a minor fraction of all cells receive any energy deposition. One must conclude that the effect is not one on autonomous cells; other, more complex factors must codetermine the dose-effect relation [41-43, 20]. The nature of these factors is still unknown; microdosimetry can serve only to exclude certain mechanisms of radiation action.

C. BIOPHYSICAL IMPLICATIONS OF MICRODOSIMETRIC DATA

The subsequent remarks deal with the somewhat more complex problem of the dose-effect relations at higher doses, where multiple events occur in the nucleus of the cell. The classical multihit or multitarget theory had no actual validity. Nevertheless, its equations were, and are still, widely used for the description of observed dose-effect relations. The simple target-theory models can also illustrate valid arguments that can be recast in the conceptual and quantitative framework of microdosimetry.

1. The Equations of the Multihit and Multitarget Theory

The target theory had considerable success in explaining the single-hit inactivation of microorganisms. It was, therefore, natural that an attempt was made [2-4] to explain also curvilinear dose-effect relations in terms of random hits in hypothetical targets. An evident weakness of the approach has been the unrealistic assumption of equal, statistically independent hit events and of hypothetical, equal targets. If these unrealistic assumptions were to be dropped, the number of free parameters would be far too large for a meaningful use of the models. Only two highly simplified models have therefore been retained.

In the *multihit model* it is assumed that the cell contains one hypothetical critical target. The target can tolerate $(n - 1)$ hits. If n or more hits occur the cell is inactivated. This is a pure Poisson process. The probability for survival of the cell is

$$S(D) = e^{-\alpha D} \sum_{p=0}^{n-1} \frac{(\alpha D)^p}{p!} \quad (96)$$

For $n > 1$ this equation yields dose-effect relations with a shoulder in an initial region where $S(D)$ decreases with dose at a less than an exponential rate. The slope of the shoulder increases with increasing hit number n . To reconcile the equation with the common observation of an initial slope of the cellular survival curves an additional factor $e^{-\gamma D}$ is multiplied into the survival probability; it is

2. FUNDAMENTALS OF MICRODOSIMETRY

interpreted as single-hit inactivation by the densely ionizing component of the radiation:

$$S(D) = e^{-(\alpha + \gamma)D} \sum_{p=0}^{n-1} \frac{(\alpha D)^p}{p!} \quad (97)$$

The multi-hit curves have the asymptotic slope $-\alpha$ or $-(\alpha + \gamma)$ in the common representation of the logarithm of survival versus dose. The curves do not (for $n > 1$) have an asymptotic tangent; in the terminology of radiation biology they are said to have an infinite extrapolation number.

The *multitarget model* postulates m hypothetical, equal targets, each of which can be inactivated by a single hit. Equal hit probabilities are assumed and also statistical independence of the hits on individual targets. The resulting equation for the survival of the cell is

$$S(D) = 1 - (1 - e^{-\alpha D})^m \quad (98)$$

Again one introduces, somewhat artificially, an additional exponential term to account for the observed initial slope of most survival curves:

$$S(D) = e^{-\gamma D} [1 - (1 - e^{-\alpha D})^m] \quad (99)$$

The multitarget curves have similar characteristics as the multihit curves. Their asymptotic slope is also $-\alpha$ or $-(\alpha + \gamma)$, but this slope is attained sufficiently fast that the curves have an asymptotic tangent that intersects the ordinate at the finite value m . The term *extrapolation number* is currently used rather than the term *target number* for this value m .

The target-theory equations are obviously crude approximations that can have only heuristic value. A multiplicity of assumptions could readily be modified, and a far more general description in terms of Markov processes contains the conventional equations as simple special cases [34]. The more general stochastic treatment presents no mathematical difficulties, but any realistic analysis requires an adequate consideration of the microscopic patterns of energy deposition for different radiations. The micropatterns determine the striking differences of the biological effectiveness of ionizing radiations, and it is evident that a biophysical analysis without microdosimetric data can have little value. It is, therefore, of interest to examine how the basic ideas of target theory can be translated into the language of microdosimetry.

2. The Threshold Model in Terms of Microdosimetry

A dose-effect relation is the expression of several stochastic processes. The energy deposition in the cell varies randomly, but the response of the cell itself, too, must be described in probabilistic terms. The multihit or multitarget models

disregard the inherent stochastic response of the cell; the dose-effect relation is interpreted merely in terms of the statistics of energy deposition. Such an interpretation is obviously incomplete and even unrealistic, and the same objection applies to any threshold model that is formulated in terms of microdosimetry. Nonetheless, one can draw certain valid conclusions from the consideration of the threshold model that can then lead to a more realistic treatment.

One could imagine a microscopic dosimeter that registers specific energy in a spherical tissue region of diameter d and responds precisely when a critical value z_c is reached. As explained in Section III, B, the probability for no response $S(D)$ is then

$$S(D) = \int_0^{z_c} f(z; D) dz = F(z_c; D) \quad (100)$$

Figure 17 represents such dependences for x rays, a spherical site of $1 \mu\text{m}$, and for various critical values z_c . In spite of the assumed threshold reaction, the resulting dose dependence is, of course, not a step function. If z_c is comparable to, or smaller than, the average event size $\bar{z}_F$ for the radiation, the dependence is nearly exponential; one deals then with a *single-hit reaction*. For larger values z_c the dose-response relations have a shoulder; i.e., they are the result of cumulative damage. For a specified z_c the curves come closer to the step function as larger site diameters d are assumed. For any observed dose-effect relation $S(D)$ one can determine the values z_c and d that make the function $F(z; D)$ agree most closely with $S(D)$. The functions $F(z; D)$ and $S(D)$ can be formally regarded as sum distributions of the dose required to produce the effect. As with any distribution, one can compute for the functions a mean $\bar{D}$ and a standard deviation (see Hug and Kellerer [34]). Since there are only two free parameters z_c and d it is sufficient to fit the two curves in terms of these two parameters. The diameter d obtained in this way is not the actual size of any critical target

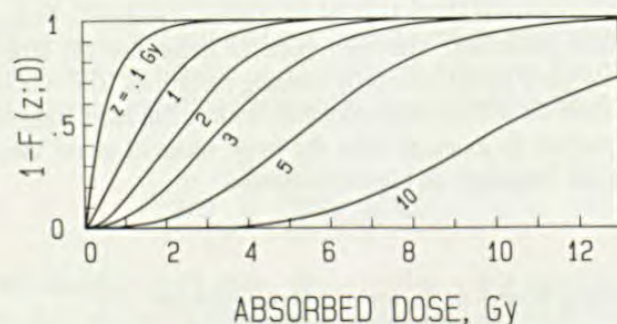


Fig. 17. The probabilities $1 - F(z; D)$ not to exceed a critical value z of the specific energy versus absorbed dose in a spherical tissue site of $1\text{-}\mu\text{m}$ diameter exposed to x rays. (Redrawn from Hug and Kellerer [34].)

or gross sensitive volume in the cell, it is merely a lower bound. The sensitive volume may be larger, or a multiplicity of small targets may be dispersed through a larger volume. Only part of the observed deviation from the step function need then be due to the stochastics of energy deposition; the other part could reflect the stochastic response of the cell, even to a fixed specific energy. On the other hand, one can exclude any model that invokes one smaller critical target or a multiplicity of critical targets contained in a region of diameter less than d ; even under the hypothetical assumption of a threshold reaction, the deviations from the step function would have to be larger than those observed. By this argumentation gross sensitive volumes not smaller than $1 \mu\text{m}$ have been inferred [34] from survival curves of mammalian cells exposed to x rays. For cellular inactivation studies this result is in agreement with the accepted belief that DNA, dispersed through the whole nucleus, is the target of radiation action. For more specific radiation effects, such as chromosome aberrations or cellular transformations, this argument by exclusion can be more informative.

3. Dependence of the Effect on Dose and on Specific Energy

In principle it is easy to generalize the formulation. Let $S(D)$ be the probability for no effect versus dose and let $S(z)$ be the same probability versus specific energy in the nucleus. The function $S(z)$ must be somewhat steeper than the function $S(D)$. The relation between the $S(D)$ and $S(z)$ is

$$S(D) = \int_0^{\infty} S(z) f(z; D) dz \quad (101)$$

If the dose-effect relation were known with sufficient precision, one could utilize $f(z; D)$ to invert the equation and determine the dependence on specific energy. In practice this is impossible because dose-effect relations are not known well enough to permit the inversion of the equation. Even if the numerical procedure were possible, the dependence of the effect on specific energy in the nucleus may provide not much more information than the dependence of the effect on absorbed dose. Thus, for sparsely ionizing radiations the fluctuations of z relative to the entire nucleus of a mammalian cell are insignificant at doses of a few grays (see Fig. 13). The effects of ionizing radiations on the cell are influenced by energy fluctuations on a smaller scale.

The attempt has therefore been made, in various early applications of microdosimetry, but also in recent studies, to search for combinations of smaller site sizes and of functions $S(z)$ that would fit observed dose-effect relations. However, it is evident that this approach, in addition to various other limitations, contains too many free parameters. An infinite number of fits is possible, which provides no real information. A simpler and less formalistic approach is, therefore, required. Such an approach can be illustrated in terms of the model of *dual radiation action*.

4. Treatment of a Second-Order Process

Lea (see [5]) and later Neary *et al.* [44] have examined the condition that cellular lesions result from the interaction of pairs of chromosomal lesions within the nucleus of the cell. They inferred a dependence of the yield of lesions on the square of the specific energy in certain sites, and they deduced from curvilinear dose dependences for sparsely ionizing radiations site sizes of a fraction of a micrometer or more. The microdosimetric analysis of dose dependences for a large variety of radiation effects and the study of RBE dose dependences for neutrons later led to largely similar conclusions [45]. In particular, high RBE values for neutrons were predicted that were subsequently verified in experiments on radiation carcinogenesis, life shortening, and chromosome aberrations. Details of these studies are not the subject matter of this chapter, but the basic microdosimetric relations for the treatment of dual radiation action can be given.

If the yield of a particular cellular damage is proportional to the square z^2 of the specific energy within a certain site, the average yield will be proportional to the mean of the square $\overline{z^2}$ of the specific energy, i.e., to the second moment of $f(z; D)$:

$$E(D) = k\overline{z^2} \quad (102)$$

With the fundamental relation [see Eq. (62) or Eq. (33)]

$$\overline{z^2} = \sigma_z^2 + \bar{z}^2 = \zeta D + D^2 \quad (103)$$

one obtains

$$E(D) = k(\zeta D + D^2) \quad (104)$$

Under the assumed dependence on the square of specific energy, the effect is, therefore, a linear-quadratic function of absorbed dose. The magnitude of the linear component is proportional to the weighted average ζ of the specific energy produced in individual events. ζ is largest for small sites and for densely ionizing radiations (see Fig. 15).

The existence of the linear component can be understood in terms of the intratrack interaction of sublesions; i.e., it can be due to the fact that a substantial amount of energy is deposited in the site even in single events. Accordingly, the energy concentration, and therefore z^2 , can be large in those sites that are affected by an event, even if the absorbed dose is small. The number of sites affected by an event is proportional to absorbed dose.

The treatment in terms of the site concept has the advantage that the specific energy and its distribution can be measured for any radiation. However, it is obviously a simplification. Radiation products will not react with equal probability with all potential reaction partners within an imagined site. The probability for the reaction will, instead, depend on the separation of the partners. In those ex-

ceptional cases where short-range electrons or heavy recoils produce the same z in a site as long-range particles, the cellular effects may be markedly different. It has also become clear from recent radiobiological investigations that short-distance interactions predominate in the intratrack effect while larger distances are involved in the cumulative cellular damage from independent charged particles (see Section VI, E).

V. The Straggling Problem and the Single-Event Spectrum

The present section deals with essential features of the single-event spectrum and with the different factors that determine it. Particular attention is given to energy-loss straggling, a problem that is of interest not only in microdosimetry but also in high-energy physics.

For sites of diameter less than roughly 300 nm no reliable methods exist for measuring the microdosimetric spectra. However, computations are possible. To refer to the case of very small sites the term *nanodosimetry* has sometimes been used. In this small region of interest, charged particles undergo few collisions. The fluctuations of *energy loss* can be computed. They are the result of a compound Poisson process, and the convolution algorithm of Section III can be utilized instead of the approximate solutions of the energy-loss straggling problem given by Landau [46], and later by Vavilov [47]. This is the topic of Section V, A.

The computation of the *energy imparted* to submicroscopic sites is more complicated. When the ranges of the δ rays are comparable to the site size, a substantial part of the energy lost by the charged particle is transported out of the site. Due to the complex configurations of δ rays, i.e., secondary electrons, exact computations are nearly impossible. One can either apply approximate corrections to the solution of the straggling problem, or one can utilize Monte Carlo simulations. The modified solution of the straggling problem is dealt with in Section V, A. Technical details of Monte Carlo simulations are outside the scope of the present chapter, and are therefore not considered. Essentials of the sampling procedures and fundamental relations for the mean values of the spectra will, however, be treated in Section VI; some of the relations hold regardless of the complexities of the particle tracks.

For larger regions with diameters exceeding several hundred nanometers the single-event spectra can be measured. Their calculation is also simpler, and certain complicating factors that are essential for small regions can be disregarded. One condition of particular interest is treated in detail in Section V, B. This condition of *short track segments* requires that the ranges of the charged primaries*

*The term *charged primaries* denotes—if one disregards certain special cases—all ionizing charged particles except the δ rays. Electrons liberated by photons, or recoil nuclei set into motion by neutrons, are regarded as charged primaries.

are much larger than the site diameter. The change of LET of the particle while traversing the site can then be neglected, and one deals with a set of random factors that can largely be separated. These factors are the LET of the particle, the chord length of the trajectory in the site, and, last but not often predominantly, the energy-loss straggling. The case of short track segments applies to moderate- or high-energy electrons, to fast heavy ions, and to the proton recoils of fast neutrons, for regions of fractions of a micrometer to several micrometers.

Another case of considerable pragmatic interest is the condition of *large track segments*. In this case the ranges of the charged primaries are comparable to or smaller than the dimensions of the site. The particles then lose substantial fractions of their energy while traversing the site, and the change of LET during the traversal must be taken into account. Incomplete traversals can also occur. For heavy particles—but usually not for electrons—it is then possible to disregard energy-loss straggling and the transport of energy by δ rays. The reason is that the maximum ranges of δ rays are small compared to the particle ranges and site diameters (see Fig. 3). Numerous computations of this type have been performed for neutrons (see, e.g., [50–52]), and a detailed discussion is therefore not required in the present context.

Energy-loss straggling is so commonly a dominant factor in microdosimetry that one can term the microdistribution of energy the result of LET plus straggling. The treatment of LET is relatively straightforward. The straggling problem demands a separate treatment that pertains also to work with particle detectors in nuclear physics.

A. ENERGY-LOSS STRAGGLING

The straggling problem will be treated for the case of short segments of particle tracks. The change in the kinetic energy and LET of the particles can then be disregarded. In microdosimetry one is interested in the *energy imparted* to a specified region, rather than the *energy lost* by the charged particle. The difference can be substantial whenever δ rays have ranges comparable to the region of interest. The straggling problem will first be treated in terms of energy lost by the particle. Subsequently the modifications will be considered that are required when one considers energy imparted to the site.

1. The Collision Spectrum and Its Convolutions

The distribution $f_s(\epsilon, \Delta)$ of energy ϵ lost by charged particles along a short track segment with average energy loss Δ is the result of a compound Poisson process. The characteristic spectrum of the process is the distribution $f_c(\epsilon)$ of energy losses in individual electronic collisions. As pointed out in Section III, the mathematical relation is entirely analogous to that between $f(z; D)$ and $f_1(z)$. The same formulae and the same computer program apply to both

problems. The subsequent remarks are therefore partly a repeat of considerations in Section III.

If an ionizing charged particle with stopping power L traverses a short length l in matter it loses on the average the energy $\Delta = lL$. The energy ϵ actually lost is a random variable. There are two random factors. The number of collisions follows a Poisson distribution; the amount of energy lost in individual collisions varies widely and is the more important stochastic factor.

Let $f_c(\epsilon)$ be the probability density of energy losses in individual collisions. In the free electron model, which is used in the solutions of Landau [46] and Vavilov [47], one has

$$f_c(\epsilon) = k/\epsilon^2, \quad \text{for } I^2/\epsilon_{\max} < \epsilon < \epsilon_{\max} \quad (105)$$

where I ($= 70$ eV for water) is the mean excitation potential of the medium, and $\epsilon_{\max}$ is the maximum energy loss in a collision. For the case of electrons—where a somewhat different spectrum applies— $\epsilon_{\max}$ is conventionally set equal to half the kinetic energy of the particle. For heavy charged particles $\epsilon_{\max}$ is roughly the kinetic energy of the particle multiplied by four times the ratio of masses of the electron and the heavy particle.

To obtain a properly normalized spectrum one must set

$$k = \epsilon_{\max} I^2 / (\epsilon_{\max}^2 - I^2) \sim I^2 / \epsilon_{\max} \quad (106)$$

With Eq. (105) the average loss δ_1 per collision is

$$\begin{aligned} \delta_1 &= \int \epsilon f_c(\epsilon) d\epsilon = 2 \ln(\epsilon_{\max}/I) \cdot \epsilon_{\max} I^2 / (\epsilon_{\max}^2 - I^2) \\ &\approx 2 \ln(\epsilon_{\max}/I) \cdot I^2 / \epsilon_{\max} \end{aligned} \quad (107)$$

This is, as will be seen, a purely formal value, which is far smaller than the actual mean energy loss per collision.

The second moment is

$$\epsilon_1^2 = \int \epsilon^2 f_c(\epsilon) d\epsilon = (\epsilon_{\max} - I^2/\epsilon_{\max}) \cdot I^2 / \epsilon_{\max} \quad (108)$$

and the weighted average, i.e., the average of the energy-weighted spectrum, $d_c(\epsilon) = \epsilon f_c(\epsilon) / \bar{\epsilon}_F$ is*

$$\begin{aligned} \delta_2 &= \bar{\epsilon}_1^2 / \delta_1 = (\epsilon_{\max} - I^2/\epsilon_{\max}) / 2 \ln(\epsilon_{\max}/I) \\ &\sim \epsilon_{\max} / 2 \ln(\epsilon_{\max}/I) \end{aligned} \quad (109)$$

*The notation $d_c(\epsilon)$ for the weighted spectrum is chosen to emphasize the analogy to the dose-weighted spectra $d(L)$, $d(y)$, and $d(z)$ of LET, lineal energy y , and specific energy z , respectively. However, the simpler notation δ_1 and δ_2 is used for the frequency average and the weighted average of the collision spectrum.

This quantity is meaningful because it is not strongly influenced by the unrealistic low-energy part of the free electron collision spectrum.

According to the fundamental relation for the compound Poisson process [see Eq. (62)] the weighted average δ_2 determines the standard deviation of the straggling distribution $f(\epsilon; \Delta)$:

$$\sigma(\Delta) = \sqrt{\delta_2 \Delta} = \sqrt{\delta_2 L} \quad (110)$$

and with Eq. (105), i.e., the free electron model, one obtains

$$\sigma(\Delta) \sim \sqrt{\epsilon_{\max} \cdot Ll/2 \ln(\epsilon_{\max}/I)} \quad (111)$$

where L is the collision stopping power and l is the length of the track segment. For simplicity bremsstrahlung and nuclear interactions are here disregarded.

The free electron model applies to knock-on collisions, i.e., to collisions with ϵ large compared to the binding energies of the electrons. For glancing collisions, i.e., energy losses ϵ comparable to I or less than I , Eq. (105) is unrealistic. The extension of the $1/\epsilon^2$ spectrum down to the energy $\epsilon_{\min} = I^2/\epsilon_{\max}$ serves merely to make the spectrum agree with the actual stopping power.

It is evident that the $1/\epsilon^2$ spectrum cannot lead to valid results for very short track segments where only few collisions occur. In microdosimetry, however—and not only in microdosimetry—this is a case of considerable interest, and it is therefore useful to deal with the exact solution by the method of Section III, which can replace the familiar approximate solutions of the straggling problem.

For a realistic treatment of the glancing collisions various quantum-mechanical approximations can be used, or measured data can be employed. A discussion of these approaches and an examination of the actual shape of the collision spectra are beyond the scope of this chapter. For the present discussion it is sufficient to consider the principles of the straggling problem. Figure 18 indicates the general nature of the difference between the $1/\epsilon^2$ spectra and the ac-

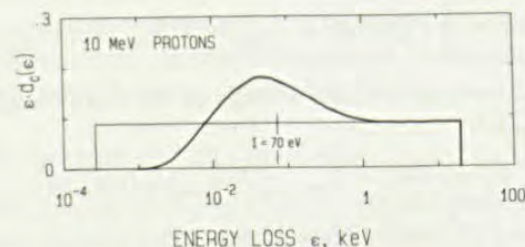


Fig. 18. Energy-weighted density of energy loss per collision relative to a logarithmic scale for a 10-MeV proton in water. The box-shaped distribution corresponds to the $1/\epsilon^2$ spectrum. The heavier line indicates a possible dependence that accounts for glancing collisions. The curves represent the fraction of energy lost by the particle per unit logarithmic interval of ϵ .

tual distributions for heavy ions of 10 MeV/nucleon. As earlier in Fig. 8 the weighted spectra relative to a logarithmic scale in ϵ are plotted as

$$\epsilon d_c(\epsilon) = \epsilon^2 f_c(\epsilon)/\delta_1 \quad (112)$$

These spectra represent the fraction of energy loss, i.e., stopping power, in specified logarithmic intervals of ϵ . For the free electron model one has a constant density on the logarithmic scale:

$$\epsilon d_c(\epsilon) = 1/\ln(\epsilon_{\max}/\epsilon_{\min}) = 1/2 \ln(\epsilon_{\max}/I) \quad (113)$$

i.e., equal logarithmic intervals of ϵ contribute equally to the stopping power. For large values of ϵ this simple rule is reasonably accurate even for realistic spectra. In many microdosimetric considerations, particularly with high-energy charged particles, it permits adequate estimates of basic features of track structure.

The parts of the spectra corresponding to glancing collisions are not well known and, as stated, they will not be considered quantitatively. The heavy curve Fig. 18 merely illustrates a possible dependence at small values of ϵ . The subsequent considerations and the resulting solution of the straggling problem apply regardless of the form of the collision spectrum.

Analytical solutions of the energy-loss straggling problem have been derived by Landau [46] and Vavilov [47] (see also [48]) on the basis of the $1/\epsilon^2$ spectrum. Landau's solution disregards both the upper and lower limits of the spectrum. It is applicable to those cases where the number of collisions is large, but it requires also that the number is sufficiently small that the probability for collisions near $\epsilon_{\max}$ can be disregarded. Vavilov has given an improved solution that accounts for the upper limit of energy loss in individual collisions. However, his solution, too, is restricted to the case of many collisions and is linked to the $1/\epsilon^2$ spectrum and certain other analytical expressions.

The solution of the compound Poisson process in Section III is equally applicable to the straggling problem. One can use the method of successive convolutions to obtain from any specified collision spectrum the straggling distribution $f_s(\epsilon; \Delta)$ for a specified mean energy loss Δ . In analogy to the procedure in Section III, a very short track segment with a mean collision number $\tau < 10^{-3}$ is considered first. The number τ is taken to be smaller by an integer power of 2 than the actual mean collision number Δ/δ_1 . The approximation of the solution for the very short segment is, in analogy to Eq. (51),

$$f_s(\epsilon; \eta) = (1 - \tau)\delta(\epsilon) + \tau f_c(\epsilon) \quad (114)$$

with $\eta = 2^{-N}\Delta$ and $\tau = 2^{-N}\Delta/\delta_1$.

The solution $f(\epsilon; \Delta)$ is then, in the same way as in Section III, obtained by

N successive convolutions:

$$\begin{aligned} f_s(\epsilon; 2\eta) &= f_s(\epsilon; \eta) * f_s(\epsilon; \eta) \\ f_s(\epsilon; 4\eta) &= f_s(\epsilon; 2\eta) * f_s(\epsilon; 2\eta) \\ &\vdots \\ f_s(\epsilon; \Delta) &= f_s(\epsilon; \Delta/2) * f_s(\epsilon; \Delta/2) \end{aligned} \quad (115)$$

As in the application to $f_1(z)$ and $f(z; D)$, it is essential to execute the convolutions on a logarithmic scale of ϵ . The algorithm described in the Appendix can therefore be used. A linear scale would require awkwardly big arrays for the representations of $f_c(\epsilon)$ and the straggling distributions, and the inconvenient lengths of the arrays make it equally impractical to use the solution in terms of Fourier transforms.

Figure 19 gives solutions for the 10 MeV/nucleon heavy ion obtained with the spectra in Fig. 18. The inadequacy of the simple $1/\epsilon^2$ spectrum at small mean energy losses Δ is evident. At larger values the shape of the solutions is mainly determined by the weighted mean δ_2 of the collision spectrum. The value 1.81 keV for the realistic spectrum is not substantially larger than the value $\epsilon_{\max}/2 \ln(\epsilon_{\max}/I) = 1.77$ keV for the simple spectrum. Accordingly, the solutions for the two spectra are similar at larger energy losses.

To further illustrate the identity of the problem of Section III and the energy-loss straggling problem, scatter diagrams are given in Fig. 20 that correspond to the curves in Fig. 19. These diagrams are entirely analogous to the D - z diagrams in Section III (see Figs. 13 and 14).

2. Correction for the Difference between Energy Lost and Energy Imparted

In microdosimetry, as in work with particle detectors, the relevant quantity is the *energy imparted* to a reference region, rather than the *energy lost* by the

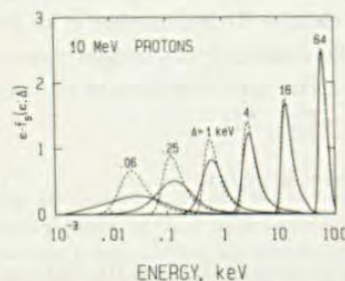


Fig. 19. Distributions of energy lost by a 10-MeV proton in a foil of thickness corresponding to the specified mean energy loss Δ . Densities $\epsilon \cdot f_1(\epsilon; \Delta)$ relative to the logarithmic scale of ϵ are plotted. The solid lines correspond to the heavier lines in Fig. 18. The dashed lines correspond to the free electron spectrum, i.e., the box-shaped distribution in Fig. 18. At sufficiently large mean energy losses both sets of distributions are in agreement; the classical solutions are then adequate.

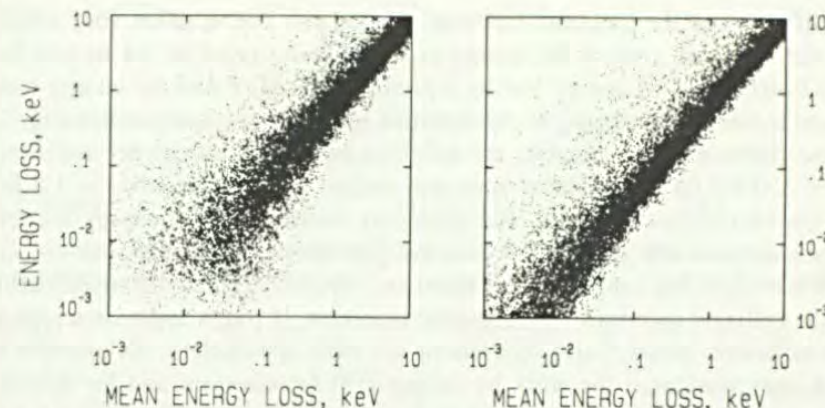


Fig. 20. Scatter diagrams corresponding to the solutions in Fig. 19. The right panel corresponds to the free electron model.

particle. The mean energy lost by a particle in a thin foil is equal to the mean energy imparted. However, even with this special geometry, the two straggling distributions differ. The reason is that high-energy δ rays transport energy out of the site and into it. The net effect is a reduction of the frequency of high values of energy lost due to individual collisions. This can significantly cut the tail, and therefore the second moment, of the collision spectrum. The width of the straggling distribution is correspondingly reduced.

A quantitative treatment would have to account for the complexities of the energy transport by δ rays. However, an approximation can at least illustrate the method to account for the difference between the two straggling distributions.

Figure 21 exemplifies the problem for heavy ions of 10 MeV/nucleon, i.e., for the same case that has been treated in the preceding subsection. In Fig. 18 an approximate modification of the $1/\epsilon^2$ spectrum at low values of ϵ has been included. This same modification is also given in Fig. 21. As stated earlier, this correction has little influence on δ_2 and on the variance of the straggling distributions. In contrast, the variance of the distributions is substantially affected by any modification of the high-energy part of the collision spectrum $f_c(\epsilon)$. A

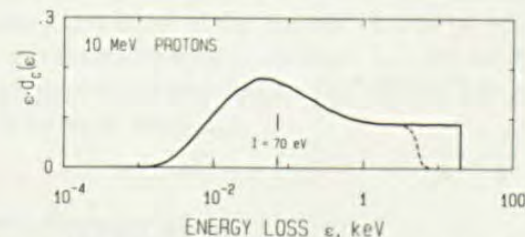


Fig. 21. The spectrum shown in Fig. 18 with a modification (dashed line) that accounts for δ ray efflux from a $1\text{-}\mu\text{m}$ -diameter site.

modification of the spectrum that may, for a $1\text{-}\mu\text{m}$ site, account very roughly for the efflux of some of the energy of δ rays is indicated by the broken line. The distributions of energy lost by a proton of 10 MeV and the energy transferred to the site, according to the modified spectrum, are compared in Fig. 22. The difference is considerable; the weighted average of energy per collision is $\delta_2 = 550\text{ eV}$ for the modified spectrum instead of the value of $\delta_2 = 1.8\text{ keV}$ for the energy-loss spectrum. The straggling distributions for energy imparted have a substantially smaller tail than the distributions of energy lost.

It is evident that complex calculations are required to obtain the modifications of the collision spectrum with adequate precision. If rough approximations are not sufficient, Monte Carlo simulations are more appropriate. An example for such computations is the work by Berger [18] for electrons and for spherical sites with diameters of $1\text{ }\mu\text{m}$ or more.

B. COMPARISON OF THE VARIOUS RANDOM FACTORS

The complexity of microdosimetric distributions makes it difficult to establish general rules for the experimental determination or the computation of these distributions. Different site sizes and different radiations require different treatment. It is nevertheless possible to derive certain guidelines to identify those random factors that are most essential in a given situation, and to discriminate them from other factors that play a minor role and may be disregarded under some conditions. The subsequent considerations are aimed at such a comparative assessment.

1. The Chain of Random Factors

If the medium is exposed to an absorbed dose D the energy imparted to a specified region is determined by a chain of random factors, as indicated in the diagram of Fig. 23. This diagram refers to the condition of small sites, i.e., of particle ranges considerably larger than the site diameter.

The first arrow indicates the random variable ν , i.e., the *number of events* or, in a simplified statement, the number of independent charged particles

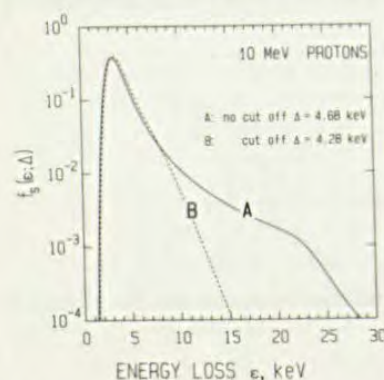


Fig. 22. Comparison of the distributions of energy lost and energy imparted for a 10-MeV proton traversing the $1\text{-}\mu\text{m}$ diameter of a spherical tissue site. The distributions are based on the spectra in Fig. 21.

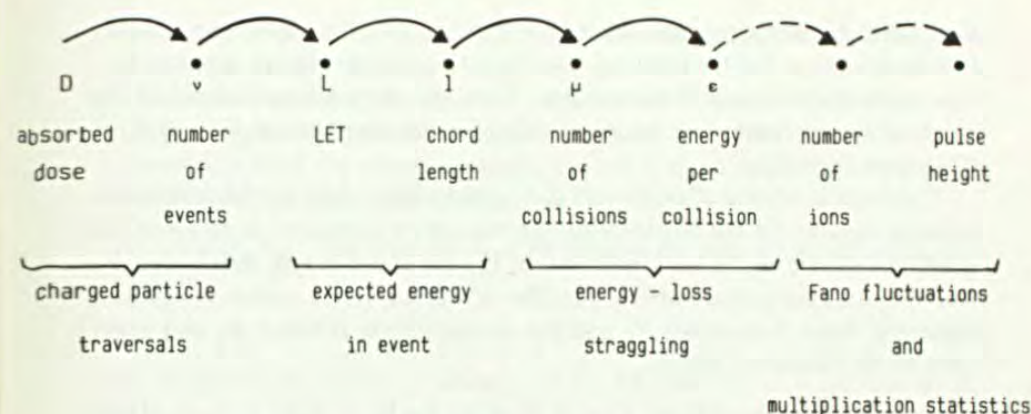


Fig. 23. Diagram of the chain of random factors that determine energy deposition in a site.

traversing the region. As stated earlier, this random variable follows a Poisson distribution.

The second random variable is the *stopping power* L of the ionizing particle. The LET distribution is designated by $f(L)$ (see Section II, B), i.e., $f(L) dL$ is the probability that a particle traversing the site has an LET between L and $L + dL$. Whenever the range of the most energetic δ rays is comparable to the size of the region, the LET is, in effect, reduced.

The expected energy imparted by an event is the product of the LET of the particle and the *chord length* in the site. The third arrow in the diagram refers to this random factor. The distribution $f(l)$ of the chord length l depends on the geometry of the exposed region. Spheres of diameter d have the triangular distribution $f(l) = 2l/d^2$, with $0 < l \leq d$. For other regions, such as cylinders or spheroids, the chord-length distributions are more complex [31, 68].

Due to the energy-loss straggling the energy actually imparted to the region of interest may deviate considerably from the product of chord length and (restricted) stopping power, $l \cdot L$. A fourth random factor is therefore the number μ of collisions along the chord, and the fifth, and more important, factor is the energy lost in each collision.

The last two arrows refer to factors that enter only into measured spectra. They are the *Fano fluctuations* of the number of ions produced for specified energy transfer, and the *multiplication statistics*, i.e., the stochastic fluctuation of pulse heights produced by a specified number of ions. The influence of these factors is, as will be seen, small.

2. The Formula for the Relative Variance

The influence of the different random factors on the single-event distribution can be assessed in terms of the relative variance, i.e., the variance divided by

the square of the mean. Utilization of a relative, i.e., dimensionless, quantity has the advantage that the resulting quantity is largely independent of particularities in the definitions of the quantities. Thus, V_L , the relative variance of the single-event spectrum, has the same value for the distributions $f_1(z)$, $f_1(\epsilon)$, or $f(y)$ [see Eq. (31)].

Under the condition of small sites and considerably longer particle ranges the relative variance of the single-event spectra can be expressed in terms of the relative variance of the LET distribution V_L , the chord-length distribution V_l , and the straggling distributions V_s . The additional terms added in brackets represent Fano fluctuations V_F and the multiplication statistics V_m and apply only to the measured spectra:

$$V_1 = V_L + V_l + V_L V_l + V_s + (V_F + V_m) \quad (116)$$

The formula is here cited without derivation [see Kellerer (33, 49)]. Its simplicity makes it a practical tool for assessing the role of the different random factors. The relative variances are additive, apart from the additional term $V_L V_l$ that enhances the joint effect of the LET and chord-length variations. In effect, one need merely compare the relative variances for the individual random factors to establish their respective importance. The magnitudes of the relative variances are, therefore, examined next.

3. Magnitudes of the Relative Variances

a. Linear Energy Transfer. V_L is the relative variance of the LET distribution $t(L)$ (see Section II,B):

$$V_L = (\bar{L}_D / \bar{L}_F) - 1 \quad (117)$$

where $\bar{L}_D$ is the dose average LET, and $\bar{L}_F$ is the track average LET. Both quantities require a suitable cutoff, if the maximum δ ray range exceeds the dimensions of the reference region. Whenever one deals with sites of the order of fractions of 1 μm or more, the cutoff has little effect on the numerical values of $\bar{L}_D$ and $\bar{L}_F$. For the mixed radiation fields produced by uncharged primaries V_L is usually larger than 1. Small values of V_L apply only in the track-segment experiments where microorganisms are exposed to monoenergetic heavy ions.

b. Chord-Length Distributions. The relative variance V_l of the chord-length distribution $f(l)$ is given by

$$V_l = (\bar{l}_D / \bar{l}_F) - 1 \quad (\text{for the sphere: } V_l = \frac{1}{8}) \quad (118)$$

where $\bar{l}_D$ is the weighted average of the chord length ($3d/4$ for the sphere of diameter d), and $\bar{l}_F$ the unweighted average ($2d/3$ for the sphere). The mean chord length $\bar{l}_F$ is for isotropic radiation fields, or for isotropic orientation of the sites, equal to $4V/S$, where V is the volume and S the surface of the (convex)

site. Figure 24 gives the values V_l for spheroids, which are often good approximations for a cell or a cell nucleus.

From the data for spheroids one concludes that V_l can be comparable to V_L only for the case of the track-segment irradiations or for sites that deviate very substantially from the spherical shape. Except in these cases the chord-length variations are of minor importance.

c. Straggling. The quantity V_s is the relative variance of the straggling distribution

$$V_s = \delta_2 / \Delta \quad (119)$$

and comprises, as shown in Section III.D, the term $(\delta_2 - \delta_1) / \Delta$ due to the fluctuations of energy loss per collision and the considerably smaller term δ_1 / Δ due to the fluctuations of the number of collisions.

As shown in the preceding section, δ_2 , the weighted average of the energy imparted per collision, depends not only on the velocity of the particle but, due to the δ -ray efflux, also on the geometry of the site. This influence of the geometry is, on the one hand, a complication but, on the other hand, a simplification. For sufficiently fast particles, i.e., particles with ranges substantially exceeding the site diameter, the maximum δ ray ranges tend to be larger than the site diameter. The geometric cutoff determines the weighted energy imparted per collision δ_2 and it is (to a first approximation) merely a function of the site diameter. Thus for the example of the preceding section a value of ~ 550 eV was deduced for a 1- μm spherical site. As a rough approximation one may assume the value $E_d/2 \ln(E_d/I)$, where E_d is the energy of the electron with range somewhat smaller than the diameter of the site.

The comparison of the relations for V_L and V_s shows that LET is the dominant random factor for mixed fields of heavy particles, such as neutron recoils, and site sizes of fractions of 1 μm or more. Straggling can nevertheless have a substantial influence on the shape of the spectra for sufficiently fast heavy ions. For slower heavy recoils with ranges comparable to the site diameter, straggling becomes insignificant. It is, therefore, justified that Caswell and

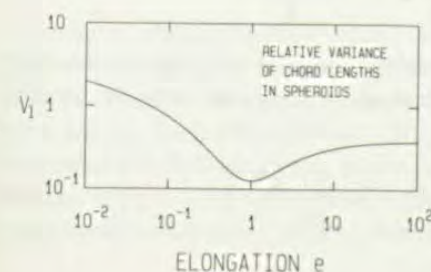


Fig. 24. Relative variance of chord length in prolate and oblate spheroids.

Coyne have disregarded straggling in computations of microdosimetric spectra for neutrons [50-52]. For much smaller sites straggling is, of course, the dominant factor.

d. Fano Fluctuations and Multiplication Statistics. For a given energy imparted, V_F is the relative variance of the number of ions produced. The relative variance of the pulse heights generated by the proportional counter for a given number of ions is denoted by V_m .

The Fano factor is smaller than the term w/Δ that would result if Poisson statistics were applicable. The symbol Δ is the energy imparted, and w is the mean energy per ionization. An approximation is

$$V_F = w/2\Delta \quad (120)$$

The influence of the multiplication statistics is somewhat larger. Under near optimal conditions a proportional counter responds with an exponential distribution of pulse heights to single electrons. One obtains then

$$V_m = w/\Delta \quad (121)$$

Since w is always much smaller than δ_2 , the two terms V_F and V_m are insignificant in microdosimetry. It follows that microdosimetric spectra are not greatly affected if one measures ionizations only, instead of all energy transfers. Computations with simulated tracks have shown [53] that, at least for electrons up to 10 keV, the differences between the exact spectra and spectra based on ionization counts are indiscernible. With adequate performance of microdosimetric instruments, measured spectra should be nearly identical to correctly computed spectra.

To summarize the comparison of the various random factors, one can give the entire formula for the relative variance of the single-event spectrum $f(y)$ or $f_i(z)$. The case of a spherical site is assumed; the modification for a different geometry with $V_i > \frac{1}{8}$ is straightforward. The value 34 eV is inserted for w :

$$\begin{aligned} V_1 &= \frac{1}{8} + \frac{9}{8} V_L + \frac{\delta_2}{\Delta} + \left(\sim \frac{50 \text{ eV}}{\Delta} \right) \\ &= \frac{9 \bar{L}_D}{8 \bar{L}_T} - 1 + \frac{\delta_2}{\Delta} + \left(\sim \frac{50 \text{ eV}}{\Delta} \right) \end{aligned} \quad (122)$$

This is equivalent to a relation for the weighted mean of the single-event spectrum. Substituting $V_1 = (\bar{y}_D/\bar{y}_F) - 1$ and $\bar{y}_F = \Delta/\bar{l}_F$ one obtains for the spherical site of diameter d

$$\bar{y}_D = \frac{9}{8} \bar{L}_D + \frac{3\delta_2}{2d} + \left(\sim \frac{3 \cdot 50 \text{ eV}}{2d} \right) \quad (123)$$

with analogous expressions for $\bar{e}_D$ and ζ . The term in parentheses applies as stated, only to experimental determinations.

Finally, one has, according to Eq. (65), the relation for the dose-dependent distributions $f(z; D)$:

$$V_z = (\bar{z}_F/D)(1 + V_1) = (1/n)(1 + V_1) \quad (124)$$

where n is the mean event number. The first term represents the (usually insignificant) influence of the Poisson fluctuations of the event numbers; the second term represents the fluctuations of event size.

VI. Geometric Aspects of the Inchoate Distribution

The computational simulation of charged particle tracks—initiated by the electron transport studies of Berger [18, 54] and first realized in full detail in the computer code of Paretzke [17, 55]—has become an important and common tool of microdosimetry (see also [19, 56-60]) and has stimulated its recent development. The method permits the assessment of the microdistribution of energy in small regions that are not, at present, accessible to experimental investigations.

The simulation of charged particle tracks is more than a substitute for deficiencies in the experimental techniques. Its theoretical implications have been equally important and have led to various new concepts and quantities in microdosimetry. These developments would require a separate treatment. Even a summary is, at present, difficult, because the mathematical foundations for some of the essential results are still inadequate; few rigorous derivations have yet been given. However, in spite of their insufficiently developed foundations, the new concepts are extensively used, and any survey of the fundamentals of microdosimetry without their consideration would remain incomplete. A brief synopsis of essentials is therefore given in this final section. Some of the new ideas have arisen from problems of sampling of simulated tracks [62]. A consideration of sampling procedures is therefore a suitable starting point.

A. UNIFORM AND WEIGHTED RANDOM SAMPLING

Simulated charged particle tracks can be utilized to derive the distributions $f(y)$ or $f_i(z)$ and the corresponding mean values. In analogy to microdosimetric measurements one may envisage a fixed site surrounded by randomly positioned tracks. Each interception of the site by a track determines then the size of an energy-deposition event. In the computations a reversed point of view is far more suitable. One can envision a fixed particle track surrounded by randomly positioned equal sites (probes), which are usually spheres. Each interception

of the track by a site determines then the size of an energy-deposition event. The resulting distributions and parameters are the same, provided correct sampling procedures are applied. There are several different sampling procedures that are equally correct but not equally efficient or convenient.

It is assumed that simulated charged particle tracks are specified by the spatial coordinates of the transfer points T_i and by their associated energy transfers ϵ_i . The sampling procedure will be explained with reference to individual tracks. Actual computations require a set of tracks, i.e., repeated random realizations. Energies and types of the particles will vary for mixed radiation fields. If one considers track-segment experiments, one deals with partial tracks. For simplicity the subsequent explanations will refer to entire tracks.

1. Associated Volume and Event Frequency

One can begin the discussion with the seemingly simple problem of determining from a simulated track the mean specific energy $\bar{z}_F$ per event in a spherical region of radius r .

To deal with the problem one needs first the notion of the associated volume A of the track.* This is the union of all spheres of radius r that are centered at transfer points. The specific energy z in a sphere of radius r differs from zero if and only if the sphere is centered within the associated volume. It is convenient to assign the value z for a sphere to the center C of the sphere; i.e., to treat z as a point function although its value depends, of course, on the energy transfers within the spherical neighborhood of C . With this formal convention one can say that the associated volume is that region where the specific energy is larger than zero.

The notion of the associated volume was introduced by Lea [5] for the specific purpose of computing event frequencies. It is, however, merely a special case of one of the basic concepts of geometric probability, namely, the *Minkowsky product* (or *direct product*) of two geometrical configurations [63, 64]. In the general definition the two geometric configurations can be arbitrary. In Lea's definition they are the spherical site S and the set T of transfer points forming the track, i.e., the inchoate distribution of energy transfers. A spherical site will be assumed in the subsequent considerations, too, but it will be evident how the results can be generalized.

A site incurs an event if it is centered within the associated volume of a track. Accordingly the event frequency equals the associated volume per unit volume of the exposed medium. Let E be the fixed initial energy of charged particles and A the mean associated volume per particle. The number of particles per unit volume and unit dose is then ρ/E , and the associated volume per unit volume

*To avoid complicated terminology the expression *associated volume* and the symbol A are used both for the domain itself and for its measure, the volume.

and unit dose is $\rho A/E$. As stated, this equals the event frequency $\phi(0)$ per unit dose:

$$\phi(0) = \rho A/E \quad \text{and} \quad \bar{z}_F = E/\rho A \quad (\rho = \text{density of the medium}) \quad (125)$$

To determine $\phi(0)$ or $\bar{z}_F$ one need therefore merely compute the associated volume of simulated tracks of specified energy. This may seem a simple procedure, but even the two-dimensional analog in Figure 25 demonstrates that an exact computation is nearly impossible. There is no practical algorithm to obtain the volumes of multiple intersections.

To derive the distribution of specific energy, or even the volume of the associated region, one must therefore use random sampling. The naive approach would be to define a box or a cylindrical region of volume B that contains the associated volume and then to choose uniformly distributed random points within this region. With a sufficiently large number of trials the associated volume is then

$$A = f \cdot B \quad (126)$$

where f is the fraction of sampling points that are contained in the associated volume.

By the same sampling procedure one can, in principle, also determine the parameter $\bar{z}_D$ or the distribution $f_1(z)$ or $F_1(z)$. For this purpose one must merely disregard those trials that lie outside the associated volume; for the sample points that are inside one must determine the specific energy. Then $F_1(z)$ is the fraction of points with specific energy up to z :

$$\text{Prob}\{z_i \leq z\} = \int_0^z f_1(z) dz = F_1(z) \quad (127)$$

The parameter $\bar{z}_D$ is obtained in terms of the mean values over all sample results in the associated volume:

$$\overline{z^2}/\bar{z} = \bar{z}_D \quad (128)$$

• TRANSFER POINTS



Fig. 25. Diagram to indicate the problem of the computation of the associated volume for a given inchoate distribution.

The frequency mean need not be determined by Eq. (125). It could also be derived as the mean of all observed z -values for the sample points in the associated volume:

$$\bar{z} = \bar{z}_F \quad (129)$$

The naive *uniform* sampling is correct, but it is too inconvenient and wasteful to be practicable. In actual computations a different method is therefore required.

2. Formulas for Weighted Sampling

With *uniform* sampling any volume element dB of the sample space has equal probability to be selected. An alternative method is *weighted* sampling, i.e., sampling where the probability of dB to be selected is proportional to the specific energy at dB . The specific energy at dB is, as has been stated, the specific energy in a sphere centered at dB .

Weighted sampling can be achieved as follows. In a first step one selects randomly an energy transfer point T_i , the selection probability being proportional to the corresponding energy transfer ϵ_i . In a second step one chooses a random point in the sphere of radius r centered at the selected random point. In this way any point in the associated volume is reached with probability proportional to the specific energy at the point.

While uniform sampling provides the density $f_1(z)$, weighted sampling provides the weighted distribution $d_1(z)$ (see Section II,F). Accordingly one derives, from the average of values z_k obtained with weighted sampling, the parameter $\bar{z}_D$:

$$\langle z \rangle = \int z d_1(z) dz = \bar{z}_D \quad (130)$$

The symbol $\langle \rangle$ denotes the average obtained with weighted sampling, and discriminates it from the averages obtained by uniform sampling.

To determine the frequency average with weighted sampling one must average the reciprocals of the observed values [see Eq. (27)]:

$$\langle z^{-1} \rangle = \int z^{-1} d_1(z) dz = \int f_1(z) dz / \bar{z}_F \quad (131)$$

and therefore

$$\bar{z}_F = 1 / \langle z^{-1} \rangle \quad (132)$$

The procedures for uniform and for weighted sampling can be formulated in a modified but equivalent form that applies also to nonspherical regions S of reference:

For uniform sampling one selects a random point P in a sufficiently large region containing the track, and then positions the center, i.e., a fixed reference point, of S at P . The direction of S is distributed randomly for isotropic sampling.

For weighted sampling one selects a random point on the track, i.e., a transfer point, and a random point in S , and one then superimposes the two selected points. The direction of S is distributed randomly for isotropic sampling.

It is readily seen that this more general formulation defines a procedure that is equivalent to the one specified for spherical sites.

The procedure for weighted sampling is required in numerical evaluations of simulated particle tracks [62, 65]. It is also essential in considerations that lead to the proximity function and its applications, and it will be discussed in this context in Section VI,C. However, before these matters are treated, certain simplified formulas for the event frequencies and the mean event sizes can be derived.

With uniform sampling one obtains the distribution $f_1(z)$ directly; however, as stated, uniform sampling is highly inefficient. With weighted sampling one obtains the distribution $d_1(z)$, but the distribution $f_1(z)$ can be obtained from $d_1(z)$:

$$f_1(z) = z^{-1} d_1(z) / \int z^{-1} d_1(z) dz \quad (133)$$

B. FORMULAS FOR THE ASSOCIATED VOLUME AND FOR THE UNWEIGHTED AVERAGES

For a complex inchoate distribution it is, as has been stated, difficult to compute the associated volume, and random sampling is therefore required. However, for the special case of linear tracks there are simple solutions and it has been found [66] that the solutions apply surprisingly well even to electrons.

1. Spherical Sites

Consider for a particle of range R and initial energy E a track that is depicted as a straight line segment with the continuous slowing-down approximation. This will be termed a *linear track*. The associated volume of this track is

$$A = r^2 \pi R + 4r^3/3 \quad (134)$$

Accordingly the event frequency per unit absorbed dose is [see Eq. (125)]

$$\phi(0) = 1/\bar{z}_F = \rho(r^2 \pi R + 4r^3/3)/E \quad (135)$$

Evidently this relation applies also for the mean range $\bar{R}$ of the particles in a mixed field and for the mean initial energy $\bar{E}$ and in this more general form it permits a convenient check for internal consistencies of experimental microdosimetric data.

The frequency mean lineal energy $\bar{y}_F$ is linked to $\phi(0)$ by the equation

$$\bar{y}_F^{-1} = \phi(0)/r^2\pi\rho \quad (136)$$

Utilizing the relation $\bar{L}_F = \bar{E}/\bar{R}$ one obtains with Eq. (135)

$$1/\bar{y}_F = \bar{R}/\bar{E} + 4r/3\bar{E} = 1/\bar{L}_F + \bar{l}/\bar{E} \quad (137)$$

Accordingly one obtains a straight line if $1/\bar{y}_F$ is plotted versus the mean chord length $\bar{l} = 4r/3$ of the site. The event frequency or the unweighted mean event sizes $\bar{y}_F$, $\bar{z}_F$, or $\bar{\epsilon}_F$ are therefore determined for all site sizes if they are known for two sizes. The knowledge of merely two values permits also the determination of the track average LET $\bar{L}_F$, the mean range $\bar{R}$, and the mean initial energy $\bar{E}$ of the particles.

Figure 26 indicates the application of the result to photons.

2. Generalization of the Result to Nonspherical Sites

The average chord length $\bar{l}$ is, for all convex bodies, equal to four times the volume V divided by the surface area S . For example, one has for circular cylinders of radius r and height h :

$$\bar{l} = 4V/S = 2rh/(r + h) \quad (138)$$

The mean intercept $\bar{s}$ of tracks with a convex site depends only on the mean chord length $\bar{l} = 4V/S$ of the site and the mean range $\bar{R}$ of the tracks (31):

$$1/\bar{s} = 1/\bar{l} + 1/\bar{R} \quad (139)$$

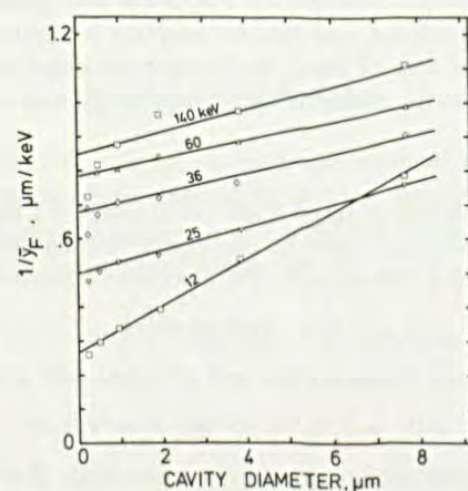


Fig. 26. Diagram of frequency average event sizes $\bar{y}_F$ for x rays according to Kliuga and Dvorak [66].

With the relation for the mean energy imparted:

$$\bar{\epsilon}_F = \bar{E}(\bar{s}/\bar{R}) \quad (140)$$

one obtains

$$1/\bar{y}_F = \bar{l}/\bar{\epsilon}_F = \bar{l} \cdot \bar{R}/\bar{E} \cdot \bar{s} = 1/\bar{L}_F + \bar{l}/\bar{E} \quad (141)$$

The result of this subsection applies, therefore, generally to all convex sites.

C. THE RANDOM INTERCEPT OF TWO GEOMETRIC OBJECTS

This subsection deals with a formula of considerable generality that permits—applied to microdosimetry—the computation of the weighted averages of the single-event spectra in sites of specified shape. The formula also demonstrates the important role of the proximity functions, or point-pair distance distributions. All considerations in this subsection are given in purely geometric terms, but without mathematical rigor. The application to microdosimetry is treated in Section VI.D.

1. Uniform and Weighted Randomness of Straight Random Lines

Uniform isotropic sampling corresponds to *uniform isotropic randomness* for the intercept of two geometric objects S and T . Weighted isotropic sampling corresponds to *weighted randomness*. The concepts are familiar, and a satisfactory mathematical theory exists in the special case where S is a convex body (in a space of arbitrary dimensions) and T is a straight random line (see, e.g., [67, 68]).

For uniform randomness one selects a random point in the space that contains S and then chooses a straight line T with random direction that passes through the point. Only those lines are considered that intersect S , and the resulting distribution of intercepts l is termed the chord-length distribution $f(l)$ for uniform randomness.

For weighted randomness one selects a random point within S and then chooses a straight line T with random direction that passes through the point. The resulting distribution of intercepts l is termed the chord-length distribution $d(l)$ for weighted randomness.*

Kingman [69] has obtained the important relation that links the densities for uniform and for weighted randomness:

$$d(l) = lf(l)/\bar{l}_F \quad (142)$$

*Uniform randomness and weighted randomness have also been termed ν -randomness and μ -randomness, and the notation $f_\nu(l)$ and $f_\mu(l)$ has been used for the densities. The notation $f(l)$ and $d(l)$ is here used to emphasize the correspondence to the frequency distributions and weighted distributions $f_1(z)$ and $d_1(z)$.

where $\bar{l}_F$ is the mean chord length for uniform randomness. $\bar{l}_F$ equals, as has been pointed out in earlier sections, $4V/S$ for convex bodies of volume V and surface S .

Equation (142) is analogous to the relations that link other densities and weighted densities, such as $f_1(z)$ and $d_1(z)$ [see Eq. (27)], and as with the other densities one obtains the relation between the weighted mean and the first two moments of the unweighted distribution

$$\bar{l}_D = \int l d(l) dl = \frac{\bar{l}_F^2}{\bar{l}_F} \quad (143)$$

The variance for uniform randomness is

$$\sigma_F^2 = (\bar{l}_D - \bar{l}_F) \cdot \bar{l}_F \quad (144)$$

2. Uniform and Weighted Randomness of Pairs of General Geometric Objects

The notion of uniform and weighted randomness can be extended to apply to arbitrary geometric objects, provided these objects meet certain minimum requirements of regularity that need not be considered in the present nonmathematical context.

Uniform isotropic randomness of the intersection of two objects S and T results if S is kept in fixed position and if a reference point, called center, of T is superimposed with a random point in a sufficiently large region containing S . Random orientation of T is assumed. If T misses S , i.e., if $T \cap S = 0$, the trial is disregarded.

The resulting density of the volumes* u of the intersection $T \cap S$ is termed the frequency distribution $f(u)$.

Weighted isotropic randomness of the intersection of two objects S and T results if a random point is chosen in S and in T , if S is kept in fixed position, and if T is shifted so that the two random points coincide. Again T is randomly oriented. The resulting density of the volumes u of intersection is termed the weighted distribution $d(u)$.

One can show [70] that the familiar relation between densities and weighted densities applies also in this general case:

$$d(u) = uf(u)/\bar{u}_F \quad (145)$$

where $\bar{u}_F$ is the mean overlap under uniform randomness. Relations for $\bar{u}_F$ are obtained from the *fundamental formula* of Blaschke and Santalò (71, 72); for a simplified, nonmathematical explanation see Kellerer [73].

*The term volume is used in the general sense of measure, i.e., the region of overlap need not be three-dimensional.

As in the familiar case of straight random lines one obtains the relation between the weighted mean intercept and the first moments of the unweighted distribution:

$$\bar{u}_D = \int u d(u) du = \frac{\bar{u}_F^2}{\bar{u}_F} \quad (146)$$

For uniform randomness the variance of the overlap u is

$$\sigma_F^2 = (\bar{u}_D - \bar{u}_F) \cdot \bar{u}_F \quad (147)$$

The importance of Eqs. (145) to (147) lies in the fact that a general and fairly simple formula exists for the weighted averages, and therefore that the variance of $f(u)$ can be obtained also.

The formula will be considered next. It contains the *proximity functions*, or *point-pair distance distributions*, of the geometric objects T and S .

3. The Formula for the Weighted Mean Intersection and Definition of the Proximity Functions

One may choose pairs of independently and uniformly distributed random points in a geometric object and determine the density $p(x)$, of distances x between the points. The function $p(x)$ is termed the *point-pair distance distribution*, and $p(x) dx$ is the probability that a pair of random points has a distance between x and $x + dx$.

For some geometries it is convenient to utilize the nonnormalized densities that result when $p(x)$ is multiplied by the measure of the geometric object. This nonnormalized distribution is called the *proximity function*. For convenience the proximity functions for the geometric objects S and T will be denoted by $s(x)$ and $t(x)$,

$$s(x) = V_S \cdot p_S(x) \quad \text{and} \quad t(x) = V_T \cdot p_T(x) \quad (148)$$

where V_S is the measure, i.e., volume, of S , and V_T is the measure of T , and $p_S(x)$ and $p_T(x)$ are the distance distributions for S and T .

A fundamental relation [70] expresses the weighted mean of the intercept in terms of the proximity functions

$$\bar{u}_D = \int_0^{x_{\max}} \frac{s(x)t(x)}{4\pi x^2} dx \quad (\text{in three-dimensional space}) \quad (149)$$

and

$$\bar{u}_D = \int_0^{x_{\max}} \frac{s(x)t(x)}{2\pi x} dx \quad (\text{in two-dimensional space}) \quad (150)$$

where $x_{\max}$ is the maximum point-pair distance in S or T , whichever value is smaller.

The somewhat lengthy proof of this remarkably general result is omitted; it has certain connections to a theorem derived by Robbins [74]. The fundamental formula of Blaschke and Santalò determines the unweighted average $\bar{u}_F$ of the random intersection, for uniform isotropic randomness. Equation (149) together with Eq. (147) determine the second moment of the intercept for uniform randomness or the variance

$$\sigma_u^2 = (\bar{u}_D - \bar{u}_F) \cdot \bar{u}_F \quad (151)$$

Proximity functions are known for regular geometric objects such as cubes [75], cylinders [76], or spheroids [68]. As an example one can give the familiar relation for a sphere of radius r ,

$$s(x) = 4\pi x^2 \left[1 - \frac{3x}{4r} + \frac{1}{16} \left(\frac{x}{r} \right)^3 \right], \quad 0 \leq x \leq 2r \quad (152)$$

Instead of the proximity function one often uses, when one deals with compact geometric objects, the *geometric reduction factor*:

$$\begin{aligned} U(x) &= s(x)/4\pi x^2 && \text{(in three-dimensional space)} \\ U(x) &= s(x)/2\pi x && \text{(in two-dimensional space)} \end{aligned} \quad (153)$$

For complicated structures the computation of the proximity function may require complex numerical integrations or even Monte Carlo simulations, but the functions can always be obtained. It is striking that the variance of the overlap of two objects or their weighted mean overlap can be expressed by a simple integration over two functions each of which depends on only one of the structures.

An application of the result to microdosimetry is straightforward and will be considered next.

D. APPLICATION TO MICRODOSIMETRY

In the preceding subsection the random intercept of two configurations has been treated purely as a geometric problem. This has made the essential result more transparent.

However, the considerations are intimately linked to the problems of energy deposition by ionizing radiations. The point-pair distance distributions and the geometric reduction factor have been introduced into dosimetric computations by Berger [77]; his treatment also contains the essence of later results. The equation for the weighted mean intercept, too, has been derived in microdosimetric studies; Eq. (149) is merely a generalization of this result.

1. Proximity Function for Charged Particle Tracks

To apply Eq. (149) to microdosimetry one must first introduce the concept of the proximity function of a particle track. This concept is entirely analogous

to that of the geometric proximity function, with the only modification being that energy replaces volume. The function $t(x)$ is the proximity function of the particle track. Analogously to the (volume) proximity function, the (energy) proximity function $t(x)$ can be understood as the distance distribution of energy transfers (weighted by their values ϵ_i) multiplied by the total energy of the track. The proximity function includes a δ function at $x = 0$ that is proportional to the weighted mean of individual energy transfers ϵ_i .

The actual computation is best explained in terms of the integral proximity function

$$T(x) = \int_0^x t(x') dx' \quad (154)$$

From a simulated track, $T(x)$ is obtained by considering all pairs of transfer points that are separated by a distance less than x :

$$T(x) = \sum_{i,k} \epsilon_i \epsilon_k / \sum_i \epsilon_i \quad (155)$$

where the summation runs over all i and over all transfer points k separated by distance up to x from the transfer point i . The value $T(0)$ determines the δ function $T(0) \cdot \delta(x)$ in $t(x)$. It results from the pairs with $i = k$:

$$T(0) = \sum_i \epsilon_i^2 / \sum_i \epsilon_i \quad (156)$$

From Eq. (155) one recognizes that $T(x)$ equals the expected energy (on the same track) in a sphere centered randomly on an energy transfer. Accordingly $t(x) dx$ is the expected energy within the distance interval x to $x + dx$ from a randomly selected energy transfer.

In actual computations the proximity function is derived by sampling all pairs of energy transfers in a sufficiently large number of simulated particle tracks. If the particle tracks have different initial energies, the proximity function is the energy-weighted average for the different tracks. Proximity functions have been computed for electrons and for track segments of heavy ions (see, e.g., [53, 62, 78–81]). Figure 27 gives the functions for electrons up to 10 keV.

2. The Formula for the Dose Average Event Size

The theorem from Section (VI,C) can be applied to the overlap of a site S with a charged particle track T . Thus one obtains the weighted average of the energy imparted per event:

$$\bar{\epsilon}_D = \int_0^{x_{\max}} \frac{t(x)s(x)}{4\pi x^2} dx = \int_0^{x_{\max}} t(x)U(x) dx \quad (157)$$

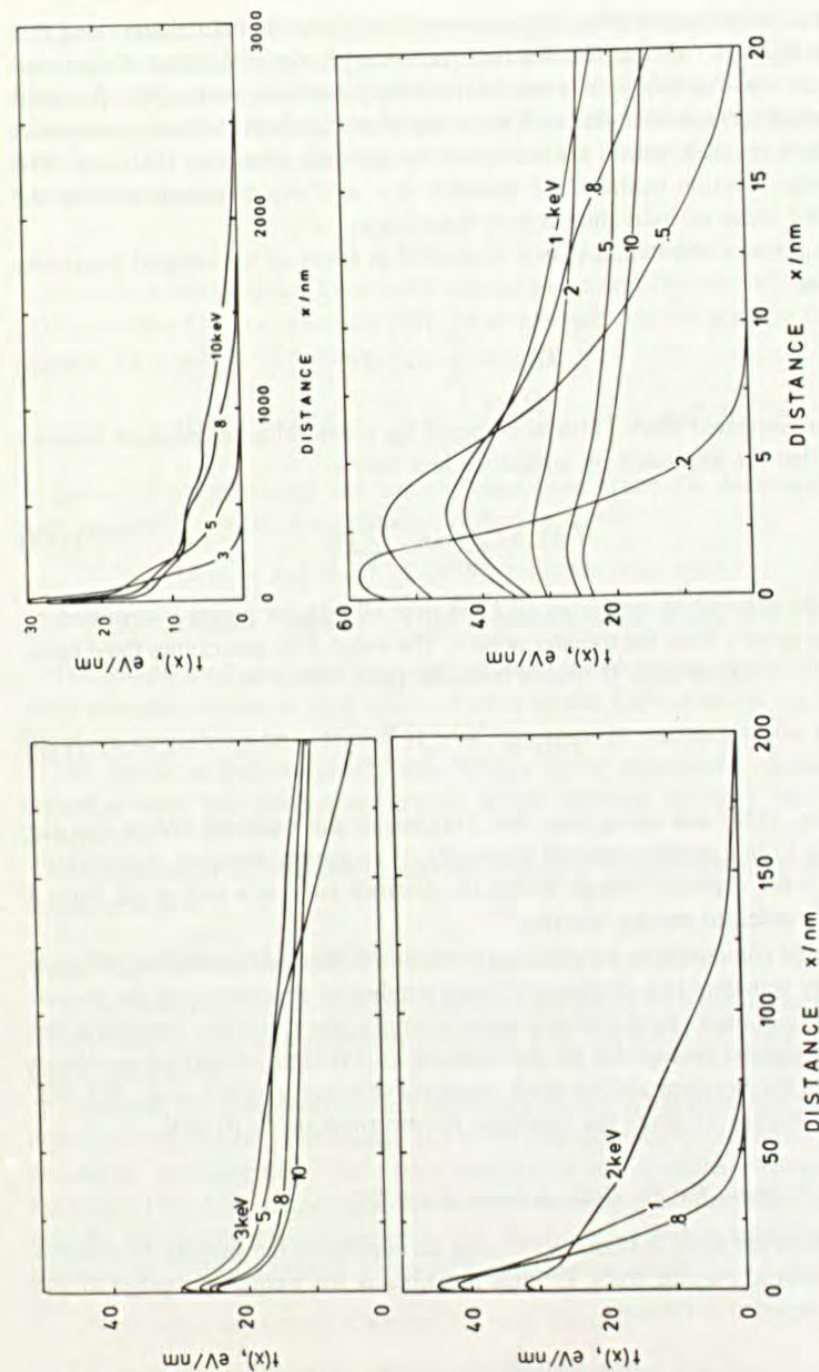


Fig. 27. Proximity functions $t(x)$ for low-energy electrons. (From Chmelevsky *et al.* [53].)

The right-hand expression utilizes the geometric reduction factor $U(x)$ of the site rather than the proximity function $s(x)$ of the site.

The upper bound of integration is the maximal diameter of the site, in the case of the sphere its diameter. With $\bar{\epsilon}_D$ one also obtains the averages $\bar{z}_D$ and $\bar{y}_D$ of specific energy or lineal energy in single events.

The result is valid regardless of the complexity of the tracks and the structure of the sites. The site need be neither spherical nor convex, nor even simply connected. The generality of the theorem permits a great variety of applications and extensions.

3. Further Considerations of the Proximity Function

If a particle of initial energy E were represented by a straight line segment of length R and constant energy transfer E/R , the proximity function would have the simple form

$$t(x) = 2E/R(1 - x/R), \quad 0 \leq x \leq R \quad (158)$$

A somewhat better approximation postulates straight line segments with energy transfer varying as a function of remaining range (continuous slowing-down approximation). One obtains then [24]

$$t(x) = 2 \int_x^R L(s)L(s-x) ds/E \quad (159)$$

where $L(s)$ is the LET at remaining range s . At sufficiently large values x the approximation in terms of Eq. (159) is good for heavy ions. For electrons the approximation fails as shown in Fig. 28.

From Eq. (159) one obtains $t(x) = 2\bar{L}_D$ for sufficiently small x . The proximity function can therefore be considered as an extension of the LET concept that takes into account additional factors, such as the ranges of the particles.

Equation (159) is appropriate for heavy particles for all values of x that exceed the maximum range of the δ rays. At smaller distances $t(x)$ is codetermined by energy-loss straggling and the radial distribution of energy around the center of the track.

For track segments the influence of straggling and radial energy distribution can be separated [24, 62]. The proximity function is the sum of a term, $t_\delta(x)$, that is the weighted average of the proximity functions for all δ rays, and a term that is equal to the proximity function of a continuous track with the same radial energy distribution as the actual track:

$$t(x) = t_\delta(x) + t_a(x) \cdot L \quad (160)$$

The index of the second term indicates the fact that it refers to an *amorphous* track that averages out δ rays. This term contains the stopping power as a factor,

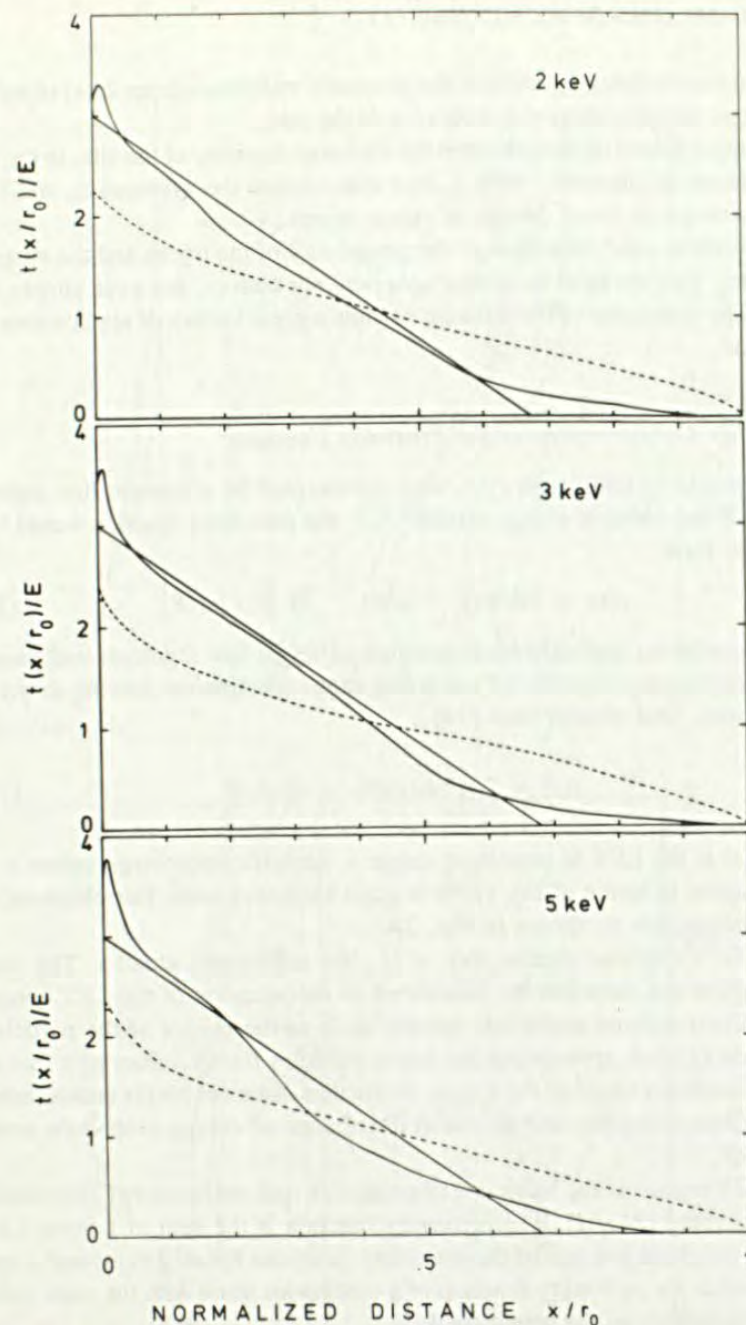


Fig. 28. Comparison of proximity functions for electrons (solid curves) with functions that result from simplified models of Chmelevsky *et al.* [53]. The dashed curves are obtained when the electron tracks are treated as straight lines with continuous energy loss according to the LET. The solid straight lines result if the tracks are pictured as straight lines with constant energy loss rate and ranges equal to $\frac{2}{3}$ of the continuous slowing-down range. The ranges and LET values are for water. The curves show that the LET approximation is unsuitable for electrons because of straggling and angular scattering.

and the other factor, $t_a(x)$, depends only on the radial dose distribution and can be calculated from it.

Examples of the proximity functions for track segments of heavy ions of 20 MeV/nucleon and their separation into the δ term and the term for the continuous track are given in Fig. 29 for short track segments of fast heavy ions.

Equation (160) can be utilized for an alternative derivation of the weighted average and the variance of the single-event distribution. One obtains with the theorem of Eq. (157).

$$\begin{aligned} \bar{y}_D &= \frac{1}{l} \int U(x)(t_\delta(x) + t_a(x) \cdot L) dx \\ &= \frac{1}{l} \int U(x)t_\delta(x) dx + L \cdot \int U(x)t_a(x) dx = \frac{\delta_2}{l} + \frac{Lx'}{l} \quad (161) \end{aligned}$$

The first term contains the weighted mean energy imparted per collision. It represents, as stated earlier with reference to Eq. (123), the influence of straggling. The second term contains the parameter x' , which equals the weighted mean chord length whenever $t_a(x)$ is sufficiently close to 2.

The comparison with Eq. (123) shows that the proximity functions make it possible to account exactly for track structure and site geometry where the earlier considerations had to be based on relatively crude approximations (see also Kellerer [82]).

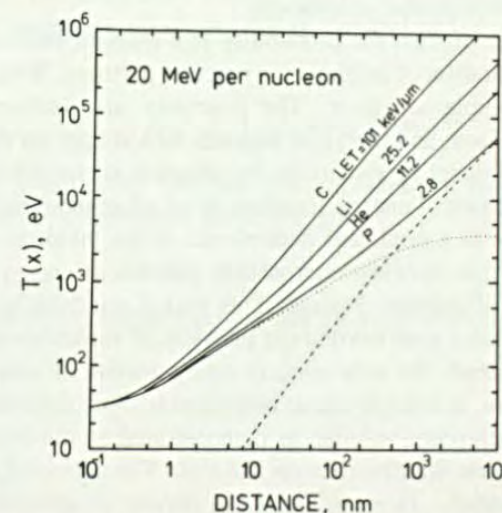


Fig. 29. Integral proximity functions $T(x)$ for different heavy ions of kinetic energy of 20 MeV per nucleon from Chmelevsky [62]. The dotted line represents the contribution $T_\delta(x)$ due to δ rays and the dashed line corresponds to the LET-dependent term for protons. For the other ions this term is shifted by a factor corresponding to the increased LET.

E. A BIOPHYSICAL MODEL

As in Section IV, a concluding section is added here that indicates, without actual assessment of radiobiological findings, the applicability of some of the microdosimetric concepts and data. The treatment is a generalization of the considerations that have been given, with reference to the site concept, in Section IV, C.

Assume that energy transfers produce a proportional number $k\epsilon_i$ of *sublesions*. Such sublesions can be molecular alterations, for example, in DNA. They can also be more complex changes, such as chromosome instabilities (breaks). Assume further that the sublesions can interact pairwise. The formation of *lesions* is then a *second-order process*.

A second-order process leads, under certain conditions, to a quadratic dependence of the yield on the concentration of reactants (sublesions). One implied condition is the absence of saturation, i.e., the absence of concentrations that are so high that there is no further proportional increase of reactants. Another condition is lack of competition between sublesions (depletion); this amounts to the postulate that the pairwise interactions of sublesions are independent processes. Lack of saturation is a realistic concept for sparsely ionizing and medium-densely ionizing particles. However, with densely ionizing particles saturation plays a role on various levels toward the observable cellular damage. Lack of competition may well be a realistic postulate for DNA single strand breaks that have relatively small probabilities of combining into double strand breaks. It may also be realistic for chromosome instabilities that are far more numerous than observable chromosome aberrations.

In a second-order process the probability of a reactant interacting is proportional to the concentration of neighboring reaction partners. With chemical reactions this is a simple situation. The reactants are uniformly distributed throughout the medium, and the yield depends then simply on the square of the concentration. With ionizing radiations the situation is characteristically different. The concentrations of energy transfers or of subsequent radiation products vary greatly, and with a nonlinear dependence of the yield on concentration a complex relation exists between microscopic patterns of energy deposition and the resultant yield of cellular damage. This makes microdosimetry necessary. However, the difficulty goes beyond the problem of variations of concentration in the medium. Instead, the very concept *concentration* of energy transfers or radiation products is, in a strict sense, inapplicable. Any derived value depends on the size of the reference volume, as demonstrated by the strong dependence of the microdosimetric distributions on site size. The notion of a concentration is therefore meaningless. However, one can invoke an *effective concentration* if one has knowledge of the scale of potential interactions between reaction partners.

2. FUNDAMENTALS OF MICRODOSIMETRY

The microdosimetric variables ϵ , z , and y are defined in terms of a test region, the site. Its dimensions determine the scale of possible interaction. According to this concept, an energy transfer, or the resulting reactant, has as potential reaction partners all transfers that happen to be located in the same region. This treatment has the advantage that the distributions of microdosimetric variables can be measured. On the other hand, it is, as stated in Section IV, an obvious simplification; the reactants within the site cannot, regardless of their mutual distances, be equally likely to interact. In many biophysical applications the resulting inaccuracy of the treatment may be of little concern. The principal aim is the derivation of a rough effective interaction distance, and this can be achieved in terms of the site concept. The concept is, however, grossly inadequate if long-range particles are compared with very short range particles. The long-range particles dissipate energy throughout the site; the short-range particles concentrate the energy in small clusters. For an adequate comparison of such disparate configurations and for any investigation of distance-dependent interaction probabilities, one needs a modified analysis.

The concept of the proximity function permits such an analysis. Although this chapter does not deal with radiobiophysical applications, the essential formulas can be given. An abbreviated terminology will be used in speaking of the *interaction of pairs of energy transfers*. This refers to the probability that sublesions are formed by the two energy transfers and that they interact to form a lesion.

It is assumed that the probability that a transfer will interact is proportional to the number of its neighbors multiplied by a distance-dependent interaction probability $\gamma(x)$. Evidently each transfer has to be weighted by the corresponding amount of energy. Accordingly the probability of interaction of an energy transfer ϵ_i is

$$p_i \sim \epsilon_i \int_0^\infty t_D(x) \gamma(x) dx \quad (162)$$

where $t_D(x)$ is a proximity function that includes the trivial dose-dependent term that is due to independent particle tracks:

$$t_D(x) = t(x) + 4\pi x^2 \rho D \quad (163)$$

It is practical to normalize $\gamma(x)$ so that its integral over the whole space is unity:

$$\int_0^\infty \gamma(x) 4\pi x^2 \rho dx = 1 \quad (164)$$

One then obtains

$$p_i = k\epsilon_i \left[\int_0^\infty t_D(x) \gamma(x) dx + D \right] \quad (165)$$

The yield $E(D)$ of lesions, arbitrarily normalized to unit mass, is then

$$E(D) = kD \left(\int_0^\infty t(x) \gamma(x) dx + D \right) = k(\xi D + D^2) \quad (166)$$

The essential result is that, with the proximity concept in the same way as with the site concept, one obtains a linear-quadratic dose dependence. The coefficient ξ of the linear term is, as with the site concept, a measure of an *effective local concentration* of energy transfers in the individual charged particle tracks of the radiation.

One can consider the earlier result for the site concept as a special case of the present formula. Assume that an energy transfer has constant interaction probability with all transfers within the same site, and zero interaction probability with transfers outside the site. The geometric reduction factor $U(x)$ is the probability that a neighboring transfer at distance x is within the site. The interaction probability $\gamma(x)$ is therefore proportional to $U(x)$. Inserting $\gamma(x) = U(x)/\int 4\pi x^2 \rho U(x) dx = U(x)/m$ into Eq. (166) one obtains

$$E(D) = kD \left(\int_0^\infty t(x) U(x) dx/m + D \right) \quad (m = \text{mass of the site})$$

$$= k(\xi D + D^2) \quad (167)$$

This is the earlier result for the site concept.

In reality both the site and the proximity aspect may jointly play a role. The function $\gamma(x)$ is then the product of two terms representing the inherent dependence on distance and the influence of the geometry of the site, $\gamma(x) = g(x) \cdot U(x)$. In an experiment it may be difficult to separate the two factors and the aim of a biophysical investigation is therefore to determine the compound function $\gamma(x)$.

The above considerations can serve as a frame of reference, but it is evident that additional factors play a role. There can, for example, be an inherently linear component in the dependence of the effect on z , and it is certain that the quadratic term in absorbed dose is, unlike the linear intratrack term, dependent on dose rate. In spite of such added complexities the formulation in terms of the proximity function leads to definite conclusions when applied to experiments with correlated heavy particles [83, 84]. It is found that $\gamma(x)$ decreases sharply at small distances, but reaches out to distances of the order of several micrometers. The result is in line with the relatively high efficiency of low-energy photons [85-89], and it illustrates the need for the analysis in terms of the proximity concept.

APPENDIX: Algorithm for the Compound Poisson Process

A. SOLUTION BY SUCCESSIVE CONVOLUTIONS

The function $f_1(x)$ is the density of the spectrum of the Poisson process. In the application of this section $f_1(x)$ equals the function $f_1(z)$. In the application to the energy-loss-straggling problem

2. FUNDAMENTALS OF MICRODOSIMETRY

$f_1(x)$ stands for the collision spectrum $f_c(\epsilon)$. The symbol x is used for the variable of the Poisson process to avoid the impression that the algorithm is linked to a particular application.

The spectrum $f_1(x)$ can extend over several orders of magnitude of the variable x . It is therefore essential that the convolution is executed on a logarithmic scale.

The solutions $f(x)$ correspond to the dose-dependent distributions $f(z; D)$ or to the straggling distributions $f(\epsilon; \Delta)$. The solutions have a finite component at $x = 0$:

$$f_0 = \exp(-n) \quad (n = \text{mean event number}) \quad (A.1)$$

The remaining density $f(x)$ contains no zero component and its norm is $(1 - f_0)$.

For the initial approximation a suitable, small mean event number ϵ is chosen and the solution is approximated by

$$f_0 = (1 - \epsilon), \quad f(x) = \epsilon \cdot f_1(x) \quad (A.2)$$

Each of the successive convolutions then replaces

$$f_0 \quad \text{by} \quad f_0^2 \quad (A.3a)$$

and

$$f(x) \quad \text{by} \quad 2f_0 \cdot f(x) + f(x) * f(x) \quad (A.3b)$$

This is continued until the desired mean event number and, accordingly, the desired mean value $\bar{x}$ are reached.

In the computer code f_0 is a scalar and $f(x)$ is represented by an array:

$$F(L) = f(x), \quad L = 1 \text{ to } N \quad (A.4)$$

where

$$x = x_0 \cdot X(L) \quad \text{with} \quad X(L) = 2^{L/N} = \exp(L \cdot U), \quad U = \ln(2)/N_2 \quad (A.5)$$

and N_2 is the integer number of grid points in an interval that corresponds to a factor of two of the random variable. One needs then $N_2 \ln(10)/\ln(2) = 3.3N_2$ points per decade of the random variable. A typical value for the start of the procedure is $N_2 = 8$; common microdosimetric spectra can then be represented by arrays with a length N of 100 to 150.

As the convolutions proceed insignificant tails of the array $F(L)$ are cut and the parameters N and x_0 are adjusted. When the arrays become too short, for example if their length falls below 60, N_2 is doubled and $F(L)$, $X(L)$, and certain auxiliary arrays that are required in the convolution are recalculated.

The convolution is the only nontrivial step in the algorithm. It is therefore considered in detail, and a BASIC version of the subroutine is included.* The formula for the convolution is

$$h(x) = \int_{x_0}^x f(x-s)f(s) ds = 2 \int_{x_0}^{x/2} f(x-s)f(s) ds \quad (A.6)$$

Changing to the new variable

$$L = \ln(x/x_0)/U, \quad x = x_0 \exp(L \cdot U) \quad (A.7)$$

*A FORTRAN listing was given in an earlier report [33] with the first description of the algorithm.

and using the functions

$$\left. \begin{aligned} H(L) &= h(x) \\ F(L) &= f(x) \\ X(L) &= \exp(L \cdot U) = x/x_0 \end{aligned} \right\} \text{ for } 0 \leq L \leq N \quad (\text{A.8})$$

one obtains from Eq. (A.6) with $ds = x_0 \cdot U \cdot X(K) dK$

$$H(L) = 2x_0 U \int_0^{L-N_2} F(W)F(K)X(K) dK \quad (\text{A.9})$$

with

$$\begin{aligned} W &= \ln[(x-s)/x_0]/U \\ &= \ln[\exp(L \cdot U) - \exp(K \cdot U)]/U \\ &= L + \ln[1 - \exp[(K-L) \cdot U]]/U \\ &= L - A(K-L) \end{aligned} \quad (\text{A.10})$$

where

$$A(J) = -\ln[1 - \exp(J \cdot U)]/U \quad (\text{A.11})$$

is an auxiliary function. To reduce computing time in the necessary interpolation $A(J)$ is split into its integer part $\text{IN}(J)$ and its fractional part $\text{FR}(J)$.

With the additional convention $FX(K) = F(K) \cdot X(K)$ one has the integral

$$H(L) = 2x_0 U \int_{K_0}^{L-N_2} F(L - \text{IN}(L-K) - \text{FR}(L-K)) FX(K) dK \quad (\text{A.12})$$

$$\text{for } N_2 \leq L \leq N + N_2 \quad (\text{A.12})$$

with

$$K_0 = \begin{cases} 0 & \text{for } L \leq N \\ \max[0, L - A(L-N)] & \text{for } L > N \end{cases} \quad (\text{A.13})$$

In the actual computation this is substituted by the sum

$$H(L) = 2x_0 U \sum_{K_0}^{L-N_2} [P \cdot F(J-1) + (1-P) \cdot F(J)] \cdot FE(K) \quad (\text{A.14})$$

with

$$P = \text{FR}(L-K) \quad J = L - \text{IN}(L-K) \quad (\text{A.15})$$

Only half of the last term ($K = L - N_2$) is included in the sum.

The following BASIC subroutine returns the array $H(L)$ with $L \times 1$ to N . The parameter x_0 is, accordingly, replaced by $2x_0$.

```

REM SUBROUTINE *** CONVOLUTION ***
FOR L = 1 TO N: FX(L) = F(L) * X(L): NEXT L:
      C = X0 * U * 2: F(N+1) = 0
FOR LL = 1 TO N: L = LL + N2: H = 0: K0 = 1
IF L > N THEN K0 = L - IN%(L - N) - 1
IF K0 < 1 THEN K0 = 1
FOR K = K0 TO LL: LK = L - K: P = FR(LK):
      I = L - IN%(LK)
      S = (P * F(I-1) + (1-P) * F(I)) * FX(K):
      H = H + S: NEXT K
H(LL) = C * (H - S/2): NEXT LL: RETURN

```

```

REM SUBROUTINE *** AUXILIARY FUNCTIONS ***
LM = 2 * N: FOR L = 1 TO LM: X(L) = EXP(L * U):
A = LOG(1 - EXP(-U * L))/U: IN%(L) = INT(A):
      FR(L) = A - IN%(L)
NEXT L: X(0) = 1: RETURN

```

B. SIMULATION OF SPECIFIC ENERGY DISTRIBUTIONS

The method of successive convolutions permits the computation of the distribution of specific energy for a given dose. If, as in the computations for Figs. 12 to 14, values of specific energy are required for many different values of absorbed dose, simulations can be more efficient. A fast method to perform such simulations is therefore described.

As explained in the first part of this Appendix, the single-event distribution $f_1(z)$ is represented by an array on a logarithmic scale of z . The integral distribution is then computed on the same scale, but in contrast to the usual convention it is summed from the tail:

$$S_1(z) = \text{Prob}\{z \geq z\} = 1 - F_1(z) \quad (\text{A.16})$$

Taking the example of 15-MeV neutrons and a 6- μm spherical site and plotting $\ln(S_1)$ versus $\ln(z)$ one obtains the lowest curve in Fig. 30.

When a random variable z , with sum distribution $S_1(z)$, is to be generated, the familiar procedure is to choose a random value S_1 , uniformly distributed between 0 and 1, and to select the corresponding z . The simulation requires, therefore, the reciprocal of the sum distribution. For this purpose values of z corresponding to a logarithmic grid of the probabilities are stored in an array:

$$z(L) \quad (L = 0 \text{ to } LL), \quad \text{with corresponding probabilities } S_1 = \exp(-L/M) \quad (\text{A.17})$$

With $LL = 200$ and $M = 16$ one covers, in this way, the range of probabilities 1 to 3.7×10^{-6} .

To obtain a value of specific energy for one event, one generates a random number S , uniformly distributed between 0 and 1, and sets $L = -\ln(S) \cdot M$. With proper interpolation one obtains then from the array $z(L)$ the random value z .

*Computing times can be reduced by utilizing fast algorithms to generate exponentially distributed random variables without computation of a logarithm.

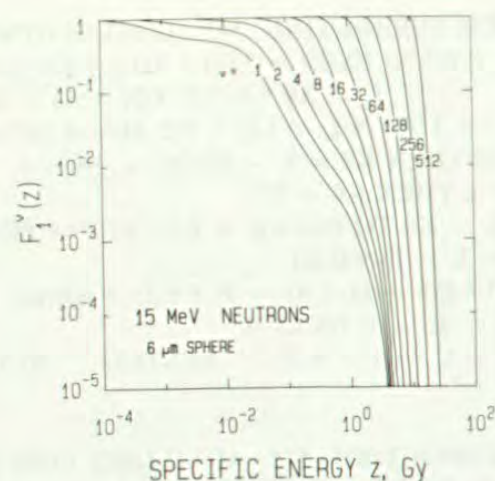


Fig. 30. Sum distributions $1 - F(z)$ for a spherical tissue region of $6\text{-}\mu\text{m}$ diameter exposed to 15-MeV neutrons. Such curves are used for the fast simulation of specific energy. The event numbers are indicated as parameters.

Simulation of random paths, such as in Fig. 10, is the naive method to obtain random values of z at specified absorbed dose D . A random path is a succession of random steps of D and z . The random steps of dose are generated by choosing a random number S and setting $\Delta D = -\ln(S) \cdot \bar{a}_p$; this corresponds to the sum distribution, $S = \exp(-\Delta D/\bar{z}_p)$, of doses ΔD to the next event. The random steps of z are obtained as described above. The process is continued until the specified dose D is reached.

The simulation of a large number of events is, however, too uneconomical. Instead one utilizes sum distributions of z obtained by successive convolutions. These distributions that correspond to the event numbers 2^N are also shown in Fig. 30. They are stored in the same way as the single-event distribution.

For a given absorbed dose one computes first the expected event number $D/\bar{z}_p$. Using the inverse of the Poisson distribution one selects then a random event number that belongs to this mean value. (For large values one uses the Gauss distribution.) The resulting event number is split into integer powers of 2. For each of the powers a random value of specific energy is selected and the values are added. The result is a random value of specific energy belonging to the dose D .

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3

Dosimetry for External Beams of Photon and Electron Radiation

ANDREE DUTREIX AND ANDRE BRIDIER

SERVICE DE RADIOPHYSIQUE
INSTITUT GUSTAVE-ROUSSY
VILLEJUIF, FRANCE

I. The Detector	164
A. Choice of the Detector	164
B. General Considerations on the Use of Air-Filled Ionization Chambers	169
C. Calibration of the Ionization Chamber at a Standards Laboratory	173
II. Dosimetry for Photon Beams	181
A. Electronic Equilibrium Conditions	183
B. Radiation Quality	188
C. Measurement of the Absorbed Dose at a Reference Point	192
D. Measurement of the Absorbed Dose at Any Point in a Phantom	199
III. Dosimetry for Electron Beams	202
A. Electron Beam Energies	202
B. Measurement of the Absorbed Dose at the Reference Point	208
C. Measurement of the Absorbed Dose at Any Point	214
IV. Uncertainties	216
A. General Considerations	216
B. Uncertainties in the Determination of the Absorbed Dose at the Reference Point	219
C. Uncertainties in the Determination of the Absorbed Dose at Any Point	221
References	223

The purpose of this chapter is to provide basic data and information for practical dosimetry in high-energy photon and electron beams. This chapter is not intended to review extensively the very abundant literature which has appeared on the subject during the last thirty years. It is intended to deal only with those

theoretical considerations which are necessary for the understanding of the methods proposed. The discussion may appear to focus on medical applications because several protocols for dosimetry in this field have been developed. However, the techniques are more generally applicable. The various dosimetry methods are briefly reviewed to show their advantages and disadvantages for dosimetry of high-energy photon and electron beams. Some recent references are given in which the reader can find more comprehensive information.

Since this chapter was written, several papers have been published on the measurement of absorbed dose. Furthermore, discussions among physicists from different countries have clarified some ideas about dosimetry. Considering these developments, some sections may have been written from a different perspective. However, the authors have decided not to modify the original text, but only to add comments when needed.

The International Commission on Radiation Units and Measurements (ICRU) has assigned the date of December 1985 for exclusive use of SI units*; only these units are used in the present chapter. The use of SI units implies the use of grays (Gy) instead of rads and coulombs per kilogram ($C\ kg^{-1}$) instead of roentgens (R).†

I. The Detector

This chapter is not a discussion of instrumentation, but of the application of instruments in dose measurement. This section on detectors is not intended to be exhaustive, rather its purpose is to provide background and a basis of comparison for several detector options.

A dosimeter consists of a radiation-sensitive part (the detector) and an electronic device used to transform the signal of the detector into an electrical signal (the reading device). Only the detectors are considered in this chapter and the reading devices are assumed to be reproducible and accurate.

A. CHOICE OF THE DETECTOR

The size of the detector must be small enough to minimize perturbations of the particle fluence but large enough to be subjected to a large number of interactions and to yield a signal that can be read with precision.

The atomic composition of the detector should be as similar as possible to the medium in which the absorbed dose has to be measured in order to reduce the perturbations of the particle fluence. The ideal situation is achieved when the

*It does not imply the use only of the fundamental units like kilogram and meter. The submultiples like gram or centimeter are fully authorized. For instance, densities can be expressed in grams per cubic centimeter, and not necessarily in kilograms per cubic meter.

†1 Gy = 100 rad; $1\ C\ kg^{-1} = 3876\ R$.

sensitive material of the detector is matched to the medium, which is the case for the ferrous sulfate dosimeter in a water phantom. For mechanical or electrical reasons, the detector is often surrounded by a container different from the detector material, e.g., the wall of an ion chamber or the vessel of a Fricke dosimeter. The atomic composition of the "wall" should ideally be matched to both the medium and the detector.

If the sensitive material cannot be matched exactly to the medium, then the atomic composition of the "wall" should be matched to either the medium or the detector itself. When the atomic composition of the wall is matched to the medium, the detector is considered a detector "without wall." When the atomic composition of the wall is matched to the detector, the correct choice of wall thickness, depending upon the secondary electron spectrum, must ensure that perturbing boundary effects will occur only within the wall.

1. Calorimeters

Calorimetry is a fundamental dosimetry method. During the past 10 years the size of the sensitive part of a calorimeter, the absorber, has been decreased to a very small mass. Furthermore, tissue-equivalent plastic or water may be used in place of graphite (Laughlin and Genna, 1966; Guiho and Simoen, 1975a; Hofmeester, 1981; McDonald *et al.*, 1981; Domen, 1982).

However, calorimeters are not generally commercially available and are used only in national or research laboratories for the determination of fundamental physical parameters.

2. Chemical Dosimeters

Chemical dosimeters, especially ferrous sulfate Fricke dosimeters, have long been used as reference dosimeters (Fricke and Hart, 1966; Guiho and Simoen, 1975b; Cottens *et al.*, 1981). The (dilute) solution used as a detector may be considered as water equivalent in atomic composition as well as in density. The detector does not perturb the particle fluence in a water medium provided the irradiation vessel containing the solution is made of a water-equivalent material. The reading (optical density) of the dosimeter can be directly converted to absorbed dose, but it is advisable to check the response of the whole system, including the photodensitometer, in a calibrated radiation beam such as a cobalt-60 γ beam. Unfortunately, chemical dosimeters present two great disadvantages:

- (i) the range of the absorbed dose which may be determined with adequate accuracy is high (e.g., 50 to 200 Gy for the conventional Fricke dosimeter);
- (ii) the minimum amount of solution needed for an adequate signal is about $2\ cm^3$, which makes it difficult to use in high dose-gradient regions.

Furthermore, the cleaning of the vessels and the preparation of the solution require great care (Fricke and Hart, 1966; Cottens *et al.*, 1981).

3. Gas-Filled Ionization Chambers

Ionization chamber dosimetry is the most common method used for absorbed dose measurements in radiation beams of x-ray generators and cobalt-60 therapy units. It is recommended in most published protocols (American Association of Physicists in Medicine, 1966; Hospital Physicists' Association, 1971; ICRU, 1969, 1972; Deutsche Institut für Normung, 1975; Nordic Association of Clinical Physics, 1980, 1981).^{*} Description of ion chambers and fundamentals of the theory have been reviewed in the literature (Boag, 1966) and are beyond the scope of the present chapter. Sections II and III deal with the determination of the absorbed dose from the dosimeter reading.

Compared with the other detectors, ion chambers are readily available, and are simple to use as field instruments. They are precise and reproducible and can be related simply to the national standards. They are the oldest radiation detectors used and ionization chamber theory is well documented. With modern electronic amplification techniques, a low signal can be detected with the needed accuracy and reproducibility. This means that the size of the sensitive cavity can be decreased to a fraction of a cubic centimeter while still covering doses from a fraction of a milligray to 10 Gy or more.

The shape of an ion chamber can be chosen to best match the characteristics of the radiation field. In homogeneous radiation fields a cylindrical (thimble chamber) or spherical cavity is usually preferred, while in high dose-gradient regions, a flat parallel-plate chamber is recommended. Not only is the volume small, but the amount of sensitive material needed (the mass of gas in the cavity) is very small, leading to good spatial resolution.

The use of ionization chambers requires certain precautions and corrections before the readings can be interpreted in terms of a dosimetric quantity. Some of these relate to atmospheric conditions and electronic characteristics and are described in Sections I, B, 2 and I, B, 3. Other corrections result from chamber design and its interaction with the radiation field. For example, the radiation-sensitive material used in most commercially available ionization chambers is air, which differs from water or tissues not only in atomic composition but also in density. The large difference in density between the air of the cavity ($1.205 \cdot 10^{-3} \text{ g cm}^{-3}$ at 20°C) and medium ($\sim 1 \text{ g cm}^{-3}$) results in a perturbation of the particle fluence when even a small volume of medium is replaced by the air-filled cavity. The perturbation corrections to be applied to the readings are discussed in Sections II, C and III, B.

^{*}Hospital Physicists' Association (1983), American Association of Physicists in Medicine (1983), and Comité Français "Mesure des Rayonnements Ionisants" (1985) were published after the present manuscript was written.

4. Radiographic Dosimeters

Radiographic dosimetry is an excellent practical method for relative dose measurement in high-energy photon and electron beams (Loevinger *et al.*, 1961; Dutreix and Dutreix, 1969; Dutreix, 1976a; Johansson and Svensson, 1982). It provides a permanent document of the dose distribution with the best possible spatial resolution: a fraction of a millimeter in the plane of the film. Because of its high spatial resolution, radiographic film is the detector of choice in regions with high dose gradient, especially in electron beams. However, certain precautions must be taken when irradiating films (Dutreix and Dutreix, 1969). The response is highly photon energy dependent, hence films must be used with caution for low- and medium-energy photons such as cobalt-60 γ rays (Marinello and Sliwinski, 1974; Johansson and Svensson, 1982). Angular dependence has been found to be negligible by Dutreix and Dutreix (1969) even for films parallel to the beam axis. Even though films are very convenient for relative measurement of dose distribution they should not be used for dose determination except where uncertainties due to energy dependence can be avoided.

5. Solid-State Dosimeters

Various types of solid-state dosimeters have been described and can be used for dosimetry. The most commonly used are thermoluminescent dosimeters and semiconductor junction detectors. Both of them have a density higher than that of water, and both are available in small sizes. However, because they are made of materials which differ from water in atomic composition and density ($2\text{--}3 \text{ g cm}^{-3}$), they perturb the particle fluence in a water phantom. Moreover, they are energy dependent with respect to absorbed dose in water. Therefore they must always be calibrated under appropriate conditions to yield absorbed dose. Thermoluminescent dosimeters are very convenient for *in vivo* measurements because they have a small volume, there is no associated voltage, they are readily available, and their signal is easily read out (Fowler and Attix, 1966; Attix, 1965; Cameron *et al.*, 1968).

Some authors have reported reproducible and accurate results obtained with semiconductor detectors in relative dose-distribution measurements. However, in our experience as well as that of some others (Naudy, 1981; Johansson and Svensson, 1982), semiconductor detectors are not only energy dependent but also dose-rate and temperature dependent. Furthermore, they may be damaged after irradiation by 100 Gy or more. Precautions must be taken to check the characteristics of such a detector before using it.

6. Comparison of the Different Dosimeters

Table I summarizes the main dosimetric properties of the various methods together with their limitations and the uncertainties involved. The various methods can be divided into three groups.

TABLE I
CHARACTERISTICS OF DOSIMETERS

Dosimeter	Main use	Limitations	Uncertainties
Calorimeter	Absolute determination of absorbed dose. Determination of basic physical parameters.	Research laboratories	Systematic: 0.24% (Guiho and Simoen, 1975a)
Ferrous sulfate dosimeter	Absorbed dose determination through measurement of physical quantities. Reference dosimeter.	Dose measured: 50 to 200 Gy. Poor spatial resolution.	In <i>G</i> -value determination: 1.20% (Guiho and Simoen, 1975b)
Ionization chamber	Absorbed dose determination through measurement of physical quantities. Reference or field dosimeter. Excellent precision.	Radiation-sensitive material (air) different from water and tissue. Perturbation of the particle fluence.	For reference dosimeter: systematic: 0.55% (Guiho and Simoen, 1975a); for field instrument, see Table IV.
Photographic dosimeter	Relative measurement of dose distribution. High spatial resolution.	Energy dependence. Poor reproducibility in the signal to dose ratio.	The uncertainty varies from a few percent to more than 10% depending on the irradiation conditions.
Solid-state dosimeters Thermoluminescent dosimeters Semiconductor junctions	<i>In vivo</i> dosimetry. Small volume.	Atomic composition $\neq$ water. Perturbation of the particle fluence. Energy, dose rate, and temperature dependence.	The uncertainty varies from a few percent to more than 10% depending on the irradiation conditions.

(i) Calorimetry is a fundamental method and the most nearly absolute method for measuring absorbed dose. In calorimetry, one measures directly (through temperature rise) the quantity of interest, namely, the energy absorbed by matter. Unfortunately it is time-consuming, difficult, and expensive, and should be left to standards laboratories and a few research laboratories.

(ii) Ferrous sulfate dosimetry and ionization chamber dosimetry are methods in which absorbed dose can be calculated from measurements of physical quantities: optical density and optical path length of the solution for ferrous sulfate dosimetry, and electrical charge and volume of the cavity for ionization chamber dosimetry. Dosimetric constants which have been determined by

calorimetry are used to calculate the absorbed dose in both methods but the uncertainty in W/e (the average energy expended in air per unit charge of ionization) is much less (0.5%) than the uncertainty in G (the yield of ferric ions per 100 eV absorbed energy, 1.2%).

In a laboratory far from a standards laboratory where it is difficult and expensive to have an ionization chamber regularly calibrated, ferrous sulfate dosimetry can serve as a reference. As ferrous sulfate solution is nearly water equivalent, it does not perturb the particle fluence and its response per unit dose is quite independent of radiation energy over a wide range. It can be used as a reference for the determination of some perturbation correction factors for ionization chambers (Johansson *et al.*, 1978) and as a supplementary check for absorbed dose determination at a reference point.

For any other purpose, ionization chamber dosimetry is the method of choice either for reference dosimetry or for field dosimetry. It is considered in detail in the following sections.

(iii) The last two methods, radiographic dosimetry and solid-state dosimetry, cannot be used for absorbed dose determination without careful calibration under the appropriate irradiation conditions. They are, however, very useful for relative dose measurements which may be done *in vivo*.

B. GENERAL CONSIDERATIONS ON THE USE OF AIR-FILLED IONIZATION CHAMBERS

1. Choice of Water as a Reference Medium

For dosimetric work, the human body has been considered water equivalent for two main reasons: first, differences in atomic composition and density between water and soft tissues are small and the accuracy in absorbed dose determination is not high enough to justify the small correction these differences introduce; second, determination of the size and composition of various internal organs is complicated, and to assume that the whole body is water equivalent is believed to be a safer basis for dosimetry. The precision of absorbed dose determination has increased and the use of computerized tomographic scanners has made it easy to determine size and density of organs.

A standard muscle tissue has been proposed as reference medium both for absorbed dose calculations and for dose-distribution determinations in neutron beams for which hydrogen content is critical [Report 26 (ICRU, 1977)]. However, for high-energy photon and electron dosimetry, the ICRU in Report 24 (ICRU, 1976) and in Report 21 (ICRU, 1972) has recommended the continued use of water as reference medium. This means that one must determine the absorbed dose to a small mass of water in a water phantom. The two main arguments for water are, first, that water is an unambiguously defined medium and easily available and, second, that medical experience has been related for a long

time to in-water dose distributions. A third argument could be that since various human tissues (such as brain, liver, bone, etc.) are different from the standard muscle tissue, corrections would be necessary anyway when accurate dose distributions in the body are needed.

2. Corrections Related to Atmospheric Conditions

Determination of absorbed dose to air in a cavity implies determination of the mass of air filling the cavity. Since most ion chambers are unsealed and open to the atmosphere, the mass of air inside the fixed volume of the cavity depends upon the temperature t and pressure p of the ambient air. If the calibration conditions are as usual, 22°C and 101.33 kPa,* the mass of air under the t and p conditions is

$$m_{t,p} = m_{22, 101.33} \cdot \frac{295.2}{273.2 + t} \cdot \frac{p}{101.33}$$

As the ionization is inversely proportional to the mass of air, the charge measured must be multiplied by

$$k_{t,p} = \frac{273.2 + t}{295.2} \cdot \frac{101.33}{p} \quad (1)$$

to give the correct dose.

The measurement of air temperature and pressure in the room where the ionization chamber is used is usually easy to perform, however, a few precautions should be taken.

Barometers are often corrected for the altitude variation of the pressure to give the pressure "corrected to sea level." Such barometers should be recalibrated to give the correct (local) pressure. Mercury barometers are the most accurate instruments and are recommended.

When ionization chambers are used in a water phantom, the air in the chamber cavity is in general in temperature equilibrium with the water. This temperature may be different from the room temperature, especially if the phantom has been filled with water a short time before the measurement (a situation to be avoided). The time necessary to assure temperature equilibrium depends on the construction of the particular chamber used and should be determined by the user.

When accurate measurements are performed, a correction must be made for the presence of water vapor in air. In some national standards laboratories, ionization chambers are calibrated in dry air conditions, and the calibration value refers to zero relative humidity. However, the number of ions produced when water vapor is present is larger than the value obtained in dry air. In a cobalt-60

*101.33 kPa corresponds to 760 mm of mercury, 760 Torr, or 1 atm.

beam, an increase of 0.3% was observed as the relative humidity increased from 20% to 70% at 20°C and 101.33 kPa (Guiho *et al.*, 1974; Niatel, 1975). This difference appears because of the variation of both the stopping power and W with the amount of water vapor present. Therefore, the charge or the dose should be multiplied by 0.997 in usual humidity conditions, assuming that the chamber calibration factor refers to dry air. Note that the National Bureau of Standards (U.S.A.) calibrates under typical laboratory humidity conditions, hence the humidity should not be corrected for by the user except if humidity conditions are far from usual (~50%).

A few commercially available ionization chambers are described in manufacturers' catalogs as "sealed chambers." These chambers are often designed for *in vivo* measurements and especially for intracavitary measurements. If a variation in pressure or temperature occurs over a few minutes, there is no variation observed in the response of these dosimeters. However, if the ionization chamber is left at a somewhat different temperature or pressure for a long time (1 h or more), a slow variation in the response may be observed, indicating that the chamber is not quite sealed. Precautions should be taken when using such chambers by checking the response to a given dose (or dose rate), for instance, with a check source (strontium-90) before and after each measurement. Calibration laboratories are equipped to check such chambers for leaks.

3. Corrections Related to Electrical Parameters

a. Recombination. As the voltage between the electrodes of an ionization chamber is increased, the current collected approaches asymptotically the saturation current which would be measured if all the ions formed by the radiation were collected. When the applied voltage is not high enough, some ions recombine with ions of opposite sign before they can reach the collecting electrode. This recombination tends to decrease the signal. Recombination can be reduced either by increasing the applied voltage or by reducing the distance between electrodes. However, by increasing the voltage over a given maximum value, a rapid multiplication of ions is observed due to the ionization by collision. This multiplication is strongly voltage dependent and must be avoided in the operation of ionization chambers.

The theory of initial and general recombination has been clearly explained by Boag (1966) for plane-parallel, cylindrical, and spherical geometry in both continuous and pulsed radiation beams. Boag defined the collection efficiency of an ionization chamber as the ratio of the measured current to the saturation current. He has given formulae to calculate the collection efficiency as a function of the following parameters:

- (i) the geometrical specifications of the ionization chamber;
- (ii) the voltage applied;

- (iii) the dose rate if the radiation is continuous; and
- (iv) the dose per pulse if the radiation is pulsed (ICRU, 1982).

More recently Boag and Currant (1980) have studied experimentally the collection efficiency in a small chamber exposed to identical recurrent pulses of radiation.

In some modern accelerators used in radiation therapy the pulsed electron beam is "swept" to cover a field size as large as $40 \times 40 \text{ cm}^2$ at 1 m from the accelerating tube window. The beam is moved in a zigzag pattern over a frame, and successive frames are interlaced to achieve good dose uniformity over the whole field. When an ionization chamber is exposed to such an electron beam, it is irradiated by full pulses when the electron beam is directed on the chamber and with low intensity or zero pulses when the beam axis is distant from the chamber. The formulae established for identical recurrent pulses do not apply to this special case and the theory has been extended by Boag (1984). The collection efficiency has also been determined experimentally by Marinello *et al.* (1976) and Majenka *et al.* (1982). Collection efficiencies as low as 0.85 have been estimated in small, commercially available ionization chambers with some swept beams. It is then essential to estimate the collection efficiency of the ion chamber used for the highest dose rate available with the accelerator on which measurements are performed. This estimation can be easily made when the applied voltage can be varied (Boag, 1982). When only a fixed voltage is available, the collection efficiency can be estimated by comparison with a dosimeter which is relatively independent of dose rate, such as a Fricke dosimeter or a thermoluminescent dosimeter.

b. Polarity. The current collected from some ionization chambers, exposed to constant dose rate, changes in magnitude when the polarity of the collecting potential is reversed. The main cause for this polarity effect is the deposition of a net charge by the radiation beam in the electrode material due to a difference between in- and out-transport of electrons. In photon beams this net charge deposition is positive near the surface in the buildup region (more electrons are removed from a small mass of material than are deposited) and is reduced to zero when transient electronic equilibrium is achieved. In noncontaminated electron beams the net charge deposition is positive near the surface because of the buildup of δ rays and negative at depths where the dose decreases when electrons are finally stopped in material (Van Dyk and MacDonald, 1972). The magnitude of the effect depends not only on the maximum energy of the electrons but also on the beam contamination by low-energy electrons, which will be stopped in the first millimeters or centimeters. This charge deposition adds to or subtracts from the charge produced in the air cavity depending on the polarity of the voltage applied.

Polarity effects are particularly noticeable in parallel-plate ionization chambers and can be reduced by a careful design (Morris and Owen, 1975; Markus, 1976; Mattsson *et al.*, 1981), in particular by reducing the electrode thickness.

When the polarity effect is significant (e.g., larger than 1%), it is necessary to perform all the measurements with both polarities and to evaluate the mean of the two readings (with positive and negative polarities).

c. Leakage Current. Irradiation of the stem, connectors, and cable may induce an "extracamerai" electrical current which increases or decreases the readings. This leakage current, when it exists, can be reduced by shielding the stem and the cable. Such an unwanted current is, in general, energy dependent (Boag, 1966).

C. CALIBRATION OF THE IONIZATION CHAMBER AT A STANDARDS LABORATORY

1. General

The first step in the determination of absorbed dose, using an ionization chamber, is the measurement of the response J_{air} of the chamber: the charge of either sign per unit mass of air inside the air cavity. This quantity could be measured directly as the ratio $Q/\rho V$, where Q is the charge measured by a calibrated charge-measuring instrument connected to the chamber, ρ is the air density, and V the active volume of the air cavity.

The active volume is the volume bounded by the limiting lines of force of the electric field along which ions drift to the guard ring. This volume may be smaller than the geometric volume of the cavity. Usually direct measurement of the active volume is too complicated.

A simple procedure is to calibrate the ionization chamber in a national standards laboratory, to obtain one known quantity for a given x- or γ -ray quality, and to use the calibration data for the evaluation of J_{air} . Most of the national laboratories use cobalt-60 γ -rays or 2-MV x rays as the principal calibration quality, and calibrate ionization chambers in terms of exposure. In fact, this process is a radiometric determination of the mass of air in the active volume of the cavity.

The exposure calibration factor N is used to determine absorbed dose in a phantom irradiated by photons following the equation recommended by the ICRU (1969, 1976):*

$$D = RNk_1k_2C_A \quad (2)$$

*A similar equation is proposed in ICRU (1972) for electron beams where C_A is replaced by C_E .

where R is the reading of the dosimeter, k_1 is the correction factor for atmospheric conditions [Eq. (1)], k_2 is a factor to correct for differences between the radiation beam used for calibration and that prevailing at the point of measurement, and C_λ is a rather ambiguous factor relating exposure to absorbed dose in the material of the phantom, including fundamental physical parameters related to radiation quality as well as correction factors related to ionization chamber characteristics.

Because of the introduction of SI units for radiation quantities and because of the eventual probability that national standards laboratories will cease to make exposure calibrations in favor of some other quantity, it appears to be necessary to revise the formalism used for the absorbed dose determination and to propose new protocols.

New protocols were adopted by the Nordic countries (Nordic Association of Clinical Physics, 1980, 1981).^{*} The ICRU (1984) has adopted the same formalism for high-energy electron beams and one may expect that a similar formalism will be proposed in the near future by the ICRU for high-energy photon beams.

In these new protocols the main steps for the determination of the absorbed dose at a point P in a medium are as follows:

(i) Calibration of the measuring device in terms of a chosen dosimetric quantity q . Determination of the q calibration $N_{q,c}$ factor where subscript c stands for calibration quality (see Section I, C, 2).

(ii) Determination of the mean absorbed dose to air in the air cavity $\overline{D_{air,c}}$. Calculation of the calibration factor $N_{D,c}$ in terms of $\overline{D_{air,c}}$. If the charge-measuring device is calibrated, determination of the mass calibration factor (see Section I, C, 3).

(iii) Determination of the absorbed dose to water at a reference depth in a water phantom (see Section II, C for photon beams and Section III, B for electron beams).

2. The Dosimetric Quantity q for Dosimeter Calibration

Dosimeters have been conventionally calibrated in terms of exposure, not only for historical reasons but also because exposure is a quantity that is easy to define and to reproduce using primary standards. It became increasingly evident in the 1950's that the physical quantity best correlated with biologic effects was not exposure but absorbed dose. In the choice of units, advantage was taken of the fact that the numerical value of the exposure measured in roentgens does

^{*}After the present chapter was written, similar protocols were adopted by the United States (American Association of Physicists in Medicine, 1983) and France (Comité Français "Mesure des Rayonnements Ionisants," 1985).

not differ by more than a few percent from the numerical value of the absorbed dose in tissue or water measured in rads, thus eliminating the possibility of too serious consequences when the two quantities were confused.

With the introduction of SI units, this similarity of units is lost, and other quantities have been proposed and are under discussion among national laboratories and international commissions.

Reich (1979) has clearly defined the terms of the problem, which can be summarized as follows. In the past it was assumed that the same quantity should be used within the calibration chain from primary standards to field instruments. However, the quantity in terms of which primary standards deliver their readings should be close to the physical effect on which the standard is based in order to require as few correction factors as possible. The aim should be to allow for high-precision comparisons between primary standards.

The quantity in terms of which field instruments are calibrated should be closest to the needed quantity: absorbed dose to water. It should allow the user to obtain the numerical value required by applying as few correction factors as possible.

Changing the quantity between primary standards and field instruments means that the conversion factor should be provided by the standards laboratories and incorporated into the calibration factor. The calibration certificate could, for instance, give calibration factors in terms of both quantities so that the values could be traced.

Exposure is currently the quantity most often used in calibration of field instruments, but probably it will be progressively abandoned by national laboratories. Four quantities are under discussion for the future:

- (i) kerma to air, "in air" (i.e., in an open beam in free space);
- (ii) kerma to water, "in air";
- (iii) absorbed dose to water, "in water" (i.e., a water phantom);
- (iv) absorbed dose to air in the air cavity of the ionization chamber used.

Adoption of the last quantity allows one to jump over the first step in following the proposed formalism (see the next section).

The calibration in terms of absorbed dose to water in water allows the user to measure the required quantity directly without any correction coefficient provided that the calibration is done for the radiation quality used. Such a calibration could, in principle, be done directly by calorimetry, avoiding the use of any correction coefficient throughout the calibration chain (except the correction for thermal defect in water). Such a direct calibration could be performed under specified reference conditions for the x-ray qualities required by the user because of the small number of different qualities used at a given institution. However, such a procedure would be difficult to apply for electron beams, since the mean energy decreases continuously with depth from a maximum at the phantom

surface and the various corrections depending upon energy could not be avoided.

The other two quantities, kerma to air in air and kerma to water in air, could be used in approximately the same way as exposure is now. However, to minimize the risk of mistakes in the correction coefficients used with the proposed protocol, kerma to air in air seems to be more convenient than kerma to water in air. In practice the most convenient quantity is the collision kerma K_{col} in air (Attix, 1979), which is more directly related to both the exposure and the absorbed dose in air, but to our knowledge it has not yet been seriously considered as a calibration quantity by national laboratories.

The three calibration quantities, exposure X , kerma to air K_{air} , and kerma to water K_{water} , are specified "in air." The words "kerma to water in air" mean that the mass of water assumed to be placed at the point of interest is small enough not to perturb the photon fluence at this point. The quantities K_{water} , K_{air} , and X are related by simple and unambiguous physical relations:

$$K_{\text{air}} = X \cdot \frac{\bar{W}}{e} \cdot \frac{1}{1 - g} \quad (3)$$

$$K_{\text{water}} = K_{\text{air}} \cdot \left[\frac{\mu_{\text{tr}}}{\rho} \right]_{\text{air}}^{\text{water}} \quad (4)$$

$$K_{\text{water}} = X \cdot \frac{\bar{W}}{e} \cdot \left[\frac{\mu_{\text{tr}}}{\rho} \right]_{\text{air}}^{\text{water}} \cdot \frac{1}{1 - g} \quad (5)$$

In SI units K_{air} and K_{water} are expressed in grays and X in coulombs per kilogram. The quotient $\bar{W}/e$ is the average energy expended in air per unit charge of ionization ($\bar{W}/e = 33.85 \pm 0.15 \text{ J C}^{-1}$ for dry air). The quantity g is the fraction of the secondary electron energy radiated as bremsstrahlung in air [equal to about $4 \cdot 10^{-3}$ for cobalt-60 γ rays (Boutillon, 1977)]. And $[\mu_{\text{tr}}/\rho]_{\text{air}}^{\text{water}}$ is the ratio of the average value of the mass energy transfer coefficients for water and air calculated for the photon energy distribution at the point of interest.

3. Determination of the Calibration Factor $N_{D,c}$

Whatever the calibration quantity, secondary electronic equilibrium must be achieved in the air cavity for the radiation quality used for calibration. Assuming the ionization chamber is calibrated in a cobalt-60 γ -ray beam, it is generally recommended that the total thickness of the wall and the added cap be equal to $0.45 \pm 0.05 \text{ g cm}^{-2}$. A total thickness of 0.65 g cm^{-2} should be necessary to exclude entirely the most energetic photoelectrons.

The calibration factor in the quantity q is $N_{q,c} = q_c/M_c$, where c stands for the calibration beam quality, and M_c is the meter reading corrected for atmospheric conditions (i.e., to 22°C , 1 atm) and for recombination losses. The

value of the quantity q at the point where the geometrical center of the ionization chamber is placed in free space, in the absence of the ionization chamber, is given by q_c .

The calibration factor needed is $N_{D,c}$, defined as

$$N_{D,c} = \bar{D}_{\text{air},c}/M_c \quad (6)$$

where $\bar{D}_{\text{air},c}$, the mean absorbed dose to the air filling the cavity, can be written

$$\bar{D}_{\text{air},c} = J_{\text{air},c} \cdot (\bar{W}/e) \quad (7)$$

where $J_{\text{air},c}$ is the total ionization charge of either sign produced in the air cavity divided by the total mass of air in the cavity. One may write $N_{D,c} = N_{m,c} \cdot (\bar{W}/e)$, where $N_{m,c} = J_{\text{air},c}/M_c$ is the mass calibration factor. When the ionization chamber is connected to a calibrated charge-measuring instrument, $M_c = Q$, the charge in coulombs, and

$$J_{\text{air},c} = (Q/(\rho \cdot V))_c, \quad N_{m,c} = 1/(\rho \cdot V) \quad (8)$$

ρ being the density of air and V the sensitive volume of the cavity. Thus, ρV is the mass of air filling the sensitive volume (i.e., the region from which the charge is collected).

The mass calibration factor $N_{m,c}$ is expressed in inverse kilograms. It is specific to each ionization chamber, and could be calculated by national standards laboratories.

Since $1/(\rho \cdot V)$ and $\bar{W}/e$ are independent of energy, the two ratios, $J_{\text{air},c}/M_c$ and $\bar{D}_{\text{air},c}/M_c$ are independent of energy. The calibration factors

$$N_{D,c} = (\bar{D}_{\text{air}}/M)_c \quad \text{and} \quad N_{m,c} = (J_{\text{air}}/M)_c$$

are then specific to each ionization chamber, and are not beam quality dependent, as long as $\bar{W}/e$ remains constant.

The mean absorbed dose to the air in the cavity $\bar{D}_{\text{air},c}$ is equal to the collision kerma $K_{\text{col,air},c}^\circ$ in a small mass of air at the point where the cavity is centered, in absence of the ionization chamber, corrected for the influence of the wall and the cap surrounding the cavity (Figure 1):

$$\bar{D}_{\text{air},c} = K_{\text{col,air},c}^\circ \cdot k_{\text{att},c} \cdot k_{m,c} \quad (9)$$

$$\bar{D}_{\text{air},c} = K_{\text{air},c} \cdot (1 - g) \cdot k_{\text{att},c} \cdot k_{m,c} \quad (10)$$

where $k_{\text{att},c}$ takes into account the absorption and scattering (attenuation) of the calibration radiation in the wall and cap materials, $k_{m,c}$ takes into account the lack of air equivalence of the wall and the cap for the calibration, and g is the average fraction of the secondary electron energy lost to bremsstrahlung in air.

Therefore $N_{D,c}$ can be related to any calibration factor $N_{q,c}$. The relations between $N_{D,c}$ and $N_{q,c}$ are detailed in Table II.

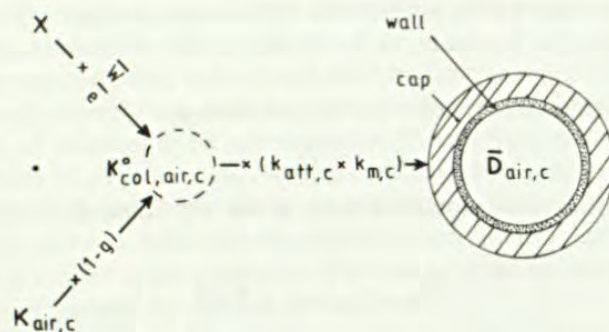


Fig. 1. Estimation of the mean absorbed dose to cavity air $\bar{D}_{air,c}$ from the calibration, in terms of exposure X or kerma to air $K_{air,c}$.

4. Correction Factors $k_{att,c}$ and $k_{m,c}$

In Eq. (10), $k_{att,c}$ and $k_{m,c}$ are factors which account for the modification to the collision air kerma at point P caused by the presence of the ionization chamber. If the wall and cap of the chamber are air equivalent, the only modification is the attenuation of the photon beam ($k_{att,c}$); $k_{m,c}$ takes account of any modification of the secondary electron fluence within the cavity when the wall and cap of the ionization chamber are not air equivalent. These two factors have been evaluated by several authors for cobalt-60 γ rays.

a. *The Factor $k_{att,c}$ for Cobalt-60 γ Rays.* Johansson *et al.* (1978) have determined experimentally the values of $k_{att,c}$ for cobalt-60 γ rays by extrapolating measured ionization to zero wall thickness, as described by Loftus and Weaver (1974). The extrapolated value was corrected by 0.5% to take into account that electrons giving rise to ionization in the cavity are generated in front of the cavity by photons penetrating through only a part of the wall. The mean of the experimentally determined $k_{att,c}$ values for the chambers with a total wall thickness of 0.6 g cm^{-2} was 0.990. The $k_{att,c}$ factor is dependent on the size and the shape of the chambers, and its value varies between 0.98 and 1.00 for different types of ionization chambers used (Niatel *et al.*, 1975; Loftus and Weaver, 1974; Almond and Svensson, 1977).

An average value of 0.990 has been proposed by the Nordic Association of Clinical Physics (NACP, 1980) for cylindrical ionization chambers with different wall and cap material combinations and a total, wall plus cap, thickness of about 0.5 g cm^{-2} .

b. *The Factor $k_{m,c}$.* When the wall and the cap are made of the same material m (Whyte, 1959; Loftus and Weaver, 1974)

$$k_{m,c} = [\bar{\mu}_{en}/\rho]_{air,c}^m \cdot [\bar{S}/\rho]_{m,c}^{air} \quad (11)$$

TABLE II

RELATIONS BETWEEN $N_{D,c}$ AND $N_{q,c}$

Calibration quantity q	Calibration factor $N_{q,c}$	Calibration factor in terms of mean absorbed dose to air cavity $N_{D,c}$	Mass calibration factor $N_{m,c}$
X	$N_{X,c}$	$N_{X,c} \cdot (\bar{W}/e) \cdot k_{att,c} \cdot k_{m,c}$	$N_{m,c} = \frac{N_{D,c}}{W/e} = \frac{1}{\rho \cdot V}$
K_{air}	$N_{K_{air},c}$	$N_{K_{air},c} \cdot (1 - g) \cdot k_{att,c} \cdot k_{m,c}$	
K_{water}	$N_{K_{water},c}$	$N_{K_{water},c} \cdot (1 - g) \cdot [\bar{\mu}_{tr}/\rho]_{water,c} \cdot k_{att,c} \cdot k_{m,c}$	

Values for $k_{m,c}$ using $[\bar{\mu}_{en}/\rho]_{air,c}^m$ values from Hubbell (1977) and $[\bar{S}/\rho]_{m,c}^{air}$ values from Berger and Seltzer (1982) have been prepared for several materials by the ICRU for their report on electron beams (ICRU, 1984). Almond and Svensson (1977) have proposed an evaluation of $k_{m,c}$ in the case where the wall is made of one material (m_w) and the buildup cap of another material (m_b):

$$k_{m,c} = \alpha \left[\frac{\bar{\mu}_{en}}{\rho} \right]_{air,c}^{m_b} \cdot \left[\frac{\bar{S}}{\rho} \right]_{m_b,c}^{air} + (1 - \alpha) \left[\frac{\bar{\mu}_{en}}{\rho} \right]_{air,c}^{m_w} \cdot \left[\frac{\bar{S}}{\rho} \right]_{m_w,c}^{air} \quad (12)$$

where $[\bar{\mu}_{en}/\rho]_{air,c}^m$ is the ratio of the mean mass energy-absorption coefficient for the wall and cap material (m) to that of the air, for the calibration beam quality (c), and $[\bar{S}/\rho]_{m,c}^{air}$ is the ratio of the mean, restricted collision mass stopping power of the air to that of the wall and cap material (m), for the calibration beam quality (c). Here α represents the fraction of the ionization due to electrons generated in the cap material and $(1 - \alpha)$ is the fraction ascribed to electrons from the wall itself. This equation assumes that the contributions of the wall and of the cap are independent. This is certainly not true but it should be correct as a first approximation.

Experimental investigations of $k_{m,c}$ values have been carried out by Almond and Svensson (1977). They have independently performed experiments with cylindrical ionization chambers with air-equivalent or tissue-equivalent walls of about 0.1 g cm^{-2} thickness and with caps made of different materials (graphite, Perspex, aluminum, etc.). The responses obtained, for the same ionization chamber, with different wall and cap combinations were normalized to a uniform air-equivalent chamber to be used as $k_{m,c}$ values. The comparison between the experimental values and the theoretical values determined from Eq. (12) shows, for all experiments, better agreement with $\alpha = 0$ than with $\alpha = 1$. It could be concluded that, when the wall thickness is equal or greater than 0.1 g cm^{-2} , the wall material is of larger significance for $k_{m,c}$ than the cap material (Almond and Svensson, 1977). This means that a large fraction of the electrons ionizing the air of the cavity are issued from the thin layer of material surrounding the cavity (Gray, 1937).

Table III gives the $k_{m,c}$ values recommended in the NACP protocol (NACP, 1980) for cylindrical ionization chambers. These values apply to ionization chambers of sizes recommended in the NACP protocol (5-mm inner diameter) and with the wall and the cap made of the same material. For some commercial ionization chambers which do not meet these specifications, as is the case for those with wall and cap made from different materials, experimental ($k_{att,c} \cdot k_{m,c}$) values can be found in the literature. For example, Johansson *et al.* (1978) and Almond *et al.* (1978) showed that a value of 0.975 ± 0.005 can be used for the common combination of a cylindrical ionization chamber with graphite wall and Perspex (plexiglas, Lucite) buildup cap.

If there is no information available in the literature, $k_{m,c}$ values for specific ionization chambers can be obtained with a reasonable approximation from the relation (11), since Almond and Svensson (1977) indicated that the composition of the inner wall is much more important than that of the buildup cap, but experimental determination of $k_{m,c}$ is to be preferred (Mattsson *et al.*, 1981).

5. Uncertainties

In the ionometric method, the ionization chamber is calibrated at a standards laboratory (secondary or national). Each step in the measurement chain, from the standards laboratory to the measurement of radiation dose in an experimental, clinical, or radiation protection application, contributes to the uncertainty in the magnitude of the dose measured, and must be considered. A detailed analysis of the different uncertainties has been made by Johansson (1982). For the estimation of the uncertainties, the authors use a modified and extended version of the Loevinger and Loftus (1977) model.

The absorbed dose at a reference point in the radiation beam at the user's institution can be determined in two ways: either directly with the user's reference instrument, or with a field instrument calibrated at the institution versus the reference instrument. The uncertainties related to the different steps encountered between the calibration of the user's reference instrument at a secondary standards laboratory and the calculation of the $N_{D,c}$ factor are given in Table IV for

TABLE III
 $k_{att,c}$ AND $k_{m,c}$ VALUES FOR CYLINDRICAL
IONIZATION CHAMBERS^a

Chamber wall and cap material	$k_{att,c}$	$k_{m,c}$
Air equivalent	0.990	1.000
Graphite	0.990	0.991
Tissue equivalent (A 150)	0.990	0.963

^aData from NACP (1980).

the two procedures. The uncertainties considered in each step are the overall uncertainties, which consist of a combination of random and systematic uncertainties. They are obtained according to a method recommended by the International Atomic Energy Agency (IAEA, 1979). The accumulated overall uncertainties for the part of the measurement chain under consideration are also given in Table IV.

The most important contribution to the overall uncertainty is due to systematic uncertainties related to calculation of the $N_{D,c}$ factor (2.6%). These potential errors derive from the recommended value of k_{att} (0.7%), the k_m value, which depends on mass stopping power and energy-absorption coefficient ratios between air and wall (1.5%), and, for a few types of ionization chambers, the lack of data about the material in the wall and buildup cap or about the k_m value for the material or combination of materials (2%).

The other main systematic uncertainties relate to the nonlinearity of the reference or field instrument (respectively, 0.5% and 1%) and to measurements of temperature and pressure (about 0.6% for pressure).

To the overall uncertainties given in Table IV must be added the random uncertainties introduced by the user into the calibration of the field instrument. These random uncertainties in reproducibility of the reference instrument, in measurement of geometrical parameters (SSD, depth, field size), and in the timer have been estimated by Johansson (1982) to be 0.8% for a cobalt-60 γ -ray beam based on experience gained in the Nordic intercomparison with his own reference instrument.

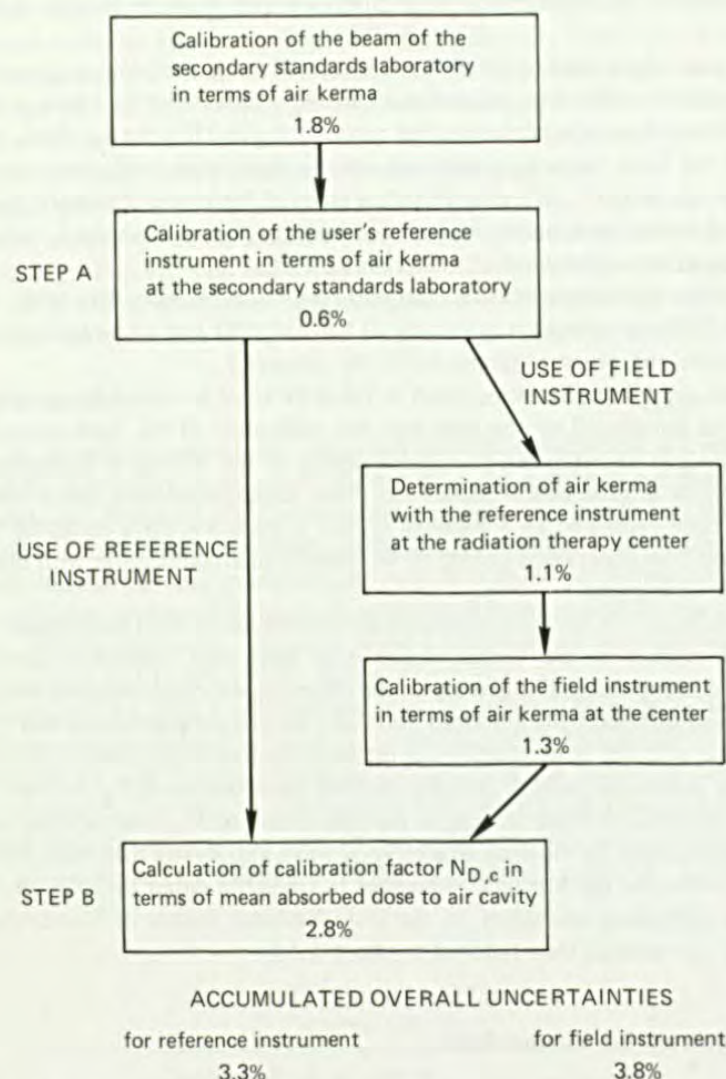
Shalek *et al.* (1976) have compared the calibration of field instruments used by some centers in the United States with their own reference instrument calibrated in exposure. The average of the experimental ratios obtained from this comparison differed by 0.5% from unity and the standard deviation was 2.7%. This value of 2.7% is comparable to the total random uncertainty (2.2%) estimated by Johansson, which includes random uncertainties in the various steps in the measurement chain leading to the calculation of $N_{D,c}$ and random uncertainties introduced by the user. Shalek *et al.* have also shown that when the calibration factor for the ionization chamber is traceable either directly, or via a regional calibration laboratory, to the U.S. National Bureau of Standards, the standard deviation is then reduced to about 1.5%.

II. Dosimetry for Photon Beams

High-energy photon fields such as Cobalt-60 γ rays and megavoltage x rays are present in radiotherapy, industrial, and research institutions. Some of the equations relating the absorbed dose to the reading of the measuring instrument imply that electronic equilibrium conditions are fulfilled. Furthermore, the coefficients used in these equations are strongly energy dependent. It is then

TABLE IV

ESTIMATED INDIVIDUAL AND ACCUMULATED OVERALL UNCERTAINTIES AT EACH STEP IN THE MEASUREMENT CHAIN LEADING TO THE CALIBRATION FACTOR $N_{D,c}$



necessary to consider the exact meaning of the words "electronic equilibrium conditions" and to estimate the "energy" of a photon beam before attempting to measure an absorbed dose.

A. ELECTRONIC EQUILIBRIUM CONDITIONS

1. Absolute Electronic Equilibrium and Transient Electronic Equilibrium

Electronic equilibrium is achieved at a point P in a medium irradiated by x or γ rays if in the sphere of radius R (equal to the maximum range of electrons set in motion by photons) centered at P two conditions are fulfilled: the medium is homogeneous and the photon radiation fluence is uniform in intensity, quality, and angular distribution.

Under these conditions a local physical effect g_1 , produced by the fluence of the secondary electrons at this point in a small mass Δm of the medium, is qualitatively and quantitatively equal to the effect g_d , which is produced in the medium along the tracks of the electrons arising from Δm including their δ rays (Fig. 2). The phenomenon most commonly considered is energy exchange, where g_d should be the collision kerma K_{col} (Attix, 1979) and g_1 the absorbed dose D . The collision kerma is that part of the kerma for which the charged particle kinetic energy is subsequently spent in collision interactions.

The identity $q_d \equiv q_1$ corresponds to an *absolute charged particle equilibrium*. Such an absolute equilibrium could be achieved only in an infinitely extended radioactive medium of uniform concentration.

On the axis of a broad high-energy photon beam the photon radiation fluence is not uniform because of geometric divergence and attenuation of the photons. The attenuation of the photon beam over a distance equal to the maximum range R of the secondary electrons is not negligible: the photon mean free path ($1/\mu$) is not markedly greater than the electron range R , e.g., for 10-MeV monoenergetic photons in water $1/\mu \approx 45$ cm and $R \approx 5$ cm.

Since the emission of the electrons, secondary to high-energy photons, is strongly in the forward direction, the absorbed dose D is higher than the collision kerma K_{col} at depths greater than R in the medium, and the decrease in the

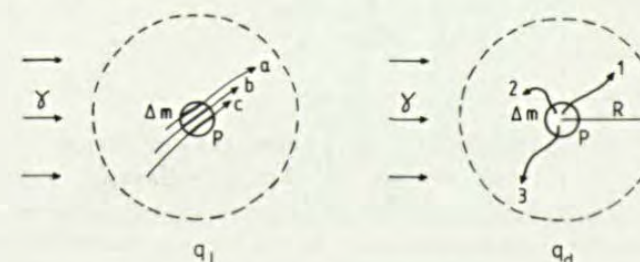


Fig. 2. Illustration of charged particle equilibrium. The local physical effect q_1 produced by the fluence of the secondary electrons (a , b , c) in the small mass Δm of medium (absorbed dose) is quantitatively equal to the effect q_d which is produced in the medium along the tracks of the electrons arising from Δm (1 , 2 , 3) (collision kerma.)

values of K_{col} and D is governed by the decrease in the photon fluence (Fig. 3). If this decrease is exponential and if there is no large change in energy distribution of the photons, the ratio D/K_{col} remains constant. Such a situation can be considered as a *transient equilibrium* (also known as quasi equilibrium) (Tubiana and Dutreix, 1958; Batho, 1968; Attix, 1979). When referring to "electronic equilibrium conditions" (CPE), one may have to accept the conditions of transient equilibrium instead, although in the latter case $D > K_{col}$, while for CPE $D = K_{col}$.

In practical situations there are two kinds of regions where one of the conditions for transient electronic equilibrium is not fulfilled:

- (i) In the vicinity of any interface between two materials, the first condition regarding homogeneity is not fulfilled, in particular near air-material interfaces.
- (ii) The second condition regarding the uniformity of the radiation fluence is not fulfilled in regions near the edges of a finite beam, or in regions very close to the radiation source.

2. Air-Medium Interfaces

Fano (1954) demonstrated that the electron fluence is not modified by changes in density alone (ignoring the so-called polarization effect). If one assumes as a first approximation that air and the medium differ only by density and that their atomic compositions can be considered the same, the transient electronic equilibrium should not be modified when passing from air to medium, provided that the required conditions are fulfilled in both air and medium. This implies that the diameter of the beam is large enough to achieve transient electronic equilibrium in air.

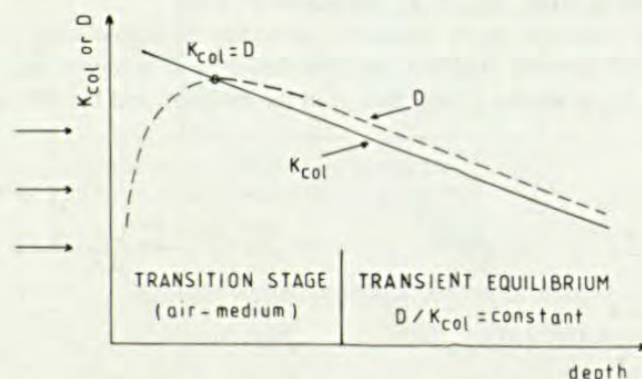


Fig. 3. Transition stage and transient equilibrium. At depths greater than the range R of the secondary electrons, the ratio D/K_{col} of the absorbed dose to medium D and the collision kerma K_{col} remain constant. This region corresponds to the transient electronic equilibrium. At depths smaller than R , the transition stage region, there is no simple relation between D and K_{col} except at the depth of the maximum dose where $D = K_{col}$.

Let us consider, for instance, a parallel beam of photons of maximum energy 10 MeV. A radius of about 5 g cm^{-2} would be necessary to achieve transient electronic equilibrium, that is to say about 50 m of air, which is far from being obtainable in practice. The electron fluence in air in a beam of radius r , $\Phi(r)$, at the interface is then much smaller than its equilibrium value $\Phi(R)$. Neglecting the photon beam attenuation and scattering in air and in medium, one can expect the following variation (Fig. 4): in air $\Phi(r)$ remains constant up to a distance R from the entrance surface of the medium; it increases up to the electronic equilibrium value $\Phi(R)$, which is reached at a depth R in medium, and remains constant up to a distance R from the exit medium surface. In the air behind the exit surface, the electron fluence decreases over a distance R . This last decrease is very rapid since the electrons originating from the medium are scattered out of the photon beam at a rather large angle.

The absorbed dose D_m in the medium can be calculated from the differential distribution of the electron fluence $d\Phi(E)/dE = \Phi_E$ by the relation

$$D_m = \int_0^{E_{max}} \left(\frac{S_{col}}{\rho} \right)_{E,m} \cdot \Phi_E \cdot dE \quad (13)$$

where $(S_{col}/\rho)_{E,m}$ is the collision mass stopping power in the medium m at the point of interest, for the secondary electrons of energy E .

Though the variation of the secondary electron spectrum with depth is not quite negligible, as a first approximation the variation of the absorbed dose in the medium is similar to the variation of the electron fluence Φ_E (see Section II,D).

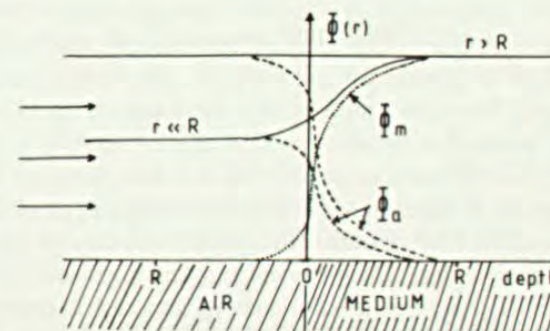


Fig. 4. Air-medium interface. The electron fluence at the interface in a beam of radius r , $\Phi(r)$, is the sum of the electrons generated in air Φ_a , decreasing gradually in the medium, and of the electrons generated in medium Φ_m , increasing gradually at depth in medium. At the interface Φ_m is equal to the electrons backscattered by the medium. In a very narrow beam Φ_a is negligible and $\Phi(r) = \Phi_m$; in a very broad beam ($r > R$) $\Phi_a = \Phi_m$ and $\Phi(r)$ is constant. In a usual beam ($r \ll R$), at the interface $\Phi(r)$ is higher than Φ_m at the interface but lower than the transient equilibrium value.

The electron fluence in air at the source distance under consideration and the absorbed dose at the medium surface depend on the following parameters:

- (i) the photon energy spectrum;
- (ii) the source distance: i.e., the air thickness passed through by the photon beam;
- (iii) the nature, thickness and position of the various components through which the photon beam is passing, including the source itself;
- (iv) the shape and nature of the collimating device in which electrons are set in motion at a given angle;
- (v) the distance from the collimating device;
- (vi) the field size.

Since the number of the parameters of influence is large, it is difficult to observe separately their role and to predict the variation of the absorbed dose in the transition region without accurate measurements. Many papers have been published showing the variation of the absorbed dose as a function of some of these parameters. Padikal and Deye (1978) and Biggs and Ling (1979) studied the depth variation of the maximum dose with field size in a high-energy x-ray beam; they clearly demonstrated that the contamination was due mainly to electrons and they examined the origin of these electrons. Figure 5 shows the variation of the absorbed dose in the buildup region for an 18-MV x-ray beam. The small depth of the maximum dose and the high skin dose measured at 4 m show the increase with source distance of the electron fluence in air as compared with the maximum electron fluence in medium.

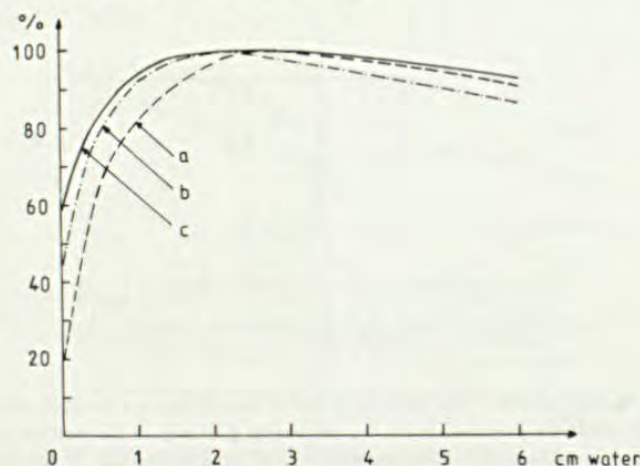


Fig. 5. Absorbed dose variation with depth in the transition region for 18-MV x rays. In usual geometrical conditions (curve a, $20 \times 20 \text{ cm}^2$, SSD 100 cm) the absorbed dose at the skin is rather low (about 20% of the maximum) while it increases with field size (curve b, $40 \times 40 \text{ cm}^2$, SSD 100 cm). In extreme conditions (curve c, field size $160 \times 160 \text{ cm}^2$, SSD 400 cm) the skin dose is as high as 60% of the maximum and the depth of the peak dose is decreased from 3 to 2 cm.

Both the number of electrons and their angular and energy distribution are modified with field size and source distance. Dosimetric parameters such as the mass stopping power are then not accurately known in the transition region because of the uncertainties in the energy spectrum of the secondary electrons. This means that the uncertainty in absorbed dose at depths smaller than the maximum range R of electrons is much larger than that under transient-equilibrium conditions (see Section II,D).

3. Lateral Electronic Equilibrium

The condition for transient electronic equilibrium concerning radiation-field uniformity implies that the distances between the point under consideration and the beam edges are larger than the maximum range R of electrons. In high-energy photon beams transient electronic equilibrium is theoretically assured on the beam axis only for broad beams, e.g., a beam radius larger than 10 cm in a 20-MV photon beam. However, an effective transient electronic equilibrium is observed in smaller beams for two main reasons:

- (i) A bremsstrahlung photon beam possesses a continuous spectrum of energies with a rather small proportion of high-energy photons, and the probability of production of electrons with an energy near the maximum photon energy is very low.
- (ii) The secondary electrons are not equally distributed in direction, and the effective range of electrons decreases continuously from the forward direction to the backward direction.

To quantify these observations, Dutreix *et al.* (1965) determined "isoinfluence curves" surrounding the volumes which contribute (through secondary electrons) 25%, 50%, 80%, 90%, or 100% of the absorbed dose at a point P . The isoinfluence curves (Fig. 6) show that the vicinity of point P has a much greater influence on the absorbed dose at P than the outer parts of the volume. Therefore the minimum volume to be considered is considerably smaller than the sphere of radius R , and it is possible to estimate the effective range R_e of secondary electrons in a given direction as the distance of the isoinfluence curve surrounding the volume contributing, for instance, 98% of the absorbed dose to the point under consideration. Such an estimation can be done either by Monte Carlo calculations or by using an experimental approach. Dutreix *et al.* (1965) approximated R_e by measuring buildup curves for beam diameters ranging from 3 to 38 mm and larger. The curves shown in Fig. 7 were measured by film dosimetry in a Lucite phantom at a large distance from the source. They are corrected for inverse square law and for the exponential attenuation of the photon beam. These curves show clearly the lack of lateral electronic equilibrium in narrow beams of high-energy photons. The effective range R_e is equal to about 2 cm in a 20-MV x-ray beam in a direction perpendicular to the beam axis. It means also that in any high-energy beam, there exists a region near the edge of

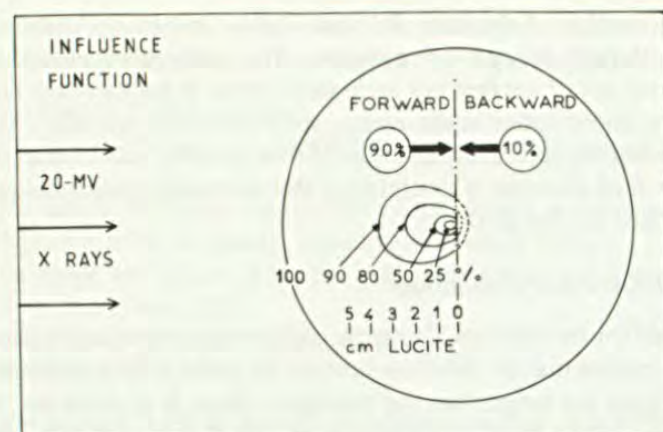


Fig. 6. The isofluence curves surround volumes which contribute to a given percentage of the total absorbed dose. The theoretical volume of tissues contributing to the absorbed dose is a sphere, the radius of which is equal to the maximum range of electrons (about 10 cm for 20-MV x rays), but the volume contributing 50% of the dose is as small as 1.2 cm for the forward electrons and a few millimeters for the backward electrons.

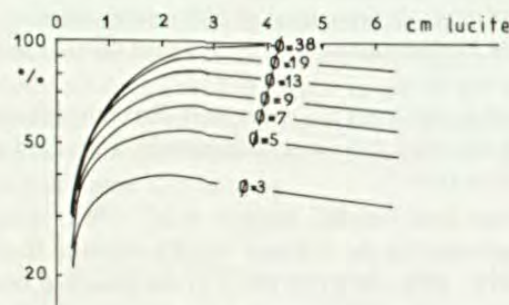


Fig. 7. Variation of the absorbed dose with depth in the transition region for narrow beams in a 20-MV x-ray beam. A diameter ϕ of 38 mm is necessary to ensure transient electronic equilibrium. For narrower beams the conditions for lateral electronic equilibrium are never fulfilled.

the beam, the width of which is equal to $2R_e$, in which the absorbed dose varies from its maximum value within the beam to a minimum outside the beam. This region cannot be reduced by decreasing the source size or improving the collimator design; it is only related to the primary photon spectrum and its width increases when the energy increases. Thus, although sometimes referred to as "penumbra," it is not a true penumbral region.

B. RADIATION QUALITY

Radiation quality must be specified to facilitate the use of published data tables such as absorption coefficients, stopping-power ratios, perturbation coeffi-

cients, etc., which depend strongly upon the radiation energy. The complete energy spectrum of the photons at the point of interest would be the most accurate quality description, but it cannot be achieved in general for accelerators.

It is then necessary to agree on an experimental method which is straightforward and unambiguous and leads to the determination of a simple parameter which can be used as an entry for published tables. Furthermore, such a parameter can serve as an index to compare physical data obtained in various photon fields and to report experimental or medical results.

1. Radioactive Sources

The atomic number and the mass number of the radionuclide used, such as cobalt-60 or cesium-137, specify the quality of a γ -ray beam since the energies of the different γ rays emitted are well known. However, radioactive sources used for medical and industrial purposes are not point sources and are contained in a shielded housing. Due to self-absorption in the source and to scattering by the source and parts of the shielded housing in the vicinity of the source, the photons emerging from the collimator present a wide spectrum, the mean energy of which is somewhat lower than the energy of the primary γ rays (Bruce and Johns, 1960; Derbowka and Cormack, 1965; Guiho *et al.*, 1973). For instance, for a cobalt-60 source with primary γ rays of 1.33 and 1.17 MeV, a mean energy of 0.9 MeV is generally accepted for the emerging beam. This energy varies slightly with the design of the source housing and with the type of collimator. Furthermore, because of the large amount of scattering in a low-Z (e.g., water) absorber, the mean energy decreases slightly when the beam penetrates it.

When using physical data calculated for monoenergetic radiations it is recommended to consider the values calculated for an energy of 0.9 MeV rather than for 1.25 MeV, which corresponds to the average of the cobalt-60 primary γ rays. When the physical data have been measured in a cobalt-60 photon beam, they can be used straightforwardly.

2. Accelerator-Produced X Rays

a. Energy Specification. In accelerators, bremsstrahlung x-ray beams are produced by accelerated electrons hitting a target. In linear accelerators, the manufacturer usually specifies the average energy and the energy spread of the electron beam. The bremsstrahlung x-ray beam presents a wide spectrum, the maximum energy of which is approximately equal to the average energy of the electron beam. Usually bremsstrahlung x-ray beams are specified by the value in megavolts (MV) of the average energy of the electron beam in mega-electronvolts (MeV). For instance, an x-ray beam produced by electrons accelerated to 20 MeV is specified as a 20-MV x-ray beam. However, the shape of the x-ray

spectrum depends upon the nature and the thickness of the target, and is strongly modified by the various filters or absorbers through which the beam passes (Podgorsak *et al.*, 1975; Brahme and Svensson, 1979).

Two different beams specified by the same maximum energy may produce quite different depth-dose distributions with differences in the depth of maximum dose as well as differences in the attenuation coefficients. This necessitates an energy specification more relevant for dosimetric considerations.

b. Attenuation in Water. For conventional, low-energy x rays, beam quality is specified by the half-value thickness measured in a reference medium in narrow-beam conditions. The reference medium is aluminum for the lowest energies and copper for medium energies. By extension, lead was proposed to specify the quality of megavolt x- and γ -ray beams. The determination of the half-value thickness in a reference medium in narrow-beam conditions for a high-energy x-ray beam is not convenient for two main reasons:

(i) Narrow-beam conditions imply that the contribution of scattered photons to the dose can be neglected. However, lateral equilibrium is needed to measure photon beam attenuation and leads necessarily to scatter that cannot be ignored. The determination of the attenuation coefficient necessitates extrapolation of the measured doses to zero diameter, thus increasing the uncertainty.

(ii) In high-Z materials, such as lead, the attenuation coefficient shows a minimum for energies around 2 to 3 MeV because of pair production for the higher energies (Fig. 8). The determination of the half-value thickness, that is to say of the corresponding attenuation coefficient, is ambiguous because it leads to two different effective energies. For instance, two different beams with mean energies equal to 10 and 1.7 MeV have the same half-value thickness of 1.26 cm of lead.

As shown in Fig. 8 the attenuation coefficient in water decreases gradually as energy increases, and reaches a minimum for an energy of about 50 MeV. Thus water is the least ambiguous reference material for quality specification. As narrow-beam conditions are not easily fulfilled, it has been suggested that broad-beam depth doses be used to specify beam quality (Dutreix, 1976b; NACP, 1980). Dutreix and the Nordic Association of Clinical Physics have suggested the measurement of an index related to the photon attenuation in the exponential part of the depth-dose curve under reference conditions. The reference conditions and the index chosen are summarized in Table V. The conditions proposed by the NACP are the simplest ones and the measurement of the index I_{10}/I_{20} is straightforward; however, this index has no direct meaning. The conditions proposed by Dutreix require the measurement of the full depth-dose curve at an infinite source-to-phantom-surface distance (SSD); but the half-attenuation thickness (HAT) represents the thickness of water required to attenuate the photon beam by a factor of two in infinite SSD conditions.

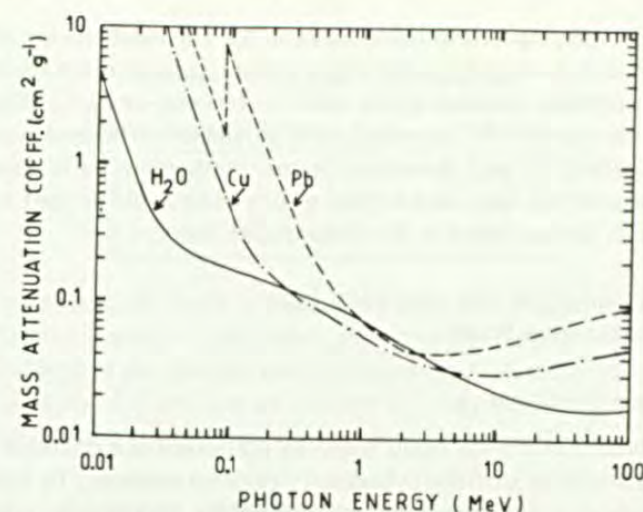


Fig. 8. Variation of the mass attenuation coefficient versus photon energy for three materials: lead, copper and water. The minimum value is reached at 3 and 8 MeV for lead and copper, respectively, and at 50 MeV for water.

TABLE V

REFERENCE CONDITIONS AND QUALITY INDEX FOR X-RAY BEAMS

Reference conditions	Dutreix (1976b)	NACP (1980)	AAPM (1983), CFMRI (1985)
	SSD	SSD	SSD
Field size	∞	10 × 10 cm ²	10 × 10 cm ²
Index	Half-attenuation thickness (HAT) ^b	I_{10}/I_{20} ^a	I_{20}/I_{10} ^a

^a I_{10} and I_{20} are, respectively, the relative ionizations at depths of 10 and 20 cm in water.

^bHAT can be measured, for instance, as $(z_{40} - z_{80})$, where z_{40} and z_{80} are, respectively, the depths corresponding to 40 and 80% relative ionizations.

If HAT is expressed in centimeters of water, the two quantities are related by the following equation:

$$I_{20} = I_{10} \cdot ((100 + 10)/(100 + 20))^2 \cdot e^{-(10 \cdot \ln 2 / \text{HAT})}$$

$$I_{10}/I_{20} = 1.19 \cdot e^{6.93 / \text{HAT}} \quad (14)$$

Dutreix and Broerse (1982), AAPM (1983), Cunningham and Schulz (1984), and CFMRI (1985) have suggested the index I_{20}/I_{10} measured in parallel beam conditions (SSD = ∞). This index brings together three advantages: it can be determined simply by measurements at two depths, it does not depend on the distance of measurement and it represents beam attenuation by 10 cm of water.

The NACP has proposed to relate I_{10}/I_{20} to the maximum energy of a theoretical x-ray beam produced in a thin target with no filtration in the beam. Figure 9 shows the relations between I_{10}/I_{20} (SSD = 100 cm) or I_{20}/I_{10} (SSD = ∞) and x-ray beam energy. The various physical parameters to be used in dosimetry could be calculated for such theoretical beams. Furthermore, it is assumed that the data measured in a beam with a given quality index could be used with minimum errors in another beam of the same quality index.

C. MEASUREMENT OF THE ABSORBED DOSE AT A REFERENCE POINT

1. Reference Conditions

The calibration of a γ -ray beam is usually performed at a reference point on the beam axis with an air-filled cylindrical ionization chamber. To increase the accuracy of the dose determination (see Section II,B), the reference point should be chosen in a region of transient electronic equilibrium in reference geometry conditions. Table VI summarizes suitable reference conditions. The ICRU in Reports 23 (ICRU, 1973) and 24 (ICRU, 1976) has proposed a depth of 7 cm for x rays between 11 and 25 MV; however, transient equilibrium is not always achieved at 7 cm depth in an x-ray beam of more than 20 MV, and 10 cm seems to be a more appropriate depth. A field size of 10×10 cm is large enough to ensure lateral electronic equilibrium even for the highest energies. In radiation therapy the reference distance to the phantom surface (SSD) is in general chosen equal to the usual treatment distance: 80 cm is often chosen for cobalt-60 γ rays and 100 cm for high-energy x rays.

The axis of the cylindrical cavity of the ionization chamber is placed perpendicular to the beam axis at the reference depth in a water phantom. When a plas-

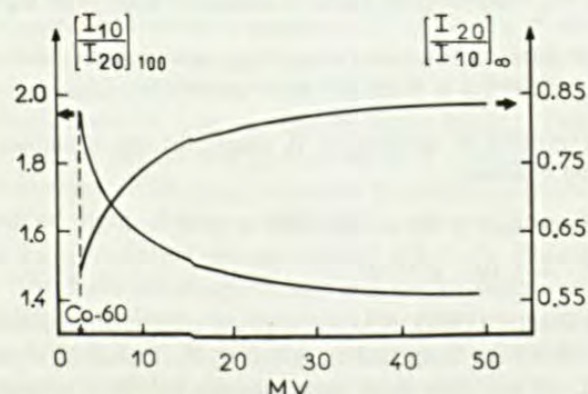


Fig. 9. Variation of the two indexes $(I_{10}/I_{20})_{100}$ and $(I_{20}/I_{10})_{\infty}$ versus x-ray beam energy in undegraded beams. [After NACP (1980) and Cunningham and Schulz (1984).]

TABLE VI

REFERENCE CONDITIONS IN PHOTON BEAMS		
Beam quality	Reference depth	Field size
Cobalt-60	5 cm	10×10 cm ²
1 to 9 MV	5 cm	10×10 cm ²
10 to 50 MV	10 cm	10×10 cm ²

tic phantom is used, the depth in centimeters must be corrected to take account of the difference in atomic composition and density of the medium (see Section II,D,2). The wall of the chamber used to measure the reference dose is assumed to be thin (less than 0.5 mm) and the chamber is to be placed in the water without buildup cap, since available caps are not always water equivalent.

2. Correction Factors to Apply to the Instrument Readings.

The absorbed dose to water at the reference point is determined in two steps: First, the determination of the absorbed dose to the air of the ionization chamber cavity, and second, the determination of the absorbed dose to water in the absence of the cavity.

a. *Step 1.* The mean absorbed dose to the air of the cavity is determined by the relation (6):

$$\overline{D}_{\text{air}, u} = M_u \cdot N_D$$

where u stands for the user's beam quality. This relation is valid for any energy since N_D is specific for the ionization chamber and independent of the beam quality as noted before. Therefore, the symbol N_D will replace $N_{D, c}$.

The quantity M_u is the instrument reading corrected for temperature and pressure ($k_{t, p}$) and by correction factors k_e related to electrical parameters: recombination losses, polarity effect, etc.

$$M_u = M'_u \cdot k_{t, p} \cdot k_e \quad (15)$$

where M'_u is the uncorrected reading.

b. *Step 2.* The absorbed dose to water at the reference depth can be calculated from the Bragg-Gray equation

$$D_{w, u} = \overline{D}_{\text{air}, u} \cdot \left[\frac{\overline{S}}{\rho} \right]_{\text{air}, u}^w \cdot p_u \quad (16)$$

$$= M_u \cdot N_D \cdot \left[\frac{\overline{S}}{\rho} \right]_{\text{air}, u}^w \cdot p_u \quad (17)$$

where $[\overline{S}/\rho]_{\text{air}, u}^w$ is the ratio of the mean restricted mass collision stopping

powers in water and in air for the user's beam quality u and p_u corrects for the total perturbation at point P brought about by the low-density air cavity as well as the cavity wall; p_u involves several factors which are discussed below.

(a) Relation (16) assumes that the chamber wall is water equivalent or that it contributes very little to the electron fluence in the cavity. The variation in the electron fluence, resulting from the lack of water equivalence of the chamber wall, cannot be calculated accurately and must be estimated experimentally. It can be determined by comparison of readings from different chambers with the same cavity size and various wall compositions. Johansson *et al.* (1978) found the same perturbation factors for chambers with 0.5-mm graphite wall and with a Lucite wall with a 50- μ m inner layer of graphite, whatever the beam energy. These results agree with the experiments by Almond and Svensson (1977) and show the dominating influence of the thin layer of matter immediately surrounding the air cavity. This finding is explained by the large component of low-energy electrons in the secondary electron spectrum (Todo *et al.*, 1982) and was demonstrated by Gray (1937), who showed that for radium γ rays generated in graphite, one-half to two-thirds of the ionization produced in a cavity is due to electrons generated within 0.2 mm of the cavity boundary. Therefore, the perturbation factor p_u depends upon the nature of the chamber wall. The ratio $p_{u,c}/p_{u,w}$ of the perturbation factors for two ion chambers of the same size but with, respectively, a graphite wall ($p_{u,c}$) and a water-equivalent wall ($p_{u,w}$) varies from 0.98 for cobalt-60 γ rays to 0.99 for 10-MV x rays and to 0.995 for x rays between 25 and 50 MV.

(b) The replacement of a small volume of water by the same volume of air leads to two perturbations. First, a perturbation of the secondary electrons has to be considered because of the differences in stopping and scattering properties between air and the condensed medium (Harder, 1968). Second, the secondary electron fluence within the cavity differs from that at the point of interest in the medium, and a correction known as the *displacement correction factor* or as the *radial displacement of the point of measurement* has to be applied.

These two perturbations can be considered separately in experiments, although they are not quite independent. In the plateau region at the peak depth of the ionization curve, a small "displacement" of the point of interest does not affect the measured ionization. Thus, the displacement correction factor may be considered as equal to unity. Therefore the first type perturbation factor can be measured at the peak depth. Comparisons of chambers with identical walls and different cavity sizes, for various beam energies have been carried out at the peak depth (Johansson *et al.*, 1978). They have shown that the perturbation of the electron fluence by the air cavity is quite negligible in photon beams. This result is expected for electronic equilibrium conditions and it is assumed that this error can be neglected when performing measurements at the peak depth. The exact meaning and appropriate methods for calculation of the displacement fac-

tor are still very controversial and can be considered from two different points of view.

In low-energy photon beams, the average dose throughout the volume of the dosimeter is equal to the dose at its center, provided the beam attenuation across the sensitive volume can be considered as linear (Burlin, 1961, 1968). However, in high-energy photon beams, it has been demonstrated that a displacement correction of some type has to be applied, which depends upon the size of the cavity and the beam energy. This correction is evident from experiments, but the interpretation is certainly not simple and is still very controversial as can be seen in the literature (Tubiana and Dutreix, 1958; Barnard *et al.*, 1964; Johansson *et al.*, 1978; Almond *et al.*, 1978; Cunningham and Sontag, 1980; Loevinger, 1981; Zoetelief *et al.*, 1981). We shall not consider here the theoretical calculations, which will differ until agreement has been reached on the interpretation of the experimental findings. The following treatments are based simply on experimental observations.

Figure 10 shows depth-ionization curves measured, respectively, with a cylindrical chamber (6 mm in diameter) and a thin parallel-plate chamber (1.5 mm thick). The readings were referred to the depth of the inner surface of the front wall for the flat chamber and to the depth of the geometrical center for the cylindrical chamber.

For points at depths where transient electronic equilibrium is achieved, a displacement correction factor p_d may be applied as a multiplicative constant to compensate for the decrease in photon attenuation and the alteration of scattering

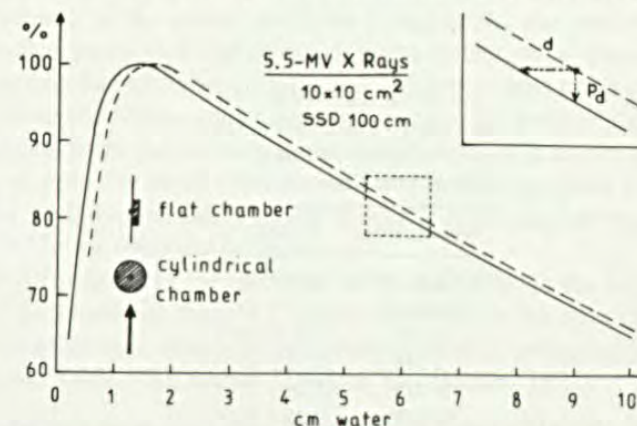


Fig. 10. Depth-ionization curves. The ionization is measured either with a cylindrical chamber (dashed line) or with a parallel-plate chamber (solid line). The depths are referred to the geometrical center of the cylindrical chamber and to the inner surface of the anterior plate of the parallel-plate chamber. The enlarged drawing at the right top shows that coincidence between the two curves can be achieved in the transient equilibrium region, either by multiplying the readings of the cylindrical chamber by a constant value p_d or by translating the dashed line upstream by a constant shift d . The coincidence in the transition region can be achieved only by translation.

resulting from the "replacement" of phantom material by the air cavity. On the other hand, to bring the two curves into coincidence at all depths requires defining an effective point of measurement for the cylindrical chamber at some fraction ($d = \alpha \cdot r$) of the radius r upstream of the geometrical center.

Direct measurements can be carried out to determine the displacement correction factor which should be applied to evaluate the dose in the absence of the ionization chamber. Measurements can be performed using chambers with medium-equivalent walls of various diameters, by comparison with a flat chamber, or by referring to a dosimeter with no air cavity, such as a calorimeter (Hofmeester, 1981) or a ferrous sulfate Fricke dosimeter.

It has been suggested to express the displacement factor p_d as a function of the cavity radius r

$$p_d = 1 - k \cdot r \quad (18)$$

Table VII shows values of k estimated for various energies by Johansson *et al.* (1978) and Zoetelief *et al.* (1981).

The radial displacement can be estimated in any part of the depth-ionization curve, either in the buildup region (Dutreix, 1976b) or in the exponential part of the curve (Johansson *et al.*, 1978; Zoetelief *et al.*, 1981). Theoretical calculations have been performed by Dutreix (1980) and Abou Mandour and Harder (1977). The gas-filled volume is considered the detector, sensing the fluence of charged particles at the point where they enter the cavity. The calculations have been made for idealized conditions in three cases which are summarized in Table VIII.

TABLE VII

VALUES OF THE FACTOR k TO BE USED TO EVALUATE THE DISPLACEMENT FACTOR p_d FOLLOWING EQUATION (18)^a

Beam quality	k^b (mm ⁻¹)
Cesium-137	$0.22 \cdot 10^{-2}$
Cobalt-60	$0.37 \cdot 10^{-2}$
5 MV	$0.4 \cdot 10^{-2}$
8 MV	$0.31 \cdot 10^{-2}$
16 MV	$0.24 \cdot 10^{-2}$
42 MV	$0.25 \cdot 10^{-2}$

^a $p_d = 1 - kr$, where r is the inner radius of the cylindrical ionization chamber.

^bThe first two values are from Zoetelief *et al.* (1981) and the others are from Johansson *et al.* (1978).

TABLE VIII

RADIAL DISPLACEMENT OF THE EFFECTIVE POINT OF MEASUREMENT UPSTREAM FROM THE CENTRAL PLANE, CENTRAL AXIS, OR CENTER OF THE CAVITY

Secondary electron distribution	Plane-parallel cavity, thickness h	Cylindrical cavity, radius r	Spherical cavity, radius r
Isotropic direction	0	0	0
Forward direction	$h/2$	$8r/3\pi$	$3r/4$
Forward contribution f	$(h/2)(f - f')$	$(8r/3\pi)(f - f')$	$(3r/4)(f - f')$
Backward contribution f'			

(i) The secondary electrons are assumed to be generated isotropically. For symmetry considerations the number of electrons entering the cavity through posterior and anterior parts of the wall are assumed equal and the effective point of measurement lies on the geometrical center of the cavity. This case corresponds to low-energy photons and the experimental data agree with the theory (Zoetelief *et al.*, 1981).

(ii) The secondary electrons are generated in the forward direction parallel to the beam axis. A simple theoretical calculation can be performed assuming that the volume of the central electrode is negligible, the number of ions is proportional to the chord length of its path in the cylindrical or spherical volume, and the electron fluence varies linearly over a distance equal to the cavity radius.

Under these conditions, calculations indicate that the average mass ionization in a cylindrical probe corresponds to the mass ionization in an elementary probe placed at an effective point of measurement displaced upstream by a distance $d = 8r/3\pi \approx 0.85r$ from the cylinder axis. The radial displacement d is equal to $\frac{3}{4}r = 0.75r$ for spherical geometry.

(iii) If a fraction f of the electrons enters the cavity in the beam direction whereas a backscattered fraction f' enters the cavity in the opposite direction, the radial displacement calculated previously should be multiplied by the factor $(f - f')$ (see Table VIII) (Abou Mandour and Harder, 1977).

Radial displacements have been measured by Dutreix (1976b), Johansson *et al.* (1978), and Zoetelief *et al.* (1981). They agree with theory, assuming a forward direction of the electrons for very high energy photons ($d = 0.85r$). The radial displacement is smaller for lower energies, implying an increase of the f' fraction (Table IX).

The displacement correction factor (p_d) may be used only for deep points (i.e., where transient charged particle equilibrium, TCPE, exists) and assumes

TABLE IX

RADIAL DISPLACEMENT d OF THE EFFECTIVE POINT OF MEASUREMENT AS A FUNCTION OF BEAM QUALITY FOR CYLINDRICAL IONIZATION CHAMBERS PERPENDICULARLY ORIENTED IN MONODIRECTIONAL BEAMS PASSING THROUGH UNIT DENSITY (e.g., WATER) PHANTOMS^a

Beam quality	d	f	f'
Cesium-137	0.23r	0.64	0.36
Cobalt-60	0.58r	0.84	0.16
6 MV	0.7r	0.91	0.09
25 MV	0.85r	1	0

^a f and f' are defined in Table VIII.

that the apparent attenuation of the beam is exponential. It is valid not only for a monodirectional photon beam but also for two opposed parallel beams or for rotational beams. It may be used neither in the buildup region nor in the plateau region. It is recommended by the Nordic Association of Clinical Physics (1980) and the American Association of Physicists in Medicine (1983) protocols to apply the displacement correction factor to the measured ionization when measuring the dose at the reference point in the TCPE region.

The radial displacement is valid only for a monodirectional photon beam. It is valid neither for two opposed parallel beams nor for rotational beams. It may be used in the buildup and the plateau region. It is, then, the only method available to measure the dose distribution along the beam axis (see Section II,D). The Comité Français "Mesure des Rayonnements Ionisants" (1985) recommends that the ionization chamber position be corrected by the radial displacement for the measurements at both the reference point and at any point in a phantom.

(c) It has been suggested (Nordic Association of Clinical Physics, 1980; American Association of Physicists in Medicine, 1983) to include the displacement correction factor into the total perturbation factor p_u . Values for p_u including p_d are given in Table X, together with the mean mass stopping-power ratios from Cunningham and Schulz (1984). $p_{u,c}$ refers to a graphite wall and $p_{u,w}$ to a water-equivalent wall for cavity diameters equal to 5 mm. For other cavity sizes p_u values should be adjusted following Table VII. For instance, when an ionization chamber 8 mm in diameter ($r' = 4$ mm) is used instead of the 5-mm one ($r = 2.5$ mm) the p_u values should be multiplied by

$$(1 - kr')/(1 - kr) = (1 - 4k)/(1 - 2.5k)$$

For a 5-MV x-ray beam, where $k = 0.4 \times 10^{-2}$, this ratio equals $0.984/0.99 = 0.99$; p_u values should be multiplied by 0.99.

The absorbed dose determined with an ionization chamber, corrected follow-

TABLE X

VALUES OF $[\bar{S}/\rho]_{air,u}^w$ AND OF THE PERTURBATION CORRECTION FACTORS p_u FOR CYLINDRICAL IONIZATION CHAMBERS OF GRAPHITE AND WATER EQUIVALENT MATERIAL AS A FUNCTION OF THE BEAM QUALITY AS EXPRESSED BY THE RATIO $[I_{20}/I_{10}]_{air}$

$(I_{20}/I_{10})_{air}$	Approximate quality	$[\bar{S}/\rho]_{air,u}^w$	$p_{u,c}$	$p_{u,w}$
0.58	Cobalt-60	1.134	0.968	0.990
0.64	4 MV	1.131	0.972	0.990
0.69	6 MV	1.127	0.979	0.992
0.74	10 MV	1.117	0.983	0.993
0.77	15 MV	1.106	0.984	0.994
0.80	20 MV	1.096	0.984	0.994
0.81	25 MV	1.093	0.990	0.994
0.82	30 MV	1.088	0.990	0.994
0.83	40 MV	1.077	0.990	0.994
0.84	50 MV	1.069	0.990	0.994

ing the NACP (1980) protocol, has been compared with the absorbed dose determined either with a calorimeter (Hofmeester, 1981) or with a ferrous sulfate dosimeter (Johansson and Svensson 1982; Mattsson *et al.*, 1982). Hofmeester compared the absorbed dose to water determined with a graphite calorimeter (D_c) and an NPL secondary standard exposure meter (D_i) in a water phantom for 13 linear accelerators in 11 different hospitals. He found that the average ratio D_i/D_c was equal to 1.002 ± 0.009 . Johansson compared the absorbed dose to water determined with a Farmer ionization chamber (D_i) and a Fricke dosimeter (D_f) in a water phantom for 33 cobalt-60 γ -ray beams and 55 accelerators. He found that the average ratio D_i/D_f was equal to 0.998 ± 0.006 .

Both of these workers have clearly demonstrated the excellent agreement which can be achieved between ionometric measurements and calorimetric or ferrous sulfate measurements. They have fully justified the method of dose determination which has been explained previously.

The various correction factors could be included in a " C_λ " or " F " factor. However, different C_λ must then be published for different ionization chambers depending upon the atomic composition of the chamber wall and the size of the cavity. It seems safer to consider separately the different parameters to be taken into account than to propose to the user a "magic" number which can easily be misunderstood or misused.

D. MEASUREMENT OF THE ABSORBED DOSE AT ANY POINT IN A PHANTOM

1. Absorbed Dose Distribution Measurements

The ultimate aim of radiation dosimetry is to determine the three-dimensional distribution of absorbed dose in the absorber. For water-equivalent absorbers,

e.g., patients or exposed individuals, an essential step is therefore the determination of the complete dose distribution in a water phantom for a few typical beam sizes and for each beam energy used.

The shape of the absorbed dose distribution in a high-energy photon beam is not very dependent on the design of the accelerator for a given quality index. However, the depth of the maximum dose (in particular for large field sizes), the dose in the buildup region, the width of the so-called penumbra region, and the flatness of the isodose surface vary from one accelerator to another. Dose distributions should therefore be measured carefully for each x-ray unit and each set of geometrical parameters.

The parameters of influence for cobalt-60 γ -ray beams are the source diameter, the collimator design, the source-absorber distance, and the collimator-absorber distance. For high-energy x-ray beams they are the energy, the target and flattening filter design, the collimator design, the source-absorber distance, and the collimator-absorber distance. The presence of filters through which the beam passes, such as trays in radiotherapy, must be carefully considered because they both absorb electrons from upstream sources and generate new electrons. The depth doses are usually normalized at the peak depth and only relative values of the doses are needed.

The corrections to apply to the dosimeter readings to evaluate the absorbed doses are N_D , $[\bar{S}/\rho]_{\text{air},u}^w$, and p_u , where N_D is specific for the ionization chamber used and independent of the beam energy, while $[\bar{S}/\rho]_{\text{air},u}^w$ and p_u are energy dependent. However, the secondary electron spectrum does not vary significantly with depth and $[\bar{S}/\rho]_{\text{air},u}^w$ and p_u can be considered as independent of the depth of measurement for depths larger than the reference depth.

Nahum (1982) calculated by Monte Carlo simulation the variation of $[\bar{S}/\rho]_{\text{air},u}^w$ in the buildup region due to the variation of the secondary electron spectrum. He found a 1.5% decrease for a 27-MV x-ray beam and a 3% decrease for a 31-MV x-ray beam, at the phantom surface compared with peak depth. The variation observed for depths larger than the peak depth is quite negligible (less than 0.5%).

As shown in Section II,C,2,(b), the displacement correction factor cannot be used at depths where TCPE is not present and it is recommended to use the radial displacement correction for dose-distribution measurement in general. For this reason the readings of the dosimeter at any depth have to be referred to the effective point of measurement, estimated from Table IX.

The absorbed dose to water $D_w(P)$ at a point P is then estimated from the absorbed dose D_R measured in the reference conditions by the relation

$$D_w(P) = D_R \cdot (PDD(P)/PDD(R)) \cdot (M_R^c/M_R) \quad (19)$$

where $PDD(P)$ and $PDD(R)$ are the percentage depth doses at point P and at the reference point R ; M_R^c/M_R is the ratio of the meter readings, with the ion-

ization chamber axis placed at the reference depth on the beam axis, for the geometrical conditions under consideration (M_R^c) and the reference geometrical conditions (M_R) (Table XI). This ratio should be determined for each set of irradiation conditions used.

The rationale in using the radial displacement is clear for measurements performed along the axis of a monodirectional beam or at points where lateral electronic equilibrium is achieved. In the penumbra region and outside the geometrical beam it is obvious that the secondary electrons are not necessarily entering the ion chamber in the forward direction; however, in these regions the dose gradient in the beam direction is very low and the position of the effective point of measurement is not at all critical.

2. Phantom Materials Other Than Water

For high-energy photon beams and in phantoms of low-atomic-number (Z) materials, interactions of photons with atoms are predominantly Compton interactions. The number of Compton interactions per unit volume of medium is proportional to the number of electrons per unit volume N :

$$\frac{N_m}{N_w} = \frac{\rho_m}{\rho_w} \cdot \frac{(\bar{Z}/A)_m}{(\bar{Z}/A)_w} = \frac{\rho_m}{\rho_w} \cdot \left[\frac{\bar{Z}}{A} \right]_m \quad (20)$$

where m refers to medium and w to water; ρ is the density.

$$\frac{\bar{Z}}{A} = \frac{\sum_i p_i Z_i}{\sum_i p_i A_i} \quad (21)$$

where Z_i is the atomic number and A_i is the mass number of element i in the composite medium m ; p_i is the mass proportion of element i in medium m .

TABLE XI

MEANING AND MEASUREMENT CONDITIONS OF THE DIFFERENT TERMS OF EQUATION (19)

Term	Quantity	Measuring conditions
D_R	Absorbed dose to water	Reference point, reference field size, reference SSD
M_R	Meter reading corrected for $k_{t,p}$ and k_e	Reference point, reference field size, reference SSD
M_R^c	Meter reading corrected for $k_{t,p}$ and k_e	Reference point, field size of interest, SSD of interest
$PDD(P)$	Percentage depth dose	Point of interest, field size of interest, SSD of interest
$PDD(R)$	Percentage depth dose	Reference point, field size of interest, SSD of interest

The depth z_m of medium m is equivalent to the depth z_w of water when

$$z_w \cdot N_w = z_m \cdot N_m$$

$$z_w = z_m \cdot (N_m/N_w) = z_m \cdot (\rho_m/\rho_w) \cdot [\bar{Z}/A]_w^m$$

Table XII gives values of $[\bar{Z}/A]_m$, $[\bar{Z}/A]_w^m$, and ρ_m for the most usual water-equivalent media.

III. Dosimetry for Electron Beams

A. ELECTRON BEAM ENERGIES

1. Introduction

The variation of physical parameters with depth in an absorber is very different in an electron beam from that in a photon beam. In a photon beam (Section II,A), the photon fluence decreases continuously with depth by attenuation processes: for depths where (TCPE) is achieved, the electron fluence decreases at the same rate as the photon fluence, but the energy spectrum of the secondary electrons does not vary significantly.

The situation is radically different in electron beams. Along the center line of a wide beam the total electron fluence increases first slowly with depth for two reasons: (1) the increased obliquity of electron tracks with depth and (2) the increased number of secondary electrons set in motion by the primaries (Kesaris, 1966). At depths greater than half the maximum range of the primary electrons, the loss of electrons is predominant and the total fluence decreases rapidly to zero. Furthermore, because of the continuous slowing down of primary electrons, the maximum energy and the mean energy of the electron spectrum decrease continuously from a maximum at the phantom surface to zero at the depth of the maximum range.

A knowledge of the electron energy spectrum is essential for absorbed dose determination, since the fundamental physical parameters, such as the stopping power of a material, are functions of electron energy.

TABLE XII

CHARACTERISTICS OF WATER SUBSTITUTES

Medium	$[\bar{Z}/A]_m$	$[\bar{Z}/A]_w^m$	ρ_m
Water	0.555	1.0	1.0
Polystyrene	0.538	0.969	0.98-1.06
Lucite	0.540	0.973	1.07-1.18
A 150	0.548	0.987	~1.12
Polyethylene	0.572	1.031	0.94-1.0

2. Description of the Energy Spectrum of the Electron Beam

The energy spectrum of the electron beam depends upon the intrinsic energy spread of the accelerator beam, but the beam is also strongly modified by scattering and energy loss in the various materials through which the beam passes (de Almeida and Almond, 1974; Brahme and Svensson, 1979). The differential distribution in energy of an electron beam after passing through a slab of material has the general shape presented in Fig. 11 (Harder, 1966). The spectrum is terminated at a given maximum energy $E_{\max}$, E_p is the modal energy, that is to say the most probable energy, corresponding to the peak of the energy distribution, and $\bar{E}$ is the mean energy of the particle fluence.

The beam leaving the accelerating tube is modified in cross section and divergence by the beam optical components. In betatrons the electron beam passes through the fringing field of the magnet, while in other accelerators it passes through the bending magnets. Furthermore, the angular and energy spread of the beam is sometimes limited by slits or apertures. The modified beam passes through the vacuum tube window (in which some scattering occurs) before final modification and collimation. This initial electron beam (referred to by index i) is different from the intrinsic electron beam at the end of the accelerating tube.

To achieve convenient dose distributions in planes perpendicular to the beam axis for clinical and other applications, the initial electron beam must be broadened and flattened. The most commonly used methods are the use of a thick, high-Z material scatterer, the use of multiple scatterers (Björngard *et al.*, 1976; Brahme and Svensson, 1976; Briot, 1982), or the use of electromagnetic beam scanning quadrupoles (Aucouturier *et al.*, 1970). These last two methods over-

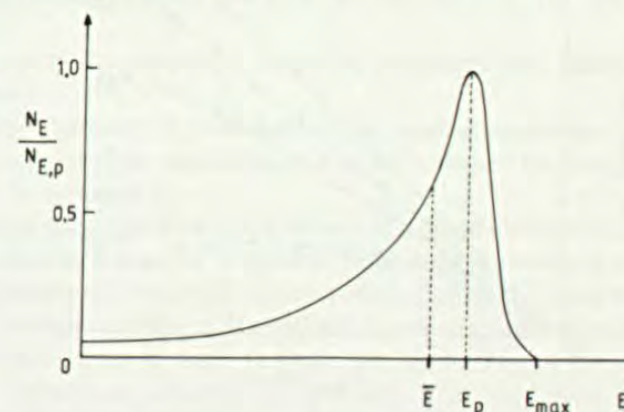


Fig. 11. Typical distribution of electrons in energy after passing through a slab of matter. The ordinates give the differential distribution in energy N_E normalized to the maximum value $N_{E,p}$, where $\bar{E}$, E_p , and $E_{\max}$ are, respectively, the mean energy, the modal energy, and the maximum energy.

come the excessive energy loss, energy spread, and x-ray contamination resulting from a single high-Z scatterer.

The use of multiple scatterers instead of a single foil allows a decrease in the total thickness of the scatterers to be used to achieve a given dose uniformity index. Figure 12 compares two electron beams ($20 \times 20 \text{ cm}^2$) with no significant difference in flatness. One is obtained with an electromagnetic beam scanning device (circles) and the other one with multiple scatterers (crosses). The figure shows that for the same modal energy at the phantom surface (evaluated from the practical range) the various range parameters are not significantly different, the contribution of low-energy electrons was not increased significantly.

After being broadened and flattened by the scatterers, the electrons usually pass through a monitor ionization chamber and through about 1 m of air before reaching the phantom surface. Furthermore, to be used for clinical and some other purposes the beam must be collimated. Some of the electrons hitting the collimator walls are scattered back into the main beam. The flatness of the beam and the shape of the depth-dose curve may vary considerably with field size and collimator distance for some collimator designs, because the collimator acts as a secondary electron source distributed along the collimator walls. Some of these collimators are designed as plate diaphragms and allow continuously adjustable field sizes (Briot *et al.*, 1973; Brahme and Svensson, 1979).

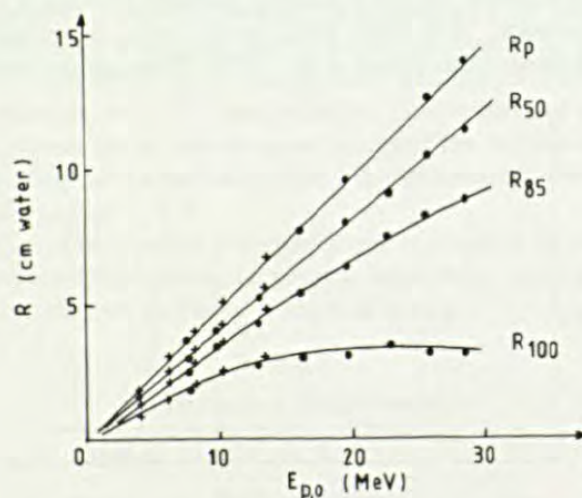


Fig. 12. Variation of range parameters with the modal energy at the phantom surface $E_{p,0}$. Crosses and circles correspond to a $20 \times 20 \text{ cm}^2$ field size in two accelerators with similar designs in accelerating tube, bending magnets, and slits, but the circles refer to an electromagnetic scanning device and the crosses to multiple scatterers (Briot, 1982). There is no significant difference in the dose profiles of the two types of beam.

The electron spectrum at the phantom surface (referred to by index 0) depends strongly upon the materials through which the initial electron beam passes, as well as upon the collimator design. However, the modal energy $E_{p,0}$ of the spectrum may be approached by the equation

$$E_{p,0} = E_{p,i} - \sum_n s_{\text{coll},n} \cdot \Delta x_n \quad (22)$$

where $E_{p,i}$ and $E_{p,0}$ are, respectively, the modal energies of the initial beam and of the beam at the phantom surface, and $s_{\text{coll},n}$ is the linear collision stopping power for the electrons of energy $E_{p,i}$ in the material n of thickness Δx_n . [In Eq. (22), the modal energy is valid only as a first approximation, and should be checked by more direct methods.]

Methods to determine the absolute energy are based on the use of either the thresholds of nuclear reactions or the Cerenkov thresholds. They are rather laborious and require sophisticated equipment. There is, however, a much simpler method which uses empirical relations between energies and range parameters.

3. General Description of an Electron Beam Depth-Dose Curve

Figure 13 shows a depth-dose curve measured in a $10 \times 10 \text{ cm}^2$ field (SSD ∞ , 19 MeV). The initial buildup is followed by a rather flat plateau where the dose is a maximum. The depth of the flat dose maximum R_{100} depends strongly upon the energy distribution of the initial beam and upon the contamination of the beam by low-energy electrons scattered by the collimator walls. In some very contaminated beams the dose is maximum at the phantom surface ($R_{100} = 0$).

The quantity R_{50} is the depth where the percentage dose is 50% of the maximum dose.

The practical range R_p is defined as the intersection between the tangent to the steepest point of the descending part of the curve and the extrapolated bremsstrahlung background D_x .

The range parameters are characteristic of a broad electron beam. They must be determined in a beam of a diameter large enough to ensure an equilibrium of the secondary and scattered electrons on the beam axis. Since the mean energy of the secondary electrons is small and the mean scattering angle of the electrons at small depths in water is small, the equilibrium of the secondary and scattered electrons is achieved for relatively small field sizes. As a rule of thumb, the field diameter ensuring equilibrium of the secondary electrons may be chosen as equal to R_p . The range parameters should be determined on a curve corrected for divergence. Figure 14 shows the measured variation of the range parameters with the modal energy for usual SSD (90 cm) and for an infinite SSD.

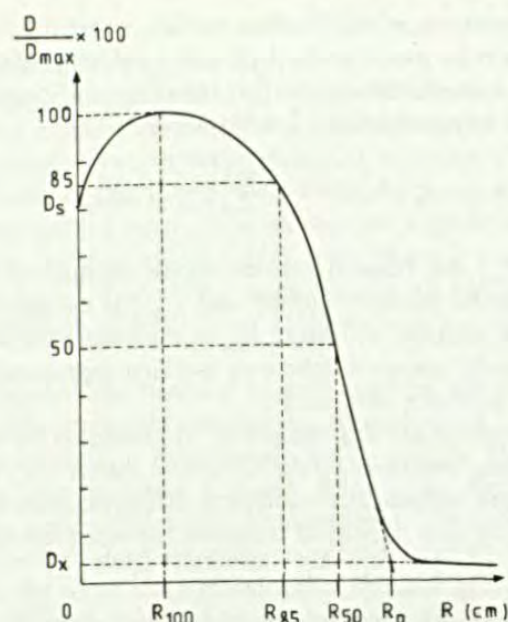


Fig. 13. Percentage depth-dose curve in an electron beam of 19 MeV (SSD = ∞ , field size $10 \times 10 \text{ cm}^2$). The main range parameters as defined in the text are shown on the curve.

The value R_{85} is not useful for dosimetry purpose but is a very important range parameter in clinical beams. It is called the *therapeutic range*. It should be larger than the deepest limit of the target volume.

4. Practical Determination of Electron Beam Energies

Empirical relations between range parameters and energy for broad beams perpendicularly incident on a phantom surface have been proposed. Markus (1961, 1964) has suggested the following relationship:

$$E_{p,0} = 1.95R_p + 0.48 \quad (23)$$

where $E_{p,0}$ is the modal energy in mega-electron-volts at the phantom surface and R_p is the practical range in centimeters of water. This relationship is valid within an uncertainty of 2% from a few mega-electron-volts to 25 MeV. To take account of the increasing importance of the radiative energy loss for higher energies, the following equation was proposed (Brahme and Svensson, 1976) to be used between a few mega-electron-volts and 50 MeV:

$$E_{p,0} = 0.22 + 1.98R_p + 0.0025(R_p)^2 \quad (24)$$

Equation (24) has been verified for clinical electron beams with various energy spreads. In both formulae R_p is assumed to be measured in centimeters of

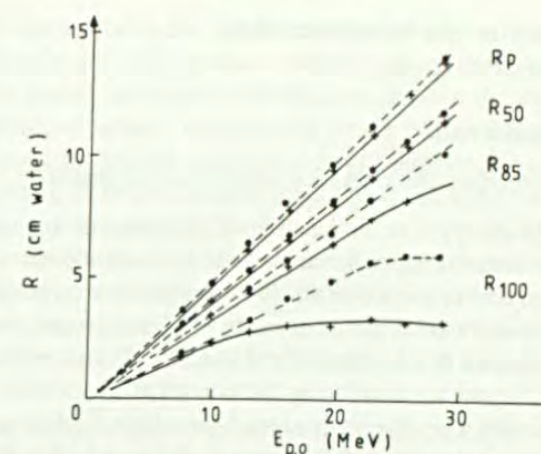


Fig. 14. Variation of the range parameters with the modal energy $E_{p,0}$. The dashed lines correspond to an infinite SSD and the solid lines to SSD 90 cm. The difference for R_p between the two sets of measurements is not more than 1 mm for energies lower than 10 MeV but reaches 3 mm at 28 MeV.

water. The corrections to be applied when R_p is measured in a plastic phantom are detailed in Section III, B.2.

A relation between the mean electron energy at the phantom surface $\bar{E}_0$ (in mega-electron-volts) and the 50% range R_{50} (in centimeters) has been proposed and verified by Brahme and Svensson (1976) on medical accelerators between 5 and 35 MeV:

$$\bar{E}_0 = 2.33R_{50} \quad (25)$$

where R_{50} is determined from an absorbed dose curve in water, corrected for divergence (infinite SSD).

Because of the contribution of the low-energy secondary electrons (Fig. 11) $\bar{E}_0$ is always smaller than $E_{p,0}$. The difference between the calculated values of $\bar{E}_0$ and $E_{p,0}$ for electron beams is related to the width of the energy spectrum.

The previous equations [(23) to (25)] relate the energies at the phantom surface to the range parameters. Harder (1966) has shown that the modal energy decreases linearly with depth:

$$E_{p,z} = E_{p,0} (1 - z/R_p) \quad (26)$$

The energy to be used in dosimetric applications when no other source of data is available is the mean energy $\bar{E}_z$ at the depth z of interest (rather than the modal energy $E_{p,z}$) but as a first approximation it can be assumed that the mean energy of the electrons also decreases linearly with depth:

$$\bar{E}_z \approx \bar{E}_0 (1 - z/R_p) \quad (27)$$

B. MEASUREMENT OF THE ABSORBED DOSE AT THE REFERENCE POINT

1. Reference Conditions

Table XIII summarizes the reference conditions suggested.

(a) As for high-energy x rays, the various published protocols recommend that absorbed dose measurements be carried out in a water phantom. However, for electron beams, and in particular for low-energy electron beams, it is often more convenient to use a solid phantom made of plastic layers, the thicknesses of which can be adjusted to a fraction of a millimeter. The corrections to be applied to scale the depths are detailed in Section III,B,2.

(b) In a beam with a small energy spread for which $\bar{E}_0$ differs only slightly from $E_{p,0}$, the depth of the point of reference is chosen equal to R_{100} . The great advantage is that the position of the ionization chamber is not critical, since at the depth of the peak, the dose variation is small. However, when the beam energy spread is wide ($\bar{E}_0$ much smaller than $E_{p,0}$) the depth of the peak is decreased and eventually the maximum dose is reached at the phantom surface. In such circumstances the minimum depth for the reference point must be chosen as indicated in column 3 of Table XIII.

(c) Most of the published protocols recommend a field size equal to $10 \times 10 \text{ cm}^2$ whatever the energy. Clinically, very low energies are, in general, used to treat small superficial tumors and the collimators available for these energies may be smaller than $10 \times 10 \text{ cm}^2$. The use of a collimator not adapted to the energy of interest could lead to large errors.

For very high energies ($>30 \text{ MeV}$), equilibrium of the secondary and scattered electrons is not achieved on the beam axis for a $10 \times 10 \text{ cm}^2$ field size and a larger field size is required.

(d) Since the dose distribution along the beam axis may vary considerably with collimator distance depending upon the collimator design, the reference

TABLE XIII
REFERENCE CONDITIONS IN ELECTRON BEAMS

Energy $E_{p,0}$	Depth of the reference point in water		Field size
	$\bar{E}_0 \approx E_{p,0}$	$(E_{p,0} - \bar{E}_0)/E_{p,0} > 0.1$	
1-5 MeV	R_{100}	R_{100}	$10 \times 10 \text{ cm}^2$
5-10 MeV	R_{100}	1 cm	$10 \times 10 \text{ cm}^2$
10-20 MeV	R_{100}	2 cm	$10 \times 10 \text{ cm}^2$
20-30 MeV	R_{100}	3 cm	$10 \times 10 \text{ cm}^2$
30-50 MeV	R_{100}	3 cm	$15 \times 15 \text{ cm}^2$

dose should be measured at the usual treatment or exposure distance. Furthermore, the various scattering materials, including air, between source and phantom modify the lateral and angular distribution of electron beams. It is possible to define an effective source, downstream on the beam axis, so that correction formulae based on the inverse square law can be applied. Any correction of the absorbed dose for a different distance should be made using the effective source distance (Briot and Dutreix, 1976; Khan *et al.*, 1978; ICRU, 1985), which decreases when the field size and the beam energy decrease.

(e) The effective point of measurement of the ionization chamber should be placed at the reference point. The meaning of the effective point of measurement and of the radial displacement has been explained in Section II,C for photon beams; these theoretical considerations are also valid for electron beams.

For flat, thin-window chambers, oriented perpendicularly to the beam direction, it is in general accepted that the effective point of measurement lies on the inner surface of the entrance window. It corresponds in the calculation to the assumption that the electrons are entering into the cavity in the forward direction. For cylindrical cavities with axis perpendicular to the beam, measurements have been performed for various diameters (Dutreix and Dutreix, 1966; Hettlinger *et al.*, 1967; Weatherburn and Stedeford, 1977; Johansson *et al.*, 1978). When results are corrected for the variation of the perturbation correction factor with depth, all the results agree reasonably well and lead to a radial displacement equal to $0.5r \pm 0.1r$, independent of energy. It has to be noted that when the reference point is defined at the peak depth in the dose plateau region, a small error in the depth causes a negligible uncertainty in the absorbed dose determination. In very-low-energy electron beams (6 MeV or less), for which the peak is sharp, it is especially recommended to use a flat, thin-window ionization chamber.

2. Phantom Materials Other Than Water

For two materials to be considered as equivalent for electron dosimetry, three physical quantities should be identical for the whole electron energy range under consideration, namely, linear collision stopping power (s_{coll}), linear total stopping power (s_{tot}), and linear angular scattering power (θ^2/l) (ICRU, 1972; Berger and Seltzer, 1982). In practice there is no ideal water-equivalent material, but any phantom material used in photon beams may be used in electron beams on condition that convenient scaling factors be applied. The simplest scaling factor to be used is given in Eq. (23) for photon beams and can be used for electron beams as a first approximation. A better approximation is achieved by the following relation:

$$z_w = z_m \cdot (\rho_m/\rho_w) \cdot (r_{0,w}/r_{0,m}) \quad (28)$$

where z_w is the thickness in centimeters of water leading to the same depth dose as the thickness z_m in centimeters of material m , ρ_w and ρ_m are the mass densities of water and material m , and $r_{0,w}$ and $r_{0,m}$ are the CSDA ranges in grams per square centimeter of water and of material m . The continuous slowing-down range r_0 is the mean range of an electron of energy $\bar{E}_0$, defined by integrating the reciprocal of the total stopping power. r_0 is equal to the path length which an electron would travel in the course of continuous slowing down, i.e., if its rate of energy loss were not fluctuating. The use of the CSDA range as a scaling factor allows the scaling of depths by means of stopping-power values averaged over the whole range of electrons. Table XIV gives the ratios of the CSDA ranges for a few water substitutes. These ratios have to be multiplied by the densities of the material to calculate the scaling factor of Eq. (28).

Johansson *et al.* (1979) have compared the various range parameters in different water substitutes and have found that the two scaling methods [Eqs. (23) and (28)] could be used with reasonable accuracy, provided the CSDA range r_0 of the water substitute did not deviate by more than $\pm 5\%$ from the corresponding range in water. Kase *et al.* (1982) have not found any significant difference in relative depth dose between polystyrene and water when scaling depths by the measured practical ranges.

One may expect a small deviation in the ionizations read at reference depths in two different materials when the linear scattering powers θ^2/l are different, since the average obliquities of the electrons are then different. Johansson *et al.* (1979) found no significant difference when the linear scattering power of the

TABLE XIV
RATIOS OF CSDA RANGES IN WATER AND IN MATERIAL m , $r_{0,w}/r_{0,m}$ FOR
DIFFERENT PHANTOM MATERIALS AS A FUNCTION OF ELECTRON
ENERGY $E_{p,0}$ ^a

$E_{p,0}$ (MeV)	$r_{0,w}/r_{0,m}$			
	Polystyrene	Lucite	A-150 plastic	Polyethylene
4	0.968	0.966	0.989	1.038
6	0.967	0.965	0.988	1.035
8	0.966	0.965	0.987	1.032
10	0.965	0.964	0.986	1.029
12.5	0.964	0.964	0.984	1.026
15	0.962	0.963	0.982	1.023
17.5	0.961	0.962	0.981	1.021
20	0.959	0.961	0.980	1.018
25	0.956	0.960	0.977	1.013
30	0.953	0.958	0.973	1.008
35	0.950	0.957	0.971	1.003

^aData from Berger and Seltzer (1982).

water substitute does not differ by more than $\pm 5\%$ from the corresponding data in water. They showed that the ionization measured in a polystyrene phantom is too low by 1.2%, whatever the energy between 5 and 20 MeV. Kase *et al.* (1982) determined the maximum ionization measured in a polystyrene phantom to be too low by as much as 3 to 4% as compared with that in water (although the relative depth dose in polystyrene is not different from water).

In 1984 two papers (Galbraith *et al.*, 1984; Mattsson and Svensson, 1984) showed that plastics behave as electrical insulators. Used in electron beams, they store a charge that may increase the chamber readings by several percent.

3. Correction Factors to Apply to the Instrument Readings

Following the same procedure as for photon beams, the absorbed dose to water at the reference point is determined in two steps (Section II,C,2), which are summarized as follows.

The absorbed dose to the air of the cavity is determined first, using relation (6):

$$\overline{D}_{\text{air},u} = M_u \cdot N_D$$

where M_u is the instrument reading, corrected to reference pressure and temperature and for recombination losses (see Section I,B,3). As the polarity effect is especially large in electron beams for some flat ionization chambers, it may be necessary to perform measurements with both polarities; M_u is then the mean of the two readings.

Second, the absorbed dose to water at the effective point of measurement for the ionization chamber under consideration is calculated by Eqs. (16) and (17):

$$D_{w,u} = \overline{D}_{\text{air},u} \cdot [\bar{S}/\rho]_{\text{air},u}^w \cdot p_u$$

$[\bar{S}/\rho]_{\text{air},u}^w$ is the ratio of the average, restricted, collision mass stopping powers in water and in air calculated over the electron energy spectrum at the position of the effective point of measurement.

If the ionization chamber is positioned as recommended with the effective point of measurement at the reference depth, then $D_{w,u}$ is the absorbed dose at the reference point.

The detector senses the electron fluence at the point where the electrons enter the cavity and not at the geometrical center of the detector. This effect is especially noticeable in the high dose-gradient regions encountered in electron beams. It is accounted for by the concept of the effective point of measurement. The differences in stopping and scattering properties of the detector, as compared with medium, perturb the electron fluence of the uniform medium. The perturbation correction factor p_u in Eq. (17) takes care of this fluence modification.

a. *Evaluation of the Mean, Restricted, Collision Mass Stopping-Power Ratio.* The Spencer-Attix formulation (Burlin, 1968) in the theory of cavity ionization uses stopping powers that include only energy losses less than a given cutoff energy Δ which depends on the cavity size. However, in this formulation, the restricted stopping powers are averaged over the whole slowing-down spectrum, including primary and secondary electrons with energies greater than Δ . In the well-known Bragg-Gray theory, the simplest formulation uses unrestricted stopping powers averaged over the slowing-down spectrum of the primary electrons alone and is independent of the cavity size.

Values of $[\bar{S}/\rho]_{\text{air}, u}^w$ for electron beams are given in Table XV as a function of depth (z) and mean energy at the phantom surface $\bar{E}_0$. These values, taken from Berger (1980) and published in ICRU Report 35 (1984) were calculated for an infinitely wide, plane-parallel beam of electrons incident upon a semi-infinite phantom using a cutoff energy of 10 keV. They represent the best approximation available for finite field sizes.

The $[\bar{S}/\rho]_{\text{air}, u}^w$ values given in Table XV are correct only if the distribution of electrons in energy at the phantom surface is characterized by a gaussian differential distribution in energy with a very narrow energy spread, that is to

TABLE XV

VALUE OF $[\bar{S}/\rho]_{\text{air}, u}^w$ FOR ELECTRON BEAMS AS A FUNCTION OF DEPTH z IN WATER AND MEAN ENERGY AT THE PHANTOM SURFACE $\bar{E}_0$ ^{a, b}

Depth z (cm)	$\bar{E}_0 = 4$ MeV $R_{p, t} = 1.9$ cm	6	8	10	12	16	20	25	30
0.0	1.066	1.042	1.023	1.008	0.996	0.976	0.961	0.945	0.933
0.5	1.083	1.051	1.030	1.014	1.002	0.983	0.967	0.951	0.939
1.0	1.105	1.070	1.043	1.024	1.010	0.989	0.973	0.957	0.944
1.5	1.125	1.090	1.059	1.036	1.019	0.996	0.979	0.962	0.949
2.0	1.134	1.109	1.076	1.049	1.030	1.002	0.984	0.967	0.953
2.5		1.125	1.093	1.064	1.041	1.010	0.990	0.972	0.958
3.0		1.133	1.109	1.080	1.054	1.019	0.996	0.977	0.962
4.0			1.130	1.109	1.082	1.038	1.010	0.987	0.971
5.0				1.127	1.109	1.061	1.026	0.999	0.980
6.0					1.124	1.085	1.045	1.012	0.991
7.0					1.126	1.106	1.066	1.028	1.002
8.0						1.119	1.087	1.044	1.015
9.0						1.124	1.104	1.062	1.029
10.0							1.115	1.080	1.043
12.0								1.107	1.074
14.0								1.112	1.101
16.0									1.108

^aData from Berger (1980).

^bTheoretical practical ranges $R_{p, t}$ are calculated from Eq. (24). $\Delta = 10$ keV, $I(\text{water}) = 75.0$ eV, and $I(\text{air}) = 85.7$ eV.

say by a mean energy $\bar{E}_0$ nearly equal to the most probably energy $E_{p, 0}$. The values indicated as theoretical practical range $R_{p, t}$ have been calculated from Eq. (24), assuming a value of $E_{p, 0}$ identical to $\bar{E}_0$.

Electron collimators used in radiotherapy are often made of low-atomic-number materials and are intended to contact the patient's skin. The electrons scattered by the applicator walls into the treated area have a lower energy than that of the primary electrons and a wider angle of incidence. They enlarge the energy spread at the phantom surface, increase the surface dose, and reduce the depth of maximum dose (de Almeida and Almond, 1974). The energy spread, the depth-dose curves, and the beam flatness may be improved by a careful design of the collimator associated with optimized scatterers (Briot *et al.*, 1973; Bjärngard *et al.*, 1976).

The measured practical range $R_{p, m}$ in a contaminated beam of mean energy $\bar{E}_0$ is larger than the theoretical one $R_{p, t}$, calculated assuming $E_{p, 0} = \bar{E}_0$. The variation in the values of the mass stopping-power ratio between the actual beam and the theoretical one can be neglected at small depths, and in particular at the depth of the reference point. However, at larger depths a correction is necessary. A simple correction can be made by scaling the depths with the ratio $R_{p, t}/R_{p, m}$. The input depth z_t to be used in the table is calculated as follows:

$$z_t = z_m \cdot (R_{p, t}/R_{p, m})$$

where z_m is the actual depth of the point of interest.

Table XVI shows the variation in the mass stopping-power ratio calculated in two different beams.

b. *The Perturbation Correction Factor.* Almond and Svensson (1977) and Kase *et al.* (1982) have verified that the influence of the chamber composition, that is to say the modification of the electron fluence by low-Z wall material, is negligible in electron beams.

The perturbation of the electron fluence by the detector is essentially due to the large difference in density between the phantom material and air (Harder, 1968). Perturbation factors have been determined experimentally with extrapolation chambers for plane-parallel chambers by Morris and Owen (1975) and

TABLE XVI

COMPARISON OF $[\bar{S}/\rho]_{\text{air}, u}^w$ VALUES AT THREE DEPTHS FOR TWO ELECTRON BEAMS SPECIFIED BY THE SAME $\bar{E}_0$ VALUE BUT WITH DIFFERENT MEASURED PRACTICAL RANGES $R_{p, m}$

	$\bar{E}_0$ (MeV)	$E_{p, 0}$ (MeV)	$R_{p, m}$ (cm)	$R_{p, t}$ (cm)	z_m (cm)	z_t (cm)	$[\bar{S}/\rho]_{\text{air}, u}^w$
Betatron	18.6	22.7	11.2	9.2	2	1.6	0.988
					5	4.1	1.023
					8	6.6	1.077
Linear accelerator	18.6	19	9.4	9.2	2	2.0	0.994
					5	4.9	1.039
					8	7.8	1.104

Markus (1976). Their measurements lead to the conclusion that the perturbation is negligible for flat chambers perpendicular to the beam when the effective point of measurement is considered as lying on the inner surface of the front wall.

Johansson *et al.* (1978) have measured the perturbation correction factors with cylindrical ionization chambers positioned at the dose maximum depth, in order to eliminate any influence of the effective point of measurement. They have shown a continuous increase of the perturbation with increasing cavity diameter and decreasing electron energy at the point of measurement. This last increase may be explained by the increase of the linear scattering-power ratio of medium and air with decreasing energy. The perturbation correction factor is then expected to increase with depth due to the energy decrease. These investigators have shown that the perturbation correction factor p_u decreases linearly with cavity radius r , i.e., $p_u = 1 - kr$. Table XVII gives k values versus the mean energy $\bar{E}_z$ of the electrons at the depth z of the reference point calculated from the experimental data of Johansson *et al.* (1978).

C. MEASUREMENT OF THE ABSORBED DOSE AT ANY POINT

1. Measurement of the Absorbed Dose Distribution along the Beam Axis

(a) The concept of the radial upstream displacement should be used for any point. Every measurement should be carried out referring the reading to the effective point of measurement of the ionization chamber. In low-energy electron beams, accurate measurements should be performed with parallel-plate ionization chambers after careful check of the polarity effect, considering the effective point of measurement as being on the inner surface of the front wall. In high-energy electron beams, where the dose gradient is less steep, accurate measurements can be performed with a cylindrical ionization chamber, referring the readings to a point upstream of the geometrical center of the chamber, at a distance equal to $0.5r$, where r is the cavity radius. However, even for high-energy electron beams, the absorbed dose at small depths (less than 1 cm) may only be

TABLE XVII

CALCULATION OF THE PERTURBATION CORRECTION FACTOR $p_u = 1 - kr$ FOR A CYLINDRICAL CHAMBER RADIUS r (IN CENTIMETERS) IN AN ELECTRON BEAM OF MEAN ENERGY $\bar{E}_z^a$

$\bar{E}_z$ (MeV)	4	6	8	10	12	15	20
k (cm ⁻¹)	0.13	0.10	0.08	0.06	0.045	0.035	0.02

^aAfter the experimental data of Johansson *et al.* (1978).

measured accurately with a parallel-plate chamber.

(b) It is often simpler to use a solid, plastic phantom instead of a water phantom, in particular for low energies with a parallel-plate ionization chamber. The depths in a plastic phantom should be scaled as explained in Section III,B,2.

(c) The relative absorbed dose distribution at depth is different from the relative ionization distribution because of the continuous energy decrease with depth in electron beams. Both the average mass stopping-power ratio and the perturbation correction factor vary significantly with energy and the ratio $D_{w,u}/M_u$ cannot be considered as constant with depth.

At each measured point it is then necessary to apply the needed correction, after the evaluation of the mean energy $\bar{E}_z$ at the depth z of the point of interest. The procedure is thus more laborious than that for photon beams. Figure 15 shows the difference between the depth-ionization curve and the depth-absorbed dose curve in a 28-MeV electron beam. The largest depth difference between the two curves is as large as 10 mm, and appears at a depth where the dose is 85% of the maximum, sometimes referred to clinically as the therapeutic range (R_{85}).

The values of the mean mass stopping-power ratios and of the perturbation correction factors are given in Tables XV and XVII, respectively, as functions of the depth and of the mean energy at the point of interest.

Readings at depths larger than the practical range should be corrected by using the correction factors for photon beams of the corresponding energy. The spectrum of the bremsstrahlung x rays generated in the phantom material is different from the spectrum of a photon beam generated in a high-Z target, but the error in the background absorbed dose, which results from assuming the latter spectrum, is not significant.

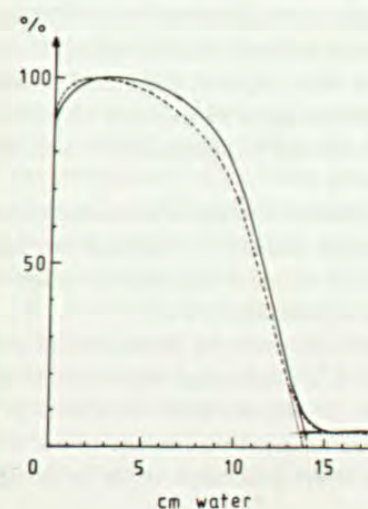


Fig. 15. Depth-ionization curve (dashed line) in a 28-MeV electron beam (SSD 90 cm, field size 20×20 cm). The solid line shows the depth-absorbed dose curve recalculated from relative ionization measurements. The largest difference (~ 10 mm) between the two curves appears at a depth approximately equal to R_{85} , that is, at the most important depth in medical applications.

2. Measurement of the Absorbed Dose at Any Point

The shape of the absorbed dose distribution in a high-energy electron beam is critically dependent upon the irradiation conditions and should be measured for each set of conditions in use. For instance, an air space of a few centimeters between the collimator end and the phantom surface may change entirely the dose distribution.

Measurements with ionization chambers using different correction factors are rather laborious. Furthermore, the position of the effective point of measurement in a cylindrical chamber is not well defined near the beam edges. Most circular flat chambers have insufficient lateral spatial resolution for accurate measurements in the penumbra region. Special flat chambers with a rectangular section of the cavity (rather than circular) have been designed to have an adequate spatial resolution.

Film dosimetry in a solid, plastic phantom is often preferred for the determination of dose distributions because it is not time-consuming and it presents the great advantage of a high spatial resolution. Greater accuracy can be achieved if the dose distribution along the beam axis is normalized to measurements made with an ionization chamber (see Section I,A,4).

IV. Uncertainties

A. GENERAL CONSIDERATIONS

The uncertainties in the determination of calibration factors for instruments were analyzed in Section I,C. The uncertainties connected to that part of the measurement chain are of the same order of magnitude for any user, since the precision achieved in any national laboratory (Niatel *et al.*, 1975) or secondary standards laboratory can be assumed to be similar throughout the world. The situation is different for the second part of the measurement chain, leading to the determination of absorbed dose to a patient or other exposed individual. These uncertainties are largely determined by the efforts spent by the local physicist.

Before any tentative estimation of uncertainties can be attempted the meaning of certain words must be clarified.

(1) *Random errors* result in variations observed when a measurement is repeated several times under the same conditions: statistical considerations lead to the calculation of the standard deviation (or random uncertainty) taken to some degree of confidence (for instance, 95% confidence level).

(2) *Systematic errors* are introduced by limitations in the measurement process. Large errors can be caused by the use of a noncalibrated measurement device (noncalibrated barometer, for instance), by the deviation with time of a measurement device (distance indicator, for instance), by a mistake of the user during the calibration process, or by an incorrect procedure or an error in a

physical parameter. Systematic errors are usually related to the quality assurance program of the institute in which the measurements are carried out. However, unavoidable systematic errors may be related to basic limitations in the accuracy of the calibration of an instrument or in the determination of necessary physical parameters.

Systematic uncertainties in the determination of the absorbed dose at a point are defined as the estimated limits of the systematic errors. These limits vary drastically among institutions depending on the quality and experience of the physics personnel.

(3) Errors can be divided into four groups:

(a) The systematic errors related to the physical parameters to be used in the absorbed dose calculation, such as perturbation factor p_a and mass stopping-power ratios.

(b) The errors related to the field instrument, such as repeatability, temperature and pressure, recombination loss, and polarity effect.

(c) The errors related to the parameters of the radiation-producing unit: SSD, field size, monitor or timer, energy variations with time, and field nonuniformity.

(d) The errors associated with the phantom and the devices used to measure dose distribution: depth, distance to the beam axis, size of the dosimeter (especially in the penumbra region), computer program (if a computer is used), etc.

Apart from group (a), the distinction between random errors and systematic errors is not easy. In an institution with a high-quality measurement program, systematic errors are likely to be negligible compared with random errors and only the latter have to be considered. On the contrary, in other situations systematic errors can be much larger than random errors. Let us consider a few examples to illustrate these general considerations.

First example: the random uncertainty in absorbed dose related to temperature measurement is about 0.2% for a random uncertainty of 0.5 K in the temperature reading. However, an error of one degree may be introduced by a thermometer of poor quality and an error of a few degrees may be introduced if the temperature of the water phantom is different from the room temperature, leading to a systematic uncertainty of 0.5 or 1% in absorbed dose.

Second example: the light simulator of the field size can be adjusted with a random uncertainty of 1 mm, leading to a random uncertainty in the dose of 0.1%. However, the light simulator may be in error by 3 mm or more if it has not been correctly adjusted, leading to a systematic uncertainty in the absorbed dose along the axis equal to 0.3%. Furthermore, in radiation therapy the critical organs near the beam edge may be overirradiated by a factor of two or more due to the error in light-beam adjustment, or to a misalignment of the lightbeam axis. An error of a few millimeters may lead to a very severe accident in some

cases, for instance, in the treatment of a head or neck tumor where the limit of the target volume lies only a few millimeters from the spinal cord. A given error, for instance, a 3-mm variation between light beam and x-ray beam, would be considered as a systematic error for the patients treated with the therapy unit under consideration because the error is exactly the same for every patient and every time an individual patient is treated.

(4) The combination of the uncertainties may follow two main models. They may be combined by simple addition, assuming a high probability for all the uncertainties to have the same sign (positive or negative). This model applies only to systematic errors and gives the upper limit of the uncertainty, and in general corresponds to an overestimation of the uncertainty. Alternatively, the uncertainties may be combined in quadrature. This model applies to random errors but it may lead to an underestimation of the uncertainties if used for systematic errors. The BIPM in a report (Bureau International des Poids et Mesures, 1980) has proposed to discontinue the use of the words "random" and "systematic" and to replace them by "category A" and "category B" uncertainties. It has also proposed to combine the uncertainties in quadrature, whatever the category, to obtain the overall uncertainty. This overall uncertainty should be multiplied by a certain constant factor, the choice of which is left to the experimenter depending upon the type of application. Its numerical value varies between one and three and should be clearly stated.

Assuming, for instance, that each of 10 parameters introduces small uncertainty equal to 0.2% and that 1 parameter introduces a large uncertainty equal to 2%, the combination by addition should lead to an accumulated uncertainty of 4%, where half the uncertainty is due to the 10 small uncertainties. When the uncertainties are combined in quadrature, the accumulated uncertainty is equal to 2.1%, where the contribution of the 10 small uncertainties is quite negligible.

In their evaluation of the total uncertainties in radiation dosimetry, Loevinger and Loftus (1977) combined uncertainties in quadrature and did not propose any distinction between systematic and random uncertainties. They considered two extreme situations, an optimal model and a minimal model. Johansson (1982) considered an optimal model and assumed that the systematic errors (apart from physical coefficients) were quite negligible as compared with random errors. He combined the systematic uncertainties in quadrature and applied weighting factors to random and systematic errors following the method proposed by the IAEA (1979).

(5) The possible consequences of uncertainties have to be considered. In radiation therapy applications, systematic uncertainties in fundamental physical quantities (e.g., stopping power) may be rather large compared with some of the random uncertainties. Since these quantities are generally used uniformly over the world, they have no direct clinical consequence because the error in the dose absorbed by tissues is identical for all patients. Clinical experience has

been and is still being gained through national and international intercomparisons, using values of dose which are based on the same values of physical parameters in every country. Any modification of a value of a physical quantity used should be clearly indicated, even if the "new data" lead to a decrease of the uncertainty, because it would unavoidably lead to a modification of the dose effectively delivered to the patients.

On the other hand, systematic as well as random uncertainties in the dose introduced at the therapy center could be deleterious to the clinical results. Furthermore, they will certainly increase the uncertainty in any comparison between patients treated with different therapy units and at other centers.

B. UNCERTAINTIES IN THE DETERMINATION OF THE ABSORBED DOSE AT THE REFERENCE POINT

The absorbed dose at the reference point (e.g., in water) must be determined by the user for each beam quality used. Measurements are performed either with a field instrument or with the reference instrument calibrated at the secondary standards laboratory. Following the proposed protocol, ionization chambers are calibrated in terms of the mean absorbed dose to the air cavity N_D . The uncertainties in the N_D factor are given in Table IV.

For cobalt-60 beams, a detailed analysis of uncertainties in absorbed dose determination at the reference point in water with an ionization chamber has been made by Loevinger and Loftus (1977) and Johansson (1982). The results of their studies are compared in Table XVIII according to the four groups proposed for the sources of errors. In Johansson's data the main uncertainties are the systematic uncertainties in the physical parameters (perturbation factor and mass stopping-power ratio) and in the nonlinearity of the field instrument. Johansson considers the other sources of error as minimal, and gives uncertainties similar to the ones proposed in the optimal model by Loevinger and Loftus. A direct comparison between the two sets of data is not easy because the investigators have not followed the same protocol for the calibration of the ionization chamber. Furthermore, Loevinger and Loftus took into account some sources of errors related to the instrument which were not considered by Johansson: in particular, calibration factor stability, energy dependence, and leakage and stem effect of ionization chambers.

We have estimated the overall uncertainties in the absorbed dose at the reference point in water by combining the various uncertainties in quadrature. The value calculated in Johansson's model (1.8%) is in fair agreement with the uncertainty calculated for the optimal model of Loevinger and Loftus (1.7%). In the minimal model the uncertainty is as high as 3.6%. The previous values concern only the uncertainties in the absorbed dose at the reference point in water, and they must be combined in quadrature with uncertainties in the calibration

TABLE XVIII
UNCERTAINTIES IN THE ABSORBED DOSE AT THE REFERENCE POINT IN WATER FOR
COBALT-60 GAMMA-RAY BEAMS^a

Sources of error	Johansson (1982)	Loevinger and Loftus (1977)	
		Optimal model	Minimal model
(a) Systematic errors			
p_a or d	0.7%	1.0%	1.5%
$(\bar{S}/\rho)_{air, u}^w$	1.0%	—	—
(b) Field instrument			
Repeatability	0.1%	0.5%	1.7%
Resolution	0.2%		1.0%
Nonlinearity	1.0%		1.0%
Temperature	0.2%		1.0%
Pressure	0.3%	0.1%	0.1%
Energy dependence	—	0.5%	0.5%
Calibration factor stability	—	0.5%	1.0%
Leakage	—	0.1%	0.2%
Stem effect	—	0.2%	1.1%
(c) Treatment unit parameters			
SSD	0.3%	0.5%	1.0%
Timer	0.2%	0.1%	1.5%
Field size	0.1%	0.2%	0.6%
(d) Phantom			
Depth	0.6%	—	—
Overall uncertainties [(a) to (d)]	1.8%	1.7%	3.6%

^aThe meaning of the different sources of error is detailed in the original papers.

factor of the field instrument (see Table IV) to obtain the accumulated overall uncertainties.

For the other beam qualities, x rays and electron beams, the overall uncertainties in the absorbed dose at the reference point in water are more serious than those for cobalt-60. Johansson has estimated these uncertainties at 2.8 and 3.8%, respectively, for x rays and electron beams. The differences observed are mainly related to the uncertainties in recombination loss, polarity effect, and energy determination, which are not considered for cobalt-60. The uncertainties concerning radiation quality are higher for electron beams because the mass stopping-power ratio and, to some extent, the perturbation factor used in air ionization chamber dosimetry are strongly energy dependent. A value of 2.5% is given by Johansson (1982) for the uncertainties in energy determination for the electron beams, as compared with 0.5% for photon beams.

Johansson and Loevinger and Loftus evaluated *a priori* the uncertainties in the various parameters. Instead of using this approach, some investigators (Johansson, 1982; Shalek *et al.*, 1976; Samulski *et al.*, 1981; Thompson *et al.*, 1978) have performed large intercomparisons between radiotherapy centers and

have analyzed the error distribution of the ratio of determined versus stated value either for some important parameter or for absorbed dose at the reference point.

Johansson (1982) compared the standard deviations calculated for the various experimental results, and showed that his results were similar (1.4% for cobalt-60 and 2.8% for high-energy x rays) to those of Samulski *et al.* (1981) (2.7% for a mixed group of cobalt-60 and x-ray units). These results were much better than the results published by Shalek *et al.* (1976) (3.4% for cobalt-60), probably showing a significant evolution with time. The standard deviation estimated by Johansson in the intercomparison of electron beams (3.2%) was higher than that for photon beams.

Postal dose intercomparisons have been performed between radiotherapy centers either on a national or on an international basis (Eisenlohr and Jayaraman, 1977; Racoveanu, 1981). The aim of such intercomparisons is to detect significant discrepancies between institutions in order to improve accuracy in radiotherapy. The stated accuracy in the IAEA-WHO intercomparison service for cobalt-60 radiotherapy machines is better than 5% at the 95% confidence level. Bjärngard *et al.* (1980) evaluated the feasibility of applying the same technique for x rays in the energy range 4–25 MV. They concluded that, for cobalt-60 γ rays, a discrepancy larger than 2% could be detected at the 95% confidence level, whereas a discrepancy larger than $\pm 3\%$ could be detected with the same confidence level for high-energy x rays.

C. UNCERTAINTIES IN THE DETERMINATION OF THE ABSORBED DOSE AT ANY POINT

In the procedure described for photon beams [Eq. (19) and Table XI] as well as for electron beams, a radiation producing unit is first calibrated for the energy under consideration at the reference point under reference conditions. In a second step, the absorbed dose at the reference point for the conditions under consideration is measured relative to the dose at the reference point under reference conditions. In the last step, the absorbed dose at the point of interest for the conditions under consideration is measured relative to the dose at the reference point under the same conditions.

The uncertainties encountered in each step should be combined to evaluate the total uncertainty. Most of the random uncertainties [types (b) and (c)] are encountered in each step and should be accumulated, for instance, the uncertainties related to temperature and pressure for field instruments, or to the adjustment of the parameters of the radiation-producing unit. On the other hand, any systematic error in one of these parameters is obviously identical during the three steps (provided the time elapsed between the various steps is not too long) and therefore should have no effect when ratios are taken.

As has been emphasized in Section IV,A, the efforts spent by the responsible physicist in the application of a quality assurance program are essential to reduce the various uncertainties. The estimation *a priori* of the size of the uncertainties does not seem to be realistic, and large intercomparison programs should be encouraged to determine the weakest points in the dose determination chain and to improve the final accuracy.

As an example, Johansson (1982) estimated *a priori* the uncertainty in field size at a radiation therapy center as 1 mm. This 1-mm uncertainty is, from our own experience, the lowest limit of the random uncertainty in dosimetric practice and does not include the systematic error in the field size indicator. Samulski *et al.* (1981) proposed a 2-mm criterion, but in their experience this value was not realistic because 40% of the units reviewed failed to meet this standard, whereas 88% of the machines checked were within a 3-mm criterion. Samulski *et al.* also observed that the rate of failure to meet their 2-mm criteria did not significantly change with time, showing that the actual uncertainty in field size is much larger than that proposed *a priori* by Johansson.

The standard deviations found experimentally in national intercomparisons are much smaller when all the radiotherapy centers follow the same protocol, because errors in physical quantities both in dosimeter calibration and in the absorbed dose determination at a point can be neglected, as they are identical for all the participants. The more complex the irradiations, the larger the errors. For instance, in their investigation Samulski *et al.* found that the adopted criterion of 5% in the absorbed dose at a target point was met by 95, 85, and 77% of the participants for reference doses with open fields, wedge fields, and irregular large fields, respectively, whereas 87% of the units were found to have output within the recommended limit of $\pm 3\%$.

In a similar way, the uncertainties are larger in electron dosimetry than in high-energy photon dosimetry, and larger for high-energy x-rays than for cobalt-60 γ rays. The sources of errors are more numerous with accelerators, than with cobalt-60 units. Deviation of the monitor response with time, asymmetries in radiation fields, and misalignment of the flattening filters are supplementary sources of error. Furthermore, in electron beam dosimetry, the need to correct with a different mass stopping-power ratio at each point is an additional source of error. Johansson found that the cumulated uncertainty calculated for the absorbed dose at any point increased from 4.9% for cobalt-60 beams to 6% for x-ray beams, and to 6.8% for electron beams. However, Samulski *et al.* did not find any significant differences between cobalt-60 units and x-ray machines (mainly 4-MV units). They have shown that the main sources of errors were, surprisingly, the mechanical parameters and not the electrical ones. These results are in good agreement with the previous results by Gagnon *et al.* (1978).

Johansson *et al.* (1982) have performed a large intercomparison between all the therapy units of the Nordic countries. For this intercomparison Johansson

recalculated the overall uncertainty considering that the systematic errors at an individual center should be considered as random errors in an intercomparison. The standard deviations calculated for cobalt-60 units, x-rays, and electrons were, respectively, equal to 1.7, 2.3, and 2.5%, in good agreement with the experimental standard deviations of 1.4, 1.8, and 3.2%.

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4

Dosimetry of External Beams of Nuclear Particles

JOHAN J. BROERSE

RADIOBIOLOGICAL INSTITUTE TNO
RIJSWIJK, THE NETHERLANDS

JOHN T. LYMAN

LAWRENCE BERKELEY LABORATORY
UNIVERSITY OF CALIFORNIA
BERKELEY, CALIFORNIA

JOHANNES ZOETELIEF

RADIOBIOLOGICAL INSTITUTE TNO
RIJSWIJK, THE NETHERLANDS

I. Introduction	230
II. Beam Characteristics	232
A. Charged Particles	234
B. Fast Neutrons	234
III. Dosimetry Methods	238
A. Gas-Filled Devices	239
B. Calorimeters	244
C. Other Detection Techniques	246
D. Beam Monitoring	249
IV. Radiation Quality	250
V. Determination of Absorbed Dose at a Reference Point	252
A. Principles of Ionization Chamber Dosimetry	255
B. Physical Parameters for Ionization Chamber Dosimetry	257
C. Displacement Correction Factor	260
D. Comparison of Different Dosimetry Methods	264
VI. Treatment Planning for External Beam Therapy	264

VII. Response of Biological Dosimeters	268
A. Basic Aspects of the Concept of Relative Biological Effectiveness	269
B. Dosimetry Based on the Induction of Chromosome Aberrations	270
C. Intercomparison of Radiation Quality of Different Fast-Neutron Beams Derived from Biological Dosimeters	272
D. The Use of Biological Dosimeters to Assess RBE Variations with Position Inside (Human) Phantoms	273
VIII. <i>In Vivo</i> Dosimetry	275
A. Experience with Neutrons	276
B. Experience with Charged Nuclear Particles	278
IX. Conclusions	278
References	281

I. Introduction

In the context of this chapter nuclear particles are taken to mean entities with rest masses many times that of an electron, notably, pi mesons (pions), protons, neutrons, helium, and heavier ions such as carbon, neon, and argon. With the exception of protons these particles can be classified as high-LET (linear energy transfer) radiation, in view of the dense energy deposition along their tracks or (in case of neutrons) that of the secondary charged particles produced [see, e.g., ICRU (1978)]. Nuclear particles are of importance for fundamental radiobiology since their application generally results in dose-effect relationships which are less difficult to analyze and to interpret than is the case for low-LET radiation. Other areas of interest in the use of these particles are radiotherapy and medical and technological diagnosis. The latter applications can provide information complementary to radiographs obtained with conventional photon radiations [see, e.g., Tobias *et al.* (1977) and Bewley *et al.* (1973)]. The introduction of high-energy linear accelerators and cyclotrons involves specific radiation protection problems (Swanson, 1979). Assessment of levels of dose equivalent outside the shielding of nuclear installations should generally be made with an overall uncertainty better than about $\pm 30\%$. In the present chapter emphasis is placed on dosimetry of nuclear particles in biology and medicine. Due to the steepness of dose-effect relations, the dosimetry for biomedical applications should be performed with an accuracy (overall uncertainty) of less than $\pm 5\%$ and a reproducibility (precision) of within $\pm 2\%$ (ICRU, 1978).

Radiotherapy is an important means for the treatment of malignant diseases. In the Netherlands approximately 50–60% of all cancer patients are subjected to radiation either alone or as part of combined treatments, e.g., in conjunction with surgery or chemotherapy (Van der Heide-Schoon, 1983). Although radiotherapy with ^{60}Co gamma rays and megavoltage x rays has been successful for a number of maladies [e.g., 90% cure (10-year survival) for Hodgkin's disease, stage I] the results for tumors of some other types have been very disappointing [e.g., 5% cure for lung carcinoma (5-year survival) and 10% cure (10-year

survival) for grade II astrocytomas]. Due to the limitations of photon radiotherapy, considerable effort has been invested over the past 15 years in research on nuclear particles with advantageous physical and/or biological properties. Over 10,000 patients have been treated with fast neutrons (Dutreix *et al.*, 1984) and more than 2000 patients with heavy charged particles in centers located all over the world. Apart from the treatment of malignant diseases, beams of protons and heavier charged particles are employed for radiosurgery. Using small collimated beams and taking advantage of the minimal scattering of the particles and the adjustable depth of penetration by selection of the incident energy, small target volumes can be irradiated with restricted dose to the surrounding tissues. Protons and helium ions have been used for pituitary ablation in cases of hormone-sensitive metastatic disease, and for treatment of pituitary tumors, central nervous system disorders, and primarily intracranial vascular disorders (Kjellberg, 1979; Fabrikant *et al.*, 1980). The number of patients treated with these radiosurgical techniques (small volume, high dose, low fraction number, short treatment time) is about 3000.

Fast nuclear particles can be produced with the aid of linear accelerators, cyclotrons, and synchrotrons either directly or through nuclear interactions. In Table I, the characteristics of the particles and the minimum energy required to treat deep-seated tumors (at depths in excess of 10 cm) are summarized. Fast neutrons are produced through the fission process or by the impinging of light ions (e.g., protons or deuterons) on targets commonly made out of deuterium, tritium, or beryllium. These latter reactions can be denoted as $i(E) + J(\Delta E)$, where $i(E)$ is the incident particle of type i with a kinetic energy E expressed in mega-electron-volts, J is the target element, and ΔE in mega-electron-volts is the maximum energy lost by the incident particle in the target. The notation $p(66) + \text{Be}(49)$ indicates a 66-MeV proton beam incident on a Be target in

TABLE I
NUCLEAR PARTICLE PROPERTIES^a

Particle	Mass	Charge	Minimum energy for therapy
Negative pion	273 m	$-1e$	60 MeV
Proton	1836 $m = 1.008\text{ u}$	$+1e$	160 MeV
Neutron	1839 $m = 1.009\text{ u}$	0	15 MeV
Helium ion	4.007 u	$+2e$	400 MeV/u
Carbon ion	12.0 u	$+6e$	250 MeV/u
Neon ion	20.0 u	$+10e$	400 MeV/u
Silicon ion	28.0 u	$+14e$	450 MeV/u
Argon ion	40.0 u	$+18e$	500 MeV/u

^a $m = 9.1096 \cdot 10^{-31}\text{ kg}$, $\text{u} = \text{atomic mass unit}$, and, $e = 1.6022 \cdot 10^{-19}\text{ C}$.

which it loses maximally 49 MeV. Pions are produced as secondary particles in nuclear interactions when a primary proton or electron beam of sufficient energy strikes a target material. For biomedical applications, proton and heavier charged particle beams are generally primary beams produced by accelerators with energies in excess of 70 MeV (the minimum proton energy to treat ocular tumors). Beams of radioactive ions such as ^{11}C can be produced by the fragmentation of a ^{12}C beam, followed by the appropriate momentum selection to separate the ^{11}C particles (Alonso *et al.*, 1979). Characteristics of several nuclear particle sources are described in more detail elsewhere (Broerse and Mijnheer, 1982a; Larsson, 1980; Raju, 1980; Salzmänn, 1984; Schimmerling, 1980). Information on the interaction processes of heavy charged particles with matter is available (Bichsel, 1968; Raju *et al.*, 1969; ICRU, 1978). An assessment of the usefulness of the different beams for biomedical purposes should be based on the spatial and temporal distributions of the dose and on their biological properties. The probable biological advantages of high-LET over low-LET radiation are related to the reduced oxygen effect, the lower amount of sublethal damage produced, and the smaller dependence of radiosensitivity with phase of the cell cycle [see, e.g., Barendsen (1971), Fowler (1981), and Hall (1982)].

The physicists responsible for the dosimetry of nuclear particle beams for therapy applications have intensified their contacts during the past few years. This has resulted in a formal cooperation in the European Clinical Neutron Dosimetry Group (ECNEU), and in Task Groups 18 and 20 of the American Association of Physicists in Medicine (AAPM) working on the therapy beam dosimetry of neutrons and heavy charged particles, respectively. These activities accomplished the formulation of two protocols for neutron dosimetry for external beam therapy (AAPM, 1980; ECNEU, 1981). These protocols provide a common basis for neutron dosimetry which is an essential requirement for evaluating and comparing the clinical results obtained at different fast-neutron radiotherapy facilities. A protocol for heavy charged particle therapy beam dosimetry is near completion. All protocols are concerned with the sequence of dosimetry procedures required to deliver and specify the absorbed dose to biological objects as summarized in Fig. 1.

The different steps involved in biomedical applications of nuclear particles will be discussed, including dosimetric techniques, spatial distribution of dose, radiation quality assessment, absolute dosimetry, treatment planning, response of biological dosimeters, and *in vivo* dosimetry.

II. Beam Characteristics

The characteristics of a fast nuclear particle beam with respect to its yield and energy spectrum determine to a large extent the irradiation techniques and dosimetry procedures to be applied. With regard to depth-dose distributions

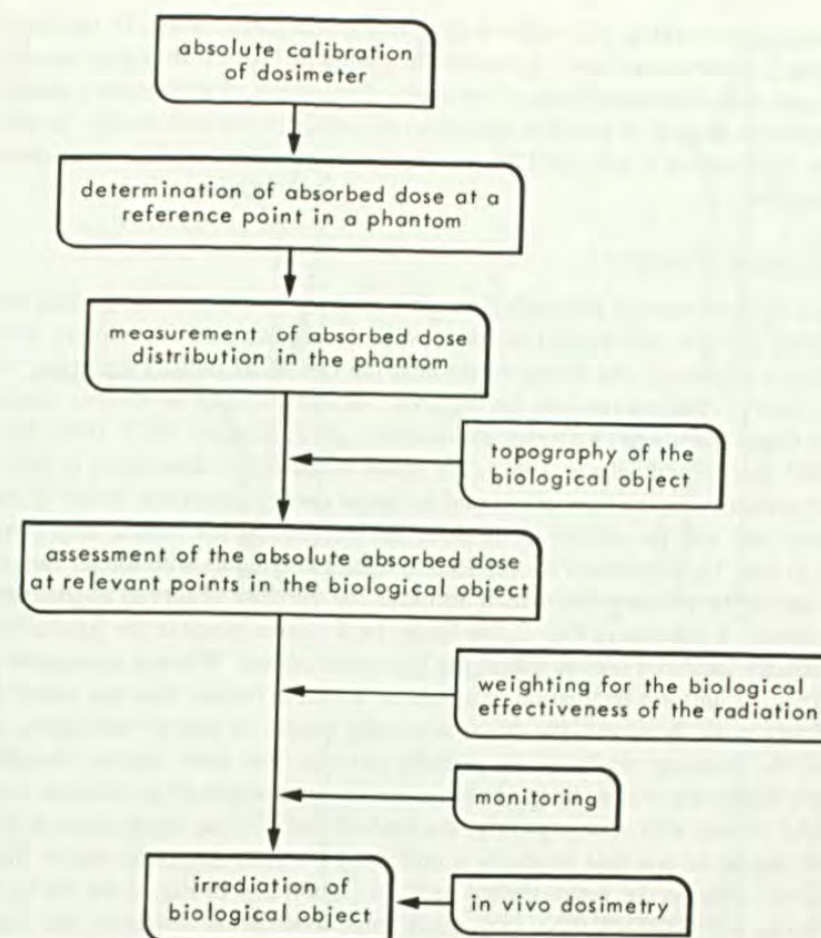


Fig. 1. Sequence of dosimetry procedures to deliver the absorbed dose to the biological object. [From Broerse and Mijnheer (1982b).]

there is a fundamental difference between charged and uncharged particles. A charged particle has a restricted range of penetration depending on its energy and type (Bragg and Kleeman, 1904). Neutrons can traverse matter over relatively long distances without interaction, and will in principle be attenuated exponentially. A complication in the use of some of these particle beams for biomedical application is their contamination with other radiation components, e.g., photons in the case of neutron beams and electrons, muons, and neutrons in pion beams (Raju, 1980). This phenomenon deserves special attention in view of the different RBE (relative biological effectiveness) values associated with the various types of radiation. The production of nuclear particles involves the use

of accelerators working with either a continuous or a pulsed beam. In the latter case (e.g., synchrotrons) special precautions should be taken with regard to saturation and dead-time corrections of detectors. The output of accelerators should be monitored in view of possible variations in beam current and energy. In this section information is provided about range-energy relations and depth-dose distributions.

A. CHARGED PARTICLES

As a charged particle proceeds through a medium its rate of energy loss (or ionization per unit path length) increases with decreasing particle velocity until reaching a maximum (the Bragg peak) near the end of its path. Tabulations of range, energy, and energy loss for different nuclear particles in various media can be found elsewhere (Andersen and Ziegler, 1977; Ziegler, 1977, 1980; Janini, 1982; Berger and Seltzer, 1982). The shape of the depth-dose curve is determined primarily by the type of ion and its initial energy spectrum. Some of the primary ions will be subject to inelastic nuclear interaction before they have come to rest. These interactions can lead to lighter fragments with longer ranges than that of the primary ion, which accounts for the dose observed beyond the Bragg peak. As shown in Fig. 2, the Bragg peak makes possible the irradiation of a strictly localized region within the biological object. When it is necessary to produce a uniform irradiation of a volume which is thicker than the width of the Bragg peak, however, the beam is usually range- or energy-modulated to spread the stopping region of the primary particle. For most nuclear charged particle beams there is a change in beam quality with depth of penetration (see Sections IV and VII). Consequently, the desired shape of the depth-dose distribution would be one that produces a uniform biological response, rather than a uniform dose, in the range-shifted peak. As illustrated in Fig. 3 for 80-MeV π mesons and 400-MeV ^{20}Ne ions, such range modulation decreases the ratio of dose values in peak and plateau regions considerably.

B. FAST NEUTRONS

The absorbed dose of fast neutrons as a function of depth shows an initial rise due to the buildup of secondary charged particle equilibrium (similar to that observed for photon beams), followed by an exponential decrease due to attenuation of the neutron beam. The depth at which the dose maximum is reached is determined by the range of the most energetic secondary protons. However, the high-LET component of the secondaries (alpha particles and heavy recoils) will attain equilibrium at much smaller depths (Broerse *et al.*, 1968). The buildup of secondary charged particle equilibrium is dependent on the experimental conditions, e.g., the properties of the collimator (material and field size). Contamination of the incident neutron beam with charged particles will influence the

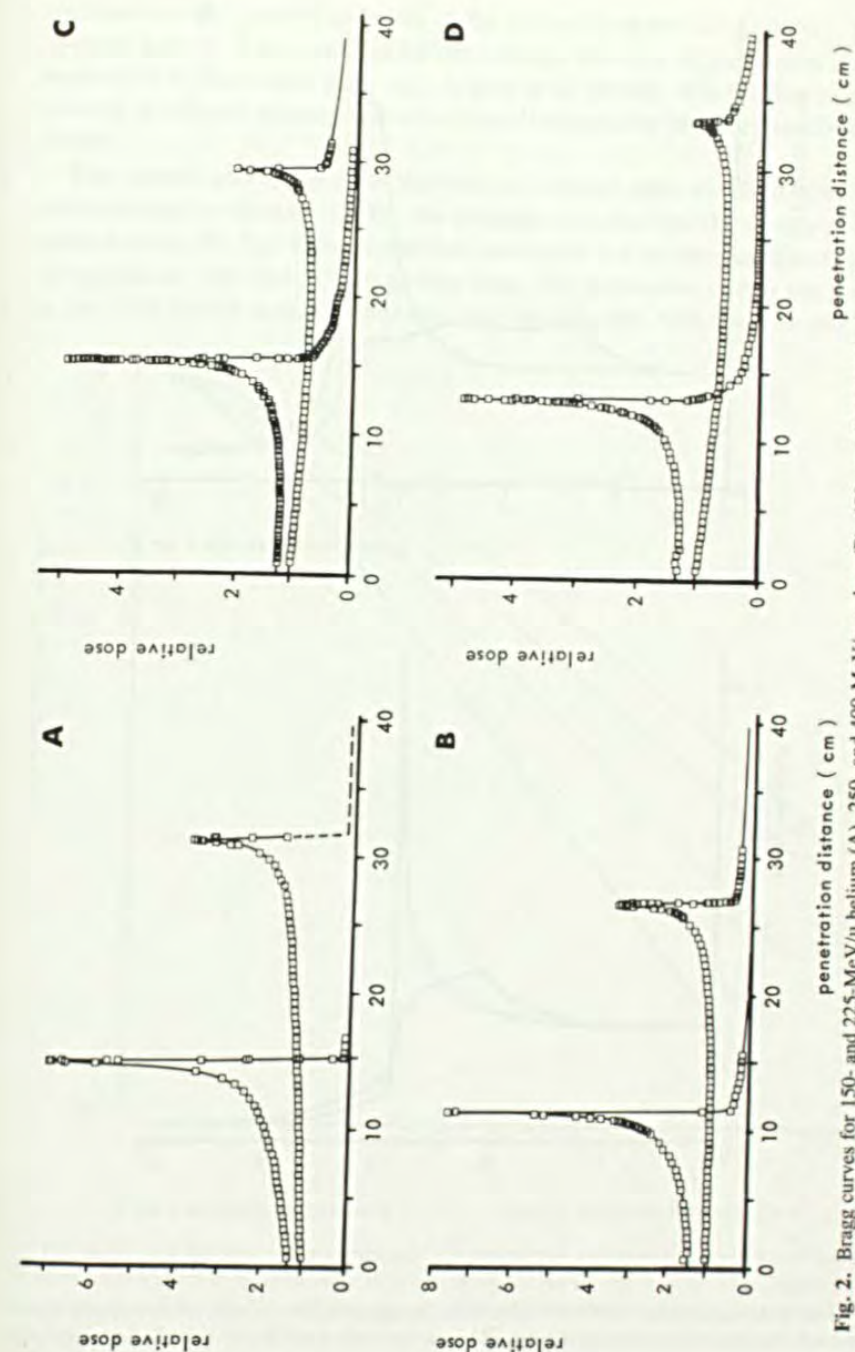


Fig. 2. Bragg curves for 150- and 225-MeV/u helium (A), 250- and 400-MeV/u carbon (B), 400- and 594-MeV/u neon (C), and 500- and 900-MeV/u argon (D) ion beams in water.

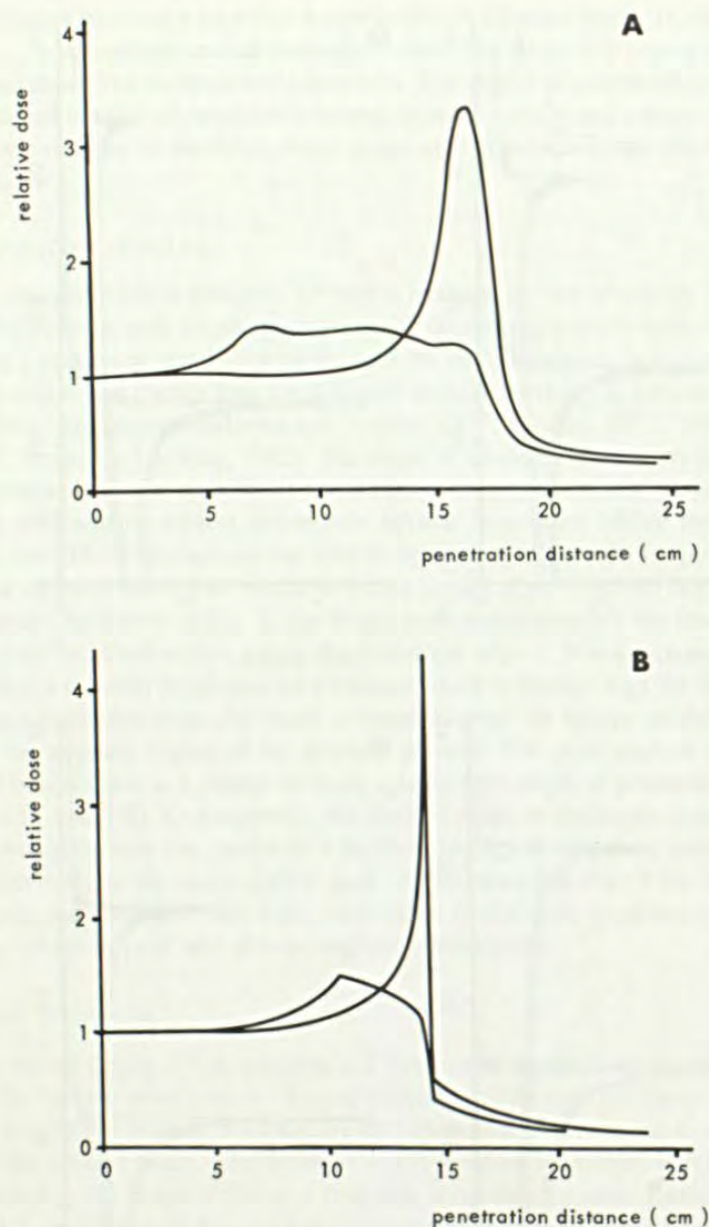


Fig. 3. Bragg curves of an 80-MeV pion beam (A) and a 400-MeV/u neon ion beam (B) in water and dose distributions of the range-modulated beams. [From Hogstrom *et al.* (1980) and Lyman and Howard (1977).]

absorbed dose in superficial layers of the object (Zoetelief *et al.*, 1978). This charged particle component can be removed by insertion of metal foils in the beam close to the surface [see, e.g., August *et al.* (1980)]. The buildup characteristics of different neutron beams have been reviewed by Mijnheer and Broerse (1979).

The central axis depth-dose distributions depend upon the field size, the source-to-surface distance (SSD), the phantom material, and the energy of the neutron beam. In Fig. 4 the penetration characteristics of three neutron beams are compared with that of ^{60}Co gamma rays. The depth-dose curves are given at the SSD values actually employed for radiotherapy with these beams. A

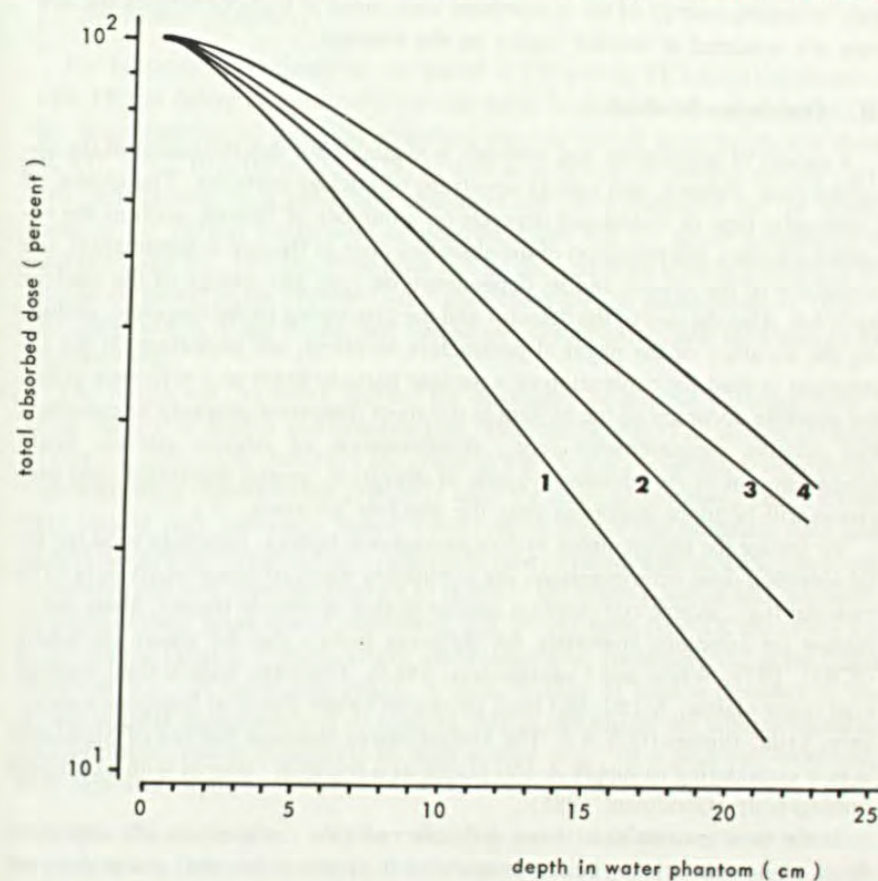


Fig. 4. Central axis depth-dose curves for ^{60}Co gamma rays and three different neutron beams in water for field sizes of approximately 10 cm $\times$ 10 cm. Curve 1 is for d(14.3) + Be (Essen, 125 cm SSD), curve 2 for d(0.25) + T (Amsterdam, 80 cm SSD), curve 3 for ^{60}Co gamma rays (80 cm SSD), and curve 4 for d(50) + Be (Louvain, 157 cm SSD). [From Mijnheer and Broerse (1979).]

summary of parameters concerning depth-dose distributions of clinically employed fast-neutron beams is given elsewhere (Mijnheer and Broerse, 1979). The relative photon contribution to the total absorbed dose in a phantom results from gamma rays produced in the target, in the collimator, and, most importantly, in the object itself. It increases with field size and depth in phantom.

With regard to the transverse absorbed dose distribution for collimated neutron beams, the dose levels outside the primary beam are higher than those obtained with teletherapy photon beams at the same depth. Relatively large penumbras are observed when neutron sources with large target dimensions are employed (see Section VI). For cyclotrons the sharpness of the beam improves with increasing energy of the accelerated ions, since at higher energies the neutrons are scattered at smaller angles on the average.

III. Dosimetry Methods

A variety of instruments and methods is available for determination of the absorbed dose, fluence, and energy spectrum of nuclear particles. The choice of a particular type of instrument depends on a number of factors, such as the required accuracy and precision of the absorbed dose or fluence determination, the sensitivity of the system and its dependence on type and energy of the nuclear particles. Also the size of the detector and the processing of the response, including the accuracy of the physical parameters involved, are important. If the instrument is used for calibration of a nuclear particle beam at a reference point, the absolute accuracy of the system is the most important property to consider. For relative measurements (e.g., determination of isodose curves, beam monitoring, or *in vivo* dosimetry) ease of operation, spatial resolution, and precision will be more important than the absolute accuracy.

To reduce the uncertainties in dose conversion factors, detectors used for total absorbed dose determinations are commonly made of tissue-equivalent (TE) material (i.e., atomic composition similar to that of muscle tissue). More information on substitute materials for different tissues can be found elsewhere (ICRU, 1977; White and Constantinou, 1982). The most widely used muscle-equivalent plastic, A-150, had been produced by the Physical Sciences Laboratory, Lisle, Illinois (U.S.A.). The United States National Bureau of Standards is now considering to supply A-150 plastic as a research material with a certified homogeneity (Goodman, 1985).

If the dose contributions from different radiation components are important (e.g., photons in neutron beams or neutrons in pi-meson beams), a twin-detector dosimetry system employing instruments with differing sensitivities for the different radiation components can be used. In some situations, the dose contributions of differing radiation components can be derived from a single instrument (e.g., a proportional counter or a high-pressure ionization chamber).

4. DOSIMETRY OF NUCLEAR PARTICLES

In this section emphasis is placed on systems applied for absorbed dose determinations in nuclear particle beams used for biomedical purposes, and on devices used for beam monitoring. The basic physical parameters employed for derivation of absorbed dose are discussed in Section V. For the assessment of radiation quality either microdosimetric instruments (e.g., proportional counters) or devices providing information on nuclear particle energy spectra (e.g., scintillation counters or activation detectors) might be employed.

A. GAS-FILLED DEVICES

1. Ionization Chambers

For biomedical applications, the use of A-150 plastic TE ionization chambers with TE gas filling is commonly considered to be the most practical method for the determination of the total absorbed dose in mixed neutron-photon fields (ICRU, 1977; Broerse *et al.*, 1978) as well as in beams of charged nuclear particles. Nonhydrogenous ionization chambers are often employed for the assessment of the photon component of neutron fields and for beam monitoring of charged particles. The absorbed dose can be derived from a measurement of the charge collected in the chamber after correction for ion recombination, density and composition of the filling gas, and shape, composition, and thickness of the chamber wall.

The ionization chambers applied for dosimetry of nuclear particles show a great variety in design and construction (Broerse, 1980). The design sketches of five TE ionization chambers are given in Fig. 5. Parallel-plate, transmission-type ionization chambers are primarily used for beam monitoring. These chambers should only introduce minor disturbances across the radiation field and should have relatively thin walls. The collection electrodes in such chambers can be subdivided into separate regions to monitor the beam position and diameter. Because of their good spatial resolution, flat, in-phantom-type TE chambers are preferable to thimble-type chambers when steep dose gradients exist in the beam direction (i.e., at the Bragg peak). A disadvantage of such a chamber is the strong angular dependence of its reading, which can hinder the interpretation of measurements under conditions of high scatter (e.g., in a phantom irradiated with fast neutrons).

a. Ion Recombination. As the collecting potential of an ionization chamber in a radiation field is increased, the current (or charge) increases until it approaches the saturation value for the given radiation intensity. The saturation current (or charge) is reached if all ions formed in the sensitive volume of the chamber are collected at the electrodes. The recombination of ions decreases with increasing collecting potential; the maximum voltage applicable is limited

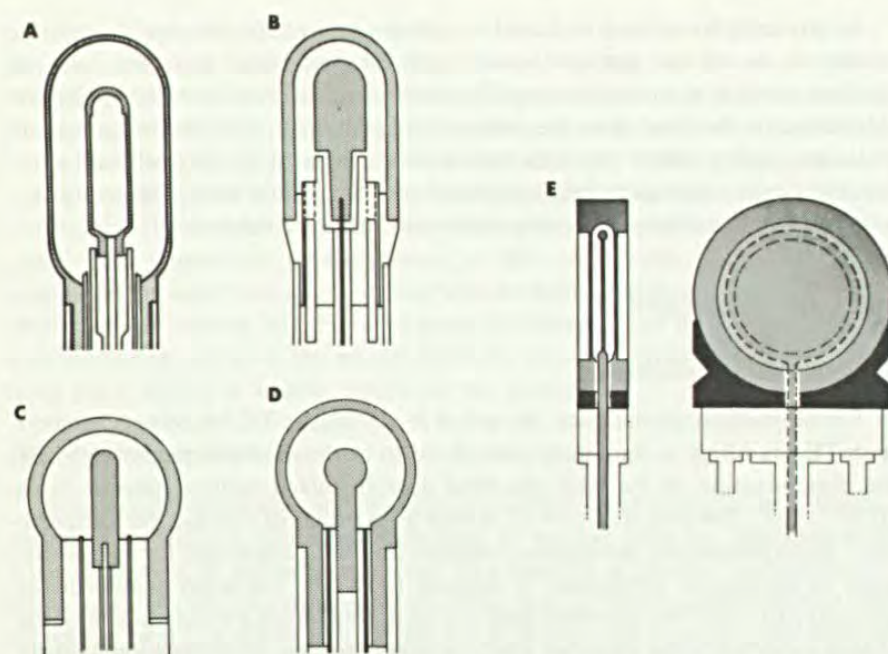


Fig. 5. Design sketches of five tissue-equivalent ionization chambers manufactured by (A) Centre d'Etude Nucléaires, Fontenay-aux-Roses, France; (B) Exradin, Warrenville, Illinois, U.S.A.; (C) Far West Technology, Goleta, California, U.S.A.; (D) Radiobiological Institute TNO, Rijswijk, The Netherlands; and (E) Columbia University, New York, U.S.A. [From Broerse (1980).]

by electron multiplication or by excessive leakage across the insulating surfaces. Ion recombination can be either initial (columnar) or general (volume). Columnar recombination involves positive and negative charge carriers formed along the track of a single charged particle. Since the probability of this type of recombination depends upon the density of the ions formed along a single track, it will be independent of the dose rate and dependent on LET. Volume recombination occurs between charge carriers after thermal diffusion and ionic drift have destroyed the initial track structure (Boag, 1966) and will be dependent on dose rate. These two types of recombination can, to a first approximation, be described as

$$i_{\text{sat}}/i(V) = 1 + v/V \quad (\text{columnar}) \quad (1)$$

$$= 1 + w/V^2 \quad (\text{volume}) \quad (2)$$

where i_{sat} is the saturation current, $i(V)$ is the measured current at the collecting voltage V , and v and w are constants. These relations are valid only for field strengths close to saturation. More detailed information concerning general (vol-

ume) recombination for high dose rates and also for pulsed radiation can be found elsewhere (Boag and Currant, 1980). An experimental and theoretical treatment of columnar recombination (Jaffé theory) is available for alpha-particles [see, e.g., Boag (1966)]. To arrive at the saturation charge, the Jaffé function $f(x)$ should be plotted against the collection field strength. However, the information required to calculate $f(x)$ is not available for TE gas, nor for all charged nuclear particles and all secondaries produced by neutrons. Experimental data on initial recombination at higher gas pressures can be found elsewhere (Zielczynski *et al.*, 1981; Zoetelief *et al.*, 1981a, 1985).

b. Chamber Wall. The reading of a chamber is determined by the ionization produced by charged particles in the gas-filled cavity. In the case of uncharged nuclear particles, secondary charged particles are created in the wall, the central electrode (usually the same material as the wall), and the gas. The relative contributions from wall material and gas depend on the primary particle spectrum. Starting with zero wall thickness, an increase in the reading with increasing wall thickness is observed due to the buildup of secondary charged particles [see, e.g., Mijnheer *et al.* (1978)]. The wall must be made thick enough to establish secondary charged particle equilibrium if the reading is to be unequivocally related to the primary particle fluence.

The ionization chamber wall causes attenuation of the primary radiation, which is partly compensated by a dose increase due to a contribution from scattered primary particles in the wall which otherwise would not have reached the sensitive volume. The resulting corrections are generally established experimentally by measurements at different wall thickness [for neutrons, see, e.g., Schraube *et al.* (1980)].

Absorbed doses are often specified in a reference material, for which, most commonly, muscle tissue with the composition recommended by the ICRU (1964) is used. To facilitate the derivation of tissue dose, ionization chambers are usually constructed from materials with an atomic composition closely matching that of the tissue. The most commonly employed TE plastic, A-150, has the correct hydrogen content but too high a carbon and too low an oxygen content (see Table II). The high carbon content is intended to make the A-150 plastic electrically conducting. In radiation fields where capture processes are important (e.g., stopping pions and low-energy neutrons) this mismatch results in conversion factors appreciably different from unity for use in deriving the dose in ICRU muscle tissue from that in the wall material. Nonhydrogenous chambers (e.g., Mg, Al, and C) have a reduced fast-neutron sensitivity and can, therefore, be used in the paired dosimetry method (ICRU, 1977) to determine the gamma-ray absorbed dose component of a neutron beam.

TABLE II

ELEMENTAL COMPOSITION (WEIGHT PERCENT) OF
DIFFERENT TE MATERIALS AND ICRU MUSCLE TISSUE

Element	ICRU muscle	A-150 plastic	A-150 gas	Methane TE gas
H	10.2	10.2	10.2	10.2
C	12.3	76.8	76.8	45.6
O	72.9	5.9	5.9	40.7
N	3.5	3.6	3.6	3.5
Ca	0.007	1.8	—	—
F	—	1.7	1.7	—
Ar	—	—	1.8	—
Other	1.1	—	—	—

c. *Filling Gases.* The chamber reading is dependent on the density and the composition of the cavity gas, which is often slowly flowed through. In general, the pressure and temperature of the gas are taken to be the same as the ambient atmospheric conditions for slow enough flow rates. The independence of ionization chamber reading on flow rate under actual measuring conditions should be verified. Often a conversion of the reading is made to a reference pressure and temperature, which is, to a first approximation, governed by the ideal gas law. For high pressures, i.e., in excess of about 5 bars (0.5 MPa), large deviations from a linear pressure dependence of the reading can occur due to a considerable contribution of initial recombination [see, e.g., Boag (1966), Zielczynski *et al.* (1981), and Zoetelief *et al.* (1981a, 1985)].

The response of ionization chambers depends on the nature of the filling gas. Depending on the application, different gases (e.g., various TE gases, air, N₂, CO₂, Ar, and Ne) are adopted. To keep the gas-to-wall absorbed dose conversion factor (see Section V) close to unity and to fulfill the Fano condition (Fano, 1954) for application of the cavity theorem, a gas is preferably selected to have an atomic composition which closely matches that of the chamber wall. Methane-based TE gas is widely used in dosimetry (Rossi and Failla, 1956). However, this gas does not match the atomic composition of either ICRU muscle tissue or A-150 plastic. There are several ways to establish a gas mixture with an atomic composition similar to that of A-150 plastic (Awschalom and Attix, 1980). Propane-based TE gas (Rossi and Colvett, 1972) can be preferable when ion recombination has to be limited, as for multiplication ionization chambers. Air-filled ionization chambers should not be used for total absorbed dose determinations in neutron beams with considerable contributions of low-energy neutrons (Schraube *et al.*, 1980). Nitrogen (a nonelectronegative gas) is frequently used in ionization chambers to monitor charged nuclear particle beams, owing to the relatively small ion recombination corrections to be made. Carbon dioxide

and argon are commonly used in nonhydrogenous chambers. The porosity of graphite chamber walls causes dilution of the flow gas (CO₂) by air unless the chamber is made airtight (Maier *et al.*, 1975; Pearson *et al.*, 1980).

2. Proportional Counters

Proportional counters (Rossi, 1968) are the most commonly employed instruments for microdosimetric measurements (see Section IV). In addition, these detectors can be used for total absorbed dose determinations as well as for the separate determination of dose components. An example of this latter feature is the separate determination of neutron and photon absorbed doses D_N and D_G , respectively, in a mixed neutron-gamma field [see, e.g., Maier *et al.* (1975)].

The principle of operation of proportional counters is similar to that of ionization chambers, i.e., under influence of an applied electrical field, ions formed in the gas cavity will drift to the electrodes. In a proportional counter, however, before collection, the number of electrons produced is increased due to electron multiplication. The effective detector "size" (e.g., expressed in grams per square centimeter across a diameter) can be made much smaller than that of ionization chambers by operation at relatively low gas pressures. The detectors most commonly employed are homogeneous and constructed from tissue-equivalent materials. Consequently, according to Fano's (1954) theorem, the fluence density spectrum and the spectrum of ionizing particles are (ideally) the same for wall and cavity, which allows the determination of microdosimetric quantities in solid media through the measurement of ionization in gases. The determination of energy imparted with a proportional counter requires knowledge of the relation between energy deposited and number of ion pairs measured. Multiwire proportional counters (constructed by stacking planes of parallel wires) are applied for monitoring charged nuclear particle beam profiles. More detailed information on proportional counters can be found elsewhere [see, e.g., Rossi (1968) and Menzel *et al.* (1978)].

3. Geiger-Müller Counters

The use of small Geiger-Müller (GM) counters as photon dosimeters in mixed neutron-photon fields was originally suggested by Wagner and Hurst (1961). A suitable lead + tin shield has to be placed around the counter to make the response to photons less energy dependent. The relative neutron sensitivity of the counters (per gray in tissue) varies between 0.1 and 3% over the neutron energy range from 0.1 to 20 MeV (Guldbakke *et al.*, 1980; Mijneer *et al.*, 1982). In view of their high thermal neutron sensitivity the GM counters should be shielded by a thermal-neutron absorber that does not emit prompt gamma radiation in a neutron capture process. A commonly chosen material is ⁶Li in the form of the metal or ⁶LiF powder. A disadvantage of GM counters can be their high sensitivity. Owing to their dead time of approximately 20 μ s, accurate

results can only be obtained in fields with photon absorbed dose rates less than about 2 mGy/min.

B. CALORIMETERS

Commonly, absorbed dose determinations with ionization chambers in nuclear particle beams show larger inaccuracies than is the case for photon beams. The inaccuracy is predominantly related to a lack of knowledge of the basic physical parameters involved in the derivation of absorbed dose from the ionization chamber reading. Since eventually almost all energy dissipated in matter appears as heat, calorimetry is in principle an absolute method for the determination of total absorbed dose, requiring thermal, electrical, and mechanical measurements only. However, the method is too laborious (and in some instances too insensitive) for routine use, and should serve rather as a reference system for calibration purposes. The absolute accuracy of calorimetry is generally limited by the uncertainty in the thermal defect, which is defined as the fraction of the energy absorbed from the beam that does not appear as heat, but is dissipated in, e.g., lattice defects or chemical changes. This is usually a few percent or less.

For photon dosimetry, highly purified graphite is usually employed as the material of choice for the construction of calorimeters. The uncertainty in the determination of absorbed dose in photon beams at a point in carbon is estimated to be smaller than 0.5% [see, e.g., ICRU (1969a)]. The conversion of absorbed dose in graphite to absorbed dose in water will result in a larger overall uncertainty of about 2%. Since the quantity of interest for biomedical applications is the absorbed dose in tissue, the composition of a calorimeter should preferably be tissue equivalent. Dosimetry with A-150 plastic calorimeters has been reported by, e.g., Bewley *et al.* (1974) and McDonald *et al.* (1981a, b). An important difference in the use of A-150 plastic instead of graphite is that both the thermal defect and the heat capacity of A-150 plastic are larger and consequently the uncertainty in the (total) absorbed dose is increased. A schematic diagram of the construction of an A-150 plastic calorimeter (McDonald *et al.*, 1976) used for dosimetry in nuclear particle beams is shown in Fig. 6. The construction of a liquid calorimeter employing a small thermistor sandwiched between two thin polyethylene films immersed in a water phantom has been reported by Domen (1980). In principle, the water in this calorimeter can be replaced by any desired liquid, although the heat conductivity and heat capacity should be taken into account as well as the thermal defect. The overall uncertainty in the determination of the absorbed dose in tissue-equivalent plastic with A-150 plastic calorimeters is about 3%. This uncertainty is reduced to about 1.5% for neutrons [see, e.g., McDonald *et al.* (1981c)] using an estimate of the thermal defect of 4% for neutrons with an uncertainty of about 1% (Goodman and McDonald, 1980). A review of radiometric calorimetry has been published elsewhere (Gunn, 1976).

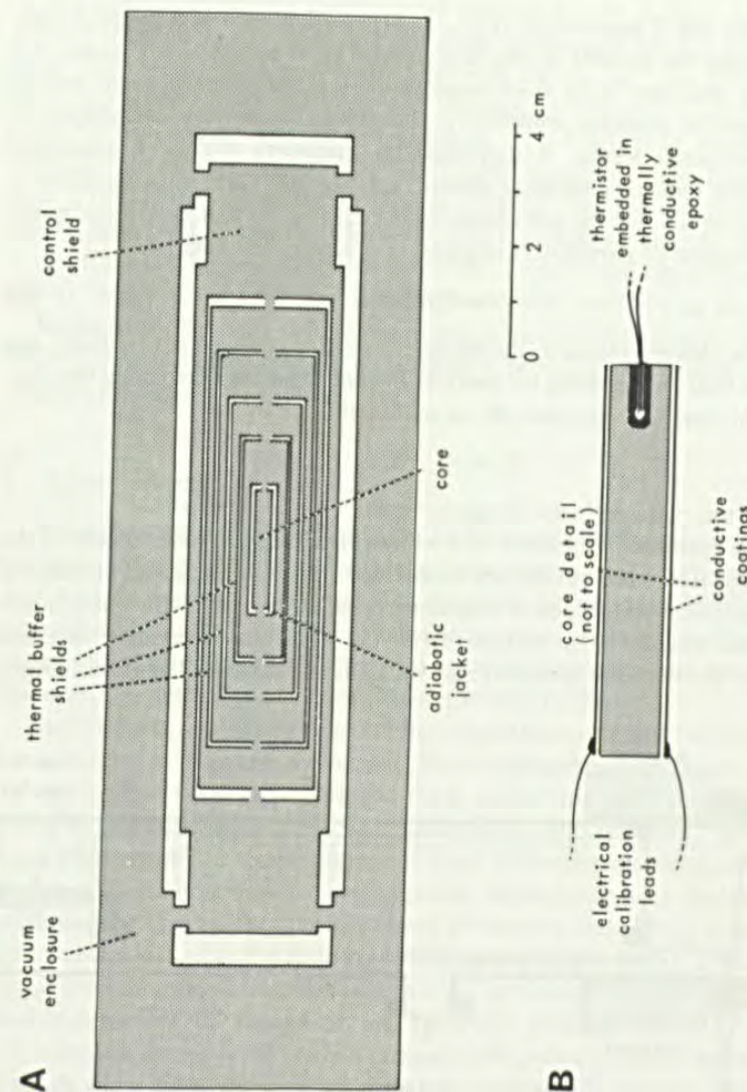


Fig. 6. Cross-sectional view of tissue-equivalent calorimeter (A) with a schematic diagram (B) of the core. [From McDonald *et al.* (1976).]

C. OTHER DETECTION TECHNIQUES

1. Fluence-Measuring Devices

The Faraday cup (Chamberlain *et al.*, 1951) is often used as a primary standard to determine the number of ions in a charged particle beam. A Faraday cup consists of an absorber block thick enough to stop the primary beam and its secondary charged particles completely. The block, generally cup-shaped to minimize backscatter losses, is supported by insulators within an evacuated chamber, which reduces ionization effects (see Fig. 7). Care must be taken in designing the cup so that the net charge collected is only that delivered by the beam. The number of particles N stopped is given by the relation

$$N = Q/(Ze) \quad (3)$$

where Q is the charge collected, Ze the average charge carried per particle, and e the electron charge. Knowing the particle fluence Φ passing through a thin layer of material, one can calculate the dose D from

$$D = \Phi \cdot S/\rho \quad (4)$$

where S/ρ is the mass collision stopping power of the material.

The charge collection efficiency of a well-designed cup is independent of the beam intensity. The true current can be derived from the measured current by introducing corrections for lack of total absorption, ionization in the gas remaining within the chamber, backscatter from the absorber, secondary electron production in the entrance window, and background current (Laughlin, 1969).

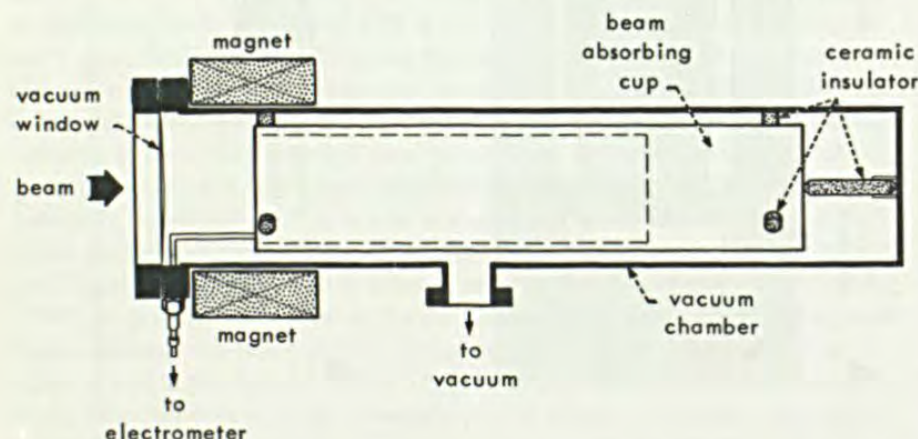


Fig. 7. Schematic diagram of a Faraday cup. The magnetic field is perpendicular to the beam to suppress the secondary electrons from the window and to prevent escape of electrons from the cup to the ground.

4. DOSIMETRY OF NUCLEAR PARTICLES

To eliminate the production of ions in the vicinity of the absorber, the Faraday cup is evacuated to about $1 \mu\text{Pa}$ ($= 10 \text{ pbar}$) (Verhey *et al.*, 1979; Pruitt, 1966; Palmieri and Goloskie, 1964).

The fluence of nuclear particles can alternatively be derived from the radioactivity induced in a suitable detector material through specific nuclear interactions. The use of several detector materials allows the crude determination of the beam composition with regard to the type and energy of nuclear particles. Cross-section values for selected reactions as a function of energy have been compiled for fast neutrons (ICRU, 1977) and charged particles [see, e.g., Smith *et al.* (1977) and Hogstrom and Amols (1980)]. Neutron-induced fission reactions can be used in analogy with activation reactions for fluence determination. A specific advantage of activation detectors is their small size (foils, pellets, and wires), which allows fluence measurements with good spatial resolution.

Methods for deriving differential fluence spectra often depend upon time-of-flight techniques [see, e.g., ICRU (1969b)] or recoil spectrometers [see, e.g., Thomas and Lewis (1981)].

2. Light-Emitting Devices

When exposed to ionizing radiation many liquid or solid substances emit visible light [see, e.g., Ramm (1966)] by a phenomenon known as scintillation. Under well-established conditions the identity of charged particles can be determined from the time dependence (decay time) of the emitted light. The use of liquid organic scintillators for fast-neutron dosimetry and spectrometry is based on the detection of recoil nuclei (ICRU, 1977).

Due to energy deposition in specific crystalline materials, electrons can become trapped at lattice imperfections. By elevating the temperature of the material the charge carriers can escape from their traps and recombine with the emission of visible light (thermoluminescence). The resulting "glow curves" (light intensity versus temperature) can have different patterns depending on the quality of the radiation under investigation. Glow-curve peak analysis has been employed for the separate determination of neutron and photon absorbed doses in mixed fields (Lucas and Kapsar, 1977; Rassow *et al.*, 1985). The characteristics of thermoluminescence dosimeters used for nuclear particle dosimetry have been summarized by Kalef-Ezra and Horowitz (1982).

Lyoluminescence is the phenomenon of light emission shown by some organic solids when dissolved in water after being exposed to ionizing radiation. The intensity of the light output is a measure of the absorbed dose in the material. The lyoluminescent materials include saccharides and amino acids, which have the advantage of compositions similar to that of soft tissue. Extensive information on the principles and applications of this technique has become available (Ettinger and Puite, 1982; Puite and Ettinger, 1982).

3. Semiconductor Detectors

Semiconductor detectors are solid-state analogs of ionization chambers, the charge carriers being the electrons and holes. The use of a solid as detector material is attractive because the sensitive layer can be very thin while still absorbing enough energy to give a good signal. The sensitivity is further enhanced because of the small amount of energy required to produce an electron-hole pair: ~ 3.75 eV in silicon (Goulding, 1965). This leads to smaller statistical fluctuations in the number of electron-hole pairs and hence improved energy resolution over that of gas-filled and scintillation counters having comparable energy absorption. An attractive feature of these detectors is that their response is proportional to the deposited energy, independent of the type of particle. Semiconductor detectors are used in measuring energy and energy losses of heavy charged particles in a spatially defined region. In regard to applications for neutron dosimetry, one should be aware of the dependence of the sensitivity on neutron energy (ICRU, 1977).

4. Chemical Dosimeters

Well-characterized and quantitative chemical changes in aqueous solutions can be employed for the assessment of absorbed dose. These chemical dosimeters have the advantage that they closely approximate the density and composition of biological tissue. When enclosed in a container simulating the shape of the biological volume of interest, the reading of the dosimeter provides information about the average absorbed dose over the specimen. Since chemical dosimeters can easily be dispatched through the mail and the absorbed dose information remains stored in a reproducible way, they have been used for dosimetry intercomparisons.

The most widely used chemical dosimeter is the ferrous sulfate system (Fricke and Hart, 1966), which consists of an aqueous solution of ferrous ammonium sulfate, sodium chloride, and sulfuric acid. Oxidation of ferrous into ferric ions is proportional to the mean absorbed dose in the sample. The change in optical density as a result of this chemical reaction can be measured with a spectrophotometer. Dosimeter sensitivity is quantitatively characterized by the G value, i.e., the number of ferric ions formed per 100 eV energy absorbed by the solution. The G value is dependent on the LET of the radiation, being approximately 15 for photons (Fricke and Hart, 1966) and about 9 for neutrons (ICRU, 1977). A major disadvantage of the Fricke dosimeter is that relatively high doses (40 to 400 Gy) are needed to achieve reliable dosimetry. However, a modified ferrous sulfate solution as proposed by Gupta (1973) with higher sensitivity can be used in the dose range of 0.1 to 30 Gy [see, e.g., Maughan *et al.* (1983)]. Among other useful chemical dosimeters are ceric sulfate solutions (Matthews, 1982).

4. DOSIMETRY OF NUCLEAR PARTICLES

5. Photographic Emulsions and Nuclear Track Detectors

Photographic emulsions can be used to give a qualitative picture of the spatial nuclear particle distribution (field mapping), to take nuclear particle radiographs (Tobias *et al.*, 1977; Bewley *et al.*, 1973), to determine the dose of charged particles, and to measure the absorbed dose of gamma rays which contaminate nuclear particle beams. General information concerning dosimetry with photographic emulsions can be found elsewhere (Dudley, 1966). For neutron dosimetry, photographic methods are applied by counting tracks of recoil protons in the emulsion. The threshold neutron energy for track registration by direct interaction is between 0.1 and 1 MeV depending on the characteristics of the material. For fast neutrons, the sensitivity of photographic materials is on the order of a few percent relative to that for photons (ICRU, 1977). The neutron sensitivity can be increased by adding a proton radiator (e.g., a sheet of polyethylene) on the side facing the neutron source, thus taking advantage of the large elastic scattering cross section of hydrogen.

Charged particles produce permanent tracks of radiation damage in insulating materials. These can be detected with an optical microscope after treating the irradiated glass or polymer with an etching agent which preferentially attacks the tracks. The number of recorded tracks per unit area of detector surface is related to the nuclear particle fluence. Each type of detector material has a minimum energy-deposition threshold for track detection (Price and Fleischer, 1971). Neutrons produce secondary charged particles in the dielectric material either by direct interaction or indirectly by means of reactions in a radiator or converter. A disadvantage of solid-state track detectors is the dependence of the response on the angle of incidence of the nuclear particles, which limits their usefulness for biomedical applications (ICRU, 1977).

D. BEAM MONITORING

The production of nuclear particle beams generally involves the use of accelerators. These complex machines and associated beam-handling systems cannot be expected to operate over long time periods at an output sufficiently constant for biomedical purposes. Therefore, continuous monitoring of the beam intensity is essential. It is preferable that a beam be monitored by more than one detector, since the ratio of the readings will enable easier identification of unsatisfactory performance. Monitors which are capable of detecting changes in the beam position and composition are preferred, but if this is not achieved, additional equipment is required to monitor beam position. Linearity and dose-rate independence are essential, as well as a constant relationship between monitor response and absorbed dose at a reference point in a phantom for a fixed beam size.

Specific instruments used for beam monitoring are parallel-plate ionization chambers, fission counters, Geiger-Müller counters, and secondary-emission monitors (SEM). The secondary-emission monitor is particularly useful in high-intensity radiation fields of charged particles, where ionization chambers cannot be used because of incomplete ion collection due to recombination (Tautfest and Fechter, 1955). In an SEM the charge measured is due to the transfer of low-energy secondary electrons between the high-voltage electrode and the collection electrode in ultrahigh vacuum ($\sim 1 \mu\text{Pa}$). With this type of monitor one avoids the problems of ion recombination and dependence of response upon temperature and pressure. The SEM is commonly calibrated against an ionization chamber at a low beam intensity.

IV. Radiation Quality

To predict the responses of biological objects subjected to ionizing radiation it is essential to determine the energy dissipation at a macroscopic level. This can be achieved by a specification of absorbed dose, its temporal distribution, and its variation from point to point in the medium. In addition, information should be provided on the microscopic distribution of the energy deposition (i.e., the radiation quality) since the effectiveness for producing a certain frequency or incidence of biological effect depends on this distribution as well. Attempts to account for radiation quality led first to the concept of linear energy transfer (LET) and its distributions, and more recently to the microdosimetric stochastic quantities lineal energy y , specific energy imparted z , and their distributions [see, e.g., Rossi (1968) and ICRU (1980)].

Tissue-equivalent proportional counters are commonly used to get a measure of the radiation quality. To obtain a single parameter characterizing radiation quality, mean values of y have been introduced, based on the frequency distribution $\bar{y}_F$ or the distribution of absorbed dose $\bar{y}_D$. Kellerer and Rossi (1972) introduced the saturation-corrected, dose-averaged lineal energy y^* . This was based on the experimental finding that an event of given y becomes less effective in producing a certain biological effect per unit dose for high y values.

If the radiation quality does not vary considerably with position in a phantom, significant variations in relative biological effectiveness (RBE) in the irradiated object are not to be expected. Large variations in microscopic energy-deposition spectra indicate the need for radiobiological experiments to assess the changes in effectiveness of the radiation field at different positions.

Radiation quality can also be related to the energy spectra of the nuclear particle beam. Neutron energy spectra as a function of depth in a phantom have been determined experimentally [e.g., with organic scintillators (Schmidt *et al.*,

1980) or activation and fission detectors (Mijnheer *et al.*, 1981a)] and theoretically (Burger and Morhart, 1979). The results obtained for different neutron beams indicate an increased contribution of lower-energy neutrons with increasing depth in the phantom. As a result of the interactions of the neutrons with the phantom material (water or tissue equivalent), the relative contribution of gamma rays to the total absorbed dose will increase appreciably with increasing depth. Microdosimetric measurements performed for different collimated neutron beams have shown that the lineal energy spectra do not change considerably with depth in the center of the beam [see, e.g., Fidorra and Booz (1981) and Hogeweg *et al.* (1978)] but that appreciable variations occur with lateral distance from the beam axis outside the primary field. Microdosimetric spectra measured at four depths in a water phantom irradiated with $d(14) + \text{Be}$ neutrons are shown in Fig. 8. The biological consequences of these findings are discussed in Section VII.

A beam of monoenergetic heavy charged particles has a distinct radiation quality. As such particles are slowed down in a medium, the average lineal energy of the primary beam increases and its distribution broadens due to an in-

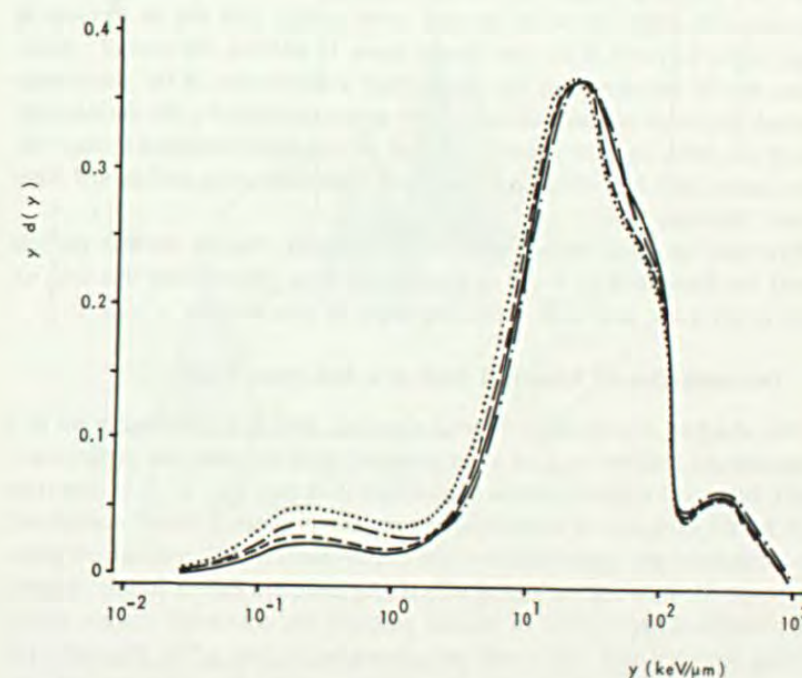


Fig. 8. Microdosimetric spectra measured at four depths in a water phantom irradiated with $d(14) + \text{Be}$ neutrons, (—)5, (---)10, (- · -)15, and (· · ·)20 cm. [From Fidorra and Booz (1981). Copyright by The Institute of Physics.]

creased rate of energy loss and an increase in the energy spread of the beam. In addition, some particles have interactions with nuclei in the absorbing medium. If the primary particles contain multiple nucleons, these interactions generally result in fragmentation of the incident particle. The amount of fragmentation depends upon the nature of the stopping medium and the initial energy and nuclear charge of the primary charged particle. Fragments will have a smaller nuclear charge and, therefore, a smaller rate of energy loss than that of the primary ion. Since the rate of energy loss varies approximately as the square of the nuclear charge, this will result in a very large range in the size of the energy-loss events. The importance of the lower energy-loss tail to the y distribution will depend primarily upon how much fragmentation of the primary beam occurs.

The interaction processes of pions with matter differ from those of heavier charged particles. Some pions are lost from the beam by decay to a mu meson. Pions that do not decay will undergo nuclear capture at the end of their range. The capturing nucleus will then fragment due to the excess energy. Charged fragments will have short ranges and a much higher y than that of the incoming pion. A pion beam, therefore, will have a narrow, low y distribution when it enters the stopping medium. This distribution shifts toward higher y and broadens with depth due to an increase in the energy loss and an increase in straggling of the pions as they are slowed down. In addition, the overall y distribution will be augmented by the broad, high y distribution of the short-range charged fragments and neutrons which are produced following the nuclear capture of the pi meson. The number and type of fragments produced through the pion capture vary depending upon the nature of the capturing nucleus and other factors (Richman, 1981).

Examples of lineal energy spectra for different charged nuclear particle beams are shown in Fig. 9. All of these beams have distributions that shift toward larger event sizes with increasing depth of penetration.

V. Determination of Absorbed Dose at a Reference Point

The absolute determination of total absorbed dose at a reference point in a homogeneous phantom (e.g., a water phantom) is an essential step in the procedures followed for specifications of absorbed dose (see Fig. 1). It is important both for the exchange of biomedical information between different centers and for comparison and interpretation of the effectiveness of different nuclear particle beams for inducing biological effects. As indicated earlier in this chapter, for biomedical applications of nuclear particles the dosimetric system should provide absorbed dose values with an accuracy better than $\pm 5\%$. Preferably the dosimetry system should be optimally sensitive, relatively easy to operate, and suitable for all types of nuclear particles. These requirements reduce the number of dosimetric techniques available for this purpose (see Section III).

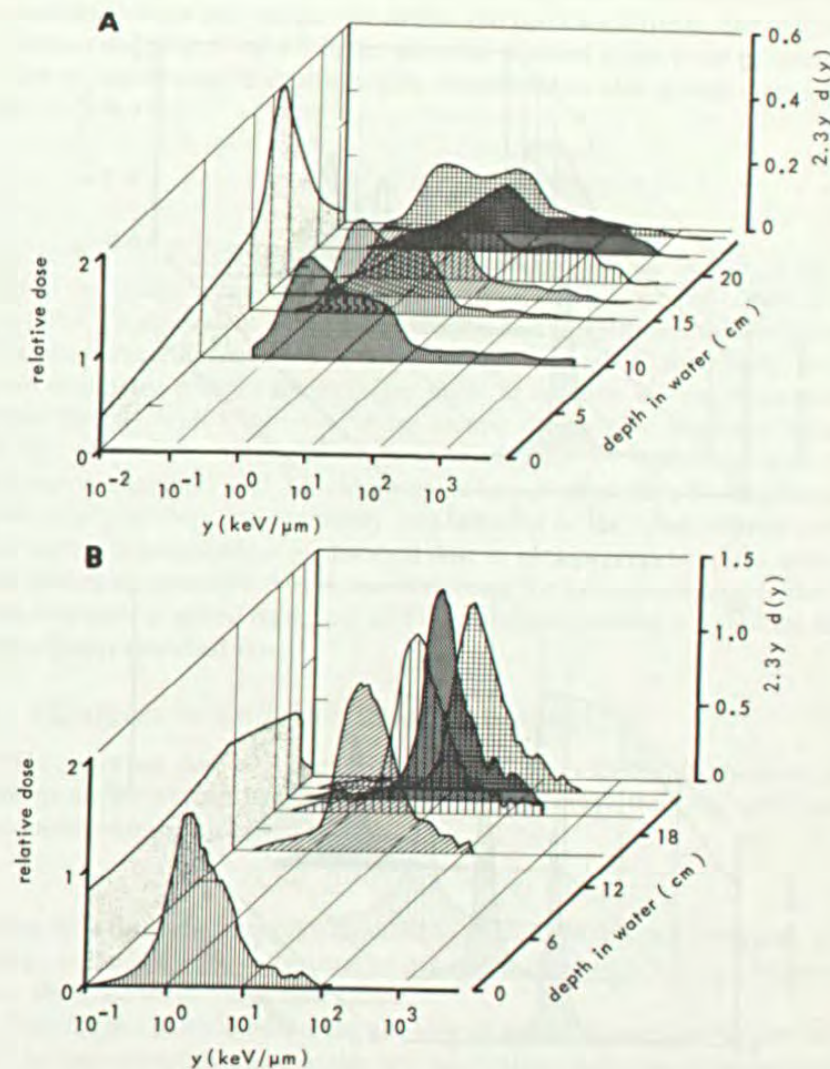


Fig. 9. Microdosimetric spectra for 230-MeV/u helium (A), 400-MeV/u carbon (B), 400-MeV/u neon (C), and 500-MeV/u argon ions (D). Spectra are shown for different positions along the central axis for which the depth-dose distribution is shown on the left of each panel. The beams shown in B, C, and D were range-modulated. [Redrawn from Luxton *et al.* (1979)].

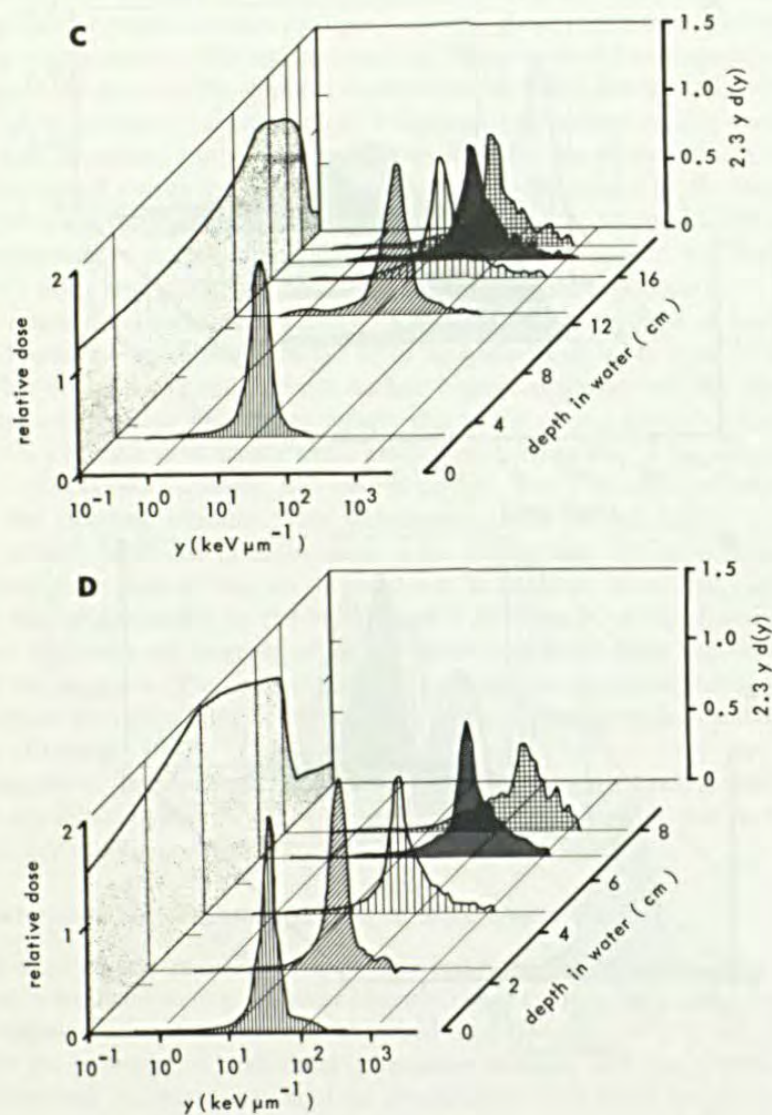


Fig. 9. (Continued)

In principle, the absorbed dose can be derived when information is available about the fluence and energies of all charged particles involved and their linear collision stopping powers S in the absorber material at the point of interest. If delta-ray equilibrium is established the absorbed dose D in a small mass is given by

$$D = \sum_i \int \Phi_i(E) S_i(E) \rho^{-1} dE \quad (5)$$

where $\Phi_i(E)$ is the energy spectrum of the charged particle fluence, ρ the density of the medium, and i an index to sum over the different types of contributing particles. In general, it is too laborious for routine purposes to determine the charged particle fluence as a function of energy in detail. Consequently, in common dosimetry practice attempts are made to measure the energy deposition more directly with a detector whose atomic composition simulates tissue as closely as possible. Examples of these instruments are tissue-equivalent (TE) ionization chambers and TE calorimeters (see Section III). Tissue-equivalent ionization chambers are commonly considered to be the most practical method for routine determinations of absorbed dose in nuclear particle beams, although the overall uncertainty is larger than is the case for TE calorimeters. In this section emphasis is placed on the use of TE ionization chambers for absolute determination of absorbed dose.

A. PRINCIPLES OF IONIZATION CHAMBER DOSIMETRY

The absorbed dose D_m in the wall material of an ionization chamber adjacent to the cavity can be derived, using cavity theory, from the total charge produced within the cavity Q :

$$D_m = Q \cdot (1/m) \cdot (W/e) \cdot r_{m,g} \quad (6)$$

where W is the average energy required to produce an ion pair in the gas, e the charge of the electron, m the mass of gas within the cavity, and $r_{m,g}$ the gas-to-wall absorbed dose conversion factor.

For charged particle beams the gas-to-wall absorbed dose conversion factor will be approximately equal to the ratio of the mass collision stopping powers in the wall and the gas of the detector. Special care is needed if the stopping power of the charged nuclear particles changes due to energy losses as the particles traverse the detector. For photon irradiation the secondary charged particle fluence will consist of electrons, while for neutrons it will, in general, consist of electrons, protons, helium nuclei, and heavier recoils such as carbon, oxygen, and nitrogen nuclei (Caswell and Coyne, 1972). If the charged particles produced by photons or neutrons within the wall of the chamber lose a negligible part of their energy in crossing the cavity, then the conditions for the

Bragg-Gray theorem are satisfied and $r_{m,g}$ is identical to $s_{m,g}$, the effective ratio of the mass collision stopping powers in the wall to that in the gas for the secondary charged particles produced in the material adjacent to the cavity. For a cavity whose size is not negligible in relation to the range of the secondary charged particles, it is necessary to make a more detailed calculation for the value of $r_{m,g}$ as a function of cavity size and neutron energy (Makarewicz and Pszona, 1978; Rubach and Bichsel, 1982; Siebert and Coyne, 1984).

The mass of the gas in the cavity can be obtained from a calibration of the chamber in a uniform photon field (e.g., a ^{60}Co gamma-ray beam) whose calibration is directly traceable to a national standards laboratory. According to various dosimetry protocols (ECNEU, 1981; AAPM, 1980) the mass m of the gas in the cavity is equal to

$$m = \frac{W_c}{e} (s_{m,g})_c \left[\frac{(\mu_{en}/\rho)_t}{(\mu_{en}/\rho)_m} \right]_c \cdot \frac{1}{N_c \cdot f_{t,c} \cdot \prod k_A} \quad (7)$$

where the subscripts c refer to the photon calibration field, μ_{en}/ρ are the mass energy absorption coefficients in tissue (subscript t) and wall material (subscript m), N_c is the photon exposure calibration factor, and $f_{t,c}$ is the exposure-to-absorbed-dose conversion factor in tissue. The factor $\prod k_A$ accounts for the attenuation and scattering of primary radiation by the wall, central electrode, and buildup cap of the chamber and also for the radiation scattered by the stem of the chamber into the sensitive part of the chamber. In addition, corrections should be made if there is any radial or axial nonuniformity in the field.

When the chamber is placed in a nuclear particle beam the charge produced within the cavity at a reference temperature and pressure can be derived from the reading R of the chamber and several conversion and correction factors k_R by

$$Q = R \cdot \prod k_R \quad (8)$$

The factors in $\prod k_R$ include the electrometer calibration factor and correction factors for ion recombination, temperature and pressure, gas flow rate, and leakage current.

For uncharged ionizing particles, the absorbed dose in a reference tissue adjacent to the cavity of the chamber D_t can be calculated from D_m using the ratio of mass energy absorption coefficients in the tissue t and wall material m assuming that there is charged particle equilibrium:

$$D_t = \frac{(\mu_{en}/\rho)_t}{(\mu_{en}/\rho)_m} \cdot D_m = \frac{K_t}{K_m} \cdot D_m \quad (9a)$$

For neutron irradiations, the ratio of the mass energy absorption coefficients can be replaced by the ratio of kerma factors K (ICRU, 1977).

For charged particles, the ratio of D_t to D_m is equal to the ratio of the mass

collision stopping power in tissue and wall material:

$$D_t = \frac{(S/\rho)_{col,t}}{(S/\rho)_{col,m}} \cdot D_m \quad (9b)$$

Equations (9a) and (9b) give the absorbed dose in tissue adjacent to the cavity whereas, for most biomedical measurements inside a phantom, the required result is the absorbed dose at the position of the (effective) center of the chamber in the unperturbed phantom. Owing to the displacement of phantom material by the cavity of an ionization chamber, the effective point of measurement can be shifted with reference to the calibration situation. For indirectly ionizing radiation, a correction for the lack of attenuation and scattering in the cavity of the chamber has to be introduced. The displacement factor k_d is defined as the ratio of the absorbed dose for an infinitely small cavity to that which is actually measured with an ionization chamber of finite dimensions taking the center of the chamber as the point of measurement [see, e.g., Zoetelief *et al.* (1980)]. For charged particle beams the correction is applied by an upstream shift of the point of measurement.

Combining Eqs. (6)–(9) yields the total absorbed dose in tissue due to nuclear particles:

$$D_t = R \cdot \prod k_R \cdot k_T^{-1} \cdot N_c \cdot f_{t,c} \prod k_A \cdot k_d \quad (10)$$

where

$$k_T^{-1} = \frac{r_{m,g}}{(s_{m,g})_c} \cdot \frac{W}{W_c} \cdot \frac{(\mu_{en}/\rho)_t/(\mu_{en}/\rho)_m}{[(\mu_{en}/\rho)_t/(\mu_{en}/\rho)_m]_c} \quad (11a)$$

and

$$k_T^{-1} = \frac{(S/\rho)_{col,t}/(S/\rho)_{col,g}}{(s_{m,g})_c \cdot [(\mu_{en}/\rho)_t/(\mu_{en}/\rho)_m]_c} \cdot \frac{W}{W_c} \quad (11b)$$

for uncharged and charged particles, respectively.

B. PHYSICAL PARAMETERS FOR IONIZATION CHAMBER DOSIMETRY

To achieve consistency in dosimetry of nuclear particles it is necessary to use a coherent set of basic physical parameters appropriate for the nuclear particle spectrum. Spectral measurements are necessary for the computation of the values of these basic physical parameters. Such measurements have been accomplished by many methods, e.g., semiconductor devices, time-of-flight techniques, proton-recoil detectors (in case of neutron radiation), and activation detectors. The available techniques are those developed over a period of years for use in nuclear particle research and are partly summarized in Section III of this chapter.

1. Average Energy Required to Produce an Ion Pair W

The parameter used to convert the charge produced within the cavity of an ionization chamber to energy deposited in the gas is the mean energy required to form an ion pair in the gas W . The magnitude of this parameter depends on the types and spectra of charged particles involved, and on the chemical composition of the gas. For neutrons, Goodman and Coyne (1980) presented detailed calculations of W as a function of energy, using the most recent data on W values for charged particles in methane-based TE gas. The results of their calculations relative to $W_c = 29.3$ eV for photons are shown in Fig. 10.

It is recommended that the ratio of W_N/W_c be taken from this figure using the differential neutron fluence spectra. If no spectral data are available a value of 1.06 can be taken as relevant for most fast-neutron beams. An uncertainty of $\pm 4\%$ in this ratio has been estimated by Goodman and Coyne (1980).

There are experimental data available on W values for heavy ions [see, e.g., ICRU (1979b)] in TE gas; however, these data are restricted to energies below 10 MeV. At higher energies, W has little dependence on energy since the competition between ionization and excitation is rather independent of energy and the amount of energy going into kinetic energy of gas atoms is negligible.

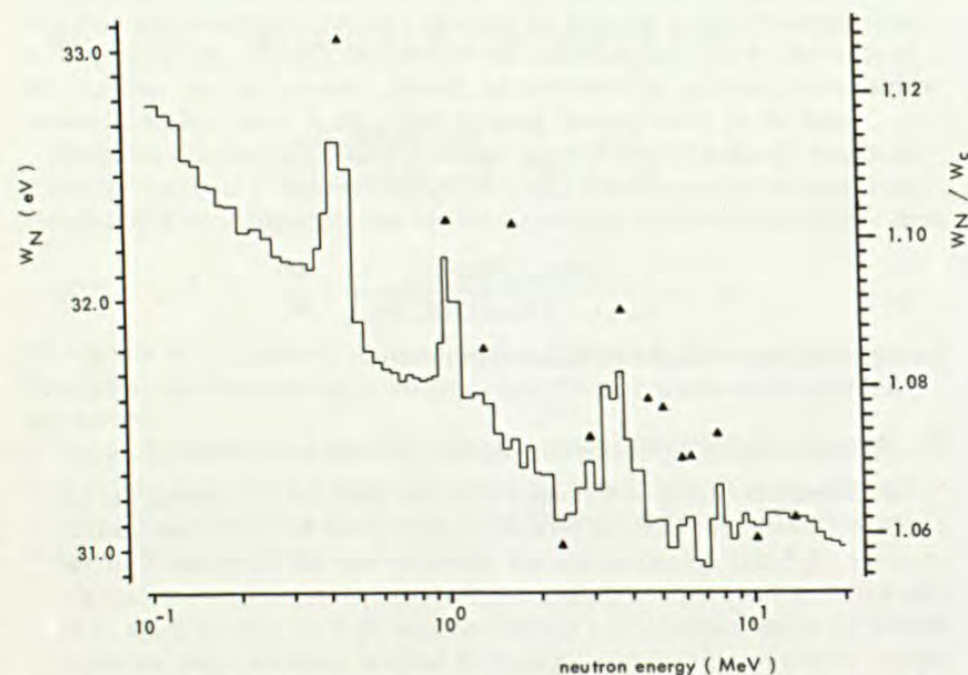


Fig. 10. Values for W_N calculated for methane-based tissue-equivalent gas by Goodman and Coyne (1980). The triangular points shown on the graph are W_N evaluated at discrete energies of prominent resonances, rather than the values averaged over energy bins shown by the solid lines.

2. Gas-to-Wall Absorbed Dose Conversion Factor $r_{m,g}$

When the Bragg-Gray conditions are fulfilled, the gas-to-wall absorbed dose conversion factor is in the simplest case equal to the ratio of the mass stopping power in the wall material relative to that in the gas, weighted by the relative abundances and energies of each type of ionizing particle in the nuclear particle beam. This is generally the case for charged nuclear particle beams. For neutrons, the gas-to-wall absorbed dose conversion factors have been compromised in their accuracy due to several basic problems: (a) adequate measurements of stopping power have not been made for the charged particles generated by fast neutrons in A-150 plastic or in TE gas; (b) the equilibrium secondary charged particle spectrum created by neutrons is not well known; and (c) the range of the low-energy heavy recoils is limited. The latter aspect results in particles classified in various categories, notably "crossers," "insiders," "starters," and "stoppers," dependent on chamber cavity dimensions (Caswell, 1966). The dose conversion factor is therefore a function of cavity size and neutron energy. Makarewicz and Pszona (1978), Rubach and Bichsel (1982) and Siebert and Coyne (1984) have calculated $r_{m,g}$ for TE ionization chambers which approximate Bragg-Gray cavities, and for chambers of finite volume. They considered the ionization due to primary charged particles (protons, and He, C, N, and O nuclei) originating in the gas relative to the total ionization; they also accounted for the possible differences in the stopping powers for solids and gases. The calculations of Rubach and Bichsel (1982), performed for discrete energies, show large variations in $r_{m,g}$ as a function of neutron energy. However, the authors quote an uncertainty of $\pm 4.5\%$ for a chamber of practical dimensions. Incorporation of these new values into existing protocols (AAPM, 1980; ECNEU, 1981) is to be expected only when more accurate data averaged over finite energy intervals become available.

3. Derivation of Absorbed Dose in Tissue from That in the Wall Material

The kerma or absorbed dose in the wall material has to be converted into kerma or absorbed dose in soft tissue, the relevant material for medical and biological applications. As can be seen from Table II, A-150 plastic (the TE plastic commonly employed as wall material of the chamber) differs from tissue in its atomic composition mainly in the carbon and oxygen content. For directly ionizing radiation, the conversion factor involves the ratio of the stopping powers in tissue to that in A-150 plastic, whereas for neutrons the ratio of the kerma factors has to be introduced. Caswell *et al.* (1980) prepared an updated version of the kerma tables included in ICRU Report 26 (ICRU, 1977). The ratios of K_t/K_m in the energy range of 0.1 to 30 MeV according to their calculations are shown in Fig. 11. For higher neutron energies there is a lack of reliable cross-section data. However, further efforts have been made to derive kerma values for this

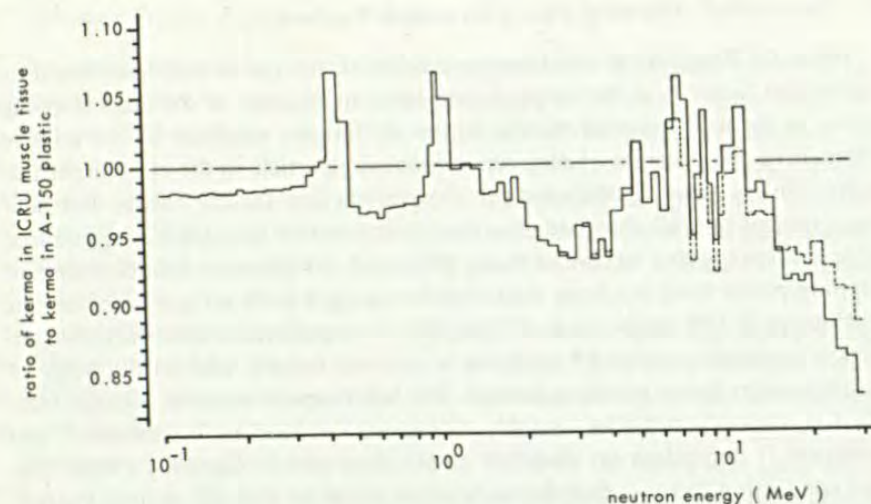


Fig. 11. Ratio of kerma in ICRU muscle tissue to kerma in A-150 plastic, (solid line) Caswell *et al.* (1980) and (dashed line) ICRU Report 26 (1977).

energy range [see, e.g., Dimbylow (1980)]. Uncertainties in the soft tissue/A-150 plastic kerma ratio increase with rising neutron energy and vary between 2.6% for a d(16) + Be spectrum to 9.7% for a p(66) + Be spectrum (Bewley, 1980).

The kerma ratios are obviously energy dependent and attempts should be made to obtain information concerning the radiation energy spectrum at the reference point. Changes in the neutron energy spectrum as a beam passes through a phantom or patient will probably not affect the kerma ratio much (Bewley, 1980).

A summary of quantitative values of W , $r_{m,g}$, and kerma ratios presently applied for three types of nuclear particles is given in Table III. The overall uncertainty in the determination of the total absorbed dose in tissue in a neutron field by use of a TE ionization chamber is given in Table IV. The uncertainties in the physical parameters are estimates for neutron beams with average energies higher than a few mega-electron-volts (ECNEU, 1981).

C. DISPLACEMENT CORRECTION FACTOR

Absorbed dose and kerma are quantities which are defined at points (ICRU, 1980), while the ionization chambers used for dose determinations are of finite dimensions. The reading of an ionization chamber is proportional to an average absorbed dose over the sensitive volume and this average value must be related to the absorbed dose at a specific point. The correction for the finite dimensions of the detector can be achieved in two ways: either a correction factor can be

TABLE III
PHYSICAL PARAMETERS CURRENTLY RECOMMENDED FOR IONIZATION CHAMBER DOSIMETRY

Parameter and symbol	Protons	Neutrons		Pions
		AAPM (1980)	ECNEU (1981)	
Average energy per ion pair formed, W (eV)	30.2	31.1-31.3	31.1-31.3	Plateau 29.5 Peak 30.1
Gas-to-wall dose conversion factor $r_{m,g}$	0.992-0.993	1.00	1.00	
Wall material-to-tissue conversion factor	0.975-0.984	0.963 ^a	0.982 ^a	Plateau 0.98 Peak 0.92

^aFor d + T neutrons.

TABLE IV
UNCERTAINTIES IN THE DETERMINATION OF
THE TOTAL ABSORBED DOSE IN TISSUE
IN A NEUTRON FIELD

Parameter	Uncertainty (%)
R	0.2
$\prod k_R$	0.3
k_d	0.5
$N_c f_{t,c} \prod k_A$	1.2
W_N/W_c	4.0
$r_{m,g}/(s_{m,g})_c$	2.0
K_t/K_m	2.6 to 9.7 ^a
$[(\mu_{en}/\rho)_t/(\mu_{en}/\rho)_m]_c$	0.1
Overall 5.4 to 10.8 ^a	

^aAccording to Bewley (1980). The lower value is valid for a d(16) + Be beam, the higher for a p(66) + Be beam.

applied when the geometrical center of the chamber is considered to be the point of measurement, or an effective point of measurement can be adopted.

The rule of thumb suggested by Kondo and Randolph (1960) for photons, that in free space the geometrical center of a chamber is the point of measurement for distances from the source in excess of five times the chamber diameter, also appears to be valid for neutrons (Zoetelief *et al.*, 1980).

The introduction of an ionization chamber into a phantom will disturb the dose distribution in and around the cavity. Differences in scattering and attenuation of the primary particle fluence in the gas-filled cavity are illustrated in Fig. 12 (Zoetelief *et al.*, 1980). The depth-dose curve is presented for a cavity

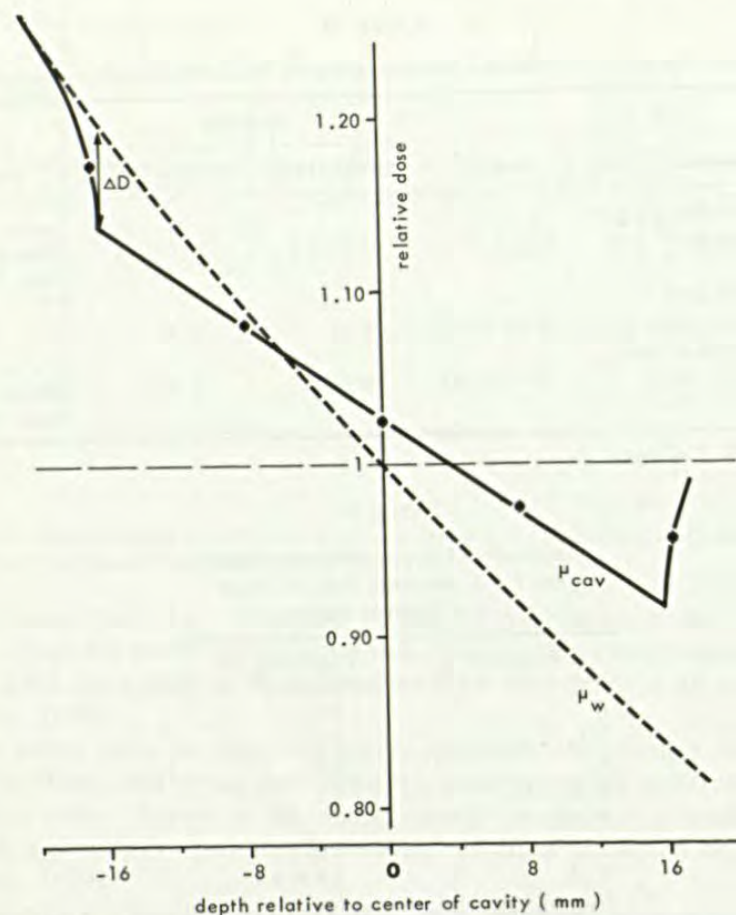


Fig. 12. Central axis depth-dose distributions in a water phantom for d + T neutrons measured in the presence (solid line) or absence (dashed line) of a Styrofoam cavity. [From Zoetelief *et al.*, (1980). Copyright by The Institute of Physics.]

simulated by a Styrofoam sphere immersed in a water phantom irradiated with 15-MeV neutrons. At the front interface of the cavity, the reduction in scattered radiation results in a sharp dose decrease. The depth-dose curve inside the cavity has a reduced slope compared to the slope in water. A buildup of the dose occurs at the rear side interface. Similar results have been obtained by Samuelsson (1977) for high-energy photons. The results presented in Fig. 12 indicate that for irradiation with 15-MeV neutrons the effective point of measurement will be displaced over some distance from the geometrical center toward the front inner surface of the cavity. The relative contributions of scatter and attenuation are different for various types of ionizing radiation and are also dependent on the energy of the photons and nuclear particles involved. For neutrons, it is preferable to use a multiplicative correction factor to account for the displacement effect

at depths beyond the dose maximum. Displacement correction factors have been inferred from measurements with ionization chambers of different dimensions by Shapiro *et al.* (1976) and Zoetelief *et al.* (1980). The latter group performed measurements with a set of three spherical TE chambers with different cavity radii for neutron and photon beams of different quality. The results summarized in Table V show that for neutrons of relatively low energy, no displacement was found, i.e., chambers of all sizes give the same ionization per unit volume when centered at the same position beyond the dose maximum. For neutrons with energies in excess of 5 MeV, the displacement correction factor k_d for a chamber of given cavity radius r , shows an almost constant value. In accordance with the results of Johansson *et al.* (1978) appreciable displacement correction factors were observed for ^{137}Cs and ^{60}Co gamma rays. A displacement correction factor of 1 was observed for orthovoltage x rays, however, for lower-energy photons (30- to 40-kV x rays), the displacement correction factor appears to be again appreciably smaller than 1. For neutron irradiations, Williams *et al.* (1982) observed results comparable to those of Zoetelief *et al.* (1980).

For heavy charged particle beams, depth-dose distributions are best measured with a thin, parallel-plate ionization chamber. The effective point of measurement is generally assumed to be the inner surface of the entry window to the sensitive volume. The effective point of measurement for other types of ionization chambers is determined by comparison with the measurements made using a parallel-plate ionization chamber. The cylindrical- or spherical-type ionization chambers should only be used with caution in regions where the dose gradients are steep (e.g., Bragg peak region) because the average dose over the sensitive volume of these chambers can then differ greatly from the dose at a point in a phantom.

TABLE V

DISPLACEMENT CORRECTION FACTORS k_d OF SPHERICAL IONIZATION CHAMBERS WITH INTERNAL RADIUS r (in mm), FOR MEASUREMENTS IN PHANTOMS WITH DIFFERENT TYPES OF RADIATION^a

Type of radiation	k_d
150-, 200-, and 300-kV x rays	$1.000 \pm 0.05 \cdot 10^{-2} \cdot r$
^{137}Cs γ rays	$1 - (0.22 \pm 0.05) \cdot 10^{-2} \cdot r$
^{60}Co γ rays	$1 - (0.37 \pm 0.04) \cdot 10^{-2} \cdot r$
Fission neutrons ($\bar{E}_N = 1$ MeV)	$1.000 \pm 0.1 \cdot 10^{-2} \cdot r$
d(2.3) + D neutrons ($E_N = 5.3$ MeV)	$1 - (0.25 \pm 0.09) \cdot 10^{-2} \cdot r$
d(0.25) + T neutrons ($E_N = 14.2$ MeV)	$1 - (0.25 \pm 0.06) \cdot 10^{-2} \cdot r$
d(0.5) + T neutrons ($E_N = 14.8$ MeV)	$1 - (0.25 \pm 0.06) \cdot 10^{-2} \cdot r$
d(50) + Be neutrons ($\bar{E}_N = 21$ MeV)	$1 - (0.21 \pm 0.05) \cdot 10^{-2} \cdot r$

^aFrom Zoetelief *et al.* (1981b).

D. COMPARISON OF DIFFERENT DOSIMETRY METHODS

For a check of the appropriateness of the basic physical parameters employed for absorbed dose determination it is very useful to compare the results of independent methods. The neutron absorbed dose values in a phantom determined with a TE ionization chamber or a TE calorimeter (D_{cal}) in combination with a neutron-insensitive detector have been compared with neutron kerma (K_{flu}) derived from fluence measurements. As indicated in Table VI, the different experimental techniques show good agreement relative to the neutron absorbed dose obtained from a tissue-equivalent ionization chamber. This implies that for neutron energies up to about 15 MeV the absolute uncertainty in the determination of the neutron absorbed dose with TE ionization chambers might be less than the values quoted in Table IV. However, for higher neutron energies absorbed doses derived from measurements with TE ionization chambers and TE calorimeters can deviate considerably from the absorbed dose in soft tissue due to uncertainties in the kerma factors.

VI. Treatment Planning for External Beam Therapy

Subsequent to the determination of the absolute absorbed dose at a reference point, information must be obtained about the relative absorbed dose distribution in a standard phantom. The materials employed for phantoms include water, tissue-equivalent liquid, polystyrene, and A-150 TE plastic. When the atomic compositions of the materials used are closely similar (especially with regard to the hydrogen content by weight), differences in depth-dose data inside different phantoms are mainly due to variation in the densities of the materials involved (ICRU, 1977). Figure 13 shows the total absorbed dose values determined inside

TABLE VI

INTERCOMPARISON OF NEUTRON ABSORBED DOSE DETERMINED WITH A TE CALORIMETER (D_{cal}) AND NEUTRON KERMA (K_{flu}) RESULTING FROM FLUENCE MEASUREMENTS RELATIVE TO RESULTS FROM TE IONIZATION CHAMBERS

Authors	Neutron beam	D_{cal}	K_{flu}	Fluence-Measuring device
Lawson and Porter (1976)	d(16) + Be	—	0.991	Fission counters
Lawson and Porter (1976)	d(0.5) + T	—	1.015	Fission counters
Mijnheer <i>et al.</i> (1981a)	d(0.25) + T	—	1.013	Fission counters
Mijnheer <i>et al.</i> (1981a)	d(0.25) + T	—	1.003	Activation foils
Thomas and Lewis (1981)	d(0.15) + T	—	0.986	Associated α -particle counting
Thomas and Lewis (1981)	d(0.15) + T	—	0.982	Telescope
McDonald <i>et al.</i> (1981c)	d(4) + Be	0.980	—	—
McDonald <i>et al.</i> (1981c)	d(49) + Be	1.003	—	—
McDonald <i>et al.</i> (1981c)	d(0.4) + T	0.997	1.021	Activation foils

4. DOSIMETRY OF NUCLEAR PARTICLES

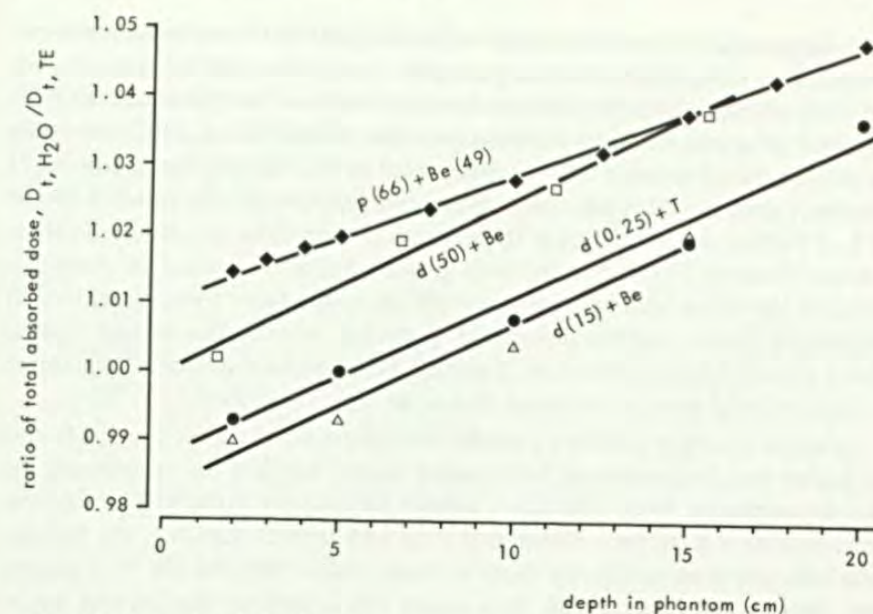


Fig. 13. Ratio of total absorbed dose measured in water and in TE liquid phantoms with one type of TE ionization chamber for various neutron beams. [From Broerse and Mijnheer (1982b).]

a TE liquid phantom (density 1.07 g/cm³) relative to those in a water phantom (Broerse and Mijnheer, 1982b). It can be concluded that the use of different liquids has the most striking effect at the greater depths and for the higher neutron energies. More information on this matter can be found in Vynckier *et al.* (1985).

For charged particle beams, the central axis depth-dose distribution is dependent on the energy and type of particles, since these quantities define the range. The extent of beam modulation determines the relative absorbed dose values in peak and plateau. Depth-dose distributions of fast neutrons depend on the neutron energy, the source-to-surface distance (SSD), the field size, and the relative contribution of gamma rays to the total absorbed dose. As shown in Fig. 4 the depth-dose characteristics of d + T neutron beams are less favorable than those of ⁶⁰Co gamma rays. Neutron beams produced by the d(50) + Be and p(66) + Be reactions have depth doses comparable to those of a 4-MV x-ray beam. In addition to the vast amount of experimental data, depth-dose distributions have also been calculated with the aid of two-dimensional discrete ordinate transport codes (Burger *et al.*, 1981b; Hehn *et al.*, 1981) and intranuclear cascade codes (Brenner *et al.*, 1981). Specific attention has also been devoted to the influence of inhomogeneities in the phantom (Pfister *et al.*, 1981).

Determination of central axis absorbed dose distributions is usually combined with transverse dose measurements at several depths to obtain isodose distribu-

tions. Because such a procedure requires considerable beam production and dose measurement time, systems for computing the dose matrix have been developed. For fast-neutron radiotherapy beams dose distributions have been derived from computer programs based on the tissue/air ratio (Shapiro *et al.*, 1979) or using a modified line decrement method (Hogstrom *et al.*, 1976). The sharpness of different beams can be compared by taking the distance off-axis between 80 and 20% of the total absorbed dose at 10 cm depth in a phantom as a criterion (Mijnheer and Broerse, 1979). For cyclotron-produced neutron beams, the sharpness generally decreases with increasing energy due to the more forward peaking of the neutron fluence and the decreasing scattering cross section of hydrogen at higher neutron energies. For d + T neutron generators a much larger penumbra is observed than that for cyclotron beams at higher energies.

At larger distances outside the useful beam the dose levels for neutron beams are higher than those obtained with photon beams, both for the neutron and the photon component. When allowance is made for changes in the RBE of the neutron component it has been shown that even with perfect shielding, the biologically effective dose outside the beam is much higher than that for ^{60}Co gamma rays (Bewley and Page, 1978). This means that in neutron therapy one has to accept a larger hazard for the patient than with megavoltage x rays. The wide penumbra and significant doses outside the penumbral region are a disadvantage of fast neutrons in comparison with the good homogeneity and minimal extraneous dose obtained with charged nuclear particles.

Dose distributions in a patient may be quite different from those in a standard phantom owing to differences in geometry, composition, and density. Computed tomography has proven to be of value in obtaining information on the patient such as body contours, shapes and densities of relevant organs, and target volume location (Goitein, 1982). Although the patient and the target volume are three-dimensional, conventional treatment planning programs are generally limited to two-dimensional isodose distributions at one or at most a few transverse sections of the patient. Sophisticated three-dimensional treatment planning techniques have been developed for pions (Pedroni, 1981), protons (Goitein, 1982), and heavy ions (Chen *et al.*, 1979). For treatment planning with neutrons, the gamma-ray component of the neutron beam must be considered separately. In addition, the contribution of neutrons scattered from the collimator and other structures placed in the primary beam cannot be neglected, contrary to the treatment planning procedure for other types of radiation. Specific problems related to treatment planning for external beam therapy with neutrons are discussed elsewhere (Burger *et al.*, 1981a).

In cooperative clinical trials such as those performed within the framework of the EORTC high-LET therapy group (Wambersie, 1982), information is collected about response of tumors at different sites and normal tissue complications. For this type of evaluation it is essential to compare the absolute dose and

the spatial dose distribution applied at the different clinics. For a number of neutron therapy facilities, a treatment planning intercomparison was organized by ECNEU. Patient cross sections for the head and neck region and for the pelvis were circulated, in which the target areas and critical organs were indicated. Figure 14 shows a summary of the treatment plans applied for a tumor of the floor of the mouth. It should be mentioned that the isodose curves represent

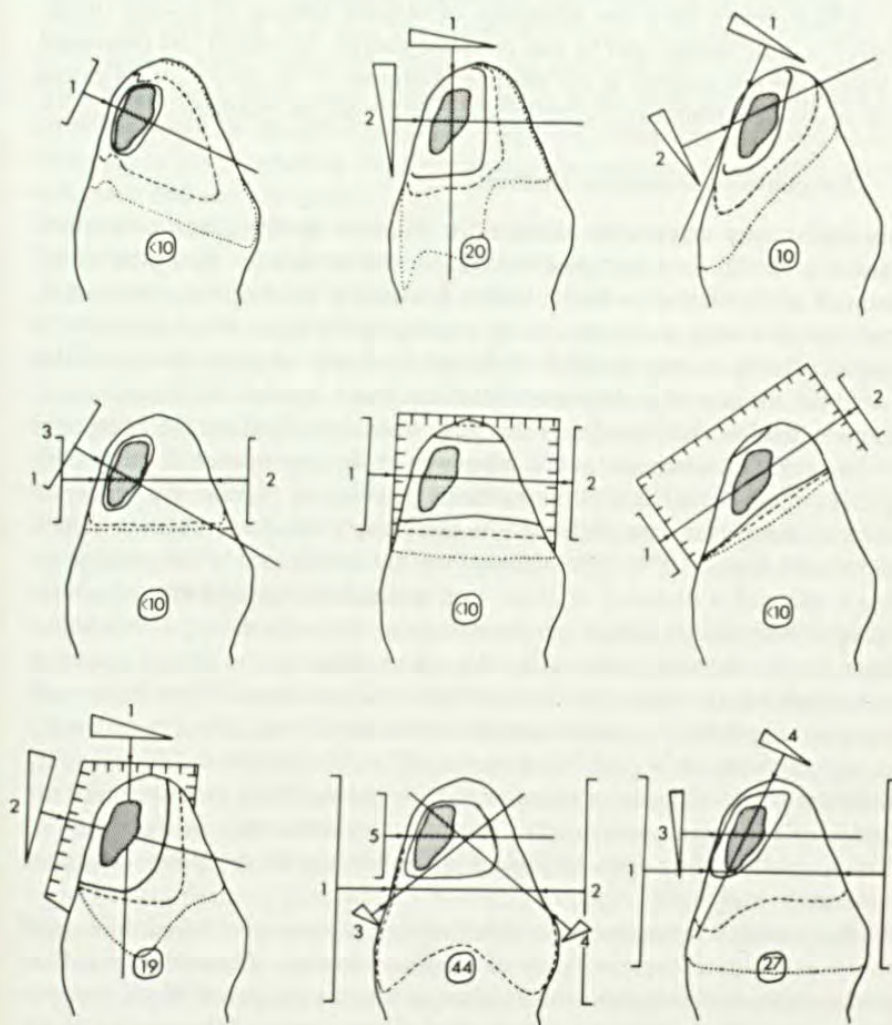


Fig. 14. Summary of treatment plans for a carcinoma of the floor of the mouth applied by different neutron therapy institutes. The 90, 50, and 20% isodose curves are indicated by solid, dashed, and dotted lines, respectively. The relative absorbed dose at the spinal cord is also indicated. [From Broerse and Mijnheer (1982b).]

values of the total absorbed dose $D_N + D_G$, the neutron absorbed dose D_N , or the biologically effective dose $D_N + \frac{1}{3}D_G$. The great variety of treatment techniques and dose specification procedures reflects the treatment policy in the individual institutes, including the application of wedge filters, bolus material, a large treatment volume compared to the target volume, or the inclusion of the lymph nodes.

The strictly localized dose distributions that can be obtained with charged nuclear particle beams have the advantage of optimal sparing of critical tissue. However, a disadvantage can be that adequate energy deposition and connected cell killing are not realized at the peripheral regions of the tumor, if its extent has been underestimated in the designation of the target volume.

VII. Response of Biological Dosimeters

Microdosimetry can provide a basis for radiation quality specification (see Section IV). It will, however, not be valid to assume that the biological effectiveness of two differing radiation beams is identical because their microdosimetric averages such as $\bar{y}_F$, $\bar{y}_D$, or y^* are the same, since the y spectrum is largely dependent on the diameter of the sensitive site adopted. This indicates the need for the use of biological systems for beam quality assessments.

In many models and hypotheses that have been developed for the interpretation of dose-effect relationships it is assumed that the dependence of relative biological effectiveness (RBE) on radiation quality is similar for different biological endpoints. For instance, in the theory of dual radiation action (Kellerer and Rossi, 1971), it is assumed that the production of two sublesions within a site with a diameter of about $1\mu\text{m}$ is the fundamental cause of a wide range of cellular effects. Other investigators [e.g., Barendsen *et al.* (1966)] have stressed that in addition to the energy deposition patterns, the precise chemical nature of primary biomolecular changes has to be considered. This latter point of view is supported by radiobiological studies involving different cell types [see, e.g., Broerse *et al.* (1977)]. Possible differences in the radiation quality of different beams of ionizing radiation can be assessed by a comparison of the response of biological systems. The intrinsic radiosensitivity as a function of radiation quality of various biological endpoints should be taken into consideration.

In this section information is provided on the response of different biological systems exposed to various types of nuclear particles. Dosimetry based on radiation-induced chromosome aberrations in human peripheral blood lymphocytes is summarized. The use of biological dosimeters for intercomparing the radiation quality of different fast-neutron beams is treated concisely. The possible differences in RBE for some biological endpoints at different positions inside biological objects are illustrated.

A. BASIC ASPECTS OF THE CONCEPT OF RELATIVE BIOLOGICAL EFFECTIVENESS

Due to the differences in the microscopic distributions of energy deposition, equal doses of different types of ionizing radiation do not produce the same level of biological effect. This has led to the concept of relative biological effectiveness (RBE), which is defined as the ratio of absorbed dose of a reference radiation to the absorbed dose of a test radiation required to produce the same level of biological effect, other conditions being equal (ICRU, 1979a). When two types of radiation produce an effect that is not of the same extent and/or nature, an RBE cannot be specified. Relative biological effectiveness should be specified only when monotonic dose-response relationships are obtained for both reference and test radiation.

Relative biological effectiveness depends not only on the microscopic distribution of energy dissipation but also on the level and kind of biological effect under investigation [see, e.g., Field (1976), Broerse *et al.* (1977), Barendsen (1979), and Hall *et al.* (1975)] and the experimental test conditions including the delivery of dose in time (repair of sublethal damage), factors affecting repair of potentially lethal damage, the oxygen concentration, temperature, stage in the cell cycle, and intercellular contact. The amount of repair of sublethal and potentially lethal damage is more pronounced for x rays (the common reference radiation) than for nuclear particles, thus affecting their RBE [see, e.g., Hall and Kraljevic (1976)]. A similar conclusion holds for dependence of radiosensitivity on stage in the cell cycle [see, e.g., Sinclair (1972)] and intercellular contact [see, e.g., Lücke-Huhle *et al.* (1979)].

Irrespective of the parameters mentioned above, the dependence of RBE on radiation quality as characterized by microdosimetric parameters is a complex one. Studies by Broerse *et al.* (1968) have shown that the track-averaged LET values of neutrons are considerably lower than those of directly ionizing particles with identical RBE. Dose-averaged lineal energy values $\bar{y}_D$ for neutron energies below 10 MeV are higher than those for neutrons with energies below 10 MeV, whereas the RBE values for the lower-energy neutrons are higher. It should be realized that a given y event becomes less effective in producing a biological effect per unit dose when y becomes too high, since part of the energy deposition is then wasted. To correct for this phenomenon, Kellerer and Rossi (1972) introduced the saturation-corrected, dose-averaged lineal energy y^* (see Section IV). However, irradiation experiments for different biological endpoints show different RBE values, whereas y^* is constant (Booz, 1979). It should be stressed that the evaluation of biological effects of different types of radiation should preferably be made on the basis of the actual dose-response functions for the biological systems concerned.

B. DOSIMETRY BASED ON THE INDUCTION OF CHROMOSOME ABERRATIONS

The induction of chromosome aberrations in cells can be used as a biological dosimetry system for ionizing radiation. Most investigators employ (human) peripheral blood lymphocytes [see, e.g., Vulpis *et al.* (1978) and Lloyd *et al.* (1976)]. The system seems particularly useful for estimating doses received by people in accidental whole-body irradiation. The detection limits for this application correspond to about 0.05 Gy of x rays, 0.1 Gy of gamma rays, and 0.01 Gy of fission neutrons [see, e.g., Purrott *et al.* (1981)].

The technique of lymphocyte cytogenetic dosimetry requires small samples of heparinized blood, from which lymphocytes are separated and stimulated to enter the mitotic cycle. After culturing for, e.g., 48 h the cells are fixed, dispensed onto microscope slides, and stained. The culturing techniques employed for (human) lymphocytes irradiated *in vivo*, as well as the sampling times, are of considerable importance [see, e.g., Van Buul and Natarajan (1980), Purrott *et al.* (1981), and Scott and Lyons (1979)]. The dose estimate for *in vivo* conditions is made on the basis of the observed yield of chromosome aberrations produced *in vitro*. In general, metaphases are analyzed for multicentrics, centric rings, and acentrics (defined as the sum of fragments and acentric rings). A summary of the formation and scoring of these aberrations can be found in the report of the United Nations Scientific Committee on the Effects of Atomic Radiation (1969). The yield of chromosome aberrations per cell, Y , is generally given [see, e.g., Edwards *et al.* (1981)] by

$$Y = \alpha D + \beta D^2 \quad (12)$$

where D is the absorbed dose and α and β are the linear and quadratic dose coefficients, respectively. In Table VII a summary of the yield of dicentric per cell for different types of radiation is given. It can be concluded from the table that the yield of dicentrics shows a considerable variation with the quality of the radiation employed.

In many situations of exposure to ionizing radiation, the dose distribution over the body will be highly inhomogeneous (especially in radiotherapy) and lymphocytes from unirradiated and irradiated regions are mixed, thus giving rise to an underestimation of the absorbed dose [see, e.g., Schmid *et al.* (1980)]. To overcome this problem Dolphin *et al.* (1973) suggested a method of estimating the blood volume by testing whether the dicentric distribution in the cells deviates from a Poisson distribution, which could be expected after homogeneous irradiation. This method seems applicable for photons [see, e.g., Kuttner-May and Traut (1981)] but not for nuclear particles, since even for uniform irradiation the dicentrics are overdispersed [see, e.g., Schmid and Bauchinger (1975)]. A study of chromosome damage in patients after therapy with d + T neutrons by Schmid *et al.* (1980) shows that the aberration yield in lymphocytes has a

TABLE VII

VALUES OF THE DOSE COEFFICIENTS α AND β FOR INDUCTION OF DICENTRICS IN HUMAN LYMPHOCYTES BY DIFFERENT TYPES OF RADIATION^a

Type of radiation	Dose rate, Gy/min	$\alpha \times 10^2$ ($\pm$ SD), Gy ⁻¹	$\beta \times 10^2$ ($\pm$ SD), Gy ⁻²	Reference
⁶⁰ Co γ rays	0.003	1.8 $\pm$ 0.8	2.9 $\pm$ 0.5	Lloyd <i>et al.</i> (1975)
⁶⁰ Co γ rays	0.5	1.6 $\pm$ 0.3	5.0 $\pm$ 0.2	Lloyd <i>et al.</i> (1975)
15-MeV electrons	1	0.6 $\pm$ 1.1	5.7 $\pm$ 0.6	Purrott <i>et al.</i> (1977)
Fission neutrons ($\bar{E}$ = 0.4 MeV)	0.002–0.07	90 $\pm$ 7	—	Vulpis <i>et al.</i> (1978)
Fission neutrons ($\bar{E}$ = 0.7 MeV)	0.5	83 $\pm$ 1	—	Lloyd <i>et al.</i> (1976)
²⁵² Cf neutrons ($\bar{E}$ = 2.13 MeV)	~0.0025	60 $\pm$ 2	—	Lloyd <i>et al.</i> (1978)
d(16.7) + Be ($\bar{E}$ = 7.6 MeV)	0.3	48 $\pm$ 3	6 $\pm$ 2	Lloyd <i>et al.</i> (1976)
d + T ($\bar{E}$ = 14.7 MeV)	0.3	26 $\pm$ 4	9 $\pm$ 3	Lloyd <i>et al.</i> (1976)
d + T ($\bar{E}$ = 15 MeV)	0.12	14.1 $\pm$ 0.7	3.8 $\pm$ 0.1	Schmid and Bauchinger (1975)
650-MeV neutrons ($\bar{E}$ = 250 MeV)	0.003	15 $\pm$ 3	25 $\pm$ 7	Vulpis and Bianchi (1982)
650-MeV neutrons ($\bar{E}$ = 250 MeV)	0.1	7 $\pm$ 4	57 $\pm$ 9	Vulpis and Bianchi (1982)
650-MeV α rays (peak)	—	14.1 ^b	2.25 ^b	Parmentier <i>et al.</i> (1975)
π^- (peak)	~0.02	23 $\pm$ 2	4.8 $\pm$ 0.9	Purrott (1975)
π^- (plateau)	~0.0125	13 $\pm$ 2	2.8 $\pm$ 1.1	Purrott (1975)

^aDoses range from approximately 0.1 to 10 Gy.

^bFor 95% confidence zones, see reference.

significant correlation with the therapeutic dose. Cytogenetic dosimetry is in principle feasible when correction factors for specific partial-body irradiations are determined. A formalism for calculation of the dose to lymphocytes in external beam therapy has been given by Ekstrand *et al.* (1981).

For partial-body irradiations, chromosome aberrations in fibroblasts might be used for dose assessments [see, e.g., Couzin (1979)]. These cells exist for long time periods in the skin and can be stimulated to divide in culture. Chromosomal aberrations have been detected from skin biopsies after therapeutic irradiation [see, e.g., Savage and Bigger (1978)]. The main disadvantages of this technique are the time required for *in vitro* cell growth between biopsy and cytological ex-

amination and the requirement for chromosome banding techniques, since the surviving aberrations are principally symmetrical chromosomal types (Couzin, 1979).

C. INTERCOMPARISON OF RADIATION QUALITY OF DIFFERENT FAST-NEUTRON BEAMS DERIVED FROM BIOLOGICAL DOSIMETERS

A prerequisite for the prediction of the responses of exposed biological specimens at different irradiation facilities is that the methods employed for absorbed dose determinations at the various institutes are intercompared. In addition, it is essential to obtain an assessment of radiation quality either by microdosimetric measurements or by employing biological dosimeters. Several studies that have been performed to estimate radiation quality of different neutron sources at various locations employing biological dosimeters are summarized below.

An intercomparison of the effectiveness of two different d + T neutron beams was made by employing the survival of colony-forming units (CFU) in mouse bone marrow as a biological dosimetry system (Broerse *et al.*, 1971). The biological systems employed by the two groups were compared using x and gamma irradiation and did not reveal significant differences. For neutrons, a difference of 25% was found between the RBE of the two beams at a CFU surviving fraction of 0.001. After correction for the differences in neutron dosimetry a 16% discrepancy remained, which had to be attributed to differences in radiation quality of the two beams, related to the presence or absence of a collimator.

Hendry and Greene (1976) summarized RBE values reported by different centers which used the intestinal microcolony assay technique for biological dosimetry. This system seems to be suitable for assessment of radiation quality. However, comparison of the RBE of different neutron beams should be performed at a certain level of survival in view of the dependence of RBE on surviving fraction. Zywiets *et al.* (1982) employed the same biological system to compare two neutron beams with mean energies of 6 and 14 MeV. In addition to the scoring of surviving jejunal crypt stem cells, the microdosimetric characteristics of the beams were measured to investigate the possibilities of replacing biological dosimeters by a suitable microdosimetric parameter. Zywiets *et al.* (1982) concluded that the differences in RBE of the two beams are within the experimental error in agreement with the ratio of y^* (see Section IV) for the two beams, indicating that y^* might be a suitable microdosimetric parameter for predicting differences in relative biological effectiveness.

Skin reactions and survival of intestinal crypt stem cells in mice were used as indicators of RBE and sparing effect of fractionation in a comparison of neutron beams generated by the d(16) + Be and d(42) + Be reactions (Bewley *et al.*, 1976). Employing the same dosimetry system, a difference of 20% in RBE between the two beams was found for doses between 3 and 20 Gy of neutrons, whereas the sparing effect of two large fractions was found to be the same.

A variety of portable biological systems, including three lines of mammalian cells cultured *in vitro*, the response of mouse skin, the survival of crypt cells in the mouse jejunum, and the loss of weight or DNA in the mouse testis, was used by Hall *et al.* (1979) to compare the RBE of fast-neutron radiotherapy beams in Japan and the United States. Apart from the mouse testis system, the biological dosimeters tend to give RBE estimates which are in reasonably close agreement. An intercomparison of the RBE of 10 neutron therapy facilities employing the survival of V-79 cells has been reported by Hall and Kellerer (1979). They state that systems and endpoints with widely different values for the RBE of neutrons give similar values for the RBE difference between two closely correlated neutron energies. Consequently the choice of a biological system for such intercomparisons may be governed largely by its portability, repeatability, and convenience.

D. THE USE OF BIOLOGICAL DOSIMETERS TO ASSESS RBE VARIATIONS WITH POSITION INSIDE (HUMAN) PHANTOMS

Differences in RBE of nuclear particles might be expected with position inside a relatively large biological object. In case of charged nuclear particles, quality changes with depth are anticipated due to LET differences between the plateau region and the (spread-out) Bragg peak. For pi mesons, in addition to this phenomenon, star formation will increase the radiation quality in the peak region. For irradiation with neutrons, variations of RBE with depth might occur owing to a decrease in (average) neutron energy with increasing penetration in the phantom. The use of beam-restricting devices (collimators) will influence the quality of the radiation due to scattering and absorption processes of the (primary) radiation. Several biological systems have been employed, including cell reproductive death, growth delay in *Vicia Faba*, chromosome aberrations, survival of CFU in mouse bone marrow, survival of mouse spermatogonia, mouse testis weight loss, and survival of mouse jejunal crypt stem cells.

An elegant method for obtaining RBE as a function of depth or position lateral to the beam axis is the use of cells suspended in a gel-medium matrix [see, e.g., Skarsgard and Palcic (1974)]. In Fig. 15 the percentage of cells, suspended in a gel medium, surviving at different dose levels as a function of depth in phantom is compared with the depth-dose distributions for three pion beams modulated in different ways (Raju *et al.*, 1979). It is concluded from this figure that variations in radiation quality occur as a function of depth even within the spread Bragg peak. The dose dependence of RBE (i.e., larger variations in RBE at lower doses) can be derived from comparison of the cell survival data with the depth-dose distributions. Differences in biological effectiveness of the different beams are evident. The data also provide information concerning how the beams must be energy modulated to obtain an approximately constant biological effect in the spread peak region. In general the dose in the proximal

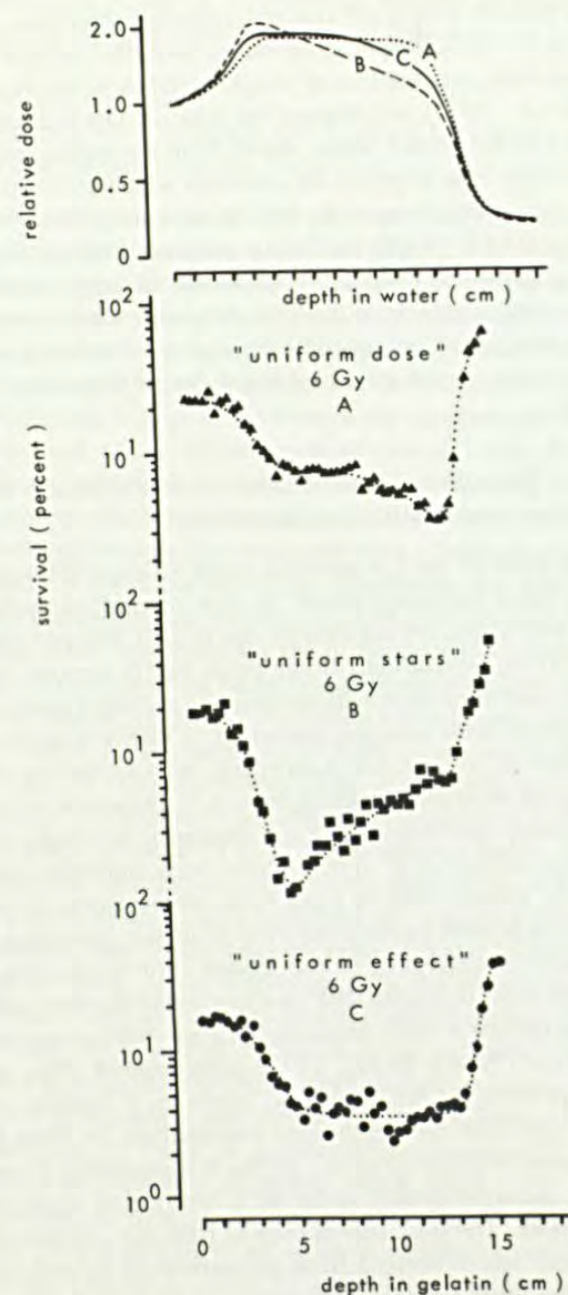


Fig. 15. Depth-dose distribution and cell killing as a function of depth for differently modulated pion beams. [From Raju *et al.* (1979).]

region must be greater than the dose in the distal region [see, e.g., Curtis (1980), Raju *et al.* (1978), and Dicello (1980)]. Scatter processes influencing radiation quality (RBE) in directions lateral to the primary beam are less of a problem for heavy charged particle (e.g., C, Ne, and Ar) beams because of the large mass of these particles than is the case for neutrons. In pion irradiations the contributions from neutrons produced in star formation processes can influence the RBE [see, e.g., Dicello *et al.* (1980)].

For several neutron sources, RBE has been investigated at various positions in the primary beam and behind the shielding of a collimator. Studies on $d(16) + \text{Be}$ neutrons by McNally and Bewley (1969) did not reveal a significant variation of RBE as a function of depth. Similar observations were reported by Hogeweg *et al.* (1978) and Mijneer *et al.* (1979) for $d + \text{T}$ neutrons, by Menzel *et al.* (1978) for $d(10.6) + \text{D}$ neutrons employing cell cultures *in vitro*, and by Geraci *et al.* (1975) for $d(21) + \text{Be}$ neutrons employing mouse testes as a biological test system. The observation of a rather constant RBE at different depths along the central beam axis is in agreement with the results of microdosimetric measurements for different collimated neutron sources (see Section IV).

The results of experimental studies with biological dosimeters at different distances from the beam axis differ for different neutron beams. Hogeweg *et al.* (1978) did not observe a significant change in RBE at lateral distances from the beam axis up to 4 cm beyond the beam edge for 6.5- and 15-MeV neutrons. A similar observation was made by Mijneer *et al.* (1978) and Hendry *et al.* (1979) for $d + \text{T}$ neutrons. Menzel *et al.* (1978) reported for $d(10.6) + \text{D}$ neutrons a slight decrease in biological effectiveness for positions outside the primary beam. Bewley *et al.* (1978) for $d(16) + \text{Be}$ neutrons and Geraci *et al.* (1975) for $d(21) + \text{Be}$ neutrons observed increases in RBE of about 30 and 60%, respectively, in the shielded areas compared to positions in the primary beam. The changes in RBE with lateral distance from the primary beam axis into the penumbral region result from two counteracting processes, namely, neutron energy degradation (higher RBE) and increasing gamma-ray contribution (lower RBE). The complex balance between the effects of neutron energy degradation and increased gamma-ray contribution will determine the net effect. This is most likely dependent on the type of neutrons and might be different for different collimator constructions [see, e.g., Burger and Morhart (1979)].

VIII. *In Vivo* Dosimetry

The purpose of *in vivo* dosimetry is to verify the accuracy of the procedures for converting absorbed doses in reference phantoms to absorbed doses in the complex biological objects actually irradiated. In addition, *in vivo* dosimeters are intended for verification of treatment planning and control procedures. Although biological systems might be applied as *in vivo* dosimeters (see Section

VII) this section will deal only with physical detectors placed on the skin of patients or inside relevant body cavities.

Detectors employed for *in vivo* dosimetry should be small, easy to operate, sensitive to the types of radiation of interest, insensitive to spectral changes, accurate, and precise. In case changes of radiation quality (RBE) within the volume of interest are important, *in vivo* dosimeters should also provide such information. This might be achieved, e.g., by employing a twin-detector technique. It could be preferable to employ detectors in which the dose information is stored, since the use of connecting cables tends to be a practical inconvenience. Such detectors might be advantageous when moving biological objects are irradiated, but disadvantageous when direct information is required to interfere with the projected course of the irradiation. In radiation therapy the disadvantage of delayed dose information is generally not a major problem, since most treatment schedules involve fractionated irradiations. Dosimeters suitable for attachment to skin or introduction into body cavities include thermoluminescent materials (TLD), activation foils or pellets, semiconductor diodes, solid-state nuclear track detectors, and photographic films or emulsions. Information on the dosimetric characteristics of these detectors is given in Section III.

A. EXPERIENCE WITH NEUTRONS

In vivo dosimetry based upon the ^{27}Mg activity induced in small pellets of aluminum has been performed by Field (1971) to assess the doses to the esophagus as well as the exit dose during treatment of carcinoma of the bronchus with $d(16) + \text{Be}$ neutrons. It was concluded from the *in vivo* dosimetry results that the isodose curves corrected for a lung density of 0.33 g/cm^3 were valid. The disadvantage of the method is that the calibration depends on irradiation time and distribution of beam intensity during irradiation.

Smith *et al.* (1976) applied silicon diodes as *in vivo* dosimeters inserted in the esophagus and bronchus of anaesthetized rhesus monkeys to measure the corrections to tumor dose resulting from intervening lung tissue. The measured-to-predicted dose correction factor as a function of the ratio between the thickness of the lung and the thickness of the overlying tissue for irradiation of *Macaca mulatta* with $d(50) + \text{Be}$ neutrons (see Fig. 16) was applied to patients treated for esophageal carcinoma. *In vivo* dosimetry with silicon diodes inserted in the esophagus of four patients showed that measured doses versus calculated doses, employing the lung correction obtained from *in vivo* dosimetry in rhesus monkeys, agreed to within 4%.

In vivo dosimetry for irradiations of tumors in the pelvic region with $d + \text{T}$ neutrons was performed by Mijnheer *et al.* (1981b) for 20 patients applying the production of ^{24}Na and ^{56}Mn in small Al and Fe pellets introduced in the rectum and on some occasions in the bladder. In addition, activation detectors were

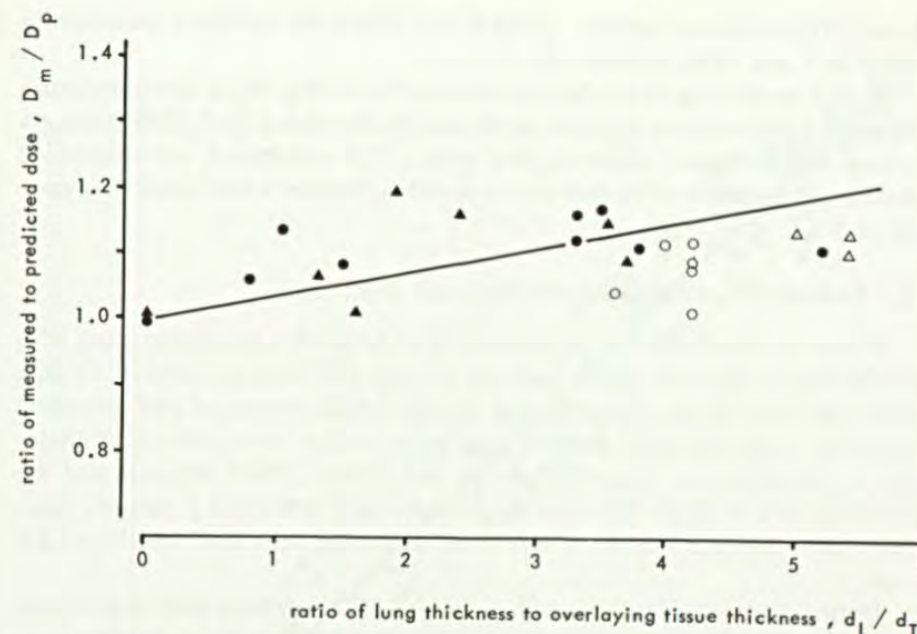


Fig. 16. Measured lung dose correction for $d(50) + \text{Be}$ neutrons using silicon diodes in rhesus monkeys. The ratio of lung thickness d_L to the thickness of overlying tissue d_T is plotted on the abscissa; the ratio of absorbed dose measured by the diodes D_m to the dose predicted D_p from isodose distributions in homogeneous media is plotted on the ordinate. (Δ , $\blacktriangle$) esophagus and ($\circ$, $\bullet$) bronchus. $D_m/D_p = 0.4d_L/d_T + 1.00$. [From Smith *et al.* (1976).]

attached to the skin. Agreement within the experimental uncertainty (about 8% at maximum in the pelvis) was obtained between the measured doses and doses calculated on the basis of measurements with ionization chambers in a water phantom.

Several systems are described for treatment verification during neutron therapy. Information on the application of a $\text{CaSO}_4 : \text{Tm}$ phosphor TLD system for *in vivo* monitoring of irradiations of patients with $d(16) + \text{Be}$ neutrons is given by Blum *et al.* (1976). The readings from the phosphor alone and the phosphor mixed with glucose provided a means to determine separately the neutron and gamma-ray absorbed dose contributions in the mixed field. The mean dose values derived from TLD relative to the isodose estimations at entrance and exit points were 1.01 and 0.86, respectively, with relatively large standard deviations of 8.8 and 17%. An analysis of the errors involved indicated that the energy sensitivity of the system is a problem.

The activation of indium (Catterall and Bewley, 1979) and of aluminum foils placed on the skin of the patients (Franke *et al.*, 1978) is routinely employed as a means of neutron treatment control. Franke *et al.* (1978) reported that the neutron yield derived from Al activation is about 9% smaller than that derived

from a TE ionization chamber, which is well within the estimated uncertainties of about 7 and 10%, respectively.

In vivo monitoring of the thermal neutron flux during boron neutron-capture therapy of brain tumors has been performed by Harasawa *et al.* (1981) using a silicon surface-barrier diode coupled with a ^6LiF radiator. It was concluded that the relative error of the monitoring is within 2% over a total irradiation period of 5 h.

B. EXPERIENCE WITH CHARGED NUCLEAR PARTICLES

In vivo measurements on the skin and in body cavities of patients using ^{24}Na production in aluminum pellets to derive the high-LET dose component of a pion beam are reported by Hogstrom and Amols (1980). Measured and calculated high-LET contributions to the total dose agreed within the statistical uncertainties in the high-dose region. Hogstrom and Irifune (1980) demonstrated the suitability of LiF (TLD-700) as an *in vivo* dosimetry system for pi mesons, since glow-curve analysis has the potential of measuring both total and high-LET dose.

During treatment of patients with pi mesons, TLD has been used as an *in vivo* dosimeter (Walder *et al.*, 1983). For intracavitary positions (e.g., bladder, rectum, and vagina) TL dosimeters (^7LiF powder encapsulated in glass) are arranged in tubes with a diameter of 5 mm. The average dose in bladder and rectum obtained from the *in vivo* dosimeters showed good agreement with the calculated dose (see Fig. 17).

Richman *et al.* (1980) measured doses of high-LET radiation, in patients during treatment with pi mesons, with thin silicon detectors placed on the skin, inside the oral cavity, and in the rectum. The authors emphasized that the measurement of high-LET doses constitutes a separate check of the treatment, and is capable of discovering deviations from the treatment plan not discernible in the total dose, which they, on some occasions, measured concurrently with a small ionization chamber. By timing the occurrence of the peak in the high-LET dose distribution with reference to the position of the range shifter, information is obtained on the water-equivalent thickness of bolus material and tissues in front of the detector.

IX. Conclusions

The possible advantages of a specific nuclear particle beam for biomedical applications are related to its dose distribution and/or biological characteristics. An attempt to compare various nuclear particle beams with respect to dose distribution and biological characteristics is shown in Fig. 18. The oxygen gain factor (defined as the quotient of the oxygen enhancement ratio of x rays and that of nuclear particles) is taken as one measure of the biological advantages of high-LET radiation. The evolution of conventional photon beams from orthovoltage

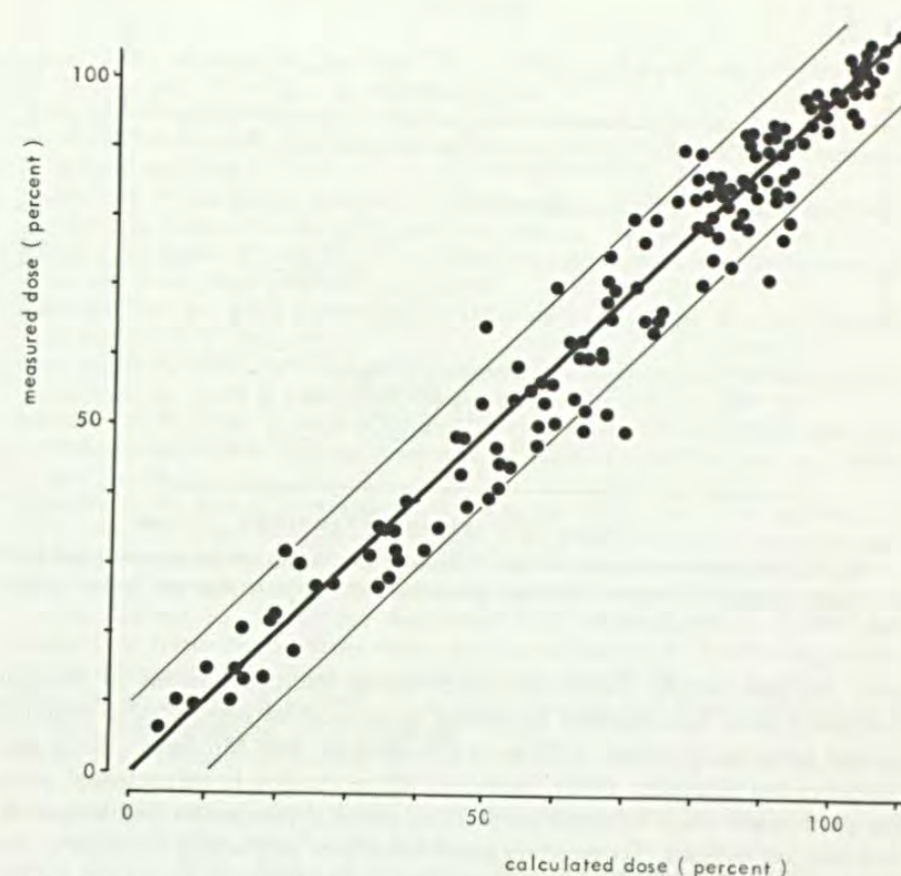


Fig. 17. *In vivo* dosimetry with thermoluminescence dosimeters placed in the bladder and rectum of patients treated with pions. [From Walder *et al.* (1983).]

to megavoltage x rays has involved considerable improvements in the dose distribution with a negligible change in the LET of the beam. Proton beams give the most advantageous dose distribution and can be produced at relatively low cost compared to the other types of nuclear particles, but their biological properties are similar to those of photons. Neutrons can be considered to be the cheapest form of high-LET radiation; however, high-energy cyclotrons are necessary to obtain depth-dose characteristics better than those of ^{60}Co gamma rays. The biological properties of neutrons are excelled by argon ions, with the additional advantage of a more appropriate dose distribution for the latter type of radiation. Pions are similar to carbon and neon ions with regard to dose distribution and biological effectiveness, but lose out because of dose-rate limitations and a less sharp edge to the beam.

Due to the great expectations of high-LET radiotherapy, the research and development of dosimetry for nuclear particles has shown remarkable progress

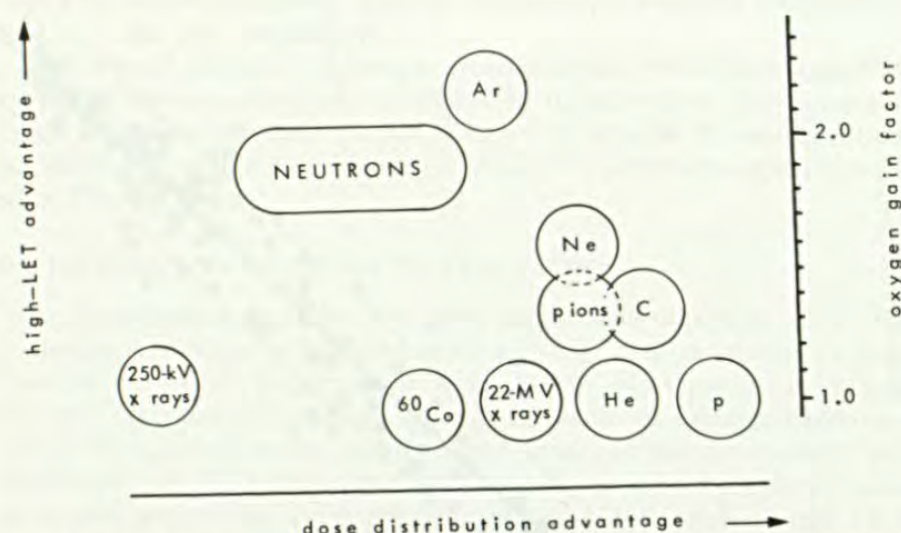


Fig. 18. The particles compared in terms of the dose distributions and the presumed biological advantages of high-LET radiation. Modified version based on an idea by Raju and Koehler. [From Hall (1982).]

over the past decade. Successive international intercomparisons of neutron dosimetry have demonstrated increasing agreement between results obtained among participants: from ± 20 to $\pm 2\%$ (Broerse and Mijneer, 1982a and Zoetelief and Schraube, 1985). However, this agreement is only attained when the participants use a coherent set of basic physical parameters and ionization chamber corrections. To ascertain good accuracies in absolute dosimetry, the knowledge of kerma ratios and W values has to be improved, especially for the higher neutron energies. The development of sensitive absolute dosimeters such as liquid calorimeters can be of great value in this respect.

Various interaction processes, e.g., multiple scattering and nuclear fragmentation, determine the depth-dose characteristics of nuclear particles. This may necessitate the use of quite complex dosimetry instrumentation and irradiation techniques, particularly for the assessment of and compensation for inhomogeneities (Goitein *et al.*, 1982). The range and energy modulation of nuclear particle beams required to obtain a homogeneous dose distribution over the target volume inevitably reduce their advantages for use in radiation biology and therapy.

Research on dosimetry for nuclear particles has provided interesting information on response of detectors for radiation of different qualities. Since high-energy accelerators, as employed for photon and electron therapy, can produce contaminating nuclear particles, this knowledge is of practical importance in that connection also.

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5

Measurement and Dosimetry of Radioactivity in the Environment

KURT LIDEN AND ELIS HOLM

DEPARTMENT OF RADIATION PHYSICS
LUND UNIVERSITY
LUND, SWEDEN

I. Introduction	292
A. Irradiation of Human Beings	292
B. Naturally Occurring Radionuclides	293
C. Human Redistribution of Natural Radionuclides	293
D. Artificial Radionuclides	293
E. Feasibility of Environmental Radioactivity Measurements	294
II. Sources of Environmental Radioactivity	294
A. Naturally Occurring Radionuclides	294
B. Radionuclides of Human-Made Origin	301
C. Air and Water Transport, Deposition, and Circulation of Radionuclides	302
III. General Aspects of Measurement Methods	305
A. <i>In Situ</i> Field Measurement	305
B. Collection and Field Treatment of Samples, Their Further Handling and Preparation	307
C. Monitoring Programs	309
IV. Analytical Procedures and Measurement of Radioactivity	310
A. General Aspects of Methods for Qualitative and Quantitative Assessment of Radionuclides	310
B. Methods for Identification and Assessment of Gamma Emitters	314
C. Methods for Assessment of Beta Emitters and Their Identification	332
D. Methods for Identification and Assessment of Alpha Particle-Emitting Nuclides	340
E. Background and Buildup or Decay Corrections	349

V. Nonconventional Methods for Assessment of Radionuclides	351
A. Neutron Activation	351
B. Mass Spectrometry	352
C. Other Techniques	353
VI. Discussion of Errors and the Need for Uniformity in Gathering and Reporting Data	353
A. Precision and Overall Uncertainty of Counting Methods	353
B. Sample Variability	355
C. Uniformity of Reported Data	356
D. International Intercomparison: The Use of Environmental Natural Matrix Standards	357
VII. Estimation of Radiation Levels Received by Human Beings	357
A. The Transfer of Radionuclides through Various Transport Chains to Humans	357
B. The Irradiation of Human Beings by Internal and External Radionuclide Sources—A Brief Overview	360
References	364

I. Introduction

A. IRRADIATION OF HUMAN BEINGS

The irradiation of human beings and the biosphere by ionizing radiation from various sources has been subject to increased concern and interest, mainly due to its possible health effects. As all human beings and all other living organisms have been exposed to such radiation since the origin of life on earth, the natural radiation dose rate could be used as a yardstick when studying the biological effects of human-made radiation. This natural radiation background is caused by cosmic radiation, radiation from radionuclides in the environment, and from some radionuclides incorporated in living tissue. Of these sources radiation from environmental and internal radioactivity contributes about 85% of the average per capita global annual effective dose equivalent, which is about 2 mSv (200 mrem), a figure adopted by the United Nations Scientific Committee on the Effects of Atomic Radiation in its last report (UNSCEAR, 1982). The main reason for this 100% increase of the hitherto accepted average value 1 mSv/yr is the improved knowledge and increased concern about environmental concentration of radon and its decay products.

Today considerable amounts of manufactured radioactive substances are deposited on the earth. The main sources for this contamination are fallout from nuclear weapons testing, releases from nuclear power plants, and medical use of radioactive pharmaceuticals. As large populations are exposed to radiation from these sources, it is considered very important to keep these contributions at a low level and, therefore, to establish continuous supervision of the presence, concentration, and transfer to the biosphere of these radionuclides.

B. NATURALLY OCCURRING RADIONUCLIDES

The most important naturally occurring radionuclides are ^{40}K ($T_{1/2} = 1.26 \cdot 10^9$ yr), ^{238}U ($T_{1/2} = 4.5 \cdot 10^9$ yr) + decay products, and ^{232}Th ($T_{1/2} = 1.4 \cdot 10^{10}$ yr) + decay products. They are present at various concentrations in almost all types of rock and soil and also in the ocean and groundwaters. Potassium-40 (^{40}K) is an isotope of potassium which is found in almost all living organisms as a necessary element for life. Uranium and thorium, as well as their decay products, have no known importance for living cells and organisms and are therefore usually found only in very small amounts in biological material. The radionuclides ^3H (tritium), ^7Be , and ^{14}C are examples of radionuclides being continuously produced by the interaction of cosmic radiation with atmospheric atoms. Tritium and ^{14}C are both incorporated in living organisms through natural processes, causing about 0.7% of the effective dose equivalent from natural background radiation to humans. The contribution from tissue-incorporated ^{40}K is about 10%. Several other known naturally occurring radionuclides, such as ^{87}Rb and ^{147}Sm are of very minor importance as sources for the irradiation of human beings and other organisms. They have found interesting applications in other fields, such as for age determination in the geosciences or as long-lived natural tracer substances.

C. HUMAN REDISTRIBUTION OF NATURAL RADIONUCLIDES

Through human activity some of the naturally existing radionuclides have created new problems as environmental contaminants. The most important ones are ^{222}Rn and its radioactive decay products, which can be present in high concentrations in the mining industry and inside modern buildings. This means that there is a rapidly increasing need for estimation and measurement of absorbed dose and dose equivalent for these radionuclides.

D. ARTIFICIAL RADIONUCLIDES

Today there are many artificially produced radionuclides present in the biosphere, most of them originating from testing of nuclear weapons. However, the increasing number of nuclear power plants and related installations in the world are releasing radioactivity at an accelerating rate to the environment. The most important radionuclides in the first case are the long-lived ^{90}Sr and ^{137}Cs , both of which rather easily enter living organisms including humans. In some cases with special food chains involved, for instance, lichens $\rightarrow$ reindeer/caribou $\rightarrow$ Lapps/Eskimos, the absorbed dose rate to people from ^{137}Cs uptake was comparable to the natural background absorbed dose rate in the period 1962 to 1970. Iodine radioisotopes, especially ^{131}I ($T_{1/2} = 8$ d), are also created at the detonation of nuclear weapons as well as being locally released in small amounts from nuclear power plants and medical facilities. The biological availability of

iodine is high and its uptake occurs predominantly in a small organ, the thyroid, which necessitates careful monitoring of iodine radioisotopes. Other environmental radionuclides of importance to people and other biological species are ^3H , ^{14}C , ^{99}Tc , ^{239}Pu , ^{241}Pu , and ^{241}Am from nuclear weapons and ^3H , ^{14}C , ^{85}Kr , ^{99}Tc , ^{133}Xe , and ^{134}Cs , among others, from nuclear installations. Induced radioactivity of corrosion products in nuclear reactors, such as ^{54}Mn , ^{59}Fe , $^{58,60}\text{Co}$, ^{65}Zn , ^{99}Tc , and $^{110}\text{Ag}^m$, is released in the cooling water. Minor releases of ^{60}Co to air have been reported. Some of these radionuclides are taken up by water-living organisms such as algae, bivalves, and fish with a possible further transfer to human beings.

E. FEASIBILITY OF ENVIRONMENTAL RADIOACTIVITY MEASUREMENTS

There is a theoretical possibility of detecting the decay of even a single radioactive atom. This extremely favorable situation is counteracted by the fact that in the environment all radioactive atoms are distributed among huge numbers of other stable atoms of many different elements, say in a ratio of 1 to 10^{20} – 10^{25} . This means that usually the activity concentration is low or extremely low, requiring carefully designed sampling methods, sample treatment, and activity analysis.

The final accuracy, reliability, and meaningfulness of the measurement of environmental radioactivity depends on several factors: representativeness of the sampling program, correct identification of radionuclides, radiochemical yield determination, source-detector geometry, low and stable detector background, uncertainty due to counting statistics, and long-term stability of detector electronics. Experience has shown that the assessment of low-level activity in distributed samples in various matrices often is a difficult task, and reliable activity calibration methods and intercomparison programs are essential.

II. Sources of Environmental Radioactivity

A. NATURALLY OCCURRING RADIONUCLIDES

About 85% of the average effective dose equivalent to humans is produced by radiation from natural radionuclides. These radionuclides are present throughout the biosphere in almost all types of rock, soil, water, atmosphere, and living organisms. Good knowledge of their distribution, concentration, physical and chemical properties, and natural separation and transport processes is necessary for accurate estimation of background radiation dose to various populations worldwide.

Frequently natural radionuclides complicate the measurement of manufactured radionuclides in the environment, usually as interfering peaks in spectrometry and/or as sources of variations of background counting rate. The levels of

natural radionuclides in the environment are summarized in some detail in the NCRP Report No. 45 (1975).

Natural radionuclides can be characterized as *primordial* (present on the earth since its creation) or *cosmogenic* (produced by interaction of cosmic rays with atoms in the atmosphere).

1. Primordial Radionuclides

The most abundant primordial radionuclides are ^{238}U , ^{235}U , ^{232}Th , and their decay products, and ^{40}K and ^{87}Rb . There are several others, much less abundant and usually with much longer half-lives, which decay to stable nuclides as do ^{40}K and ^{87}Rb . The average activity concentration of these radionuclides is less than 1% of that for ^{87}Rb , and their interference in activity measurements and their dosimetric importance are vanishingly small. The isotopic abundance of the potassium isotope ^{40}K is $1.18 \cdot 10^{-4}$ (0.0118%). It is also present in all living organisms and contributes 10–20% of the total natural background radiation dose to humans. The isotopic abundance of the pure β emitter ^{87}Rb is 27.9%, but the element rubidium is found in much smaller concentration than potassium in the earth's crust (less than 1%); further details are given in Table I.

The heavy elements uranium and thorium give birth to three radioactive series, which start with ^{238}U , ^{235}U , and ^{232}Th , respectively. Detailed knowledge of the properties of members of these series is very important when analyzing the actual radionuclides and their relatives in the transuranic series. The three series are displayed in Figs. 1, 2, and 3.

a. Earth's Crust. The average abundance of uranium and thorium in the earth's crust is low. However, the elements are present almost everywhere on the earth. The concentration in rocks and soil varies substantially, as shown in Table II. These radionuclides and their many decay products are therefore all likely to be present in environmental samples (biologic, geologic, hydrologic, etc.) through dust deposition, wash-out, weathering, sedimentation, and biological and other transfer processes (Holm *et al.*, 1981). It is important to have a good knowledge of the geological structure of rocks and soils within and around

TABLE I

PRIMORDIAL RADIONUCLIDES

Radionuclide	Half-life (yr)	Isotopic abundance (%)
^{40}K	$1.26 \cdot 10^9$	0.0118
^{87}Rb	$4.8 \cdot 10^{10}$	27.9
^{232}Th	$1.4 \cdot 10^{10}$	100
^{235}U	$7.1 \cdot 10^8$	0.7
^{238}U	$4.5 \cdot 10^9$	99.3

Atomic number	Nuclide	Half-life	Type of decay
92	^{238}U	$4.5 \cdot 10^9$ yr	α
90	^{234}Th	24.1 d	β^-
91	$^{234}\text{Pa}^m$	1.2 min	β^-
92	^{234}U	$2.5 \cdot 10^5$ yr	α
90	^{230}Th	$8.0 \cdot 10^4$ yr	α
88	^{226}Ra	1600 yr	α
86	^{222}Rn	3.8 d	α
84	^{218}Po	3.1 min	α, β^-
85	^{218}At	~ 2 s	α, β^-
86	^{218}Rn	$3.5 \cdot 10^{-2}$ s	α
82	^{214}Pb	26.8 min	β^-
83	^{214}Bi	19.7 min	β^-, α
81	^{214}Tl	1.3 min	β^-
84	^{214}Po	$1.6 \cdot 10^{-4}$ s	α
82	^{210}Pb	22.3 yr	β^-
83	^{210}Bi	5.0 d	β^-, α
81	^{210}Tl	4.2 min	β^-
84	^{210}Po	138 d	α
82	^{206}Pb	stable	

Fig. 1. Nuclides of the ^{238}U series. [After Pentreath (1980).]

areas where monitoring or other programs are performed. In some cases, e.g., in rocks, the decay products of a series are in radioactive equilibrium with their precursors. In Table II the activity concentration of the listed radionuclides is given. Assuming full decay-product equilibrium, the total activity concentration is the sum of the activity concentrations of the individual decay products.

b. Water. In soil, water, and air the various members of a series often behave differently, both physically and chemically. A classical example is the soluble ^{238}U – ^{234}U in slight disequilibrium in ocean water, where the decay product ^{230}Th precipitates in the sediments (as do other Th isotopes), thus affecting the

Atomic number	Nuclide	Half-life	Type of decay
92	^{235}U	$7.0 \cdot 10^8$ yr	α
90	^{231}Th	25.5 h	β^-
91	^{231}Pa	$3.3 \cdot 10^4$ yr	α
89	^{227}Ac	21.8 yr	β^-, α
87	^{227}Th	21.8 min	β^-
90	^{227}Ra	18.7 d	α
88	^{223}Ra	11.4 d	α
86	^{219}Rn	4.0 s	α
84	^{215}Po	$1.8 \cdot 10^{-3}$ s	α, β^-
85	^{215}At	$1.0 \cdot 10^{-4}$ s	α
82	^{211}Pb	36.1 min	β^-
83	^{211}Bi	2.2 min	α, β^-
84	^{211}Po	0.52 s	α
81	^{207}Tl	4.8 min	β^-
82	^{207}Pb	stable	

Fig. 2. Nuclides of the ^{235}U series (the actinium series). [After Pentreath (1980).]

TABLE II

MASS CONCENTRATION AND ACTIVITY CONCENTRATION OF ^{40}K , ^{238}U , AND ^{232}Th IN SOME ROCKS AND SOIL

Type of rock	^{40}K		^{232}Th		^{238}U	
	mg/kg	Bq/kg	mg/kg	Bq/kg	mg/kg	Bq/kg
Igneous						
Granite	5	1200	17	70	3	35
Basalt	0.9	230	4	15	1	12
Sedimentary						
Limestone	0.3	70	2	8	2	25
Sandstone	1.3	300	3	11	1.5	18
Shale	3.0	700	12	45	4	50
Soil						
World average	1.4	370	6	25	2	25

Atomic number	Nuclide	Half-life	Type of decay
90	^{232}Th	$1.4 \cdot 10^{10}$ yr	α
88	^{228}Ra	5.8 yr	β^-
89	^{228}Ac	6.1 h	β^-
90	^{228}Th	1.9 yr	α
88	^{224}Ra	3.7 d	α
86	^{220}Rn	55.6 s	α
84	^{216}Po	0.15 s	α
82	^{212}Pb	10.6 h	β^-
83	^{212}Bi	60.6 min	α, β^-
84	^{212}Po	$3.0 \cdot 10^{-7}$ s	α
81	^{208}Tl	3.1 min	β^-
82	^{208}Pb	stable	

Fig. 3. Nuclides of the ^{232}Th series. [After Pentreath (1980).]

distribution of subsequent decay products. The concentration of radionuclides in ocean and other seawater is quite low, partly due to the sedimentation processes and partly due to dilution in the huge water volume. For instance, the concentrations of potassium and uranium are 0.35 g/liter (11 Bq/liter) and 3 μg /liter (36 mBq/liter), respectively. In groundwater ^{222}Rn frequently appears at high concentrations (10–200 Bq/liter), particularly in deep well water, but its precursor ^{226}Ra is found in much lower concentrations.

The transfer of primordial radionuclides to the hydrosphere is strongly influenced by various chemical and physical properties of the rock–soil–water–air system. Strong local deviations from the general pattern of transport and deposition have been observed.

c. *Air.* Particularly variable sources of environmental radioactivity are the gaseous decay products ^{222}Rn and ^{220}Rn . Their emanation rate from soil is strongly dependent on the barometric pressure, the moisture content, and the snow cover. Their short-lived decay products are easily precipitated by rain, resulting in strong time variations of their activity concentration in surface air. Simultaneously, the deposition of ^{222}Rn + decay products on the ground increases, thereby causing an increase in the environmental γ -ray absorbed dose rate for some hours after a rain shower. Great local fluctuations of radon concentration in air occur, both with time and position, mostly related to the emanation

tion rate from the ground and to variations in atmospheric stability. For instance, a morning peak and an afternoon drop may occur, correlated with inversion conditions. Typical values of ^{222}Rn levels in air are above ground—1–10 Bq/m³; above ocean water—0.04 Bq/m³.

Indoor levels are strongly influenced by the emanation rate from building materials, basement construction, and underlying ground. Increased ventilation will reduce radon and its short-lived decay-product concentration considerably.

The ^{220}Rn ($T_{1/2} = 55$ s) activity concentration at the ground surface has been found to be of the same order as ^{222}Rn but decreases rapidly with height, due to its rapid physical decay.

The long-lived decay products of ^{222}Rn (^{210}Pb , a β emitter and ^{210}Po , an α emitter) are present in the atmosphere. They are also released during combustion of coal and have to be considered in assessments of the background radiation dose equivalent to humans.

2. Cosmogenic Radionuclides

Cosmogenic radionuclides are produced when cosmic rays interact with atmospheric atoms through spallation or neutron capture. Although more than 20 different radionuclides have been identified as being produced in this way, only a few of them contribute a measurable absorbed dose to human beings: ^3H , ^{14}C , ^{22}Na , and ^7Be .

The production rate of ^{14}C is 2.5 atoms/cm² s, resulting in a global inventory of $6.8 \cdot 10^4$ kg of ^{14}C with a corresponding activity of $1.15 \cdot 10^{19}$ Bq. Of this amount 94% is found in the oceans and their sediments. The production rate of ^3H is about 0.4 atoms/cm² s, giving a global inventory of about $2 \cdot 10^{18}$ Bq.

Tritium and ^{14}C are found in all living organisms due to uptake of ^3HHO and $^{14}\text{CO}_2$, as well as in the atmosphere and hydrosphere. Of the average human background effective dose equivalent, ^{14}C contributes about 0.7%; the contribution from ^3H is much smaller, and it is negligible for ^{22}Na and ^7Be .

Redistribution of Radionuclides by Action of Human Beings

Several types of conventional building materials contain naturally occurring radionuclides. High levels of uranium and thorium with their decay products are found in shale-based light concrete, causing both an increased external irradiation of persons and increased inhalation of radon and its decay products. Wooden houses and houses constructed of limestone-based bricks usually contain small amounts of radioactivity. However, through emanation from the ground radon isotopes may enter all types of houses, and in such areas efficient ventilation (of the order of 0.5 to 2 exchanges of air per hour) of the buildings is required if a low concentration of radon and its radioactive decay products is to be maintained.

Gypsum boards or blocks made of waste products from the superphosphate fertilizer industry contain relatively high ^{226}Ra concentrations, and the fertilizer itself retains the uranium present in the raw material phosphate rock (Eisenbud, 1973). Additional uranium is thus deposited on all fields fertilized by superphosphate. Further details are given by UNSCEAR (1977). However, the conclusion is that the average global dose equivalent contribution to humans by this "redistribution" process is small.

Underground mining is almost inevitably accompanied by increased radon concentration. Specially designed ventilation is required to reduce the concentration of radon and its decay products to levels in accordance with radiation protection recommendations; for instance, fresh air should not pass through crushed rocks. At present the International Commission of Radiological Protection (ICRP, 1981) recommends an annual limit of intake (ALI) of radon and any mixture of its short-lived decay products giving rise to an energy imparted to 0.02 J. For a working time of 2000 h/yr the derived air concentration, DAC, is about 1 Bq/liter. Other examples of technological redistribution are tailings from uranium mining and milling operations used as building materials, and also the release of radon when tapping geothermal energy sources and natural gas for fuel.

Coal and oil contain radionuclides such as ^{238}U , ^{232}Th , and their decay products, which can be emitted to the atmosphere by fly ash from coal- and oil-fired power plants and deposited in their surroundings. According to UNSCEAR (1977) the radionuclides ^{210}Pb , ^{228}Th , and ^{232}Th appear to be the most important contributors to the collective dose commitment to humans in this redistribution process.

4. Areas of High Natural Radioactivity

There are some regions in the world where unusually high concentrations of ^{238}U , ^{232}Th , and decay products are observed. Best known are the Kerala coast strip of India and one coastal (Rio de Janeiro) and one inland area (Minas Gerais) of Brazil. Others are the Lazio and Campania provinces of Italy, granite areas of France, and some districts of Iran, Madagascar, and Nigeria. In Kerala a 55-km strip contains the thorium mineral monazite with an average ^{232}Th content of about 9%. The population is about 70,000 persons, and the terrestrial absorbed dose rate to air was estimated to be 1.3 $\mu\text{Gy/h}$ compared with 0.03 to 0.07 $\mu\text{Gy/h}$ for 95% of the world population. The monazite coastal area in Brazil shows similar data, and still higher values of the absorbed dose rate at selected spots. High concentrations of naturally occurring radium are found in local groundwater, both natural springs and deeply drilled wells. Natural springs have been used as spas for curative effects on various illnesses by bathing or drinking or by breathing radon-containing air. In certain mineral waters ^{226}Ra

concentration as high as 4 kBq/liter has been measured, which is about 10^6 times greater than that for normal public water supplies (~ 4 mBq/liter).

Water from deep wells is sometimes an important source of radon available for inhalation by humans, together with its radioactive decay products, and should be carefully monitored if used by large population groups. As an example a Finnish community outside the capital Helsinki can be mentioned, where the average radon concentration of drinking water from wells was about 2.0 kBq/liter with a range from 0.1 to 15 kBq/liter (Asikainen and Kahlos, 1980; Kahlos and Asikainen, 1980).

B. RADIONUCLIDES OF HUMAN-MADE ORIGIN

1. Releases from the Nuclear Fuel Cycle

Under ideal conditions fission products should be trapped in the fuel elements in a reactor during fission of ^{235}U and ^{239}Pu . By damage to the fuel elements through corrosion and radiation, volatile fission products such as isotopes of krypton, xenon, iodine, and also ^3H can be released to the cooling water. By neutron interaction with corrosion products trapped on the surface of the fuel elements, activation products such as ^{60}Co , ^{65}Zn , and $^{110}\text{Ag}^m$ will be present in the coolant. For a boiling water reactor, radioactive gases may then be released during condensation of the cooling water.

In a pressurized water reactor leakage from the primary coolant is more important because of the high pressure. This is especially the case for ^3H , which is not easily trapped by filters. In a boiling water reactor the releases to air of radioactive noble gases are dominant. Examples of releases from Swedish reactors during one year are given in Table III.

TABLE III

RELEASES FROM SWEDISH NUCLEAR POWER STATIONS: REPORTED VALUES FOR 1981 IN GIGABECQUERELS (GBq) PER GIGAWATT-YEAR (GW-yr) PRODUCED ELECTRICAL ENERGY

	BWR		PWR	
	Air	Water	Air	Water
^3H	—	$0.03 \cdot 10^6$	—	$0.11 \cdot 10^6$
$^{14}\text{C}^a$	420	—	50	—
^{131}I	0.58	12	0.44	9.3
^{137}Cs	0.016	25	0.004	98
^{54}Mn	0.047	8.3	0.001	20
^{60}Co	0.54	180	0.069	330
$^{85}\text{Kr}^m$	$0.54 \cdot 10^6$	—	2800	—
^{133}Xe	$1.2 \cdot 10^8$	—	$0.28 \cdot 10^6$	—

^aData from 1979–1980.

The releases from a reactor depend on type of construction, energy production rate, age, etc. During reprocessing of nuclear fuel releases of both fission and activation products occur to air and water. The releases are generally large compared with those from nuclear reactors, especially in the case of fission products. Since the fuel elements to be reprocessed have been stored for some time, most short-lived radionuclides have decayed.

2. Releases from Testing of Nuclear Weapons

Testing of nuclear weapons has released mainly fission products. However, isotopes of americium (especially ^{241}Am), plutonium (including unfissioned ^{239}Pu), and neptunium, as well as ^{14}C , have also been released. Most of the fallout radionuclides were deposited on the ground as a result of the weapons tests conducted in the atmosphere during 1957–1958 and 1961–1962. A nuclear weapons test ban agreement was signed by the United States, the United Kingdom, and the USSR early in 1963 and since then these countries have performed only underground explosions. France and the Peoples Republic of China did not sign the agreement and have performed several nuclear test explosions in the atmosphere since 1963.

3. Releases from Medical Use of Radioisotopes

In nuclear medicine radionuclides such as $^{99}\text{Tc}^m$, ^{131}I , ^{125}I , ^{75}Se , ^{133}Xe , and others are used. Minor releases from hospitals of these radionuclides and of their long-lived decay products and impurities occur. Excretion by patients is by far the dominant source of releases to the environment of medically used radionuclides (Erlandsson and Mattsson, 1978).

4. Other Sources

Research laboratories and special industries such as radiopharmaceutical manufacturing may release radionuclides such as ^3H , ^{14}C , ^{125}I , and others. Incidents with aircraft carrying nuclear weapons and with satellites have contributed to the contamination of the environment with Pu and Am isotopes.

C. AIR AND WATER TRANSPORT, DEPOSITION, AND CIRCULATION OF RADIONUCLIDES

As indicated in Section II,A, primordial radionuclides are usually strongly bound to rock, soil, and sediment. The cosmogenic radionuclides are, of course, subject to much more migration. Manufactured radionuclides, by the very nature of their production processes, are almost all involved in various transfer and circulation patterns, as briefly discussed in Section II,B.

1. Primordial Radionuclides

Some geophysical and geochemical processes cause migration of primordial radionuclides, particularly by weathering of rock and soil, transportation by rivers, and sedimentation in lakes and oceans. For instance, the high ^{232}Th concentration in the beach sand of some coastal districts of Brazil is a result of such processes in the mountain range paralleling the coast (UNSCEAR, 1977). Another example is the release of ^{222}Rn as well as ^{226}Ra to spring water, causing a considerably increased background radiation observed at some places in France and Iran. There may be many other sites in the world not yet studied. In the hydrosphere, transfer and circulation of radionuclides occur more frequently. Fractional precipitation and sedimentation of members of a decay series have been illustrated in Section II,A: dissolved ^{238}U and ^{234}U in ocean waters decay to ^{230}Th ($T_{1/2} = 8.0 \cdot 10^4$ yr), which precipitates rapidly to the bottom sediments. The radionuclide ^{230}Th decays to ^{226}Ra , which in turn tends to dissolve in the water. This process causes a vertical gradient of the ^{226}Ra concentration, being about 4 mBq/liter in the bottom water and about 4 μBq /liter in the ocean surface water (Koczy, 1958, 1960).

Potassium plays an important role in the biosphere. Therefore ^{40}K is subject to a continuous recycling both in living organisms and as a result of transfer from living to dead material. However, potassium (and thus ^{40}K) is under homeostatic control and the internal absorbed dose from ^{40}K in humans is rather constant and independent of varying amounts in foodstuffs and diet.

2. Cosmogenic Radionuclides

The cosmogenic radionuclides ^{14}C and ^3H are converted to CO_2 and ^3HHO , then rapidly transferred to the biosphere and hydrosphere. In living organisms a constant relation between ^{12}C and ^{14}C is thus maintained (at a constant production rate of ^{14}C), and in dead organisms the decaying ^{14}C makes possible the well-known ^{14}C dating method. It is somewhat surprising to find 94% of all existing ^{14}C present in the ocean waters. Tritium participates in the normal water cycle. Freshwater seems to have a somewhat higher natural ^3H concentration than ocean water, 0.2–0.8 Bq/liter versus 0.1 Bq/liter, respectively. The annual absorbed dose to humans has been estimated to be 0.01 μGy for ^3H , as compared with 10 μGy for ^{14}C . Because of its short half-life of 53 d, 70% of the ^7Be is present in the atmosphere. Its regular presence in rain and deposition on the ground makes this radionuclide useful as a tracer and yield determinant in environmental processes.

3. Manufactured Radionuclides

Artificial radionuclides are distributed worldwide, mainly as fallout from nuclear weapons testing and releases from the nuclear industry, as discussed in

Section II.B. Most of the short-lived fallout radionuclides ($T_{1/2} < 1$ yr) have now (1984) disappeared. Today the most important artificial radionuclides in terms of absorbed dose to humans are ^{90}Sr and ^{137}Cs . Plutonium and americium isotopes have also been extensively studied but have been shown to be less important as their transfer to biota is small. On a long-term basis ^{129}I ($T_{1/2} = 1.7 \cdot 10^7$ yr) and ^{99}Tc ($T_{1/2} = 2.1 \cdot 10^5$ yr) on the ground and ^{14}C ($T_{1/2} = 5730$ yr) and ^{85}Kr ($T_{1/2} = 10.8$ yr) in the atmosphere might be the dominating contributors of the dose equivalent to humans.

Carbon-14, ^3H , and ^{85}Kr are also produced by the nuclear industry. The oceans and freshwater are natural sinks for ^3H and ^{14}C but ^{85}Kr will continue to circulate in the atmosphere. If no trapping of ^{85}Kr occurs, it is estimated that its air concentration will approach 10 kBq/m³ in the middle of the twenty-first century. The inhabitants near nuclear fuel reprocessing plants will receive a skin dose of about 1–2 mGy/yr from the release of ^{85}Kr and it will be necessary to consider effective removal of ^{85}Kr from the gaseous wastes.

Some of the short-lived fallout and waste radionuclides such as ^{131}I may be rapidly transferred to man via deposition on vegetation grazed by cows, whose milk then contains a substantial fraction of the ingested ^{131}I . This may be transferred to milk consumers and their thyroid glands. Direct uptake by inhalation of gaseous iodine compounds is another transport route to be considered for monitoring actions. The transfer and circulation of the long-lived ^{90}Sr and ^{137}Cs ($T_{1/2} = 28$ and 30 yr, respectively), which are also waste products of the nuclear industry, have been extensively studied during the last 25 years. Both are now present in almost all individuals on the earth, mainly as a result of intake of fallout-contaminated food.

The remaining ground deposition of ^{90}Sr in 1982 was about 1 to 2 GBq/km² in the northern hemisphere and 0.5 to 0.7 GBq/km² in the southern hemisphere, with the bulk of the activity (~90%) remaining in the uppermost 30-cm layer of the soil. The difference between cultivated and uncultivated soil is small (Aarkrog, 1979). ^{90}Sr is chemically related to calcium. About 90% of the uptake of ^{90}Sr by humans occurs through milk, vegetables, and bread, while other types of food are of minor importance. About 2% of the ingested ^{90}Sr is retained by an adult, but in a child retention reaches 10–15%. The average biological half-life in humans is about 10–15 years, mainly because of the deposition of ^{90}Sr in the skeleton. A high Ca intake, e.g., via milk consumption, reduces the uptake of ^{90}Sr . The actual deposition of ^{137}Cs in 1982 in the northern hemisphere was about 2–3 GBq/km². This nuclide is more efficiently bound to clay minerals in the soil than is ^{90}Sr . Therefore most of its activity is present in a thinner soil layer than that for ^{90}Sr . The soil texture, however, plays an important role in the vertical distribution pattern and the availability of ^{137}Cs for the vegetation. The main route to humans is via milk, meat, and

grain; here ^{137}Cs behaves similarly to potassium. Almost 100% of the ^{137}Cs content of these food items is taken up by the GI tract. Human metabolism of ^{137}Cs results in a mean biological half-life of about three to four months for adults, but much less for babies and young children.

Available data concerning the movement of all important radionuclides through the environment have been subject to a comprehensive review by Coughtrey and Thorne (1983); as of this writing three volumes have been published and two additional ones on actinide elements should be available in 1985.

III. General Aspects of Measurement Methods

A. *IN SITU* FIELD MEASUREMENT

High-pressure ionization chambers and plastic scintillators are capable of measuring total γ -ray exposure rates in the energy interval 0.3–2 MeV. Gamma spectrometry with large NaI(Tl) detectors is useful for monitoring purposes, but it is not possible to discriminate between the contribution of small quantities of artificial radionuclides and the complex natural background spectrum. The development of large Ge(Li) detectors with high-energy resolution has made it possible to measure small amounts of natural and artificial radionuclides in the field (Beck *et al.*, 1971; Finck *et al.*, 1976). In this way very low exposure-rate levels may also be recorded. With mobile Ge(Li) or Ge(hp) (hp = high purity) spectrometric equipment it is possible to perform rapid determination of environmental γ radiation fields *in situ*. With knowledge of the activity distribution profile it is possible to quantify the concentration of γ -emitting radionuclides in soil and air.

1. Radionuclides in Soil

Soil radionuclide measurements require a suitable choice of sites for measurements. They should be free from disturbances; a plane horizontal area for a good approximation of 2π geometry should be selected with no overwash or flooding expected. A good vegetation cover is also recommended by some investigators (Eisenbud, 1973) if fresh fallout is to be measured. A cultivated site area has the advantage that natural and aged fallout radionuclides become uniformly distributed in the plowed layer (0–20 cm) in contrast to uncultivated soil, where the artificial radionuclide concentration often decreases exponentially with depth. The selected plots for *in situ* ground measurements should also be suitable for vegetation and soil sample collection, which requires an area of about 0.25 m² for each vegetation sample and about 0.03 m² for each core of soil.

2. Radionuclides in Air

With Ge(Li) spectrometry it is also possible to determine the atmospheric concentration, and its temporal variation, of individual γ -emitting radionuclides. As substantial interference from radionuclides on the ground may easily occur, it is important that a site with a stable and low soil-related background is selected. This interference is significantly reduced if measurements can be performed on a lake or at sea. The air flow at the site should not be affected by buildings or other tall objects, which cause local turbulence for some wind directions. The same type of site is required if *in situ* measurements of particle-bound radionuclides are made by drawing known amounts of air through a filter. The filter can be automatically interchangeable and measured directly by a suitable detector system. To a certain extent radionuclides emitting α , β , and γ radiation can be measured by this method. For a complete analysis of the radionuclide content of the filter material, a laboratory analysis is usually required.

Monitoring of normal or accidental stack releases from nuclear installations presents special problems. Usually the main interest concentrates on radionuclides of the noble gases xenon and krypton. Wind speed and direction, weather type (Pasquill A-F), terrain, and height of release are important parameters for selection of detector position. At a short distance from the stack, say, 200 m, and a stack height of 100 m, the geometry of the released plume normally is quite simple. At greater distances the shape and height of the plume depend very much on weather conditions, thus complicating the calculations of plume activity concentration.

When an accidental stack release of radionuclides occurs, it is important that the rate of released radionuclides, the direction of the plume, and its spatial distribution can be determined as fast as possible. The sampling time should not exceed 5 min. A model for a permanent monitoring detector system, covering a reactor station with both BWR and PWR power plants, has been studied at our department. We found that a suitable detector distance from the reactor station was 3 km, that 38 detectors along a half circle were required if at least one detector was to detect the plume for all weather conditions and types of accident, and that one of four detectors should be energy dispersive [e.g., NaI(Tl), 7.5 cm diam $\times$ 7.5 cm], the others measuring only exposure.

3. Radionuclides in Water

Radionuclides discharged to the aquatic environment by the coolant and liquid effluents from nuclear power reactors are not easily monitored directly *in situ* due to very great dilution. The releases from waste treatment plants are generally irregular in time, weekly or monthly or during periods of maintenance procedures. In contrast, the monitoring procedures are based on measurements of concentrated, large-volume environmental samples, including biota, as well as of regularly collected samples from waste storage tanks, etc.

B. COLLECTION AND FIELD TREATMENT OF SAMPLES, THEIR FURTHER HANDLING, AND PREPARATION

For the collection of a representative sample for radioactivity measurements there are a few general rules to be stated.

- (i) The sample, as well as the number of its separate parts (individuals, items), must be large enough to obtain a statistically reliable result.
- (ii) The site should not be disturbed by uncontrolled human activity or pollution affecting the process to be studied.
- (iii) The site should contain sufficient sample material for long-term studies.

1. Water

The volume required for water samples depends on which radionuclides will be measured and their expected concentrations. For instance, measurement of uranium in seawater requires only 1 liter or less, but measurement of ^{241}Am (from global fallout) requires 100 liters or more.

It is often difficult to take a representative water sample, especially if the contamination is caused by a local (point) source. The releases from this source might also vary considerably with time, as will the water concentration. In some cases it is important to distinguish between total and filtered water. Often a filter pore size of 0.45 μm is suitable. The knowledge of both physical and chemical forms of the radionuclides are of interest.

For sampling at different depths commercially available special bottles of different volume up to 250 liters, have been developed, e.g., Nishkin or Bodman bottles. An apparatus for filtration at different depths also has been constructed (Krishnaswami *et al.*, 1976). Some kind of preconcentration in the field is an advantage if large volumes are needed. For instance, sorption of radionuclides onto different chemicals, such as cesium onto ammonium molybdophosphate and transuranium elements onto aluminum dioxide, has been employed (Nevissi and Schell, 1975). Freshwater can be evaporated in the laboratory. When large volumes are required, precipitation techniques are to be preferred, especially for seawater with its salt content. Plutonium, americium, and other actinides can be precipitated with mixed hydroxides and carbonates of calcium and magnesium by adding sodium hydroxide to the sample (Hodge *et al.*, 1974).

2. Biota

For biological samples the number of individuals included in a sample and/or the sample size must be large enough. For terrestrial samples such as vegetation and soil, materials from a specified area can be collected to determine the inventory per unit area. It is recommended that foreign matter be removed from biological samples in connection with the sampling. Sometimes the samples have

to be dissected into different parts, ground, and homogenized. At collection, note should be made about weather conditions, snowcover, screening from tree canopies, disturbances from animals or humans, and other interfering parameters of interest and importance for interpretation of the results. If samples have been frozen, additional water will be formed at melting. If the sampling expedition is carried out for several days or weeks, preservation of the samples is needed. Freezing by solid CO_2 can be employed; however, it is somewhat more convenient to use ethanol in closed cans. We have successfully tested the use of sodium disulfite in collection of marine biological material (5 g/kg of wet sample). At drying and ashing one must recognize the risk of losing low boiling point elements such as iodine (boiling point 184°C), cesium (678°C), etc.

One compound can be more volatile than another. As an example, Tc_2O_7 (boiling point 130°C) can be converted to the more stable TcO_4^- by treating the samples with ammonia. The samples can then be ashed at 550°C without losses.

From a biological point of view it might be an advantage to use concentration values on a wet weight basis, but in most cases the use of a dry weight basis gives more consistent results; the recommended drying schedule is 24 h at 105°C .

3. Soil and Sediment

The collection of soil and sediment cores poses several problems. The risk of contamination of deeper layers from higher surface activity is obvious. For sediments the resuspension and mixing of the upper layer is difficult to avoid. Compression of the cores at collection makes accurate depth determination impossible. When sediment cores are frozen for storage, radial migration of elements has been observed (A. Aarkrog, personal communication). Special techniques (often sophisticated) for sediment coring have been developed (Kannevorff and Nicolaisen, 1973).

For further treatment the cores are cut in sections, dried, and homogenized. In chemical separation procedures it is generally not necessary to perform complete dissolution (with HF) of silica as demonstrated by international intercomparison exercises. It is sufficient to leach the samples with aqua regia.

4. Air

Radon and isotopes of other noble gases can be trapped on activated charcoal at low temperature. The monitoring and measurement can sometimes be performed directly on the charcoal trap. When heated, the gases are released and can be transferred to another counting vial or reconcentrated and brought to the laboratory.

For measurement of particulate radionuclides, including radon decay products, filter samples are taken. The pore size is one to several micrometers

in general. Control of the flow rate is important to prevent breakage of the filters.

Experience has shown that accurate determination of the amount of air drawn through a filter during field conditions is difficult. The activity of ^7Be in the filter can be used as a normalization factor (Ingemansson, 1982).

If radiochemical separation and dissolving of the filters with acids is desired, fiberglass filters should be avoided. Commercially available filters of polystyrene (Microsorban) give low ash weight and are suitable.

For direct measurement of α -particle emitters, Teflon filters give the best energy resolution due to a more superficial deposition of the airborne particles.

C. MONITORING PROGRAMS

The redistribution of natural radioactivity by human action and the releases of artificial radionuclides from nuclear installations, from nuclear weapons tests, and from the use of radionuclides in hospitals, laboratories, and factories have made it necessary to establish local, regional, or global survey programs. The primary objective of such surveys is to provide assurance that the dose equivalent to humans is below some prescribed level, as recommended by ICRP or specified in governmental regulations or guidelines. This guidance may relate to annual intakes of radioactive material, and usually implies that a few radionuclides dominate (critical radionuclides) via some special pathways (critical pathways) involving the transfer processes to critical groups of people: for instance, local deposition of ^{131}I and the pathway ^{131}I -grass-cattle-milk-children from a nuclear installation, or global fallout of ^{137}Cs and the pathway lichen-reindeer meat-reindeer breeders from nuclear weapons testing.

The primordial radionuclides are sometimes subject to considerable redistribution by humans, as discussed in Section II,A,3. This requires at least some short-term survey of changes in levels of radon and its decay products in air indoors and outdoors at locations where people are regularly present. Uranium, thorium, and their decay products in dust particles (aerosols) in various industries should also be observed.

The design of a routine monitoring program outside the boundaries of a local installation should be related to the results of a preoperational investigation, giving information on critical radionuclides, pathways, and groups of people, and also on the preoperational level of radioactivity in the environment. There may be natural anomalies to be considered. Attention should also be given to levels of radionuclides other than those to be monitored but which may influence the interpretation of the measurements, e.g., radon in groundwater and ^{137}Cs and ^{90}Sr in soil. Operational surveys of surface deposition, soil content, vegetation contamination, and food content are then performed concerning critical radionuclides such as ^{65}Zn , ^{131}I , ^{137}Cs , and ^{60}Co . Other released radionuclides may

also be of interest if large populations are involved. During normal operation long-lived radionuclides may be sampled annually or quarterly, or composite samples analyzed. Short-lived radionuclides should be sampled at intervals of no longer than two to three half-lives. If large fluctuations occur in the release rate, continuous sampling and analysis may be necessary, requiring rapid sample treatment and measuring methods.

Fallout from nuclear weapons tests has been studied in many countries, and there is, in fact, a global monitoring program working. The results have been periodically surveyed in considerable detail by UNSCEAR since its establishment in 1955 and in reports published later (UNSCEAR, 1958, 1962, 1964, 1966, 1969, 1972, 1977, 1982).

Besides standard sampling programs for soil, water, air, and some foodstuffs, some bioindicators also have been studied, such as lichen, moss, algae, shellfish, etc. Sludge from community waste systems is also an indicator of interest. It is then important that the relationships for estimating dose equivalent commitments to exposed persons are known.

For all monitoring programs of common interest it is of great importance that comparable and reliable methods of sample collection and analysis be employed. Representative samples have to be available for the entire survey period. Routine environmental monitoring programs should also be supplemented by radioecological studies to obtain more information on transfer mechanisms and transfer coefficients of radionuclides. This allows for confirming or correcting the assumptions on which assessments of dose equivalent commitments are based.

IV. Analytical Procedures and Measurement of Radioactivity

A. GENERAL ASPECTS OF METHODS FOR QUALITATIVE AND QUANTITATIVE ASSESSMENT OF RADIONUCLIDES

For quantitative measurement of radionuclides it is often necessary to concentrate the elements by physical and/or chemical methods. This is almost always the case for α and β emitters due to the limited range of these particles. Preconcentration is also done for γ spectrometry when large samples are required at low activity concentrations. For low-energy γ rays attenuation in the sample is another limiting factor. Precipitation from large volumes of water is an example which involves both physical and chemical concentration processes. Many procedures for separation and source preparation can be found in the "EML Procedures Manual" (Volchok, 1981).

The recovery in radiochemical procedures is seldom 100% and is not even constant. Some kind of yield determination is then needed. This is best accomplished by adding to the sample a radioactive isotope with suitable physical half-life and decay characteristics and which can be brought into the same chemical

form as the radionuclide of interest. (See Table IV.) Such isotopes are not always available, and stable isotopes are sometimes used instead. The radiochemical yield is then determined by other types of measurements such as atomic absorption. In some cases even a chemically similar element (stable or radioactive) can be used, for example, rhenium for ^{99}Tc , ^{133}Ba for ^{226}Ra , and ^{244}Cm for ^{241}Am .

Final source preparation has quite different requirements, depending on the type of measurement. Alpha-particle counting and spectrometry require very thin sources to avoid energy straggling and consequent poor energy resolution. Figure 4 gives an example of the broadening and energy degradation of an α peak from a thickly deposited source (Holm and Persson, 1981). The most frequently used method to obtain good sources for α spectrometry is electrodeposition. Good radiochemical separation from other α -particle emitters with similar energies is necessary. For the measurement of β -particle emitters the radio-

TABLE IV
STANDARD REFERENCE MATERIAL FOR γ -RAY POINT SOURCES^a

Parent radionuclide	Half-life	Photon energy (keV)	Photon probability per decay
^{241}Am	433 ± 2 yr	26.35	0.0258 ± 0.0022
		59.54	0.363 ± 0.004
^{109}Cd	453 ± 2 d	88.04	0.0373 ± 0.0006
^{57}Co	270.9 ± 0.6 d	122.06	0.8559 ± 0.0019
^{139}Ce	137.65 ± 0.05 d	165.85	0.8006 ± 0.0013
^{203}Hg	46.59 ± 0.05 d	279.19	0.815 ± 0.008
^{51}Cr	27.704 ± 0.002 d	320.08	0.0980 ± 0.0010
^{113}Sn	114.9 ± 0.1 d	391.69	0.6490 ± 0.0020
^{85}Sr	64.85 ± 0.03 d	513.99	0.980 ± 0.010
^{207}Bi	38 ± 3 yr	569.67	0.978 ± 0.005
		1063.62	0.74 ± 0.03
		1770.22	0.073 ± 0.004
^{137}Cs	30.0 ± 0.2 yr	661.65	0.899 ± 0.004
^{94}Nb	$(2.03 \pm 0.16) \cdot 10^4$ yr	702.63	1
		871.10	1
^{54}Mn	312.5 ± 0.5 d	834.83	0.999760 ± 0.000002
^{88}Y	107 ± 1 d	898.02	0.934 ± 0.007
		1836.04	0.9935 ± 0.0003
^{65}Zn	244.1 ± 0.2 d	1115.52	0.5075 ± 0.0010
^{60}Co	5.271 ± 0.001 yr	1173.21	0.99900 ± 0.00020
		1332.46	1
^{22}Na	2.602 ± 0.002 yr	1274.54	0.99940 ± 0.00020

^aThese sources are available from the National Bureau of Standards, Washington, D.C. (U.S.A.). This table is adapted with permission from Report No. 58 (NCRP, 1978).

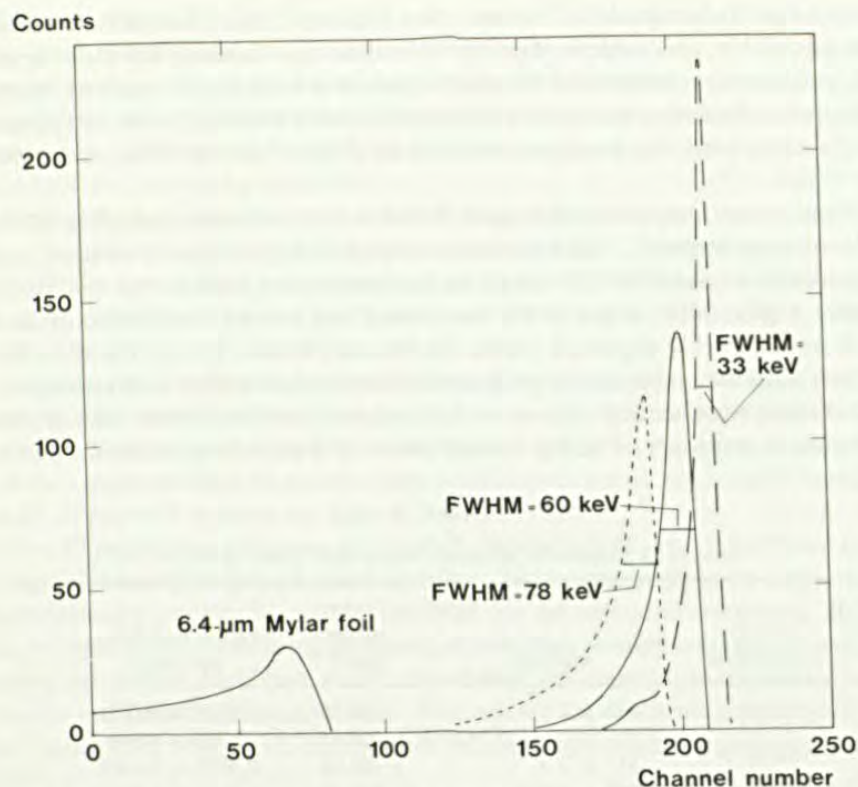


Fig. 4. Broadening and energy distortion of α spectra from $^{239} + ^{240}\text{Pu}$ sources of various thicknesses. The peaks representing FWHM 60 keV and 78 keV result from badly produced sources. The peak at FWHM 33 keV represents what normally can be obtained after successful separation and electroplating (450-mm², surface-barrier silicon detector). The left-hand spectrum is obtained by covering the source giving FWHM 33 keV with 6.4- μm Mylar foil. [From Holm and Persson (1981).]

chemical purity is also important because the particles are not monoenergetic and the counting devices have poor energy resolution, or none at all.

Alpha and β sources are often prepared as disks. For α -particle emitters such sources should be clean. Self-absorption corrections are generally not required when performing α spectrometry. For β emitters, correction for self-absorption must be done when gravimetric yield determinations or other methods causing thick deposits are used. To give good counting efficiency the diameter of the disks should fit the sensitive area of the detectors if possible. Liquid scintillation counting is generally the method of choice for β emitters with low maximum energies.

The source preparation for γ emitters is dependent on the type of detector, detector configuration, and γ energy. Samples are generally dried, ground,

homogenized, and put into containers for measurement. The detectors are calibrated for one or several configurations of the final source container. For radionuclides emitting high-energy photons, sensitivity can be increased by ashing and/or compressing the samples to obtain maximum activity in a given volume. Thin samples over larger surfaces are used at low γ energies, or a radiochemical procedure such as that for α - and β -particle measurement is performed.

The α -particle emitters are identified by specific α energies (one or several) characteristic of the radionuclide. For surface-barrier detectors or gridded ion chambers, the detector efficiency does not vary with α -particle energy.

The identification of β -particle emitters is more complicated. It must be done by β spectrometry and using a Fermi-Curie plot to determine the maximum energy. Sometimes studies of physical half-lives or decay products are possible for identification and determination.

Energy calibration can best be done using radionuclides that emit conversion electrons. For many types of detectors, such as GM gas-flow counters, no energy calibration is obtained, and energy discrimination can only be done by placing absorbing material between source and detector. Although liquid scintillation counting is probably the most frequently used counting system for β -particle emitters, counting systems differ considerably with respect to sample preparation and efficiency calibration, and will be dealt with later.

Many α - and β -particle emitters also emit photons. However, their energies, or numbers per disintegration, may not be suitable for practical detection.

Gamma-ray spectrometry offers a nondestructive method of sample analysis and this method should be used when possible. The identification of γ emitters is more complicated than that for α -particle emitters since the number of possible energies and radionuclides is much greater. Good energy resolution is then important.

Detector efficiency varies with energy; correction for various densities and elements in the source matrices must also be considered. For calibration, radionuclides with accurately known γ energies and absolute intensities are mixed and homogenized in a matrix material closely simulating the actual source matrices. In modern γ spectroscopy, evaluation of pulse-height distribution, identification, source concentration, etc., is done by computer.

Solid-state detectors used for α -particle spectrometry (surface-barrier, ion-implanted, diffused-junction detectors) are mounted in a lighttight chamber which is evacuated. The shielding provided by this chamber is enough to keep the background low (~ 1 count/day) in any specific energy region.

For β -particle counting, shielding with lead and/or reduction of background counts through the use of coincidence and anticoincidence techniques are used. Backgrounds depend on detector sizes and type of instrumentation.

Heavy shielding of the detector is the most efficient means for reducing the background for γ spectrometry. Low radionuclide content of surrounding material (shielding barriers, detector components, building material, air, etc.)

secures a low and stable background. Anticoincidence and coincidence techniques can be used for specific radionuclides. As for all types of detectors, the background depends also on sensitive areas or volumes and construction material of the detectors themselves.

B. METHODS FOR IDENTIFICATION AND ASSESSMENT OF GAMMA EMITTERS

1. *In Situ Determination by Gamma Spectrometry:* *Exposure Rate Measurement*

As mentioned in Section III,A, *in situ* γ spectrometry sometimes makes it possible to identify and directly determine ground and air concentrations of γ -emitting radionuclides.

The mobile spectrometry systems of interest for such measurements are based on large-volume Ge(Li), Ge(hp), or NaI(Tl) detectors connected to a multichannel analyzer (MCA) with sufficient number of channels and recording equipment (Fig. 5). The system should cover an energy range of up to 3 MeV (^{232}Th , 2.62 MeV) if only natural radionuclides are to be studied. If measurement of direct radiation from a nuclear power reactor is also to be considered, this energy range should be extended to 8 or 9 MeV (^{15}C , 5.3 MeV; ^{16}N , 7.11 MeV). The activity concentrations and exposure rates of γ -emitting radionuclides surrounding the detector are evaluated from the full energy absorption peak areas in the pulse-height distribution. This requires a proper calibration of the energy and angular efficiency of the detectors for photons of various energies and directions, as discussed below.

a. Energy Resolution. The resolution of a Ge(Li) system is markedly superior to that of a NaI(Tl) system, as illustrated by FWHM values at 1 MeV of 2–3 keV and 60–70 keV, respectively. Therefore Ge(Li) should be used if the contribution of small amounts of artificial radionuclides has to be determined in the presence of the complex natural background. A disadvantage is the large amount of data produced, as well as the need for continuous cooling of the detector by liquid nitrogen. A Ge(hp) detector requires cooling only while being used for measurement, but cooling down takes some hours and hampers rapid use. A closed-end coaxially drifted Ge(Li) detector with an active volume of 60 cm³ provides a full energy peak efficiency at 1.33 MeV of 10–15% of the corresponding efficiency of a 7.5-cm-diam $\times$ 7.5-cm NaI(Tl) detector, as measured with a point source at a 25-cm distance. A NaI(Tl) system can be operated with less expensive electronics. Its poorer resolution requires only a 400-channel MCA and less data handling.

Larger detector volumes and higher intrinsic efficiency makes it possible to perform meaningful measurements within 10 minutes in situations where the limited resolution of the NaI(Tl) system is acceptable. The NaI(Tl) detector re-

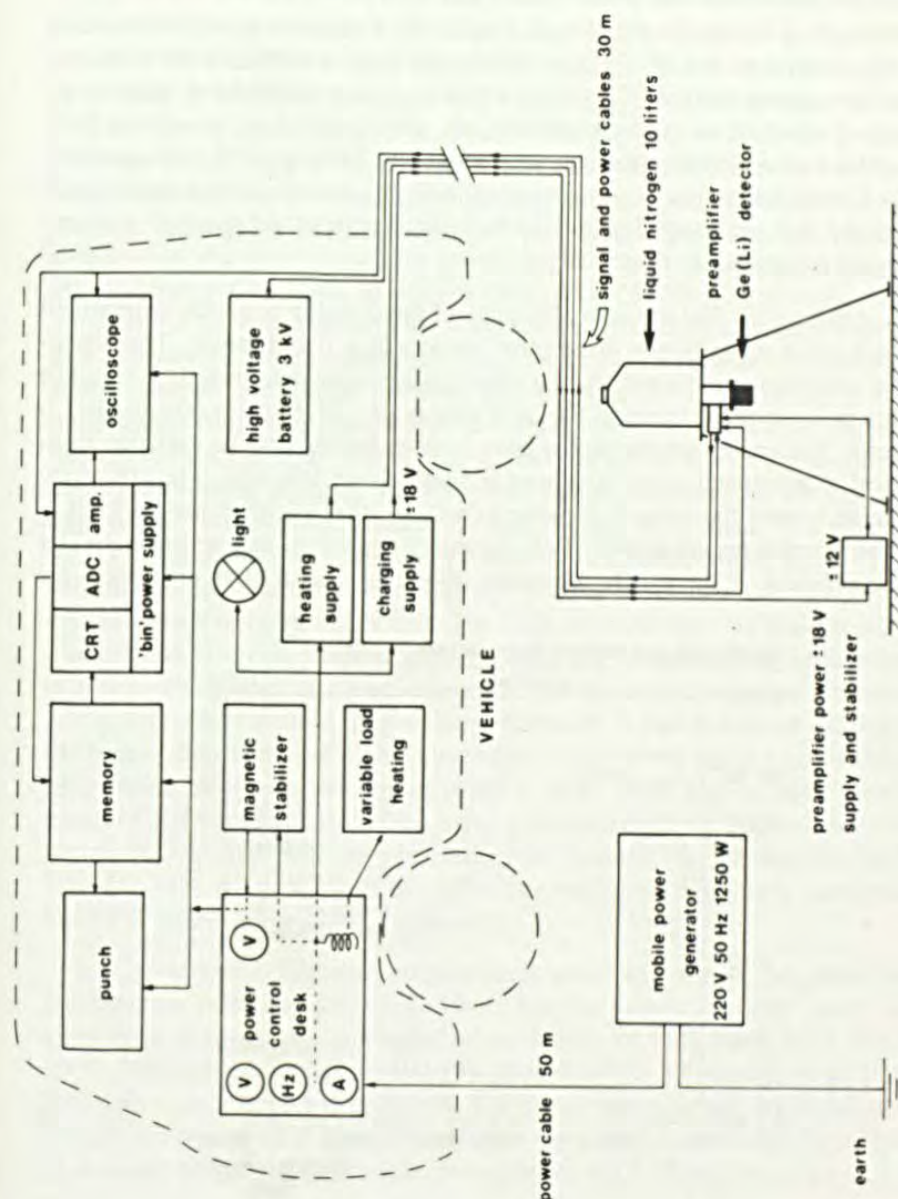


Fig. 5. System diagram of field Ge(Li) γ spectrometric equipment. [From Finck *et al.* (1976).]

quires no cooling, but the crystal material must not be subjected to rapid temperature changes, which can create cracks and spoil the detector.

Normally a Ge detector is mounted vertically 1 m above ground underneath the liquid nitrogen dewar; 10 liters of nitrogen may be sufficient for 3 days of field use without refilling. The field-effect transistor amplifier is directly attached to the detector cryostat and requires stabilized electric power (± 12 V from, for example, NiCd batteries with a capacity for at least 12 h of operation if the normal continuous charging from a mobile generator fails). It is advisable to supply the operating voltage for the detector from a separate, specially designed battery pack.

b. Efficiency. The intrinsic efficiency of the detector is usually determined by calibration using standardized point sources at a 1-m distance. The dependence on photon energy and angular direction has to be determined. A result of such a calibration is shown in Fig. 6 (Finck *et al.*, 1976) for a 60-cm³ Ge(Li) detector. The energy dependence is considerable, but the angular variation from vertical to horizontal γ -ray directions is small above 100 keV. The efficiency variation around the vertical detector axis (azimuthal angle) is insignificant.

The detector response under field geometry conditions can be illustrated by the coefficient $\dot{N}_F/\phi$ which represents the total registration efficiency for

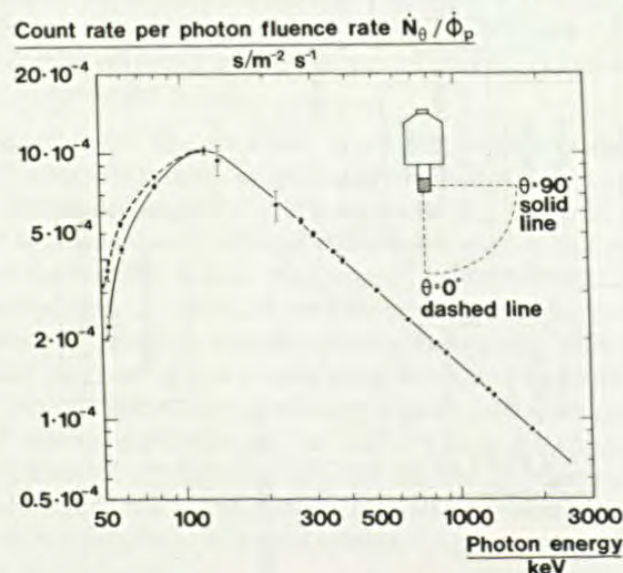


Fig. 6. Intrinsic efficiency $\dot{N}/\phi$ of 60-cm³ Ge(Li) detector at angles of incidence $\theta = 0^\circ$ and $\theta = 90^\circ$, as a function of photon energy. The quantity $\dot{N}$ is the count rate of a representative full energy peak of a radionuclide giving the primary photon fluence rate ϕ_P at the detector. [From Finck *et al.* (1976).]

primary photons. Here $\dot{N}_F$ is the total count rate 1 m above a plane surface and ϕ is the photon fluence rate at the detector. The coefficient $\dot{N}_F/\phi$ is a function of the energy and distribution of the primary photons. The most important parameters affecting the photon distribution are the source activity distribution in the ground and energy of emitted γ rays. Mass attenuation coefficients and densities of soil and air are also influential. The distribution of artificial radionuclides in the ground may usually be characterized as plane (on the surface), exponential, or uniform. If A is the soil activity concentration of a radionuclide, ϕ/A can be calculated. Finck *et al.* (1976) have shown that the angular distribution of ϕ at 1 m above the ground surface has a maximum for θ between 70° and 90° for all types of distributions. At 88.8° – 89.7° a strongly marked maximum is shown, especially for plane source distributions, at all photon energies. This means that the most probable primary photon originating from a plane source, and incident on the detector, has traversed 50–150 m of air. It is thus important to emphasize that the area directly below the detector contributes very little to the total photon fluence rate ϕ . Natural radionuclides are usually uniformly distributed in depth, whereas fresh fallout radionuclides are distributed only on the surface. For instance, for superficially distributed ^{137}Cs (662 keV) 90% of ϕ originates from distances within 60 m (corresponding area $\sim 11,000$ m²). If the same activity is uniformly distributed in the soil, this distance decreases to 8 m (area ~ 200 m²). The large area surveyed by the detector has a smoothing effect on minor area inhomogeneities of radionuclide concentration A in soil, which otherwise lead to uncertainties in standard sampling procedures. A major problem in field spectrometry, however, is that ϕ depends strongly on the type of distribution of A , and one single measurement will not give sufficient information on A as a function of depth d , $A(d)$. With two or more measurements of different angular sections using a shielded detector, approximate distribution information will be obtained. For instance, in the angular section $\theta = 80^\circ$ – 90° , ϕ for 662 keV is 15% of the total ϕ for a uniform distribution, but 58% of that for a plane distribution.

c. Corrections. The soil composition is not crucial for the calculation of ϕ for photons with energy 0.2–7 MeV, but for lower energies more exact knowledge is necessary. A change of air density of 11% (0°C , 1027 mbar to 20°C , 1000 mbar; 1 mbar = 100 Pa) gives rise to a 3% change of ϕ/A at 60 keV with a plane source distribution. For higher energies and depth-distributed activity the change of ϕ is smaller. Similar corrections are needed for a change of detector height: 10 cm at 1 m gives rise to a 3% change in ϕ/A .

d. Conversion to Exposure Rate $\dot{X}$. The coefficient $\phi/\dot{X}$ has been tabulated by Beck (1972) for different ground source distributions. For decay series, $\dot{X}$ refers to the total exposure rate of all radionuclides included in the chain. The

coefficient can generally be written as

$$\frac{\phi}{\dot{X}} \sim \frac{\sum_{j=1}^m (d\phi/dt)_j}{\sum_{j=1}^m (e/\bar{W}) \int_0^{E_j} (\mu_{en}/\rho)_{air} [{}_p(d\phi/dt)_j + {}_s(d\phi/dt)_j] E dE} \quad (1)$$

where ${}_p(d\phi/dt)_j$ is the primary photon fluence rate with energy ${}_p E_j$ at the position of the detector and ${}_s(d\phi/dt)_j$ is the fluence rate of the scattered photons, traceable to the j th photon in the decay. The quantity $\bar{W}$ is the mean energy expended in air per ion pair formed, $(\mu_{en}/\rho)_{air}$ is the mass energy absorption coefficient of air at photon energy E , and e is the charge of the electron. The ratio $\phi/\dot{X}$ depends on the same parameters as ϕ/A . As seen from Eq. (1), a variation in the fluence due to a change of the source distribution will only affect the quantity $\phi/\dot{X}$ through altering of the scattered fluence and the primary fluence of other energies. Therefore $\phi/\dot{X}$ is less sensitive to a change of source distribution than ϕ/A , varying only 25–30% between a plane and a depth-distributed source.

2. Preparation and Measurement of Large Sources

Environmental samples containing γ -emitting radionuclides can frequently be assayed by preparation of sources using nondestructive methods and measurement of emitted γ rays. The high penetration of γ rays permits comparatively simple source preparation and the use of large sources. Normally the measuring methods of choice are based on γ spectrometry, thus giving good or even excellent selectivity in discriminating among different radionuclides.

a. Preparation of Sources. The field and laboratory treatment of bulky environmental samples may involve drying, grinding, evaporation, wet-ashing, etc., all performed in such a way that the resulting product still contains all radionuclides of interest and is ready for preparation of suitably shaped sources.

Some materials may require proper preparation measures to prevent bacterial decomposition. Increased measuring efficiency for low-density material is obtainable by compressing the source in a hydraulic press at a reproducible pressure to a standard format fitting a measuring can. As the normal geometric form of a NaI(Tl) or Ge detector is cylindrical, the shape of a source should be matched accordingly. Usually the treated sample material is placed in a cylindrical container (vial, can) of suitable height and diameter and made of plastic material (PVC, etc.) free of radioactive contaminants. In the case of large samples another useful source geometry is obtained with the Marinelli beaker, where the source is distributed around the detector (May and Marinelli, 1960); this configuration is sometimes called "hat geometry." A close-fitting lid should always be used, and it should be carefully checked that the source is homogeneously packed and fully fills the container. Weight and volume of a

source should always be recorded. The content of ^{238}U and ^{232}Th can be determined by measuring γ -emitting decay products, but this requires that the source container be sealed fully gastight and stored at least one month before measurement. Uranium-235 (the actinium series) cannot be determined by this method due to interfering γ -ray energies from the ^{238}U series.

b. Measurement. Large NaI(Tl) crystals offer high detection efficiency for γ rays. If the radionuclides of interest are few, and identified in advance, γ spectrometry with this scintillator is still a very useful method for assaying environmental radionuclides. As it is difficult to predict the counting efficiency for most arrangements of detectors and large sources, a calibration of each system and radionuclide is necessary. It should be performed using a source with known activity in a matrix simulating the environmental samples as closely as possible. For instance, sawdust of the same density as ground vegetation material can be homogeneously spiked by water-dissolved γ emitters such as ^{137}Cs .

Usually more than one radionuclide is present in the sample, which requires some kind of "spectrum stripping" analysis. This is conveniently done for spectra with well-defined photopeaks by a least-squares method programmed for computer use with linear or nonlinear regression technique (Salmon, 1961; Heath *et al.*, 1967). The method requires availability of experimentally determined standard spectra for all the different radionuclides involved, a well-stabilized counting system, and a low and stable background.

The poor resolution of NaI(Tl) spectrometers makes interpretation of spectra involving unknown radionuclides rather difficult. With more than 10 times better resolution, Ge(Li) and Ge(hp) semiconductor detectors improve considerably the possibilities of identification and quantitative determination of γ -emitting radionuclides. The total efficiency of a Ge detector is considerably lower than that for NaI(Tl), due to lower atomic number (32 for Ge and 53 for I) and smaller size (50–100 cm³). However, with a Ge detector the counts S in a full energy peak are compressed into a much smaller band than with NaI(Tl), giving a correspondingly much lower background B and resulting in a substantially increased figure of merit $(S - B)^2/B$. Thus at present most measurements of environmental low-activity samples are performed using Ge spectrometry with counting efficiencies at 1.3 MeV of the order of 5 to 40% relative to a 7.6-cm-diam $\times$ 7.6-cm NaI(Tl) detector at a source distance of 25 cm.

Large source geometries require energy and efficiency calibrations in the same way as discussed above for NaI(Tl) detectors. Evaluation of Ge spectra is simplified compared with evaluation of NaI(Tl) spectra, since the essential information (γ -ray energy and photon fluence rate) is represented by the position and magnitude of individual photopeaks, as indicated by Fig. 15 in Section IV,B,5. Computer programs such as that illustrated in Fig. 7 have been developed for direct analysis of Ge spectra (4000–8000 channels) using an energy-

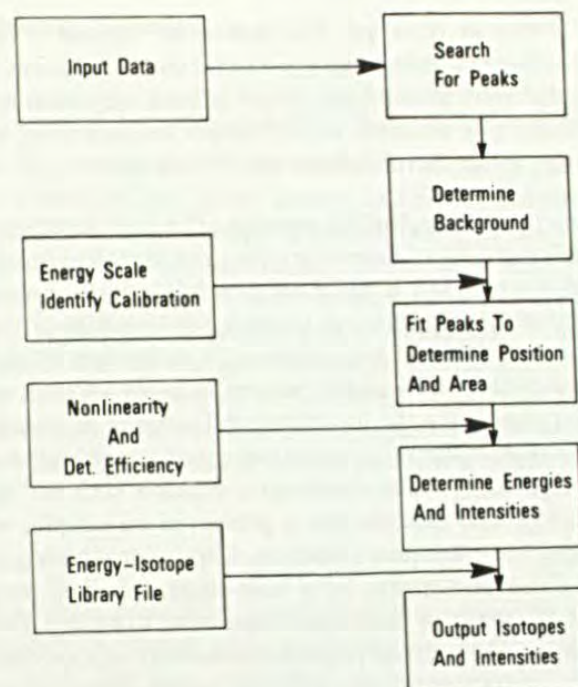


Fig. 7. Block diagram of computer programs used in the analysis of Ge(Li) γ -ray spectra for quantitative radionuclidic analysis of radioactive samples. [From NCRP (1978).]

radionuclide library file and representing peaks by combined gaussian and exponential functions and a suitable function for the background continuum (often linear). Instrumental linearity and stability are important; multichannel systems should preferably show linear response within ± 0.3 channel.

c. Whole-Body Counting. A particularly large source is the human body itself. As most studies and monitoring programs of radioactivity in the environment are related to efforts to protect people against radiation and to minimize the transfer of radionuclides to human beings, the direct measurement of radioactivity in the human body is a natural complement to other environmental studies of radioactivity. After 1950 it was realized that nuclear weapons tests resulted in a worldwide spread of radionuclides in the biosphere. Some of these nuclides have reached humans through different food chains and contaminated air. Of special interest are radioisotopes of Sr, Cs, I, and Pu and other transuranium elements. The γ -emitting isotopes ^{134}Cs , ^{137}Cs , and ^{131}I have photon energies in the range 300 to 700 keV, which are easily detected outside the body, while the pure β emitters ^{89}Sr , ^{90}Sr , and ^{14}C are not detectable by external *in vivo* counting. Plutonium-239 with its 13- to 20-keV photons is detectable, if located in the lungs, using specially designed detectors. It is now well established

that the internal collective dose commitment to the population from all fallout components is, to a large extent, delivered by ^{90}Sr , ^{137}Cs , ^{131}I , and ^{14}C . Cesium-137 (γ , 662 keV) and ^{131}I (γ , 364 keV) have therefore been subject to whole-body counting measurements in various counting geometries in low-background rooms. An often used method is the 42-cm chair geometry (Marinelli, 1957) shown in Fig. 8.

A large NaI(Tl) scintillation detector, usually 20 cm diam $\times$ 10 cm, views most of the body at approximately the same distance (legs exempted). This results in a roughly constant response for various parts of the body located at the same depth. An illustration of the minimum detectable activity (MDA) for ^{137}Cs measured as the full energy peak of 662-keV photons with this detector is given by the following data: about 20 Bq are detected at a confidence level of 0.997 and normal ^{40}K content (2 g K/kg body weight) with a measuring time of 30 min and a stable and carefully measured background counting rate of $57 \pm 4/\text{min}$ (3 SD) using the energy interval 600–720 keV.

Iodine-131 is present primarily in the thyroid and a more efficient measurement is usually made by positioning the detector about 10 cm from the neck of a person lying horizontally. By this method, a 10-min measurement will give an MDA of 2.5 Bq for ^{131}I .

In both cases the whole-body counting system must be properly calibrated. A known activity given to a patient for medical purposes may be measured several times in a suitable time period, during which excreta are collected and measured. The remaining net activity in the body is then obtained. Another

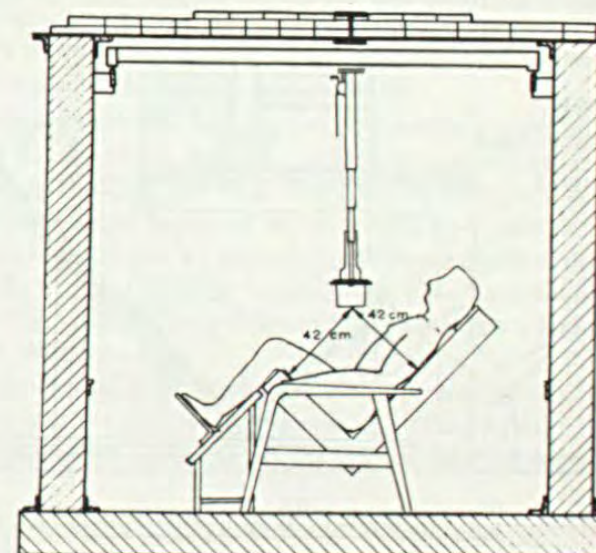


Fig. 8. Schematic representation of the detector-subject arrangement in a "42-cm chair." As shown, 42 cm is the distance from the detector to the seat and backrest of the chair.

calibration method utilizes specially designed anthropomorphic phantoms filled with a solution of the actual radionuclide, but simpler phantoms based on filled bottles of various sizes combined to simulate the body are also useful. Field measurements of γ -emitting radionuclides in the body have been performed with a simpler but transportable counting system and background shield (Naversten *et al.*, 1963; Miettinen, 1964), shown in Figs. 9 and 10.

A more simplified method utilizes only the body itself as a background shield; the detector is positioned in the lap of a sitting person bending slightly over the detector (Palmer, 1966). An MDA of about 200 Bq for ^{137}Cs is then obtained in a 10-min measurement. The lap method is, of course, more sensitive to changes in counting geometry and internal variations of the location of the actual radionuclide. For ^{137}Cs determination the overall uncertainty is estimated to be $\pm 20\%$.

Considerably more sophisticated systems have been developed for laboratory applications. A nearly 4π liquid scintillation counter with the person positioned in a 45-cm well has been described (Van Dilla *et al.*, 1967). Large plastic blocks of scintillating material are used today in several laboratories, giving 2π to 3π solid-angle geometry. If the contaminating radionuclide is known in advance, one or two radionuclides, for instance, ^{137}Cs and ^{40}K , are determined in a very short measuring time, 1 to 2 min, with a relative standard deviation of 3 to 5%.

3. Preparation and Measurement of Small/Thin Sources

The reason for working with small or thin γ -ray sources is either that only small samples are available or that the sample material must be treated radio-

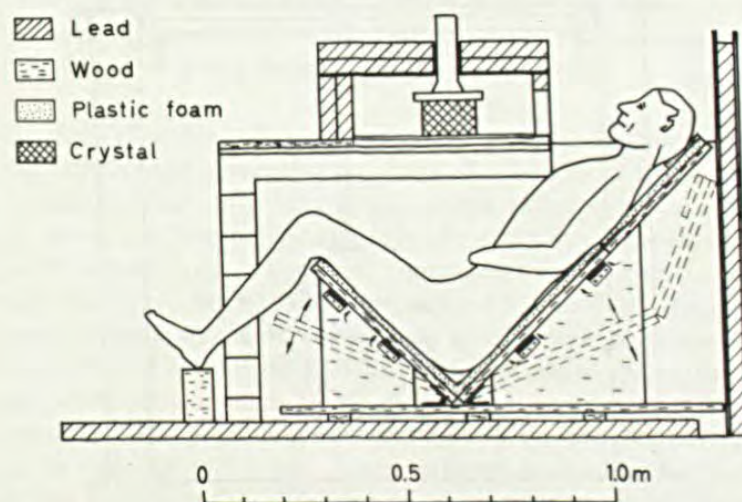


Fig. 9. Profile view showing the geometry of a semiportable whole-body counter with a folding chair. The lead shield of the side walls is not shown. [From Naversten *et al.* (1963).]

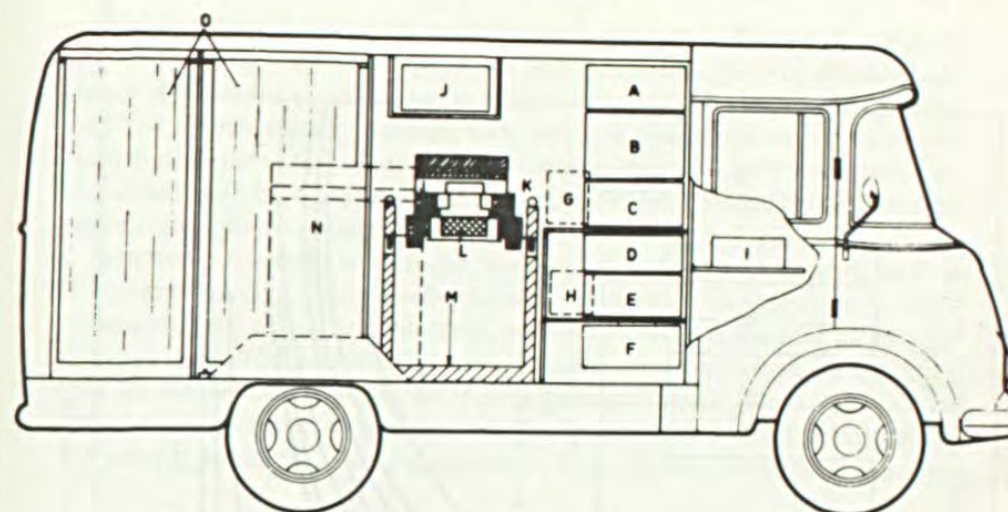


Fig. 10. General layout of instruments and facilities in a whole-body counting truck. A-F: Multichannel analyzer units; G-I: recording units; J: high-voltage set; K: 8-cm lead surrounding crystal; L: 12.5-cm-diam $\times$ 7.5-cm NaI(Tl) crystal; M: lead shield, 4 cm; N: sliding lead door, 4 cm (dashed lines); O: dressing cubicles. [From Miettinen (1964).]

chemically to concentrate the radionuclide or to separate interfering radionuclides and matrix material. Since it is almost always desirable not to destroy the original sample material, a nondestructive measurement method should always be considered even for small samples before further decomposition is started. Small sources with low activity can often be measured with sufficient accuracy in a NaI(Tl) well counter with good efficiency. For some purposes a Ge(Li) well counter may be even more efficient.

Typical performance of a NaI(Tl) well detector for a point source is illustrated in Fig. 11 (Knoll, 1979). From the diagram it is possible to estimate the efficiency for a source filling the well. The counting efficiency is close to 100% for a point source at the bottom of the well emitting γ photons in the range 10 to 120 keV. This means that a 1-Bq source will be measured with a SD of ± 0.13 and ± 0.04 Bq in 1 and 10 min, respectively, if the background is negligible. At higher energies the efficiency drops and above 400 keV a larger detector with thicker walls should be used.

A correction for self-absorption in the source matrix must be considered, especially for low γ energies. The appearance of sum peaks from cascade γ photons is sometimes a problem, but can also be used for separation of radionuclides.

Small sources of other shapes should be counted in a reproducible position in close proximity to the detector, which should have a high efficiency for the γ energies to be measured. A disk-shaped source is partly simulated by a point

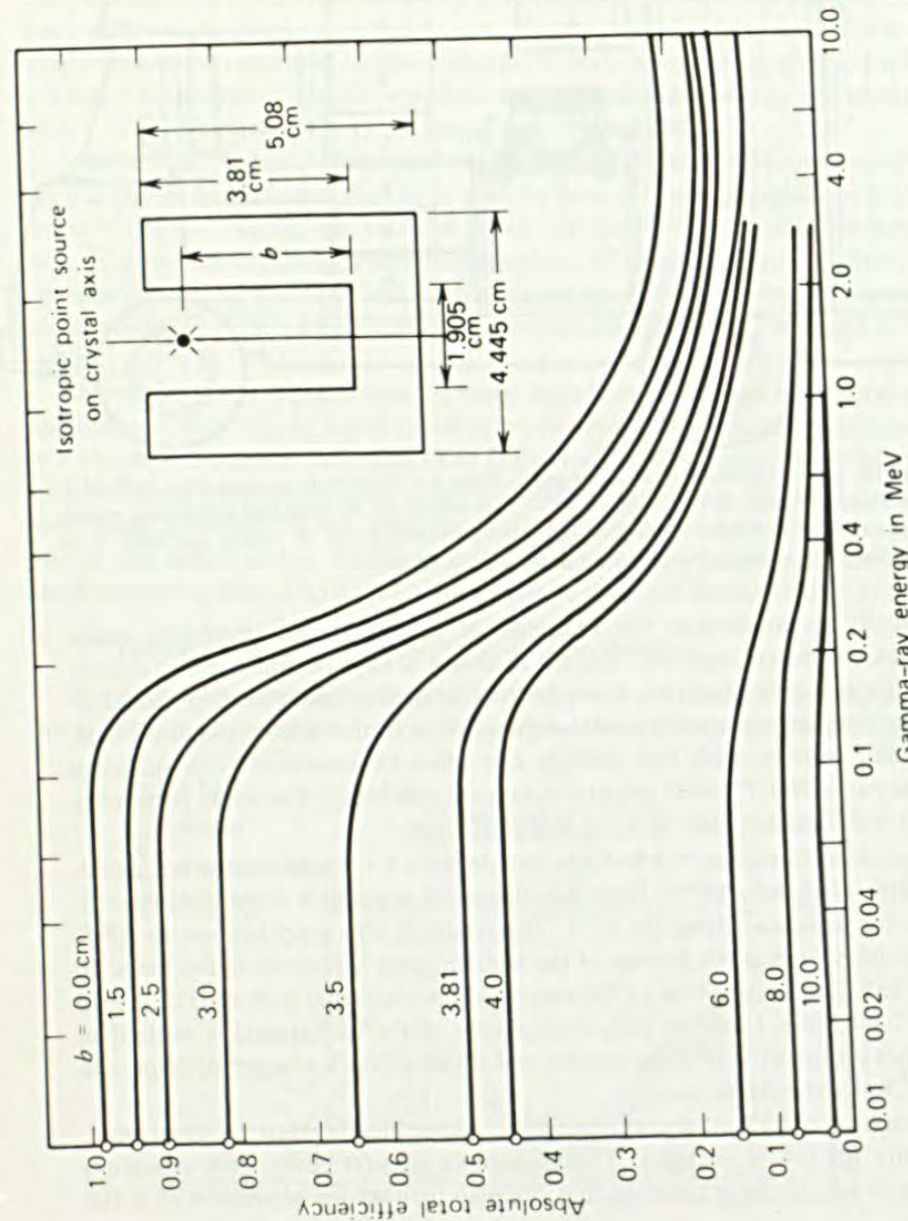


Fig. 11. The absolute total efficiency calculated for point γ -ray sources in a NaI(Tl) well-type detector with the dimensions shown. The parameter b is the source height above the well bottom. [From Knoll (1979). Copyright by John Wiley & Sons, Inc.]

source, and the efficiency obtainable for such a source is illustrated in Fig. 12 (Knoll, 1979) for a specified cylindrical NaI(Tl) crystal. The 50% absolute efficiency obtained for a source close to the detector surface ($\sim 2\pi$ geometry) drops very fast with increasing distance. Very thin disk-shaped sources (e.g., evaporated freshwater samples) are thus desirable as they will give negligible self-absorption and can be positioned very close to the NaI(Tl) detector; Ge detectors are less suitable for close contact.

Spectrometry should be included in almost all γ -measuring systems because it is rare when only a single radionuclide is present. With spectrometric techniques, the area within the peak for a specified γ energy is used for the activity determination. The peak-to-total ratio (the photofraction) plays an important role for the NaI(Tl) detector, as is illustrated in Fig. 13 (Knoll, 1979). The photofraction is that fraction of photon interaction events which appear in the full energy peak area. If several γ -emitting radionuclides are present in a sam-

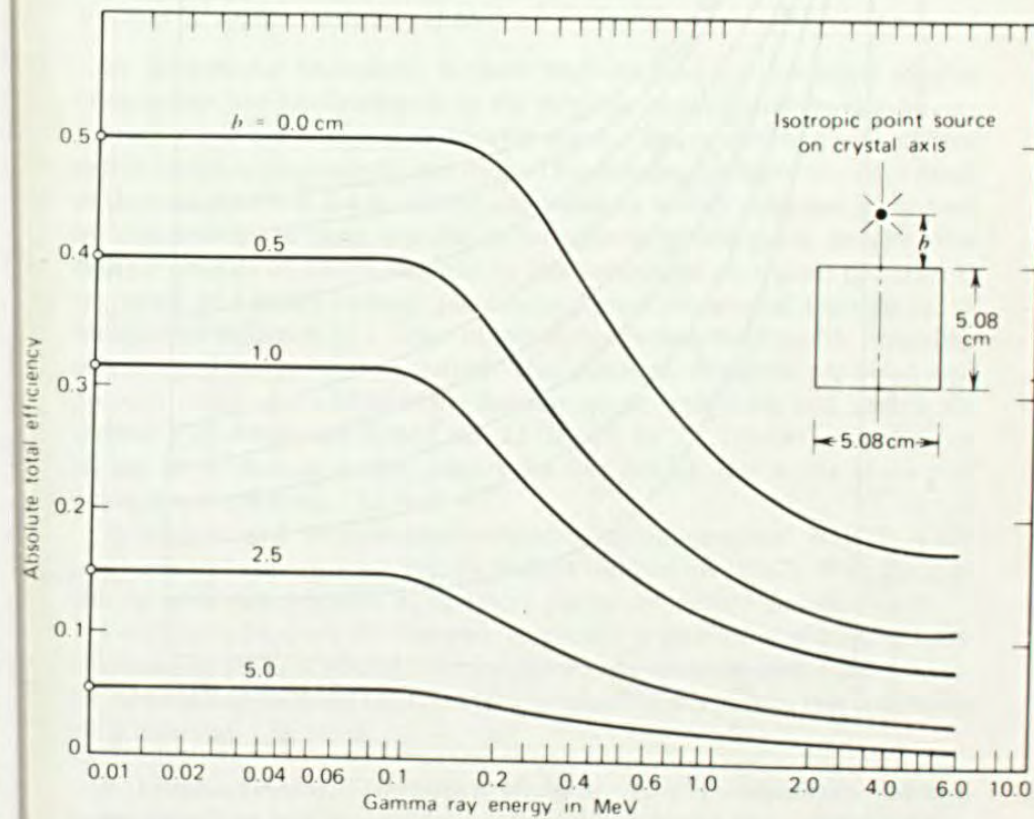


Fig. 12. The absolute total efficiency calculated for a 5.08-cm-diam $\times$ 5.08-cm NaI(Tl) detector at different source locations. [From Knoll (1979). Copyright by John Wiley & Sons, Inc.]

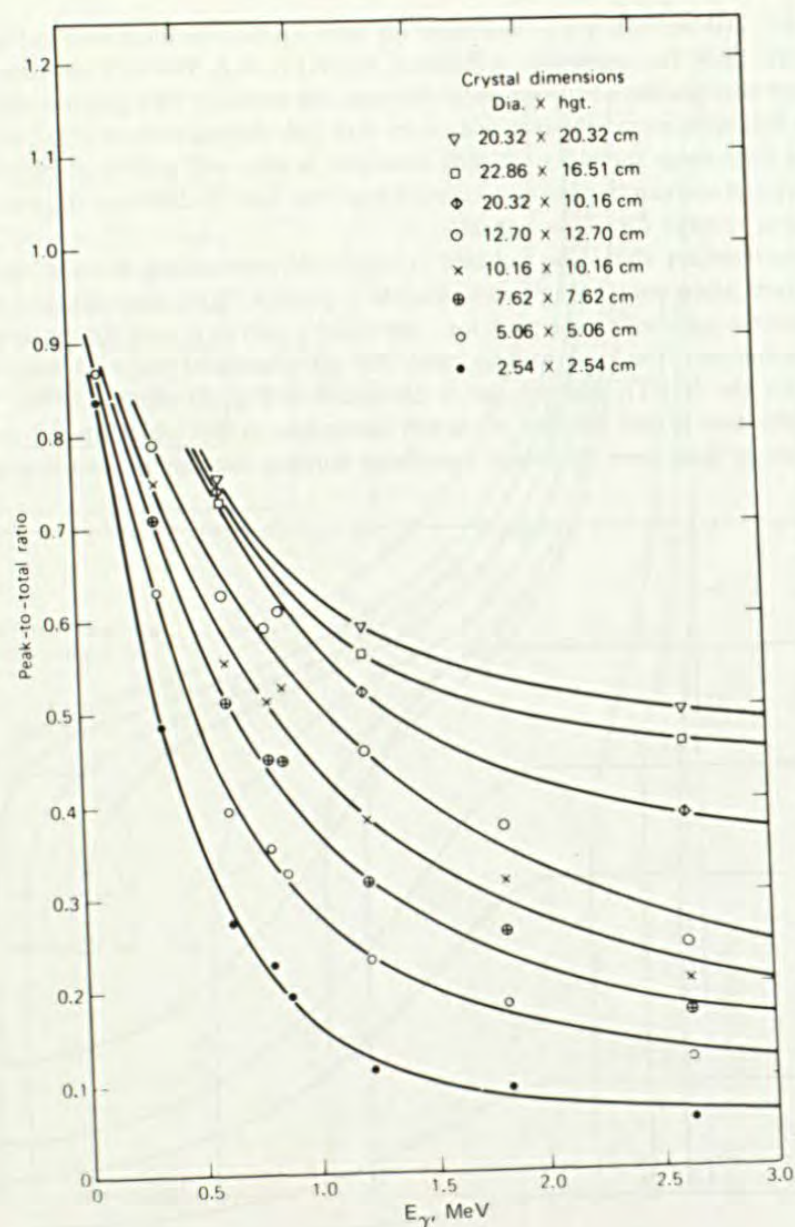


Fig. 13. Peak-to-total ratio (the photo fraction) for various cylindrical NaI(Tl) detectors for a point γ -ray source 10 cm from the detector surface. [From Knoll (1979). Copyright by John Wiley & Sons, Inc.]

ple, the use of a Ge detector may be the only way to obtain a complete quantitative measurement of the concentration of the various radionuclides in the sample.

In Fig. 13 the crystal dimension (7.62 cm diam $\times$ 7.62 cm) is the standard NaI(Tl) dimension for comparison with semiconductor detectors, particularly Ge(Li) or Ge(hp). These detectors show the same high intrinsic efficiency as NaI(Tl) at γ energies below 100 keV; at higher energies their lower efficiency is compensated by the much better resolution and small background in the narrow photopeak.

Low-energy x rays, such as the Mn K_{α} photons at 5.9 keV from a ^{55}Fe source, have to be measured by Si(Li) spectrometry or by proportional gas-flow detectors after radiochemical separation. Small and thin sources are usually necessary for coincidence or anticoincidence detector systems; they will be discussed in Section IV,B,4.

4. Special or Nonconventional Methods

a. Background Reduction. Sources prepared from environmental samples often contain low-level radioactivity and therefore measurements have to be performed with a low-background detector system. Heavy shielding well adapted to this system is the most efficient method for reducing background. Since much of the background is due to cosmic-ray muons, a further reduction is obtained by surrounding the main detector by an anticoincidence guard detector. For NaI(Tl) crystals of 12-cm diameter or less an annulus of organic phosphor is suggested as a nearly optimal guard arrangement for general applications. A background reduction by a factor of two to five is thereby obtained, depending on the γ -ray energy region involved. This technique should be combined with properly designed shielding of the detector system with γ -ray and neutron absorbers. For details, see Report No. 22 (ICRU, 1972). The use of aged air or N_2 gas in the lead cave also reduces the background, due to the absence of radon decay products, ^{85}Kr , etc.

A sophisticated anticoincidence-shielded multidimensional NaI(Tl) γ -ray spectrometer has been developed by Nielsen and Perkins (1967). With this system up to 12 radionuclides in air filters can be determined simultaneously.

For Ge(Li) detectors the Compton continuum is prominent and can obscure low-intensity peaks of other γ energies. An annular anticoincidence guard detector surrounding the main Ge(Li) detector can efficiently reduce this continuum by as much as a factor of 10.

b. Gamma-Gamma Coincidence Methods. A γ - γ coincidence counting technique will be very specific for a radionuclide emitting two γ photons in cascade. Since the background may be reduced substantially (10 times or more) with such a technique, the sensitivity of the method is also improved. A simple

type of γ - γ coincidence is obtained as a sum peak from a source in a well crystal or from a source close to the surface of a NaI(Tl) detector, or in between two adjacent detectors. The method can be illustrated by ^{60}Co , which emits two γ photons in cascade at 1.17 and 1.33 MeV, thus giving a sum peak at 2.50 MeV, which makes it possible to separate ^{60}Co from ^{59}Fe with its separately emitted γ photons of 1.10 and 1.29 MeV.

A common method is to use two coaxially arranged detectors [NaI(Tl) or Ge(Li)] as shown in Fig. 14 (Mattsson, 1971). In this case the system was used for assay of ^{22}Na in environmental samples of lichen, reindeer meat, and blood. With a crystal size of 7.5 cm diam by 5.0 cm and cylindrical source dimensions of 6.4 cm diam by 2.5 cm it was possible to obtain an MDA of 20 mBq in a 1000-min counting period. The sum condition was set at 1.02 MeV, which means that a pulse from detector A was registered only if it was in coincidence with a sum pulse for A and B in a 180-keV window around 1.02 MeV ($2 \cdot 0.511$ MeV) and also in coincidence with a pulse from detector B in a 90-keV window around 0.51 MeV.

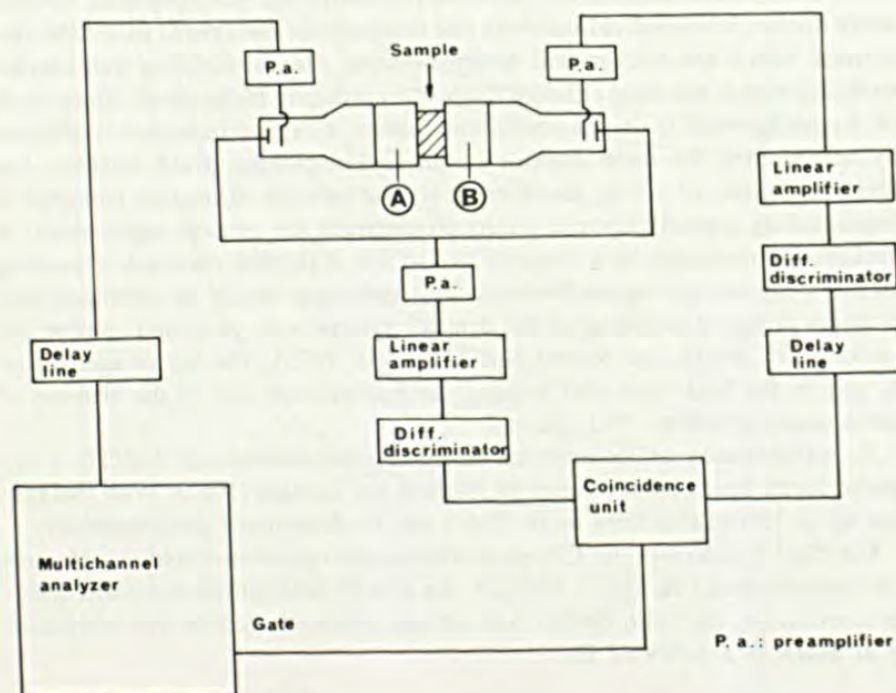


Fig. 14. Block diagram of a sum-coincidence circuit for two coaxially arranged NaI(Tl) detectors. [From Mattsson (1971). Reproduced from *Health Physics*, Vol. 23, pp. 223-230, 1971, by permission of the Health Physics Society.]

By counting straightforward coincidences between two cascade γ photons from ^{134}Cs (605 and 796 keV) it was possible to determine small amounts of this radionuclide in the presence of a much larger ^{137}Cs content in reindeer meat (Lidén and Andersson, 1962), using the same detector geometry as in Fig. 14. This measurement proved that ^{134}Cs was also present in humans, as suspected from an 800-keV bump in the whole-body γ -ray spectra of reindeer breeders.

c. The Phoswich Detector. Scintillators with different decay times can be optically coupled to a single PM tube, and the shape of the output pulse can then be used to distinguish events that have occurred in only one scintillator from those that occur in both. This combination of two detector materials, a phosphor "sandwich," is often called a "phoswich."

Often NaI(Tl) (1 to 3 mm thick) and CsI(Tl) (5 cm) with pulse decay times of 0.23 and 1.05 μs , respectively, are used together. With this shape-discrimination technique, low- γ -energy pulses (0-100 keV) are obtained only from NaI(Tl) and high- γ -energy pulses (0.1-2 MeV) only from CsI(Tl). Coincident events in both crystals will fall in between these two distributions, thus disturbing neither of them. In comparison with a single NaI(Tl) crystal having the same thickness, a background reduction by a factor of 5 to 10 is obtained with the phoswich detector for 50- to 70-keV photons. The phoswich detector has proved useful for low-level counting of ^{241}Am (60 keV) and ^{239}Pu (17 keV) in human lung. An alternative method for this rather difficult measuring problem uses large-area proportional flow counters with krypton or xenon, and anticoincidence guard detectors of the same type, as integrated parts of the detector system. The last two methods are still subject to further improvements.

5. Standardized Sources for Calibration and Intercomparison: Reference Sources

Measurement of environmental radioactivity is often the first step in a sequence eventually leading to estimates of absorbed dose to human beings. This requires knowledge of the absolute activity of the samples being measured, and this in turn means that the counting system has to be calibrated with standardized sources of the actual radionuclides.

For γ emitters the first step is to obtain a proper energy calibration of the recording system, irrespective of detector type. A useful standard for determination of an accurate energy versus pulse-height scale is a mixed radionuclide solution in a small vial of 5-cm³ volume. Figure 15 shows γ -ray spectra of such a source obtained by a NaI(Tl) well crystal and a Ge(Li) detector, illustrating the large differences in resolution and efficiency.

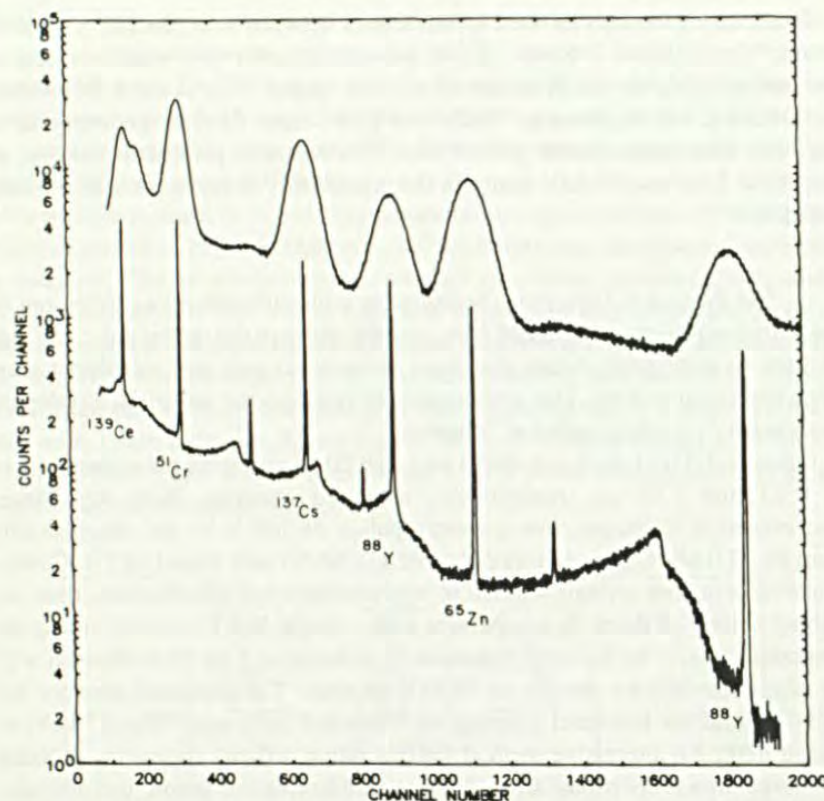


Fig. 15. Gamma-ray spectra of a 5-ml, mixed-radionuclide solution source within a 12.7-cm NaI(Tl) well crystal (upper curve) and at the face of a 60-cm³ Ge(Li) detector (lower curve). Counting time is 2000 s. [From NCRP (1978).]

Another useful radionuclide for energy calibration is ¹⁵²Eu with $T_{1/2} = 13.2$ yr and a number of γ photons of suitable energies and intensities (0.122–0.964 MeV). It should be observed that high counting rates may cause inconveniently high rates of accidental coincidences.

As most environmental samples with γ -ray-emitting radionuclides are measured in the shape of large sources, the calibration of the counter should preferably be made with a standardized source of the same shape and matrix as the unknown source. Standardized solutions of specific radionuclides are available from some international and national laboratories and commercial suppliers. A quantitative sample is taken from such a standard, usually by weighing and then dispersing the solution in a dilution flask and adding carrier solution. The diluted source can be used as a standard that simulates environmental sources in water solution or biological samples with a high water content. The solution can also be quantitatively transferred to an inactive matrix material, intended to simulate

the true environmental source matrix as closely as possible (density, atomic composition, geometry), and then kept in a homogeneously packed and closed container of the same shape as the one for the actual source measurement. Use of matrix material such as sterilized soil, dried vegetation powder, and sawdust has been reported.

Small, solid standardized or reference sources of γ -emitting radionuclides are usually more easily prepared. They are normally made as thin deposits on thin plastic disks, covering an area of about 5–100 mm², by evaporation of a solution sample under an infrared lamp. A wetting agent may be added for improved source uniformity. Detailed precautionary measures are described in various publications, for instance, the National Council on Radiation Protection and Measurements Report No. 58 (NCRP, 1978).

Efficiency calibration with small standardized sources is particularly important for semiconductor detectors, as they usually have a poorly defined sensitive volume in contrast to NaI(Tl) crystals, and thus are not suitable for accurate theoretical efficiency calculations. Sets of standards are available commercially; a frequently used set from NBS (U.S.A.) is shown in Table IV (NCRP, 1978). A calibration of the full energy peak efficiency of a 50-cm³ coaxial Ge(Li) de-

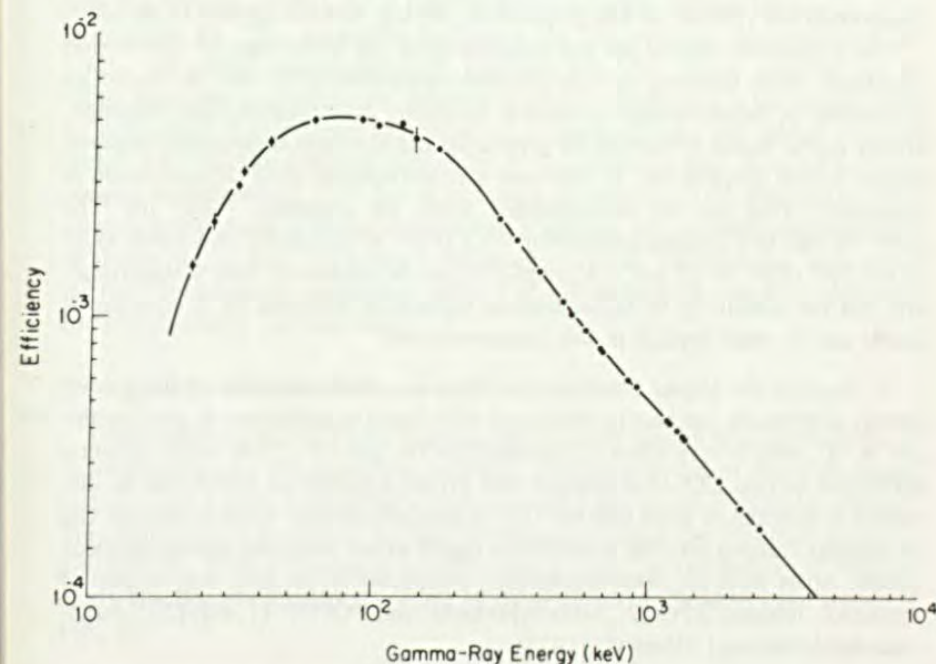


Fig. 16. Full energy peak efficiency measured as a function of photon energy for a 50-cm³ coaxial Ge(Li) detector. Source-detector distance is 10 cm. [From NCRP (1978).]

tector is shown in Fig. 16 (NCRP, 1978) measured with point sources at a 10-cm distance.

Interlaboratory comparison programs for quality control have been conducted by the IAEA and other organizations, and they have revealed many analytical errors, particularly in low-level activity measurements of environmental materials. A special problem in such a project is the difficulty of obtaining an inactive matrix (see also above) for simulating the changes of background spectra by absorption and scattering caused by a large source, and/or testing contamination introduced by sample treatment and source preparation procedures.

C. METHODS FOR ASSESSMENT OF BETA EMITTERS AND THEIR IDENTIFICATION

1. Sample Treatment and Preparation of Suitable Sources for Various Measurement Procedures

Samples of water, air, and biota require quite different sampling, concentration, and homogenization procedures for further treatment. Some elements or compounds are volatile, so that evaporation, drying, and ashing may be difficult.

The β -particles emitted are not monoenergetic (as in the case for conversion electrons). Most counting devices are also constructed with little or no energy resolution. A radiochemical procedure involving precipitation, ion exchange, and/or liquid-liquid extraction for separating the elements is generally required before source preparation. In this case a radiochemical yield determination is necessary. This can be accomplished with, for example, $^{99}\text{Tc}^m$ for ^{99}Tc ($E_{\beta}^{\max} = 292 \text{ keV}$), stable cesium for ^{137}Cs ($E_{\beta}^{\max} = 514 \text{ keV}$), and stable nickel for ^{63}Ni ($E_{\beta}^{\max} = 67 \text{ keV}$). Cesium-137 can be measured with γ spectrometry, but the sensitivity of radiochemical separation followed by β counting is better and is often applied at low concentrations.

a. Sources for Liquid Scintillation Counting. Radionuclides emitting low-energy β -particles can best be measured with liquid scintillation. A good example is ^{14}C , which very often is measured as the gas CO_2 . The most common method is to trap CO_2 in a solution thus giving a carbonate which can be dissolved in toluene. A good trap for CO_2 is fenylethylamine. Carbon dioxide can be directly trapped into the scintillation liquid where fenylethylamine has been added. Apart from fenylethylamine such a scintillation solution may consist of methanol, toluene, PPO (2, 5-difenyloxazol), and POPOP {1,4-di-[2-(5-fenyl-oxazolyl)]-benzene} (Horrocks, 1974).

For water solutions the source preparation is usually made as an emulsion of water in toluene. For instance, Triton X-100 (isooctylphenoxypolyethoxy-ethanol) can be used. As little water as possible should be used in order to obtain small water drops to minimize self-absorption.

Biological samples are either dissolved in a secondary solvent (salts of ammonium) or ashed, in which case H_2O and CO_2 can be trapped (^3H , ^{14}C measurement). For heterogeneous samples which cannot be dissolved, a solution can be simulated by a gel which is liquid when agitated and otherwise solid (Carb-O-Sil, or Instagel). Also filter papers and electroplated samples can be placed on the bottom of the scintillation vial and the scintillation solution added.

Different vials are used for liquid scintillation counting. There are certain requirements for these: (a) they should not be affected by the samples or the scintillation solution, (b) the transmission of the scintillation light should be good, (c) the natural and induced background should be low, (d) the radionuclide should not be adsorbed, and (e) the cost should be low. The vials are mostly made of glass or polyethylene. Polyethylene vials give a lower background but are more easily affected by scintillation liquid.

b. Sources for Gas Multiplication Detectors. Measurements with GM or proportional counters are common at higher β -particle energies. The main advantages are that the background is lower and counting efficiency is less variable from sample to sample than with liquid scintillation counters. The source preparation is often electrodeposition or evaporation, which give rather homogeneous and thin sources. Some kind of specific precipitation is also common, e.g., copper sulfide for ^{99}Tc , strontium carbonate for ^{90}Sr , cesium chloroplatinate for ^{137}Cs , and nickel dimethylglyoxime complex for ^{63}Ni . These sources can be rather thick and calibration and correction for self-absorption in the sample must be performed. If significant quantities of a stable element are added (such as nickel for ^{63}Ni) the counting efficiency by gas counters will decrease with increased radiochemical yield due to increased attenuation of the β -particles. (The same is true if liquid scintillation counting is applied, due to increased quenching as shown in Fig. 18). The sources are usually covered, for example, with a Mylar foil. Beta-particle absorption in this foil will reduce the counting efficiency for low- β -energy radionuclides.

2. Liquid Scintillation Counting of Beta Emitters

Good performance of liquid scintillation counting results because the counter is windowless and the geometric counting efficiency is almost 4π . This means that we have a high probability of counting the low-energy β -particles emitted. The amplification system of a liquid scintillator counter can be either linear or logarithmic. With logarithmic amplification a large energy range can be covered. The linearly amplified spectra of three different β emitters are shown in Fig. 17.

a. Solvent. Solvents should (a) effectively transfer their excitation energy to the scintillator, (b) be transparent for the scintillation light, (c) be chemically stable and unaffected by the presence of the scintillator, (d) be temperature independent, and (e) be able to dissolve the radioactive sample.

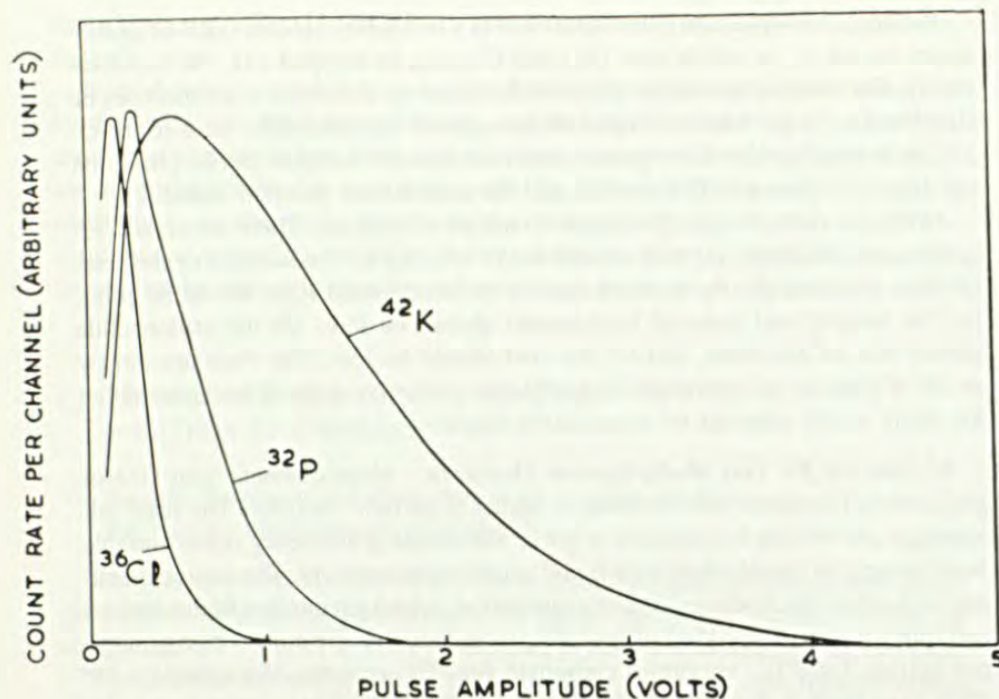


Fig. 17. Pulse-height spectra obtained from liquid scintillation counting with bialkali PM tubes for ^{36}Cl , ^{32}P , and ^{42}K ($E_{\beta}^{\text{max}} = 0.71, 1.71, \text{ and } 3.6 \text{ MeV}$, respectively). [Reprinted with permission from Parker and Elrick, "Liquid Scintillation Counting" (E. D. Bransome, Jr., ed.). Copyright 1970, Grune & Stratton, Inc.]

Several solvents, such as toluene, benzene, and 1,2,4-trimethylbenzene, fulfill most of these requirements but their ability to dissolve the sample is limited. A secondary solvent can then be used. This, however, reduces the light exchange and the pulse height.

b. Scintillator. For the scintillator we can set the following requirements: (a) the probability to emit a light photon at excitation should be large, (b) the emitted photon should have a wavelength matching the maximal sensitivity of the photocathode, (c) the scintillator must be chemically stable and not change the characteristics of the solvent, and (d) it must have small self-absorption of the scintillation light.

There are a large number of such scintillators and they are all aromatic hydrocarbons. Examples such as PPO and POPOP were mentioned earlier; POPOP is a secondary scintillator. A secondary scintillator is used to get a wavelength of emitted light for which the photocathode is more sensitive, as well as to decrease the effect of colored samples. The absorption spectrum of the secondary scintillator must closely match the emission spectrum of the primary scintillator.

c. Quenching. There are processes which result in a decrease of the maximal counting efficiency. If the sample is not in solution, but consists of small particles, part of the β -particle energy will be absorbed in the sample. This is most important for low-energy β particles such as those from ^3H .

When a secondary solvent is used, excitation energy can be lost if the secondary solvent cannot transfer energy. This chemical quenching causes energy to be transferred to molecules other than the scintillator. Examples of strongly quenching substances are nitrates and iodides. Optical quenching occurs if the sample is colored or contains light-absorbing particles. The pulse height decreases because of quenching, resulting in an increase of the minimum detectable energy. This means that the counting efficiency decreases for β -particle emitters. Correction for quenching is often necessary since the counting efficiency may vary among samples, cf. Fig. 18. This can be done by

- (1) use of an internal standard,
- (2) sample-channels ratio,
- (3) external-standard gross counts, or
- (4) external-standard channels ratio.

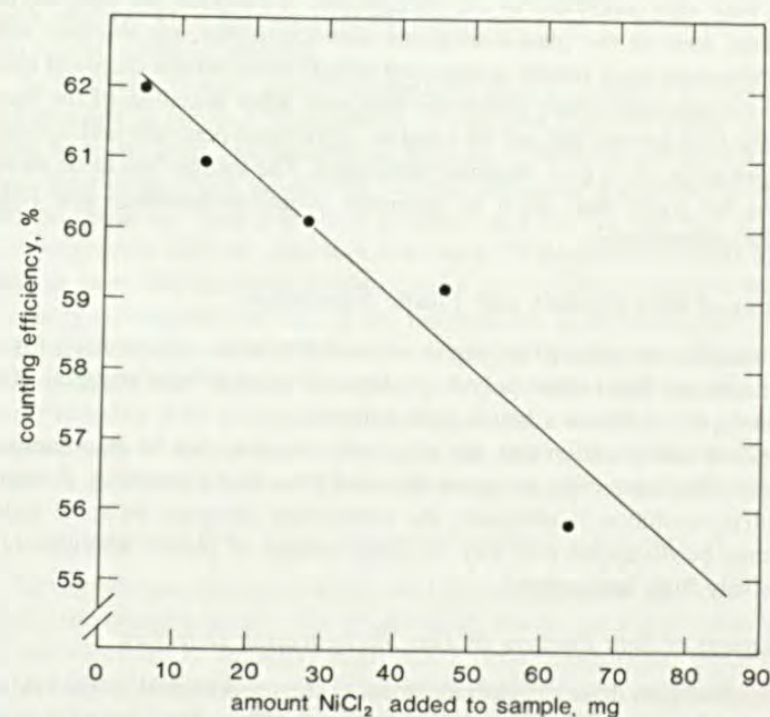


Fig. 18. Counting efficiency as a function of NiCl_2 concentration in liquid scintillation counting. [After Riedel *et al.* (1980).]

The method using an internal standard means that a known activity of the same radionuclide is added and the measurement repeated. The counting efficiency of each individual sample can then be calculated. With quenching, a shift of the pulse-height distribution is obtained. The channel-ratio correction method implies that the sample is measured in two pulse-height channels. From this ratio and a calibration of the pulse-height shift, the counting efficiency can be calculated. A γ -emitting radionuclide is used in the external-standard correction method. Interaction will give electrons which will give excitation light. The counting efficiency for the external standard changes with volume of scintillation solution and geometrical changes. These problems can be solved by use of the external-channels ratio method, which is similar to that described above for sample-channels ratio. For further details the reader is referred to the book by Horrocs (1974), "Application of Liquid Scintillation Counting," and its chapter "Quench Correction Methods."

d. Background. Thermal noise from the photomultipliers is a source of background. This is, to a great extent, eliminated by coincidence connection of the photomultipliers. Radioactivity in scintillator vials and other construction material, as well as natural radioactivity and other radiation sources close to the apparatus, may also contribute to the background. To reduce the contribution from external sources the photomultipliers and scintillator are shielded with lead. Chemiluminescence results in emission of light from certain chemical reactions. The chemiluminescence decreases with time after initiation of the reaction, thus the background will not be constant. Hydrogen peroxide and strongly alkaline substances often give chemiluminescence. The sample has to be stored in darkness for more than 24 h to eliminate chemiluminescence and light-induced phosphorescence.

3. Counting of Beta Emitters with Plastic Scintillators

Plastic scintillators coupled to one or several PM tubes can be used for α , β , and γ counting. Most often polyvinyltoluene is used as base material. This material is doped to obtain a better light output.

Any desired configuration and size of plastic detectors can be manufactured rather easily. Small detectors are generally used for α and β counting. A coarse spectrometric resolution is obtained; the conversion electrons of some radionuclides may be displayed this way. A disadvantage of plastic scintillators is their relatively high background.

4. Assessment of Beta Emitters by Gas Multiplication Detectors

Gas multiplication detectors operate in either the proportional or the GM region, often with a continuous flow of counting gas. The advantage with a continuous gas flow is that the consumption of quenching gas does not limit the lifetime of the detector. A guard detector which operates in anticoincidence with

the main detector may be used, as well as shielding around the detectors, to reduce background counts. The source is placed close to the window, which consists of Mylar film ($0.1\text{--}1\text{ mg/cm}^2$).

At very low β energies the window will reduce the counting efficiency. The source can also (depending on detector construction) be placed inside the detector, and it is then possible to reduce the low-energy limit to 250 eV. This is lower than what is possible with semiconductors or scintillation detectors. Frequently used detector gases are 90% argon + 10% methane or 96% helium + 4% isobutane for proportional counters, and 99% helium or argon + 1% isobutane for continuous gas-flow GM counters. Bromine can be added as quenching gas for noncontinuous gas-flow counters. The background is of the order of 0.005 s^{-1} for a 2.5-cm-diameter window and a counting efficiency $>40\%$. Varying backscattering from the source mount caused by use of different materials must be corrected for. Due to backscatter from the source-supporting material, a higher counting efficiency than that given by the geometrical conditions may be obtained. The background will be due to surrounding material and sources, but also to radionuclides in the construction materials of the detector device. If a radiochemical procedure is involved, the reagents can also contribute to the background. Variation in counting efficiency can occur for thick sources. Thus the experimental conditions must be maintained as constant and reproducible as possible.

As stated previously, great care must be taken to ensure that only the desired radionuclide is counted. Elimination of different nuclides likely to be present must be tested in the radiochemical procedure. Sources can be recounted after some time to see if any short-lived radionuclides were present at the initially performed counting. Pure β -particle emitters such as ^{99}Tc and ^{90}Sr are measured advantageously with the detectors described. Proportional counters offer an advantage over GM counters in that one can test by spectrometry if the maximal β energy corresponds to that of the radionuclide to be measured.

Gas multiplication detectors may be employed in the assay of gaseous sources such as ^3H , ^{14}C , or ^{133}Xe by introducing the radioactive gas directly into the detector together with a suitable counting gas. Such counters are generally constructed with a cylindrical cathode and an axial wire as anode. Internal-gas counting is a 4π method which avoids problems of source self-absorption.

5. Assessment of Beta Emitters by Semiconductor Detectors

Semiconductor detectors can be used for measurement of β -particle emitters in environmental samples. The good energy resolution makes counting of conversion electrons an attractive approach. Silicon detectors can be used but must be cooled with liquid nitrogen. In addition, detector sizes are small, geometrical conditions are unfavorable, and the counting efficiency is low.

It is more convenient to use a silicon surface-barrier detector, a diffused-junction detector, or an ion-implanted detector, such as are normally used for

α -particle spectrometry or spectrometry of other heavy charged particles. Different detector sizes and depletion depths are available.

Detectors slightly contaminated after low-level α spectrometry are acceptable. Such a counting system will be inexpensive and easy to use. The background is of the order of 0.05 s^{-1} (30–300 keV) and the counting efficiency is about 25%. The background is, to a large extent, due to the construction material of the detectors. Special low-noise and low-background detectors of other construction materials are available. In Fig. 19a are shown the pulse-height distributions of conversion electrons from $^{99}\text{Tc}^m$ and ^{51}Cr , and in Fig. 19b the pulse-height distribution of the pure β -particle emitter ^{99}Tc . Stainless-steel disks were used as source holders. The shape of the pulse-height distribution in the low-energy region is markedly influenced by the backing material.

6. Special Methods Such As the Use of Cerenkov Effect, Coincidence Technique, and Beta-Particle Spectrometry

a. Cerenkov Effect. A charged particle can have a higher velocity v than light in a transparent medium. Electromagnetic radiation called Cerenkov radiation is emitted if $v > c/n$, where c is the velocity of light in a vacuum and n is the refractive index of the medium. The method of Cerenkov counting in aqueous solutions can only be used for β particles having energies greater than 0.26 MeV. The sample is placed in a liquid scintillation vial and counted in a manner similar to that employed in liquid scintillation counting. Much of the Cerenkov radiation emitted is in the UV region and is absorbed by the glass window of the phototube and in glass vials. Cerenkov radiation is also emitted anisotropically, which reduces the counting efficiency. The detection efficiency can be improved by use of wavelength shifters. The counting efficiency increases with increasing β -particle energy, but is lower than that for normal liquid scintillation counting. The advantage of the method is the easy source preparation and absence of chemical quenching. Optical quenching occurs, however. Since the pulse height is lower in Cerenkov counting than with the use of scintillation solution, the energy window in which counting is performed should be adjusted accordingly.

b. Coincidence Counting. Coincidence counting is applied to nuclear decays involving two (or several) transitions in prompt succession. Two detectors are then required. A third counting channel records those events from the two detectors that occur in coincidence. Background counts may be strongly reduced and radionuclide identification improved; usually a smaller MDA is achieved, in spite of a sometimes reduced counting efficiency. One of the most usual applications of coincidence counting is the determination of activity by coincidence counting of β - γ decay. The pulse length must be large enough to avoid loss of coincidence. The rate of accidental coincidences increases with pulse width.

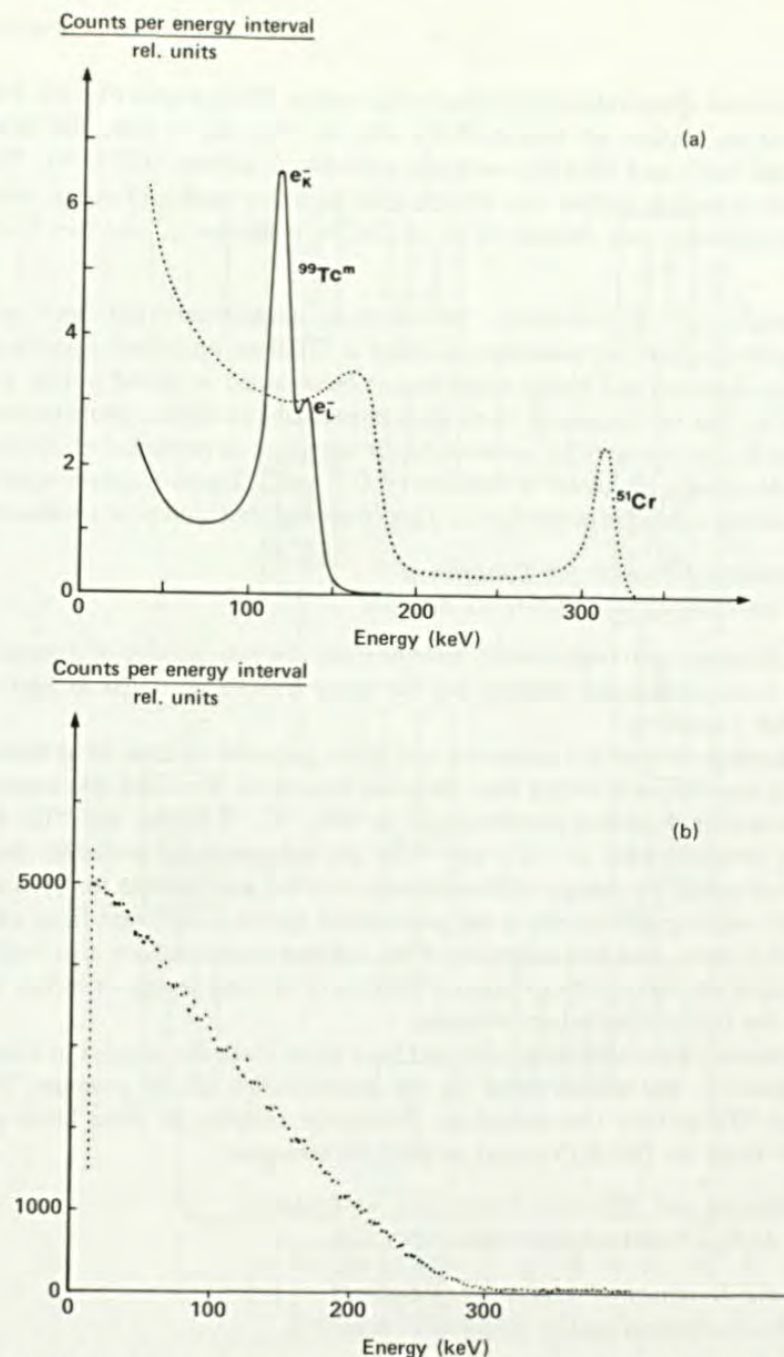


Fig. 19. Pulse-height distribution (a) of conversion electrons from $^{99}\text{Tc}^m$ ($e_K^- = 119 \text{ keV}$, $e_L^- = 137 \text{ keV}$) and ^{51}Cr ($e_K^- = 315 \text{ keV}$, $e_L^- = 324 \text{ keV}$) and (b) from β particles of ^{99}Tc . The dotted line in (b) indicates energy discrimination for noise elimination: e_K^- and e_L^- of ^{51}Cr are not resolved. [From Holm *et al.* (1982).]

This requires an optimization of the electronic system, illustrated in Fig. 20. The method can be applied for radionuclides such as ^{134}Cs ($E_\gamma = 605, 796 \text{ keV}$, $E_\beta^{\text{max}} = 662 \text{ keV}$) and ^{60}Co ($E_\gamma = 1.17, 1.33 \text{ MeV}$, $E_\beta^{\text{max}} = 314 \text{ keV}$). The coincidence counting method can also be used for other modes of decay, such as γ - γ coincidence (see Section IV,B,4.), EC- γ coincidence, and α - γ coincidence.

c. *Beta-Particle Spectrometry.* Beta-particle spectrometry has been performed with proportional counters operating at different pressures. Spectra of conversion electrons and Auger electrons can be obtained as shown in Fig. 21. The activity can be calculated from known internal conversion probabilities. Beta-particle spectrometry by semiconductor detectors is presented in Section IV,C,5. As already discussed in Sections IV,C,2 and 3, liquid or plastic scintillation counting offers the possibility of β spectrometry, but with poor resolution.

7. Standardized Sources for Calibration and Intercomparison: Reference Sources

It is, of course, also important to determine the absolute activity of β -particle emitters in environmental samples for the same reasons as stated in Section IV,B,5 for γ emitters.

Beta-particle emitters are measured as sources prepared on disks or as liquids for liquid scintillation counting after chemical separation. Standard disk sources of different pure β -particle emitters (such as ^{63}Ni , ^{14}C , ^{36}Cl , ^{90}Sr , and ^{99}Tc) but also β - γ emitters (such as ^{137}Cs and ^{60}Co) are commercially available. Such sources are useful for energy calibration purposes but less suitable for determination of counting efficiencies if the geometrical design is different from what otherwise is used. Standard solutions of the isotopes mentioned are also available allowing any laboratory to prepare sources of desired design or to use the solution for liquid scintillation counting.

International intercalibration exercises have taken place for samples of water, soil, vegetation, and animal tissue for the determination of, for example, ^{14}C , ^{90}Sr , and ^{99}Tc activity concentrations. Reference samples of these kinds are available from the IAEA (Vienna) or NBS (Washington).

D. METHODS FOR IDENTIFICATION AND ASSESSMENT OF ALPHA-PARTICLE-EMITTING NUCLIDES

1. Sample Treatment: Preparation of Sources by Radiochemical and/or Physical Procedures

Samples for analysis of α -particle emitters are generally wet- or dry-ashed. When analyzing ^{210}Po wet-ashing is necessary because polonium is a fairly volatile metal. Water samples are evaporated or a precipitation is performed to preconcentrate the elements. The gases ^{222}Rn and ^{220}Rn are collected by trapping in activated carbon or extracted from water into an organic solvent.

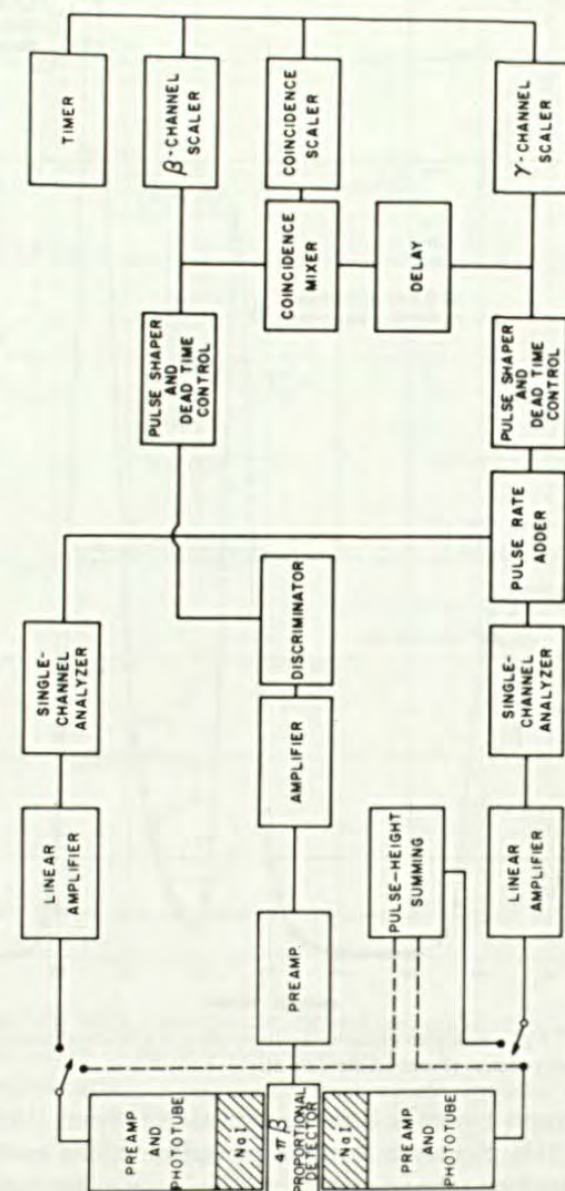


Fig. 20. Block diagram of the basic components of a typical 4π β - γ coincidence system. [From NCRP (1978).]

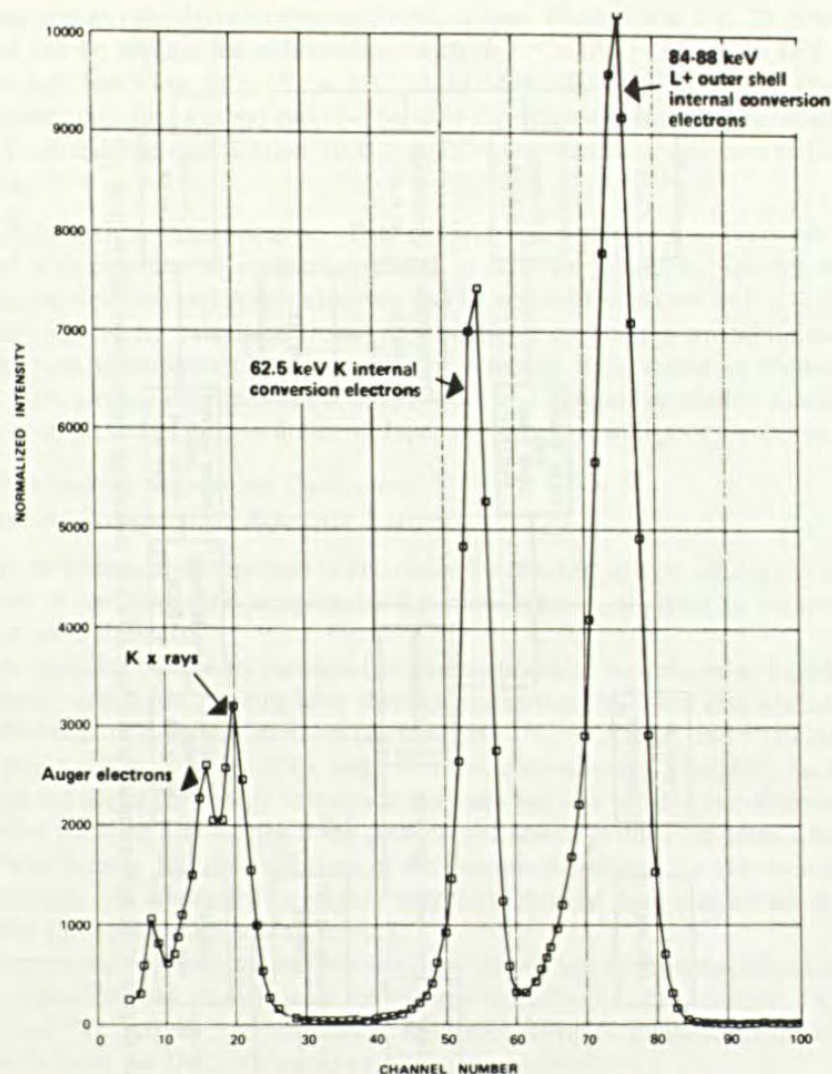


Fig. 21. ^{109}Cd - $^{109}\text{Ag}^m$ conversion-electron spectrum taken in a 30-cm-diameter, cylindrical, 4π gas-flow proportional counter. [From NCRP (1978).]

For most elements a more or less complicated radiochemical procedure must be performed. This procedure involves precipitation, ion exchange, and/or liquid-liquid extraction. The separation of ^{210}Po is somewhat easier since polonium spontaneously deposits onto disks of silver, copper, or nickel from various media (Flynn, 1968; MacKenzie and Scott, 1979). The methods are destructive, requiring planning to accomplish sequential separation of several α -particle emitters. A flow sheet for the sequential separation of plutonium, thorium, and americium is shown in Fig. 22. A disadvantage in the analysis of α -particle emit-

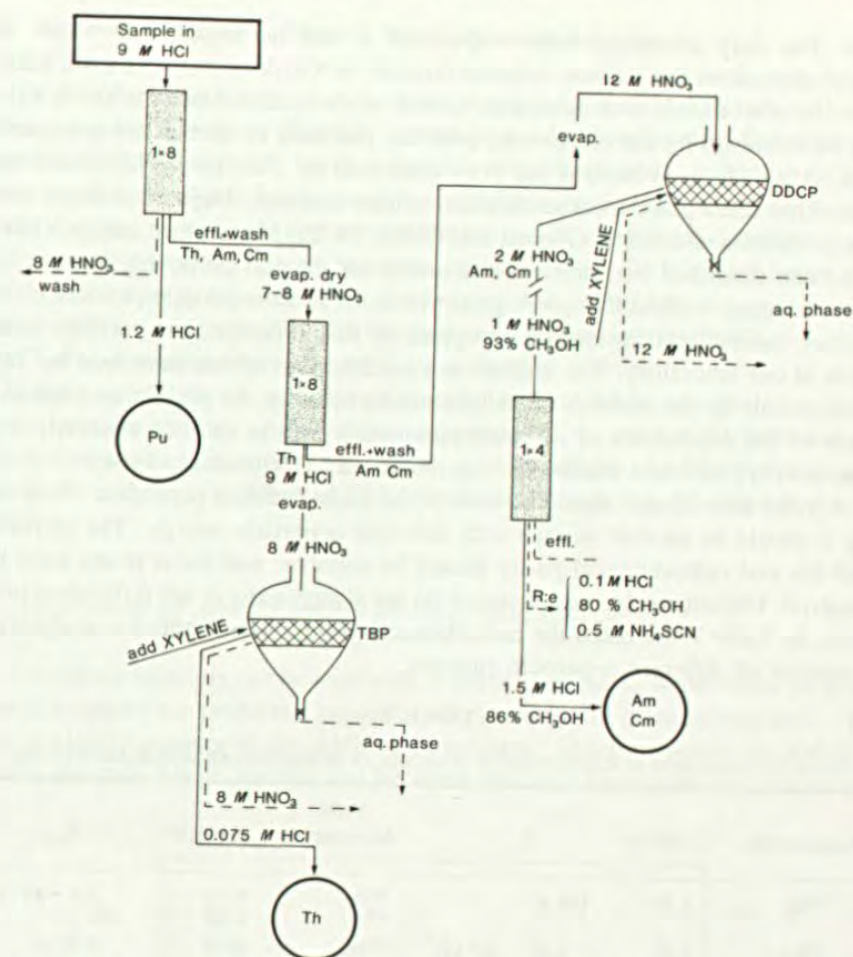


Fig. 22. Flow scheme of a radiochemical separation procedure for Pu, Am, and Th. [From Holm and Persson (1981).]

ters is the need for such time-consuming and complicated radiochemical procedures. An advantage is the low detection limit.

As emphasized earlier, α sources for spectrometry must be very thin, preferably less than $0.5\ \mu\text{m}$. Thick sources can be used with very specific precipitation methods, such as barium sulfate for radium and lanthanum fluoride for certain actinides. Then a simpler counting device can be used (ZnS and photomultiplier). The gases ^{222}Rn and ^{220}Rn , as well as some of their short-lived decay products, are also α -particle emitters. They are usually measured in a bottle, coated on the inside with ZnS, in contact with a photomultiplier.

During the final source preparation, after radiochemical separation, electrolysis is often used since it produces, in general, thinner deposits than evapora-

tion. The only advantage with evaporation is that no losses need occur. In electrodeposition the cathode material consists of a disk of stainless steel, platinum, or other metals such as copper, nickel, or aluminum. Since the anode must not be dissolved by the electrolytic process, platinum or carbon are most suitable. A widely used method has been described by Talvitie (1972) where the deposition takes place from ammonium sulfate medium. Deposition from oxalate medium is also useful (Puphal and Olsen, 1972). Many other methods have also been described but they are less useful for general purposes.

A universal method that gives good yields for most α -particle emitters is, of course, desirable. Considerable progress in this direction has recently been made at our laboratory. The method is a modification of that described by Talvitie, mainly by the addition of sodium sulfate solution. At present an extensive study of the dependence of different parameters such as cathode material, current density, different media, etc., is under way (Hallstadius, 1984).

A yield determinant should be used in the radiochemical procedure. Preferably it should be another isotope with different α -particle energy. The physical half-life and radiochemical purity should be suitable, and the α peaks must be resolved. Unfortunately, such isotopes do not always exist or are difficult to produce. In Table V are listed the radiochemical yield determinants for analysis of a number of different α -particle emitters.

TABLE V

YIELD DETERMINANTS IN RADIOCHEMICAL ANALYSIS OF α -EMITTING HIGH-Z RADIONUCLIDES^a

Radionuclide	E_α (MeV)	$T_{1/2}$	Yield determinant	E_α (MeV)	$T_{1/2}$
²¹⁰ Po	5.30	138 d	{ ²⁰⁸ Po ²⁰⁹ Po ²³⁶ Pu ²⁴² Pu	5.11	$2.9 \cdot 10^5$ yr
²³⁹ Pu	5.16	$2.41 \cdot 10^4$ yr		4.88	103 yr
²⁴⁰ Pu	5.16	6570 yr		5.77	2.85 yr
²³⁸ Pu	5.50	87.7 yr		4.90	$3.8 \cdot 10^5$ yr
²⁴¹ Am	5.48	433 yr	{ ²⁴³ Am ²⁴⁴ Cm ²⁴³ Am ²⁴¹ Am	5.27	7370 yr
²⁴² Cm	6.11	163 d		5.80	18.1 yr
²⁴⁴ Cm	5.80	18.1 yr		5.27	7370 yr
²³⁴ U	4.76	$2.5 \cdot 10^5$ yr		5.48	433 yr
²³⁵ U	4.40	$7.0 \cdot 10^8$ yr	{ ²³² U ²²⁹ Th	5.32	72 yr
²³⁸ U	4.20	$4.5 \cdot 10^9$ yr			
²²⁸ Th	5.42	1.91 yr			
²³⁰ Th	4.69	$8.0 \cdot 10^4$ yr		4.8-5.05	7300 yr
²³² Th	4.0	$1.4 \cdot 10^{10}$ yr	{ ²³³ Pa (β , γ) ²³⁹ Np (β , γ) ²²⁴ Ra ²²⁵ Ra (β , γ)		27 d
²³¹ Pa	5.0	$3.3 \cdot 10^4$ yr			2.35 d
²³⁷ Np	4.78	$2.1 \cdot 10^6$ yr		5.69	3.7 d
²²⁶ Ra	4.78	1600 yr		5.72	14.8 d
			{ ²²³ Ra 		11.4 d

^aValues of E_α and $T_{1/2}$ are taken from "Table of Isotopes," Lederer *et al.* (1978).

2. Assessment of Alpha Emitters by Scintillation Counting

Due to the short range of α particles, many scintillation detectors with entrance windows, such as NaI(Tl), cannot be used. By using thin ZnS disks in contact with a photomultiplier it is possible to perform total α counting. It was also pointed out above that measurement of Rn is performed in a bottle coated with ZnS and placed on a PM tube. Both Rn and its short-lived decay products are α -particle emitters, which increases the sensitivity. Such a scintillation device has no energy resolution but can be useful in some chemical separations, as for ²¹⁰Po, or for specific precipitation techniques, such as BaSO₄ for ²²⁶Ra and LaF for a solution containing a single known actinide.

Liquid scintillation is sometimes a useful method when isotopes of both β - and α -particle emitters of an element are measured. In spite of the poor energy resolution one can discriminate between α and β particles. An illustrative example is the counting of ²⁴¹Pu and ²³⁹+²⁴⁰Pu together, see Fig. 23 (Darrall *et al.*, 1973).

3. Assessment of Alpha Emitters by Gas-Filled Detectors

Gas-filled detectors can be used with α -particle emitters. Windowless proportional counters are preferred because energy resolution often is required. The gas is usually argon with the addition of methane, which increases the stability of the counter. Large sources can be measured with high counting efficiency.

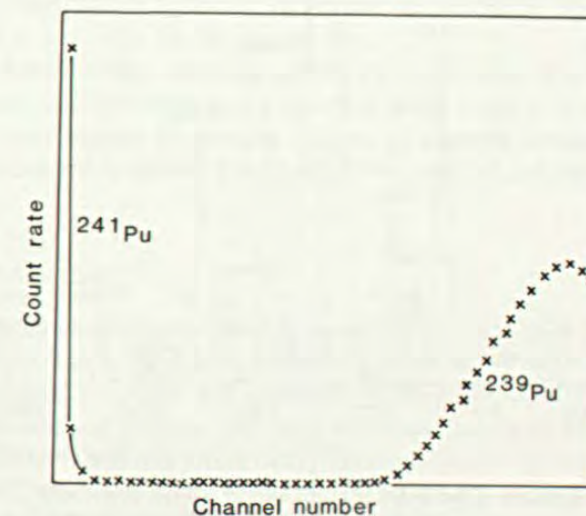


Fig. 23. Liquid scintillation spectrum, at a single logarithmic gain setting, of a mixture of the β emitter ²⁴¹Pu and the α emitter ²³⁹Pu. Activity ratio is 40 to 1. [From Darrall *et al.* (1973). Laboratory of the Government Chemist.]

Geiger-Müller counters can also be used, windowless or with thin windows. Their use is, however, limited to what has been mentioned in Section IV,D,2 about total α counting and specific separations and precipitations. Gas scintillation counters have also been successfully applied by using the light emitted from noble gases at the passage of an α particle.

4. Alpha-Particle Spectrometry

It is generally not possible to perform spectrometry of all α -particle emitters in a source at the same time because many α -particle emitters interfere with each other. It is very difficult to obtain sufficient elimination of stable elements existing in macroamounts; however, a group separation of actinides has been obtained by Holm and Fukai (1977). Since the lanthanides are also coseparated this limits the amount of sample to be processed to obtain a source for good α spectrometry (Fig. 24).

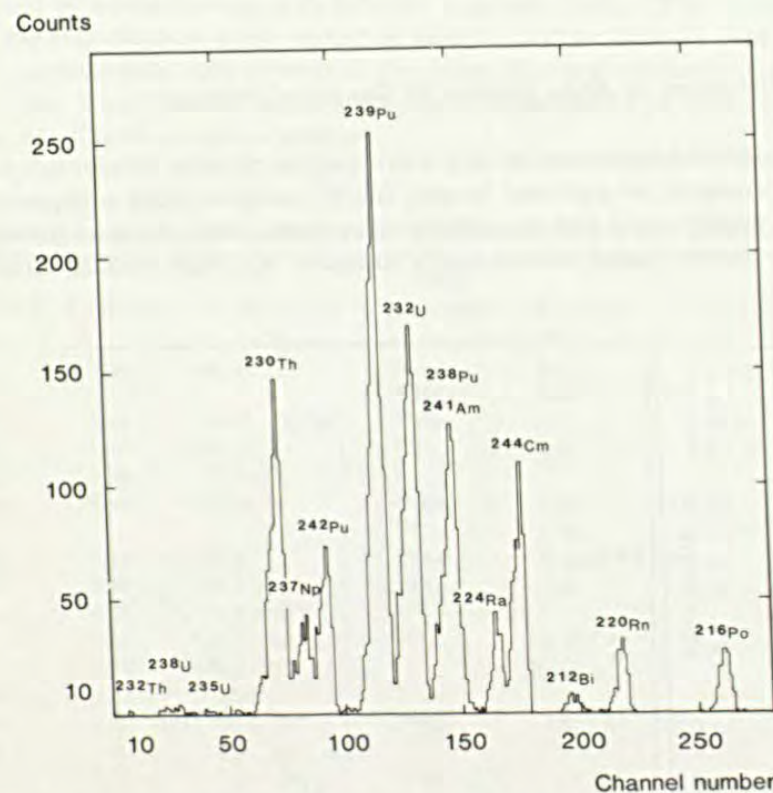


Fig. 24. Alpha spectrum of the IAEA seaweed sample AG-I-1 spiked with ^{230}Th , ^{232}U , ^{237}Np , ^{242}Pu , and ^{244}Cm (1 g seaweed; 310 min counting; $^{239} + ^{240}\text{Pu}$ was designated as ^{239}Pu). [From Holm and Fukai (1977). *Talanta* 24, Method for multi-element alpha spectrometry of actinides and its application to environmental radioactivity studies. Reprinted with permission. Copyright 1977 Pergamon Press, Ltd.]

Currently silicon surface-barrier detectors are most often used, but gridded ion chambers also give adequate resolution for spectrometry and can be used with relatively large area sources. Most elements of interest emit α -particles in the energy region 3.95 (^{232}Th) to 8.8 MeV (^{212}Po). Alpha-particle emitters with lower energies, such as ^{146}Sm ($E_\alpha = 2.46$ MeV) and ^{148}Gd ($E_\alpha = 3.18$ MeV), have been used for energy calibration and stabilization of α spectrometers.

Quantitative evaluation of α -particle spectra is simpler than that for γ -ray spectra, since the detector efficiency does not vary with energy. Correction for peak tailing into a lower energy region of interest must be made. The energy resolution improves with increasing source-to-detector distance and with decreasing detector diameter but then the counting efficiency is also lower. Since most α -particle emitters are not monoenergetic, often two or more α energies from one radionuclide are present which cannot be resolved. The most important condition is to obtain clean sources without a thick deposit of foreign matter. Poorly prepared sources can never be compensated for by better electronics or computer analysis. The detector system is calibrated with sources of known energies. We have found a mixed source of ^{233}U ($E_\alpha = 4.82$ MeV) and ^{244}Cm ($E_\alpha = 5.80$ MeV) to be suitable. Americium-241 is often used as a reference.

The background of the detectors can be kept as low as $(0.3-2) \cdot 10^{-5} \text{ s}^{-1}$ for an energy interval of 120 keV in the α -energy region of 4 to 6 MeV. Contamination on the detector may occur, for instance, by evaporated polonium isotopes, if spectrometry is performed under vacuum. Contamination by recoil nuclei can also happen, for instance, by ^{224}Ra from ^{228}Th . To keep background low, high-activity sources should not be used and detectors should be carefully handled. Contamination by recoil nuclei can be avoided by applying a positive potential of a few volts on the source disk.

Several detectors are normally connected via routing systems to one multichannel analyzer. Counting times are often long, 1 day to 1 week. In Fig. 25 a typical α spectrogram of thorium isotopes in a source prepared from a large volume of seawater is shown. Thorium-229 was used as radiochemical yield determinant.

5. Special Methods

Some special photographic emulsions can detect tracks of α -particles and fission fragments, but require long exposure periods at low activities. The tracks in the developed emulsion are counted. Alpha-particles also leave etchable tracks in plastics and glasses. The latter technique has been widely applied for measurement of radon and radon decay products in air. In such measurements plastic films coated with cellulose nitrate are frequently used. The evaluation of such measurements is subject to significant uncertainties partly caused by the rather complicated calibration procedure. Further details are given by Becker (1972).

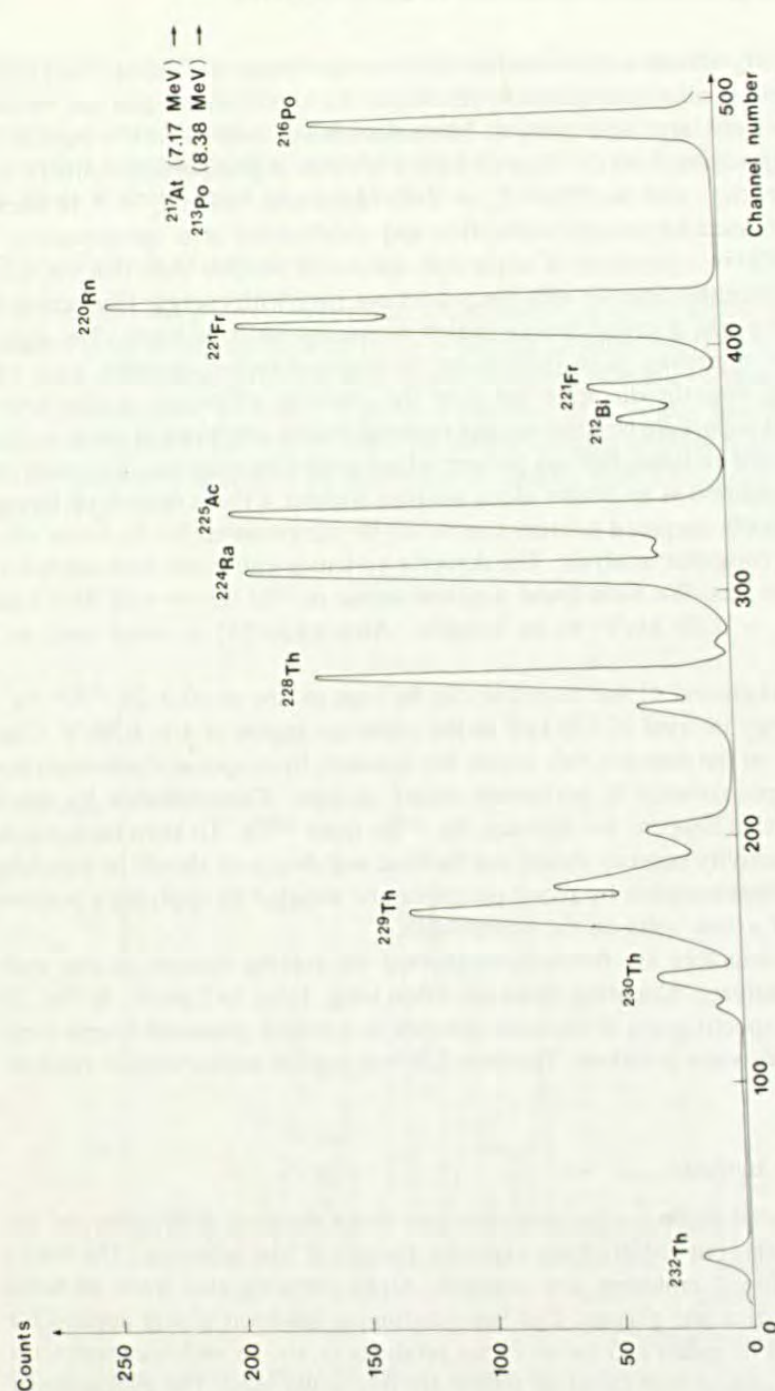


Fig. 25. Alpha spectrum of a natural thorium sample, with ^{229}Th added as yield determinant, separated from seawater. 300-mm², surface-barrier silicon detector, 100- μm depletion depth, and 5.7 keV per channel. [From Holm and Persson (1981).]

6. Standardized Sources for Calibration and Intercomparison: Reference Sources

Electrodeposited sources of, for example, ^{239}Pu , ^{241}Am , and ^{244}Cm are commercially available, which are calibrated with an overall uncertainty of 1% to 10%. Calibrated solutions of ^{241}Am , ^{242}Pu , ^{229}Th , ^{210}Pb (in radioactive equilibrium with ^{210}Po), and many other α -particle emitters also can be obtained. One can prepare calibrated sources from these solutions by evaporation of known amounts. Calibration can be checked with detectors having previously known counting efficiencies, or with 2π proportional or GM counters. Usually it is difficult to obtain a detector calibration giving an overall uncertainty better than 5%.

Specifically prepared sources and solutions for intercalibration exercises have been used much less for α -particle emitters than for γ -emitting radionuclides. Intercomparisons based on samples of different matrices such as sediments, water, algae, soil, etc., have been performed. Many of these samples are now available from NBS or IAEA as reference materials.

E. BACKGROUND AND BUILDUP OR DECAY CORRECTIONS

1. Dead Time

In general we want to derive an expression for the true interaction rate n of a detector system. A recorded count rate m is always lower than or equal to n . True events which occur during a dead period τ of the detector system are not recorded.

The fraction of the total recording time the detector is dead is given by the product $m\tau$. The rate at which true events are lost is then $nm\tau$. But because the expression for rate of losses can also be written $n - m$, then

$$n - m = nm\tau \quad \text{or} \quad n = m/(1 - m\tau) \quad (2)$$

This is what is called the *nonparalyzable* model, where a fixed dead time is assumed to follow each true event occurring during a "live period."

For the *paralyzable* model an event during the dead period, although not registered as a count, extends the elapsed fraction of the dead time by another period τ . The corresponding expression can be derived to be

$$m = ne^{-n\tau} \quad (3)$$

For low count rates ($n \ll 1/\tau$) the following approximation can be written for both nonparalyzable and paralyzable models:

$$m = n/(1 + n\tau) \cong n(1 - n\tau) \quad (4)$$

and

$$m = ne^{-n\tau} \cong n(1 - n\tau) \quad (5)$$

It should be pointed out that the built-in dead-time correction of most multichannel analyzers is found to be rather unreliable.

2. Background Correction

In absence of decay during the counting period, the background count rate n_b is subtracted from the observed counting rate after correction for dead-time losses. The true net count rate N is then

$$N = [m/(1 - m\tau)] - n_b \quad (6)$$

Variations and stability of background must be controlled.

3. Decay Corrections

In the absence of significant background and dead time but with significant decay during the counting interval, it can be shown that the count rate N_0 at time t_0 , the beginning of the counting interval, is given by

$$N_0 = [\lambda t / (1 - e^{-\lambda t})] \cdot m \quad (7)$$

where t is the duration of the counting interval and λ is the decay constant. If the count rate is assigned to the midpoint of the counting interval the influence of decay during that interval is much reduced. As an example, the observed rate exceeds the midpoint value by only 0.5% for an interval of half a half-life. An expression treating the combined effects of dead time, decay, and background has been derived by Axton and Ryves (1963) and is given by

$$N_0 = \left[\frac{m}{1 - m\tau} - n_b \right] \cdot \left[\frac{\lambda t}{1 - e^{-\lambda t}} \right] \cdot \left[1 + \frac{m(\lambda t)^2}{12} \right] \quad (8)$$

The third term is a correction term which is only significant at high initial counting rates.

The count rate is generally converted to a reference time R , which is often the earlier time of collection. Thus

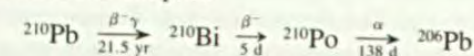
$$N_R = N_0 e^{\lambda T} \quad (9)$$

The elapsed time T , from the reference time to the counting time is taken as negative if the reference time is later than the time of counting.

4. Buildup Corrections

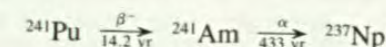
Correction for buildup of decay products is often included in the calibration. Such is the case in the measurement of ^{222}Rn , where the short-lived decay products also are counted and in reality improve the possibility of measurement.

Of more importance is the case when the activity concentration of the radionuclide to be determined has changed between sampling and measurement, due to buildup from a precursor. An illustrative example is measurement of ^{210}Pb and ^{210}Po in environmental samples:



An increase or decrease of ^{210}Po during the time from sampling to measurement can occur. The problem can be approached in two ways: (1) ^{210}Pb and ^{210}Po are both measured at a time t_1 after sampling, or (2) ^{210}Po is measured at a time t_1 after sampling, following chemical separation of ^{210}Po . We then allow ^{210}Po to "grow up" during a time $t_2 - t_1$, after which ^{210}Po is measured again at time t_2 , as a decay product of ^{210}Pb , which thus can be determined quantitatively.

Another example is measurement of ^{241}Am in samples also containing plutonium isotopes; ^{241}Am is the decay product of ^{241}Pu :



Especially serious is the case when ^{241}Am concentration is much lower than that of ^{241}Pu . In seawater, for example, ^{241}Am concentrations might be of the order of 2 $\mu\text{Bq/liter}$, and those of ^{241}Pu of the order of 300 $\mu\text{Bq/liter}$. During a year of storage the ^{241}Am concentration will then increase by about 25%. As a further complication, the low-energy β -emitting radionuclide ^{241}Pu is difficult to determine in environmental samples.

The general formula for calculating the buildup of a single decay product is given by

$$A_2(t) = A_1(0)[\lambda_2/(\lambda_2 - \lambda_1)](e^{-\lambda_1 t} - e^{-\lambda_2 t}) + A_2(0)e^{-\lambda_2 t} \quad (10)$$

where $A_2(t)$ is the activity of the decay product at time t , $A_2(0)$ the activity of the decay product at time 0, $A_1(0)$ the activity of parent radionuclide at time 0, λ_1 the decay constant for parent radionuclide, and λ_2 the decay constant for the decay product.

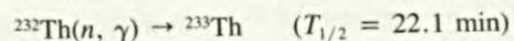
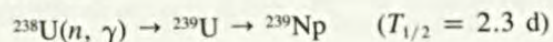
V. Nonconventional Methods for Assessment of Radionuclides

A. NEUTRON ACTIVATION

An activation technique for measurement of radioactive isotopes is sometimes used when the physical half-life is long and thus the specific activity is low, or the decay characteristics are unfavorable for conventional counting. The possibility of performing this type of measurement depends on cross sections, fluence rate, and energy spectral distribution of the neutrons.

The radionuclide produced, which will be subject to counting, must have a much shorter half-life than the target nuclide. For environmental samples a radiochemical separation sometimes has to be performed before the neutron activation. The material of the sample holders must be carefully selected to avoid creating undesired activity. The thermal neutron fluence rate required is of the order of $10^{14}/\text{cm}^2 \text{ s}$. The irradiation time depends upon what is practicable in order to obtain enough activity of the produced isotope, but may be limited by the physical half-life of the product. A few examples of usable reactions are given:

- (i) Measurement of ^{238}U ($T_{1/2} = 4.5 \cdot 10^9 \text{ yr}$) and ^{232}Th ($T_{1/2} = 1.4 \cdot 10^{10} \text{ yr}$) by the reactions



^{239}Np and ^{233}Th can be measured by β or γ counting. The other more short-lived uranium and thorium isotopes present in the sample will not be measured, which is a disadvantage compared to α spectrometry.

- (ii) Technetium-99 ($T_{1/2} = 2.1 \cdot 10^5 \text{ yr}$) has been successfully measured after chemical separation and activation by the counting of ^{100}Tc ($T_{1/2} = 15.2 \text{ s}$) (Foti *et al.*, 1972).

- (iii) The measurement of ^{237}Np ($T_{1/2} = 2.1 \cdot 10^6 \text{ yr}$) is a good example where neutron activation gives better minimum detectable activity (MDA) than α spectrometry. The reaction $^{237}\text{Np}(n, \gamma) \rightarrow ^{238}\text{Np} \rightarrow ^{238}\text{Pu}$ ($T_{1/2} = 86 \text{ yr}$) is used. The maximal ingrowth of ^{238}Pu takes about 14 d and ^{238}Pu is then measured with α spectrometry. The troublesome peak interference between ^{237}Np and ^{238}U is also avoided.

- (iv) Iodine-129 measurements are currently being made in a number of laboratories. The procedure involves isolation of iodine from the sample, followed by irradiation with thermal neutrons to yield the product ^{130}I ($T_{1/2} = 12.4 \text{ h}$) (Kaye *et al.*, 1980).

B. MASS SPECTROMETRY

In special cases mass spectrometry can be applied with advantage for measurement of certain radioactive isotopes, if the specific activity is low and conventional counting and activation techniques are unfavorable. A radiochemical separation must be done before the application of mass spectrometry. A mass-spectrometrically suitable yield determinant should be present or added to the ion source, as the spectrometer and its detector system are adapted to isotope-ratio measurements. This system is based on multiplication of electrons released by heavy ion interactions with the first dynode of a dynode chain in a PM tube.

An example of the usefulness of mass spectrometry is measurement of samples contaminated with artificial and natural uranium isotopes. The long-lived ^{236}U ($T_{1/2} = 2.4 \cdot 10^7 \text{ yr}$), ^{235}U ($T_{1/2} = 7.1 \cdot 10^8 \text{ yr}$) and ^{238}U ($T_{1/2} = 4.5 \cdot 10^9 \text{ yr}$) may be determined in this way. Uranium-236 is especially difficult to measure by α spectrometry due to peak interferences. Another example is measurement of ^{242}Pu and ^{240}Pu in samples contaminated with other plutonium isotopes. Plutonium-242 ($T_{1/2} = 3.8 \cdot 10^5 \text{ yr}$) has low specific activity and ^{240}Pu is impossible to resolve from ^{239}Pu by α spectrometry. In addition, ^{241}Pu , a low-energy β emitter, may also be determined (Krey *et al.*, 1976). Neptunium-237 ($T_{1/2} = 2.1 \cdot 10^6 \text{ yr}$) has also been measured by mass spectrometry because complete decontamination of ^{234}U , which has similar α -particle energies, is difficult (Noshkin *et al.*, 1974).

Carbon-14 is now frequently determined with mass spectrometry, which often is advantageous compared to conventional counting technique.

C. OTHER TECHNIQUES

Activation techniques using high-energy photons or protons can also be used in special cases. Reactions of the (γ, n) and (p, n) type are most likely to be applied.

The techniques using x-ray fluorescence and proton-induced x rays are generally applied for long-lived radionuclides. Counting is performed with Si(Li) detectors or planar Ge(Li) or Ge(hp) detectors. The advantage of these techniques is that they are nondestructive and the x-ray fluorescence method can be applied *in vivo*.

Autoradiography may be used to determine the distribution of activity in organs and plants. Exposure time of films might be long. It is not a quantitative method, and is seldom used for environmental activity levels.

VI. Discussion of Errors and the Need for Uniformity in Gathering and Reporting Data

A. PRECISION AND OVERALL UNCERTAINTY OF COUNTING METHODS

(1) *The precision*, or the reproducibility, of measurements is defined as the extent of agreement between measured individual values of one and the same source. It is possible to produce results of high precision but of low accuracy, e.g., if the standards used for the calibration of the counting systems are in error (Aarkrog, 1979).

The counting uncertainty is used to specify the quality of the physical measurement, and it is estimated as the Poisson standard deviation. If the count rate is $r = N/t \text{ s}^{-1}$, where N is the number of counts and t the counting time, then

this measured count rate r is subject to statistical fluctuations. The standard deviation of r is a function of t and can be expressed as

$$\sigma_r = \sqrt{r/t} = (1/t)\sqrt{N} \quad (11)$$

One generally has to consider also a background count rate. If the source and background counts are additive the standard deviation of the net counts is

$$\sigma = \sqrt{\sigma_T^2 + \sigma_B^2} = \sqrt{r_T/t_T + r_B/t_B} \quad (12)$$

T refers to the counting of the sample plus background (total) and B to the counting of the background.

In the case of a single determination σ estimates the error associated with the counting procedure. To estimate the overall error of the determination errors from sampling and chemical operations must also be included.

An estimate of the overall error of the determination is one standard error of the mean (SE) based on multiple determinations:

$$SE^2 = \frac{1}{n(n-1)} \sum_{i=1}^n (X_i - \bar{X})^2 \quad (13)$$

where $\bar{X}$ is the mean value, X_i is each individual value obtained, and n is the number of observations. It should be noted that a confidence level of 68.3% applies to the confidence interval $\bar{X} \pm SE$; $\bar{X} \pm 2 SE$ gives a confidence level of 95.5%.

In summarizing a set of data $X_1, X_2, \dots, X_n$ the following three quantities should be presented:

(a) the average

$$\bar{X} = \frac{1}{n} \sum_{i=1}^n X_i$$

(b) an estimate of the standard deviation (SD) of the actual set of data $X_1, X_2, \dots, X_n$

$$SD = \sqrt{\sum_{i=1}^n \frac{(\bar{X} - X_i)^2}{n-1}} \quad \text{or} \quad SD = SE \cdot \sqrt{n} \quad (14)$$

(c) the number of measurements n .

(2) *The accuracy or overall uncertainty* of a measurement is defined as the extent of agreement between the observed and the true value. The accuracy can only be determined by comparison with measurement (calibration) of standard sources of the same composition as the sample being determined. This is often difficult to achieve. To reduce the influence of possible contamination in standard sources these are spiked with relatively large amounts of specified radio-

nuclides, which, however, depreciates their value as standards for low-level analysis.

It is recommended that the overall uncertainty be expressed as the arithmetic sum of the uncertainties due to random and assessable systematic errors, i.e.,

$$+(t \cdot SE + \delta), \quad -(t \cdot SE + \delta')$$

(ICRU, 1968) where $t \cdot SE$ is the confidence interval for the mean at a specific probability level, t is Student's t factor, and δ and δ' are the positive and negative limits of estimated maximum values of conceivable systematic errors. Table VI gives the factor t by which the computed SE must be multiplied to derive the desired confidence interval.

B. SAMPLE VARIABILITY

The radioecological variability depends upon four variables; radionuclide, time, sample species, and location.

The coefficient of variation cv is often used in the comparison of amounts of variation in populations:

$$cv = \sqrt{s^2}/\bar{X} \quad (15)$$

TABLE VI
STUDENT'S t FACTOR^a

$n - 1$	Values of t for confidence levels		
	90%	95%	99%
1	6.314	12.706	63.657
2	2.920	4.303	9.925
3	2.353	3.182	5.841
4	2.132	2.776	4.604
5	2.015	2.571	4.032
6	1.943	2.447	3.707
7	1.895	2.365	3.499
8	1.860	2.306	3.355
9	1.833	2.262	3.250
10	1.812	2.228	3.169
15	1.753	2.131	2.947
20	1.725	2.086	2.845
30	1.697	2.042	2.750
60	1.671	2.000	2.660
120	1.658	1.980	2.617
∞	1.645	1.960	2.576

^aFrom ICRU Report No. 12 (1968).

where s^2 is the standard sample variance

$$s^2 = \left[\sum_{i=1}^n (\bar{X} - X_i)^2 \right] / (n - 1) \quad (16)$$

For vegetation samples, air and dust movements, disturbances by animals, washing by surface runoff, and other factors are more variable near the ground.

In a study by Remmenga and Whicker (1967), samples from trees seemed to be more uniform while samples from low-growing vegetation were more variable. A similar observation was made by Holm *et al.* (1982) comparing samples of lichens and mosses growing on ground and rocks: sample content of radionuclides mainly delivered by precipitation, such as ^{210}Pb and ^{137}Cs , showed less variation than the content of those transported exclusively by resuspension and flood (water submersion), such as uranium and radium.

In the study by Remmenga and Whicker (1967) an increase in variability was noted in the early growing season and a decrease of variability in late season. They also found that an occasionally high concentration in one sample of each species was rather frequent. Their results indicate that sample variability can be reduced by increasing the number of field samples and decreasing the number of analyses per sample.

C. UNIFORMITY OF REPORTED DATA

The usefulness of published data is seriously impaired if there is a lack of uniformity in reporting results. To improve the possibility of scientific comparison and utilization of published data, the following recommendations are made.

(i) The internationally defined name of the physical quantity describing radioactivity or spontaneous nuclear decay is *activity*. If A atoms decay per second in an amount of radioactive nuclide in a particular energy state at a given time, then, by definition, the activity of that amount is A . In the International System of Units, SI, the unit of activity is the reciprocal second and since 1975 the special name of this unit is *becquerel*, symbol Bq (ICRU, 1980). The old special unit of activity is the curie (Ci), which is equal to $3.7 \cdot 10^{10}$ Bq (exactly). It is highly desirable that the coherent unit $1 \text{ Bq} = 1 \text{ s}^{-1}$ be used for all quantitative data on activity.

(ii) The activity concentration of environmental samples should preferably be given as becquerels per kilogram on a dry weight basis for soil and for plant and animal tissues. The use of a wet weight basis increases the variability of data for many types of samples, as their water content often varies with weather conditions at sampling or type of preservation before weighing the sample. In intercalibration exercises a dry weight basis is used (except for liquid samples). Drying at 105°C is used as reference temperature, usually for 24 h.

(iii) The amount of sample material should, as far as possible, be related to an environmental quantity, such as plant material per unit area of ground or bottom, particle or plankton content per unit volume of air or water, and number of animals per unit area and/or volume (air, water, soil, etc.).

(iv) Although the situation is somewhat more complicated for samples collected on filters, the results could be given as becquerels per cubic meter of filtered volume. Pore size and type of filter should be specified. Concerning samples of water, it must be clearly stated if filtered or total water was analyzed.

(v) For the presentation of data in diagrams it is difficult to come to a general agreement. For example, depth of soil or sediment columns can be given in units of length or in units of area density (kilograms per square meter, grams per square centimeter, etc.). The final presentation of diagrams is often personally designed. However, it is strongly desirable that the same international system of quantities, units, and reporting systems be accepted and used to make comparison and interpretation of data simple and meaningful.

D. INTERNATIONAL INTERCOMPARISON:

THE USE OF ENVIRONMENTAL NATURAL MATRIX STANDARDS

Standardized sources and reference materials have been discussed previously. The primary role of a radioactivity standard is to enable the results of measurements to be communicated. The International Atomic Energy Agency in Vienna and Monaco and the Environmental Measurements Laboratory in New York have organized intercomparison programs for quite a long time for various radionuclides in sample matrices of different types.

It is not a serious matter if all the laboratories participating in an intercomparison share a minor systematic error in their estimate of the true value for the activity content in an environmental sample. It is more important that the results from the various laboratories are consistent. When low activities are to be measured in materials that require extensive processing during sample treatment and source preparation, there is a great need for comparative studies of standardized samples incorporating known activities in similar materials.

A major problem in the preparation of the required low-level sample materials (vegetation, soil, water, etc.) is the difficulty of obtaining an inactive matrix with which to mix aliquots of the standardized source. Further studies of this problem should be undertaken.

VII. Estimation of Radiation Levels Received by Human Beings

A. THE TRANSFER OF RADIONUCLIDES

THROUGH VARIOUS TRANSPORT CHAINS TO HUMANS

In the environment, transport of radionuclides occurs by action of several specific processes and biogeochemical cycles. This may sometimes change the

external irradiation of humans. If radionuclides enter food chains, they will become transferred to humans, thus giving rise to internal irradiation of the human body and its organs.

Examples of dominant transport processes are turbulent atmospheric or hydrospheric dispersion, wet or dry deposition, resuspension from ground surfaces, sorption (absorption and adsorption), and uptake by living organisms. Transfer to humans may then occur via direct inhalation of radionuclides as gases or aerosols, or by ingestion of radionuclide-containing food or water. Some ingested or inhaled radionuclides are easily absorbed, especially those which are chemically equal or analogous to elements of biological importance. Others, such as those which are highly insoluble or biologically nonavailable, will pass through the gastrointestinal tract, but may be trapped in the lungs.

A quantitative assessment of the irradiation of individuals or populations based on data for known or possible releases of radionuclides from specified sources requires knowledge of the transfer efficiency of each radionuclide for each successive step (or compartment) in the chain of transport processes leading to contamination of humans. To estimate the dose equivalent commitment (H_c) from a given source, UNSCEAR (1972, 1977) has treated this transfer problem by using a compartment model represented schematically in Fig. 26.

Since H_c from a given source is the integral over infinite time of the per capita dose equivalent rate $\dot{H}$, resulting from the radionuclide input in the environment, steps in Fig. 26 can be described in terms of the quotient of the infinite time integral of the appropriate quantity in step j of Fig. 26 to the infinite time integral of the appropriate quantity in the preceding step i . These quotients define the transfer coefficient P_{ij} in the pathways from input of radionuclides into the environment to the subsequent irradiation of humans:

$$P_{ij} = \frac{\int_{-\infty}^{\infty} M_j(t) dt}{\int_{-\infty}^{\infty} M_i(t) dt} \quad (17)$$

The transfer coefficient P_{ij} relates compartment j and the preceding compartment i , and $M_i(t)$ and $M_j(t)$ are the appropriate quantities (e.g., activity concentration, deposition density rate, etc.) in the respective compartment at time

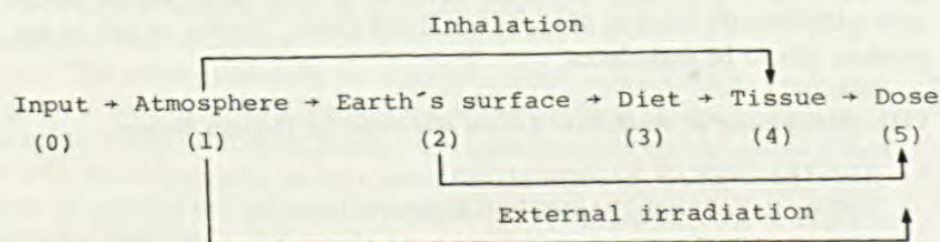


Fig. 26. The UNSCEAR compartment model for estimating the transfer coefficients in a chain of transport processes. [From UNSCEAR (1977).]

t . The total transfer coefficient P for the processes indicated in Fig. 26 is then given by

$$P = (P_{01} \cdot P_{12} \cdot P_{23} \cdot P_{34} \cdot P_{45}) + (P_{01} \cdot P_{14} \cdot P_{45}) + (P_{01} \cdot P_{12} \cdot P_{25}) + (P_{01} \cdot P_{15}) \quad (18)$$

A certain P_{ij} can usually be determined from transfer functions for the actual quantities M_i and M_j ; these functions are usually obtained by regression analysis of reported data.

Strontium-90 in the diet was described by UNSCEAR (1982) with the following function:

$$C(i) = b_1 \cdot \dot{U}(i) + b_2 \cdot \dot{U}(i-1) + b_3 \sum_{m=1}^{\infty} e^{-\lambda_s \cdot m} \cdot \dot{U}(i-m) \quad (19)$$

where $C(i)$ is the average ^{90}Sr concentration in the diet for the year i , $\dot{U}(i)$ is the annual deposition density that year, and $\dot{U}(i-1)$ is the annual deposition density in the previous year. The summation is carried out over the deposition of all preceding years (m), each weighted by an exponential term describing the combined physical decay of ^{90}Sr and any decrease in the availability to plants of ^{90}Sr in soil. The factors b_1 , b_2 , and b_3 and the effective mean life of available ^{90}Sr , λ_s^{-1} , can be derived from reported data by regression analysis. The units for λ_s are reciprocal years and for b_1 , b_2 , and b_3 they are $\text{Bq yr kg}^{-1}/(\text{Bq m}^{-2})$. UNSCEAR (1977, 1982) gives several tables of such data for ^{90}Sr , ^{137}Cs , and other radionuclides for various regions of the world.

For ^{90}Sr values of the various transfer coefficients P_{ij} defined by Fig. 26 are estimated by UNSCEAR (1982) based on both regional and global data on ground deposition, soil and water concentration, uptake by marine biota, etc. From data taken in Argentina, Denmark, and New York City (USA) on ^{90}Sr in total diet, P_{23} was estimated to be $4 \cdot 10^{-3} \text{ Bq yr kg}^{-1}/(\text{Bq m}^{-2})$. The same data also give an average value of the transfer coefficient linking diet and human bone, P_{34} , of 38 Bq yr/kg in bone per Bq yr/kg in diet. Other similar data on various radionuclides and their transfer factors were presented by UNSCEAR (1977, 1982).

The current status of the application of various models used for radiological assessments has been analyzed in the US by the NCRP (1984a). The report recommends that future research emphasize model validation and model simplification.

Models and quantitative prediction of radionuclide transport in ecosystems often make use of the empirical parameter *concentration ratio* CR, simply defined as

$$\text{CR} = (C) \text{ in compartment of interest} / (C) \text{ in reference compartment} \quad (20)$$

where C is the true activity concentration at equilibrium conditions (Whicker and Schultz, 1982). Measured CR values frequently display considerable variability and should be used with care. Radionuclide concentration in water is often influenced by uptake in suspended particulate matter necessitating a filtering procedure, usually by 0.45- μ m millipore filters. The CR values for plant-soil systems show great variations, for instance, CR for uranium is 10^{-1} – 10^{-4} .

Another parameter frequently encountered in the modeling of radionuclide sorption processes is the term *observed ratio* OR, defined as

$$OR = \frac{\left(\frac{\text{activity concentration of radionuclide}}{\text{concentration of analogous element}} \right)_2}{\left(\frac{\text{activity concentration of radionuclide}}{\text{concentration of analogous element}} \right)_1} \quad (21)$$

where 1 may refer to soil and 2 to plant or animal tissue, or 1 may refer to food or milk and 2 to plasma, etc. If no discrimination occurs in the transport from 1 to 2, OR is equal to unity. If $OR < 1$ for the relation $^{90}\text{Sr}/\text{Ca}$ this means that ^{90}Sr is partly discriminated against as compared with Ca. For modeling of ^{90}Sr passage through food chains the OR relationship has been applied successfully. However, the usefulness of OR for $^{137}\text{Cs}/\text{K}$ applied to the ^{137}Cs transport in the biosphere has not been convincing.

The specific activity SA of a radionuclide given as activity per mass unit of its stable element, for instance, $^{210}\text{Pb}/\text{Pb}$ Bq/g, has been found to vary in a transport chain (Persson, 1974). The specific activity ratio $(SA)_2/(SA)_1$ is therefore sometimes a useful parameter in the application of transport models.

B. THE IRRADIATION OF HUMAN BEINGS BY INTERNAL AND EXTERNAL RADIONUCLIDE SOURCES— A BRIEF OVERVIEW

A large amount of data on human exposure to ionizing radiations has been collected, analyzed, and presented by the United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR) since its establishment in 1955. UNSCEAR has received reports from a large number of UN State Members as well as from individual scientists and research institutes. The Committee will also continue to review the levels of radiation to which the world population is, or may become, exposed, and to improve the assessment of risk and detriment entailed by exposure to radiation (UNSCEAR, 1977). Various classes of exposure are presented separately; in the 1982 report the following sources are analyzed:

- (1) Natural radiation sources
- (2) Technologically modified exposures to natural radiation

5. MEASUREMENT AND DOSIMETRY OF RADIOACTIVITY

- (3) Radon and thoron and their decay products
- (4) Nuclear explosions
- (5) Power generation from nuclear fission
- (6) Medical use of radiation
- (7) Occupational exposures

In this context (1)–(5) will be briefly discussed.

Natural radiation sources have always exposed the entire population of the world to ionizing radiation. Internal and external irradiation from radionuclides are the main sources; at sea level cosmic rays contribute about 30% of the total absorbed dose from natural sources, i.e., 15% of the corresponding total effective dose equivalent, H_E .

Table VII (from UNSCEAR, 1977) presents the various contributions to the annual per capita absorbed dose for populations living in areas of normal radiation background. The average annual absorbed dose to the gonads, the whole lung, the cells lining the bone surfaces, and the red bone marrow is estimated to be of the order of 1 mGy. In Table VII the values in parentheses refer to estimates made in the UNSCEAR 1972 report; the differences are due to improved knowledge and analysis of reported data. Furthermore the natural exposure varies substantially from place to place, higher absorbed doses being received at high altitudes or in regions with enhanced natural radioactivity, such

TABLE VII
ANNUAL PER CAPITA ABSORBED DOSE (mGy) FROM NORMAL EXPOSURE TO
NATURAL SOURCES OF RADIATION^a

	Gonads		Whole lung		Bone-lining cells		Red bone marrow	
External irradiation								
Cosmic rays	0.28	(0.28) ^b	0.28	0.28	(0.28)	0.28	(0.28)	
Terrestrial radiation	0.32	(0.44)	0.32	0.32	(0.44)	0.32	(0.44)	
Internal irradiation								
Potassium-40	0.15	(0.19)	0.17	0.15	(0.15)	0.27	(0.15)	
Radon-222 (with daughters)	0.002	(0.0007)	0.30	0.003	(0.0008)	0.003	(0.0008)	
Other nuclides	0.02	(0.014)	0.055	0.091	(0.043)	0.04	(0.019)	
	0.78	(0.93)	1.10	0.84	(0.92)	0.92	(0.89)	

^aFrom UNSCEAR (1977).

^bFigures in parentheses refer to estimates made in the 1972 report. All values and the totals are rounded to two significant figures.

as uranium-containing granite rocks, thorium-containing monazite sand (cf. Section II,A,3), increased radon levels, etc. Individuals of rather large population groups may receive absorbed doses within the interval 0.5 to 2 mGy/yr; the number of persons receiving 5 mGy/yr or more is limited (probably a few million).

Today the inhalation of ^{222}Rn and its decay products is of increasing concern. About one-third of the lung absorbed dose is caused by α -particles, which may have a relative biological effectiveness (RBE) about 20 times higher than that for β and γ radiation in the body. Rapidly increasing knowledge about the release of radon plus decay products from the ground and their presence in indoor air in connection with low ventilation rates (0.1 to 0.2 exchanges per hour) indicates that annual absorbed dose D in the lung could reach levels of 10 to 20 mGy; with a quality factor Q of 20 the corresponding dose equivalent, $H = Q \cdot D$, becomes about 200 to 400 mSv. With an organ weighting factor w_T of 0.12 for the lung the effective dose equivalent, $H_E = w_T \cdot H$, becomes 24 to 48 mSv. Recently isolated cases of ^{222}Rn concentrations in indoor air of 10,000 Bq/m³ have been reported from some countries, in comparison with the usually anticipated normal outdoor concentration of 5 to 10 Bq/m³. This means that if the normal level of ^{222}Rn + decay products could cause an absorbed dose to the epithelial cells of the tracheobronchial tree of 2 mGy/yr (UNSCEAR, 1977) this extreme level of 10 kBq/m³ could cause an H_E of up to 4.85 Sv/yr, depending upon the time spent breathing the air. It is thus obvious that the radon problem could be serious in many locations, and methods for reliable measurements, absorbed dose estimates, and efficient reduction of indoor concentrations should be further developed and improved.

UNSCEAR (1982) discussed the radon problem in detail in a separate Annex to its report (pp. 141-210) and reevaluated the detrimental effect of radon in terms of effective dose equivalent H_E . This gives a substantially increased prominence to radon decay products in comparison with Table VII. In its 1982 report UNSCEAR estimates the global average H_E from natural radiation sources to be 2 mSv/yr, half of which originates from inhalation of radon decay products; further details are given in Table VIII (UNSCEAR, 1982).

Sources of radon and the dosimetry of its decay products in the general environment and in mines in USA have also been studied in considerable detail by the National Council on Radiation Protection and Measurements (NCRP, 1984b).

The radon problem is an example of technologically enhanced exposure to natural sources. Further examples are passengers in aircraft and the use of phosphate fertilizers. UNSCEAR estimates the corresponding annual collective absorbed dose to be about 3000 person-Gy in the first case, from increased cosmic-ray intensity, and about 300 person-Gy in the second case, caused by the ^{238}U decay series in the phosphate mineral. For all natural sources the same quantity is estimated to be $3 \cdot 10^6$ person-Gy for the world population.

TABLE VIII

ESTIMATED ANNUAL EFFECTIVE DOSE EQUIVALENTS (mSv) FROM
NATURAL SOURCES OF RADIATION IN AREAS OF "NORMAL" BACKGROUND^a

Source	External irradiation	Internal irradiation	Total
Cosmic rays			
Ionizing component	0.28		0.28
Neutron component	0.02		0.02
Cosmogenic nuclides		0.015	0.015
Primordial nuclides			
Potassium-40	0.12	0.18	0.30
Rubidium-87		0.006	0.006
Uranium-238 series	0.09	0.95	1.04
Thorium-232 series	0.14	0.19	0.33
Total (rounded)	0.65	1.34	2.0

^a From UNSCEAR (1982).

Power generation from nuclear reactors for production of electrical energy has continued to increase in the 1970s but may level off in the 1980s. The total installed nuclear generating capacity at the end of 1981 was about 162 GW(e) from 279 reactors operating in 23 countries. The projected capacity by the year 2000 is about 1300 GW(e). UNSCEAR has tried to assess the collective dose commitments per MW-yr electrical energy due to releases of radioactive substances from the various steps involved in nuclear power production. Very long-lived radionuclides, such as ^{238}U and ^{129}I , are difficult to discuss in terms of this quantity due to the extremely long exposure time of millions of years. Carbon-14 ($T_{1/2} = 5730$ yr) creates similar problems, but here UNSCEAR (1977) found a 500-year period to be realistic for calculation of its collective dose commitment. For further details the reader is referred to the UNSCEAR reports (1977, 1982). The conclusion in these reports is that the nuclear industry could produce a total collective dose commitment in the future not exceeding 0.03 to 0.06 person-Gy per MW-yr electrical energy.

Nuclear explosions have caused a global distribution and deposition of several artificial radionuclides. Of particular importance for the irradiation of humans are ^{90}Sr , ^{137}Cs , ^{131}I , and ^{14}C , though other radionuclides such as ^{99}Tc , ^{239}Pu , and ^{241}Am may be of increasing concern. The huge amount of fallout data delivered to UNSCEAR since 1955 has been continuously analyzed by successively improved methods. The parameters of the compartment model discussed above in Section VII,A have been reevaluated, and refined transfer functions and transfer coefficients worked out, to enable the Committee to estimate regional and global collective absorbed doses and absorbed dose commitments. The interested reader is advised to study the very extensive material published in the series of UNSCEAR reports.

The conclusion in the 1982 report is that the total global collective effective dose equivalent commitment (H_E)_c from nuclear test explosions up to the end of 1981 is $3 \cdot 10^7$ person-Sv. This is equivalent to about 4 years of natural, normal background irradiation of human beings. So far 10% of this (H_E)_c has been delivered, corresponding to 0.4 years of natural background. The rest, 90%, will be delivered throughout the complete decay of the released radionuclides, ^{14}C being responsible for almost the entire contribution after the year 2000.

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6

Internal Dosimetry for Radiation Protection*JOHN R. JOHNSON*

BIOMEDICAL RESEARCH BRANCH
 ATOMIC ENERGY OF CANADA LIMITED RESEARCH COMPANY
 CHALK RIVER NUCLEAR LABORATORIES
 CHALK RIVER, ONTARIO
 CANADA

I. Introduction	369
II. Basic Concepts	370
A. Dose Equivalent	370
B. Effective Dose Equivalent	371
C. Committed Dose, Annual Limit on Intake (ALI), and Derived Air Concentration (DAC)	372
III. Specific Effective Energy	376
A. Bone	377
B. GI Tract	379
C. Age and Sex Considerations	380
IV. Models of Radionuclide Transport in Humans	380
A. Lung Model	381
B. GI Tract Model	384
C. Metabolic Models	385
V. Individual Monitoring and Dose Assessment	388
A. Conditions Requiring Individual Monitoring	388
B. Derived Investigation Levels	389
C. Dose Assessment	405
VI. Summary	406
References	406

I. Introduction

The International Commission on Radiological Protection (ICRP) issued a major revision to its basic recommendation in 1977 (ICRP, 1977). These revisions resulted in a need to revise the documents dealing with radiation protection

produced by other organizations [International Atomic Energy Agency, (IAEA), 1980, 1982]. The revised recommendations, and updated metabolic models (ICRP, 1975) and dosimetric procedures [Snyder *et al.*, 1974, 1975; International Commission on Radiation Units and Measurements (ICRU), 1979], also led to a review by ICRP of the concepts used in internal dosimetry and of its recommendation on internal dosimetry models and model parameters (ICRP, 1979). This chapter reviews the concepts, models, and model parameters used in that review, and describes how they can be used in internal contamination monitoring programs. Included in the chapter is reference to ongoing work in improving the models and model parameters, particularly since 1979, and to ongoing work on extending these models to include age groups other than adults.

II. Basic Concepts

A. DOSE EQUIVALENT

The *dose equivalent* is defined as (ICRP, 1977; ICRU, 1976)

$$H = DQN \quad (1)$$

where D is the absorbed dose, Q the quality factor (ICRU, 1976), and N the product of all other modifying factors. Currently ICRP does not recommend values different from unity for N (ICRP, 1977) but previously had assigned a value of 5 to N for doses from radioactivity (excluding ^{226}Ra) in bone (ICRP, 1959) to account for possible distributional effects. The new methodology of bone dosimetry, which requires a more detailed calculation (see Section III,A), makes this factor unnecessary.

The absorbed dose rate $\dot{D}_j$ to a mass M_j from activity $A(t)$ can be calculated from

$$\dot{D}_j(t) = KA(t)\bar{E}_j/M_j \quad (2)$$

where K is the constant of proportionality and $\bar{E}_j$ the average energy absorbed in M_j per nuclear transformation. The source of activity $A(t)$ is not necessarily located in the mass M_j . Useful units for this equation are given in Table I.

The basic SI units are seldom used in radiation protection because $\bar{E}_j$ is normally given in MeV. Throughout this chapter the SI (acceptable) system of units will be used ($\text{Gy} = 100 \text{ rad}$, $\text{Sv} = 100 \text{ rem}$).

In general, the dose following the intake of radioactivity to an organ or tissue j (the target organ) will include contributions from radioactivity in more than one organ or tissue (the source organs). The total dose equivalent rate to the target organ can be calculated by

$$\dot{H}_j(t) = \sum_i S_{ij}A_i(t) \quad (3)$$

TABLE I
USEFUL UNITS FOR EQ. (2)

	$\dot{D}_j(t)$	$\bar{E}_j$	$A(t)$	M_j	K
Traditional	rad d^{-1}	MeV	μCi	g	51.2
SI (basic)	Gy s^{-1}	J	Bq	kg	1
SI (acceptable)	Gy d^{-1}	MeV	Bq	kg	1.38×10^{-8}

where $A_i(t)$ is the activity in source organ i at time t and S_{ij} is the factor that converts activity in a source organ i to dose equivalent rate in target organ j . The S_{ij} contain all the factors (K , Q , N , $\bar{E}_j$, and M_j) in Eqs. (1) and (2) and are described in more detail in Section III.

B. EFFECTIVE DOSE EQUIVALENT

The ICRP introduced the concept of a *weighted*, or "effective," dose equivalent in its 1977 recommendations (ICRP, 1977). This concept gives a method of adding doses to various organs and tissues by weighting these doses according to their estimated relative probability of producing "stochastic effects" (ICRP, 1977) in the exposed individual. The effective dose equivalent rate is given by

$$\dot{H}(t) = \sum_j W_j \dot{H}_j(t) \quad (4)$$

where W_j are the weighting factors given by ICRP. They are listed in Table II.

These weighting factors are based on the best estimates of the United Nations Scientific Committee on the Effects of Atomic Radiation (UNSCEAR, 1977) of the risk of radiation-induced fatal cancers and of serious genetic-related ill health in the first two generations following irradiations (the so-called stochastic effects). The ICRP (1980) has reviewed these factors in light of criticism that all cancers and all genetic effects have not been accounted for, and concluded that the uncertainties in the data do not allow better estimates to be derived even if these added risks were taken into account. The review by the Biological Effects of Ionizing Radiation Committee (BEIR, 1980) supports this conclusion (Myers, 1982).

The usefulness of the effective dose equivalent is that it gives a logical method of adding the contributions to risks from radiation doses to various organs or tissues for the purpose of comparing annual doses to annual dose limits, or for calculating collective doses for studies such as environmental assessments or cost/benefit analyses. However, the uncertainties in the W_i (and risk factors) should not be overlooked when calculating risks to individuals.

TABLE II
WEIGHTING FACTORS W_j FOR USE IN EQ. (4)

Organ	Weighting factors ^a
Gonads	0.25
Breast	0.15
Red bone marrow	0.12
Lung	0.12
Thyroid	0.03
Bone surfaces	0.03
Each of the five target organs not listed above that receive the highest dose	0.06
Whole body	1.00

^aFrom the ICRP (1977).

C. COMMITTED DOSE, ANNUAL LIMIT ON INTAKE (ALI), AND DERIVED AIR CONCENTRATION (DAC)

The ICRP (1977) has defined a special quantity, called the *committed (effective) dose equivalent*, for use in occupational exposure. It is the dose equivalent received in the 50 yr following a single intake of radioactivity. That is, if $\dot{H}_j(t)$ is the dose equivalent rate to organ j following an intake at $t = 0$, the committed dose equivalent is

$$H_{50,j} = \int_0^{50 \text{ yr}} \dot{H}_j(t) dt \quad (5)$$

and the committed effective dose equivalent is

$$H_{50} = \sum_j W_j H_{50,j} \quad (6)$$

Referring to Eq. (4), a single acute intake of radioactivity I_0 will result in an $H_{50,j}$ given by

$$H_{50,j} = I_0 \sum_i S_{ij} \int_0^{50 \text{ yr}} R_i(t) dt \quad (7)$$

where $R_i(t)$ is the fractional retention in organ or tissue i following the single intake. The fractional retention functions recommended by ICRP for Reference Man (ICRP, 1975, 1979) are usually used unless other information or models are available. Examples of other information are *in vitro* solubility studies and retention functions derived from past exposures to the material in question. In addition, published models* other than those recommended by the ICRP are available for a limited number of radionuclides and these may be more represen-

tative for given situations. A review of all these models is beyond the scope of this chapter.

The committed dose has two features that make it a useful quantity in radiation protection.

(1) The committed dose from all intakes in a year is essentially independent of the pattern of intake over the year, as can be demonstrated mathematically (Johnson, 1977). The largest difference in the committed dose from different intake patterns that can occur is for a radionuclide with a very long retention in the body. For these types of radionuclides, an intake at the beginning of a year will give at most a 2% higher committed dose than one occurring at the end of the year if the committed dose is calculated to the same time.

Since the committed dose from all intakes in a year is essentially independent of the intake pattern, a secondary limit, namely, the *annual limit on intake* (ALI), can be defined. This quantity is the intake that will result in a committed dose equal to the annual limit on dose. For occupational exposure, the non-stochastic (ICRP, 1977) ALI is given by

$$ALI_{NS} = 0.5/\hat{H}_{50,j}^{\max} \text{ Bq} \quad (8)$$

where 0.5 is the annual limit on dose equivalent (Sv) for occupational exposures to any organ (ICRP, 1977) and $\hat{H}_{50,j}^{\max}$ is the committed dose equivalent per becquerel intake to the organ receiving the highest dose. Similarly, the stochastic ALI (ICRP, 1977) is given by

$$ALI_S = 0.05/\hat{H}_{50} \text{ Bq} \quad (9)$$

where 0.05 is the annual limit on effective dose equivalent (Sv) for occupational exposures and $\hat{H}_{50}$ is the committed effective dose equivalent per becquerel intake.

It is common to use the smaller of the two ALI's from Eqs. (8) and (9) in radiation protection for planning purposes and for calculating such quantities as *derived investigation levels* (DIL) and *derived air concentrations* (DAC). Derived investigation levels are discussed in Section V.B. The DAC is the constant concentration of a radionuclide in air that will result in an intake by Reference Man (ICRP, 1975) of an ALI in a working year. That is,

$$DAC = ALI/(2000 \times 1.2) \text{ Bq/m}^3 \quad (10)$$

where 2000 h is the assumed working year, and Reference Man is assumed to breathe at an average rate of 1.2 m³/h during work.

*As an example, the Medical Internal Radiation Dose (MIRD) Committee of the Society of Nuclear Medicine has published recommended models for specific compounds of some radioactive isotopes of Se, Ga, Tc, Au, I, Hg, Xe, and Fe. Information about these publications can be obtained from the Society of Nuclear Medicine, 211 East 43rd Street, New York 10017.

The DAC is derived using the same concepts as used to derive the maximum permissible concentrations in air (MPC_a) given in ICRP Publication 2 (ICRP, 1959). However, the MPC_a was the concentration in air that would result in the dose to the critical organ (ICRP, 1959) being equal to the annual limit on dose to that organ at equilibrium* (or after 50 yr of exposure if equilibrium was not reached). Minor differences between the MPC_a given in ICRP Publication 2 (ICRP, 1959) and the DAC in ICRP Publication 30 (ICRP, 1979) result from the changes in recommended dose limits for organs and the introduction of the concept of effective dose equivalent. The change in the recommended quality factor (ICRP, 1977) for alpha particles from 10 to 20 also results in changes for alpha-emitting radionuclides, as does the change in the dosimetry methodology for bone (Section III, A). However, the major differences between the MPC_a of ICRP Publication 2 and the DAC of ICRP Publication 30 (such as those for the alkaline-earth elements, uranium, and thorium) arise from changes in metabolic models and model parameters (Johnson, 1983a; Ford and Watson, 1979; Watson, 1982).

(2) The other useful feature of the committed dose from a single intake I_0 is that it is numerically equal to the dose rate at equilibrium (or after 50 yr if equilibrium is not reached) for intakes at a constant intake rate numerically equal to I_0 (Johnson, 1982a). That is, the dose equivalent rate during an intake $I(t)$ is

$$H_j(t) = \sum_i S_{ij} \int_0^t I(\tau) R(t - \tau) d\tau \quad (11)$$

which for a constant intake has the same form as Eq. (7). This means, for instance, that the dose rate will, after 50 yr of intake (or before, if equilibrium is reached) at an annual intake rate of the ALI, be numerically equal to the committed dose from a single intake of one ALI. This point is illustrated in Fig. 1, which is plotted for annual intakes at the ALI for a radionuclide with an effective half-life of 3 yr. After 3 yr, the dose rate is one-half the annual limit on dose, after 6 yr, it is three-quarters the annual limit on dose, etc.; after 15 yr (5 half-lives) equilibrium is essentially reached, and the annual dose will equal the annual limit on dose. The time for this to occur is, of course, dependent on the effective half-life of the radionuclide in question. For example, ^{131}I , with an effective half-life of about 7 d, will reach equilibrium in less than 2 months and hence the dose rate will equal the annual limit on dose after less than 2 months of exposure at the DAC [see Eq. (10)]. Plutonium-239, on the other hand, with a long mean residence time in the lung, skeleton, and liver (ICRP, 1979), will not reach equilibrium even after 50 yr of exposure at the DAC. However, since the integral in Eq. (7) is to 50 yr, the annual dose will be equal to the annual limit on dose in the fiftieth year of exposure.

*Equilibrium is reached when the activity in an organ does not increase with further chronic exposure. That is, the rate that activity is excreted from the organ during chronic intake by an individual is equal to the rate at which it is taken up by the organ.

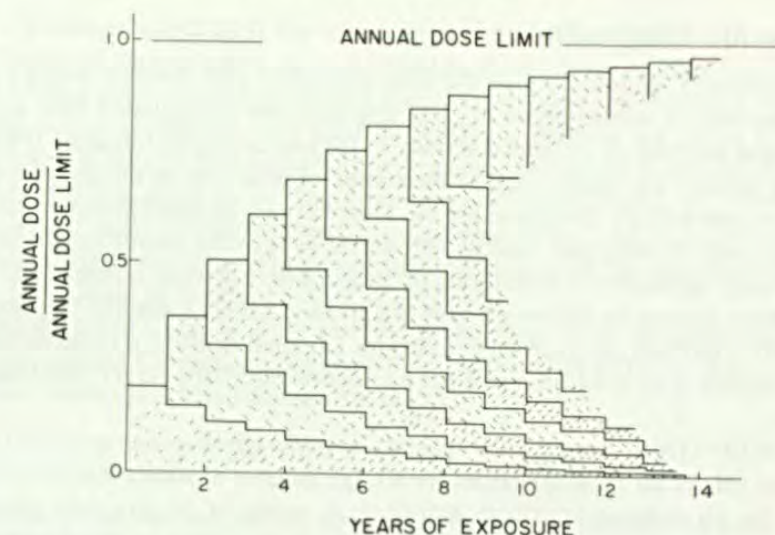


Fig. 1. The ratio of annual dose to annual dose limit for an annual intake (of a radionuclide with an effective half-life of 3 yr) equal to the ALI. The first shaded portion gives the annual doses from intakes in the first year, the second shaded portion the annual doses from intakes in the second year, etc.

While the committed dose (and the secondary limit of the ALI) is a useful quantity for controlling internal contamination, its indiscriminate use can lead to difficulties. For example, difficulties arise if all of the committed dose is assigned to an employee's radiation record in the year that the exposure occurred. Subsequent use of this dose in a risk estimate might well overestimate the risk from this exposure, if the exposure was to a radionuclide with a long effective half-life. In this case, the actual dose in the year of exposure could be as low as one-fiftieth (for a very long effective half-life) of the committed dose and this overestimation, coupled with a latency period of up to 30 yr, would greatly overestimate the risk, particularly if the exposed individual was middle-aged or older. Another difficulty that arises from assigning committed dose to the year the exposure occurred is that the actual dose in subsequent years could exceed the annual limit on dose. This would result if an intake of an ALI of a radionuclide with a short effective half-life occurred, or external exposures equaling the annual limit on dose occurred in years following an intake of a radionuclide with a long effective half-life, although both the likelihood of this occurring, and the amount by which the annual dose limit was exceeded, is small (ICRP, 1977). Because of these difficulties, it has been recommended (Health and Welfare Canada, 1980) that the accumulated internal and external doses in a year, plus the committed dose(s) from intakes in that year, be summed and compared to the annual limit for dose control purposes, but only the accumulated doses be recorded in an individual's records.

III. Specific Effective Energy

The factors S_{ij} of Eq. (3) give the dose equivalent rate in target organ j per unit of activity in source organ i . They also give the accumulated dose in the target organ per unit of integrated activity in the source organ (Johnson, 1982a).

These factors are useful because their units of dose rate to the target organ (Sv/d) per unit activity (becquerel) in the source organ, or dose (Sv) to the target organ per unit of integrated activity (becquerel day) in the source organ, make them directly applicable to monitoring programs where activity is measured. Tables of these factors for Reference Man are given in MIRD Pamphlet 11 (Snyder *et al.*, 1975) (in rads per microcurie hour) for a limited number of radionuclides, and by Snyder *et al.* (1974) (in rem per microcurie day) for a larger set of radionuclides.

The ICRP (1979) has used the quantity *specific effective energy* (SEE) and has given values for these quantities for a large number of source-target combinations for all radionuclides with half-lives in excess of 10 min (and shorter-lived daughters of these radionuclides) in the Supplement to Publication 30 (ICRP, 1979). These factors give the absorbed energy in the target organ j per nuclear transformation in the source organ i and per gram of target organ (MeV transformation⁻¹ g⁻¹). They are related to the S_{ij} (in Sv/Bq d) of Eq. (3) by

$$S_{ij} = 1.38 \times 10^{-5} \text{SEE}(j \leftarrow i)$$

$\text{SEE}(j \leftarrow i)$ can be calculated from (ICRP, 1979; ICRU, 1979)

$$\text{SEE}(j \leftarrow i) = \sum_k \text{SEE}_k(j \leftarrow i) \quad (12)$$

where

$$\text{SEE}_k(j \leftarrow i) = Y_k E_k \text{AF}(j \leftarrow i)_k Q_k / M_j \quad (13)$$

and Y_k is the yield of radiations of type k per transformation of the radionuclide under consideration, E_k the average, or unique, energy of radiation k (mega-electron-volt), $\text{AF}(j \leftarrow i)_k$ the average fraction of energy absorbed in the target organ j per emission of radiation k in source organ i , Q_k the quality factor appropriate for radiation k , and M_j the mass of the target organ in kilograms.

The large set of SEE_k factors for the radionuclides considered in ICRP Publication 30 (ICRP, 1979) were calculated at the Oak Ridge National Laboratory (ORNL) (Watson *et al.*, 1980). These data can be combined and converted to S_{ij} factors for easy use in monitoring programs and in other calculations (Johnson and Dunford, 1983; Dunford, 1982).

In the ICRP calculations of SEE_k it is assumed that $\text{AF}(j \leftarrow i)_k$ is unity for all charged particles in all organs with the exception of bone and the GI tract (see Sections III,A and III,B). That is, with these exceptions the only target or-

gan for charged particles is the source organ itself. The approximation that all the energy of charged particles is absorbed in the organ (excepting bone and the GI tract) is a good one except for very small organs [see, e.g., Johnson (1982b)].

The values of $\text{SEE}_k(i \leftarrow j)$ used by ICRP for photons are given in Reference Man (ICRP, 1975) and $\text{SEE}_k(i \leftarrow j)$ values for fission neutrons were estimated from data given by Dillman and Jones (1975) and Ford *et al.* (1977). The accuracy of these factors is generally considered to be good, assuming that the mathematical phantom used in the calculations to describe the distribution of organs and tissues in Reference Man is adequate.

The absorbed fraction of the charged particle energy in the target tissues of the bone and the GI tract was not taken to be unity by ICRP (1979). Their recommended methodology is summarized below.

A. BONE

The target tissues in bone are assumed to be the bone surfaces (a mass of 120 g, consisting of a 10- μm layer of cells covering all bone surfaces) and red bone marrow. Source organs are assumed to be mineral bone, with activity uniformly distributed throughout its volume, and bone surfaces, with activity uniformly distributed over the surfaces of both trabecular and cortical bone. This procedure is considerably different from the practice previously recommended by the ICRP (1959), which has been used extensively in radiation protection calculations since 1959. At that time it was recommended that the average dose to the skeleton be calculated and a factor of five used in the calculation of dose equivalent (see Section II,A) to account for possible distributional effects.

Absorbed fractions of beta and alpha energies recommended by the ICRP (1979) for the various source-target combinations are given in Table III. The reader is referred to ICRP Publication 30 and the original references given there for details of the calculations used to derive these values. To use Table III, the distribution of radioactivity between the various bone regions used as source organs must be known. Some models of radionuclide retention [see Section IV; see also ICRP (1973) and Johnson and Myers, (1982)] calculate the activity in trabecular and cortical bone and on bone surfaces explicitly. Other models, including most recommended by ICRP (1979), assume that all the radioactivity is either uniformly distributed throughout mineral bone (bone volume) or uniformly distributed on bone surfaces, depending on whether the element in question is known or assumed to be a bone volume or bone surface "seeker" (ICRP, 1979). In addition, the ICRP (1979) has assumed as a general rule that isotopes of volume-seeking elements with half-lives less than 15 d, and in some cases longer-lived ones (e.g., ²³⁰U, see below), are distributed on bone surfaces. This procedure is usually conservative, as some activity will be in mineral bone by this time, and therefore a smaller fraction of the energy is deposited in the

TABLE III
ABSORBED FRACTIONS FOR DOSIMETRY OF RADIONUCLIDES IN BONE RECOMMENDED BY ICRP

Source organ	Target organ	Type of Radionuclide				
		α emitter uniform in volume	α emitter on bone surfaces	β emitter uniform in volume	β emitter on bone surfaces $\bar{E}_\beta > 0.2 \text{ MeV}^a$	β emitter on bone surfaces $\bar{E}_\beta < 0.2 \text{ MeV}^a$
Trabecular bone	Bone surfaces	0.025	0.25	0.025	0.025	0.25
Cortical bone	Bone surfaces	0.01	0.25	0.015	0.015	0.25
Trabecular bone	Red bone marrow	0.05	0.5	0.35	0.5	0.5
Cortical bone	Red bone marrow	0.0	0.0	0.0	0.0	0.0

^a Average energy of each β decay group.

target organs (see Table III). When a radionuclide is assumed to be in bone volume, 20% of the radioactivity is taken to be in trabecular bone and 80% in cortical bone, and when a radionuclide is assumed to be on bone surfaces, the radioactivity is taken to be equally divided between trabecular and cortical bone (ICRP, 1979). It can be seen from Table III that large errors in the dose to the target organs of the skeleton (bone surfaces and red bone marrow) will occur if the wrong source organ is chosen for alpha-emitting and low-energy beta-emitting radionuclides. As an example, the dose to red bone marrow and bone surfaces per becquerel of ingested ^{230}U (half-life = 20.8 d) has been calculated using GENMOD (Johnson and Dunford, 1983) with the assumption that it is either uniformly distributed in bone volume or on bone surfaces. The resulting doses, as shown in Table IV, differ by about a factor of 20.

The ICRP (1975, 1979) has conservatively assumed that this radionuclide is uniformly distributed on bone surfaces, whereas in fact some of it will have been taken up by mineral bone before it all decayed. The actual doses in this case could be as much as a factor of 20 lower than those calculated by the ICRP.

B. GI TRACT

The target organs are considered to be the walls of the four compartments of the GI tract (see Section IV,B). When the source "organs" are the contents of these compartments, the ratio $AF(j \leftarrow i)_k/M_j$ of Eq. (13) is taken to be

$$AF(j \leftarrow i)_k/M_j = \frac{1}{2} \nu/M_i \quad (14)$$

for charged particles.

The factor $\frac{1}{2}$ is used because the dose from charged particles at the boundary between a mass containing uniformly distributed radioactivity and one with no radioactivity is approximately one-half the dose from charged particles at the center of the mass, provided the effective radius of the mass is larger than the range of the particles. The factor ν is introduced to account for the reduction

TABLE IV
DOSE FROM INGESTED ^{230}U ASSUMED TO BE EITHER
UNIFORMLY DISTRIBUTED THROUGHOUT BONE VOLUME
OR ON BONE SURFACE

Source organ	Dose to target organ ^a , Sv per Bq intake	
	Red bone marrow	Bone surfaces
Bone volume	1.1×10^{-8}	1.8×10^{-7}
Bone surfaces	2.8×10^{-7}	3.4×10^{-6}

^a Data from Johnson and Dunford (1983).

in dose to the walls because of the mucous layer covering the inside surfaces of the GI tract. ν is taken to be 1.0, 0.1, and 0.0 for beta-particles, alpha-particles or fission fragments, and recoil nuclei, respectively.

C. AGE AND SEX CONSIDERATIONS

The factors [S_{ij} or $SEE(i \leftarrow j)$] listed in MIRD Pamphlet 11 (Snyder *et al.*, 1975), ORNL-5000 (Snyder *et al.*, 1974), and ICRP Publication 30 (ICRP, 1979) are all calculated for Reference Man. Work is being done to calculate age-dependent factors by first defining age- and sex-dependent mathematical phantoms (Christy, 1980; Eckerman, 1981). From these one can estimate absorbed fractions in the source-target combinations for various ages by using either build-up factors (Henrichs *et al.*, 1982) or Monte Carlo techniques (ICRP, 1975). To date no large listing of age-dependent factors comparable to those in ICRP Publication 30 (ICRP, 1979) has been published. Johnson and Dunford (1983) have used approximate methods to extrapolate the values of $SEE_k(i \leftarrow j)$ for Reference Man (Watson *et al.*, 1980) to a one-year-old Reference Infant with unspecified accuracy. In this context, it should be noted that although large sets of S_{ij} or $SEE(i \leftarrow j)$ factors would be useful for calculations such as genetically significant doses and doses to the female breast, the fact that they are not available is not of vital importance in routine radiation protection practices for two reasons. First, the dose to the source organ is the major contributor to the effective dose equivalent, and this can be easily calculated from Reference Man values by a simple correction for the mass and absorbed fraction as a function of mass (Johnson and Dunford, 1983). Second, the uncertainty in the S_{ij} value is usually small compared to other uncertainties, such as the distribution and retention of the radionuclide, that go into the calculation of dose.

IV. Models of Radionuclide Transport in Humans

The values of $A_i(t)$ [Eq. (3)] and of I_0 and $R_i(t)$ [Eq. (7)] are usually estimated from measured or calculated intakes, body or organ contents, and/or excretion rates with the aid of models describing the distribution and retention of the radionuclide under consideration. The best techniques for estimating these quantities vary considerably among radionuclides, and have been reviewed for a limited number of radionuclides in ICRP Publication 10 (ICRP, 1968). A general model applicable to most radionuclides has been developed by ICRP for this purpose, and parameters of this model for Reference Man have been recommended (ICRP, 1979). This general model consists of three components: a lung model, a GI tract model, and "metabolic" models describing distribution and retention in organs. These models are described separately below. Rate constants and half-lives mentioned in these sections are the biological ones. Radioactive half-lives would have to be included in any calculations of $A_i(t)$ or $R_i(t)$.

A. LUNG MODEL

A lung model for use in internal dosimetry was developed by the ICRP Task Group on Lung Dynamics (ICRP, 1966). It is shown schematically in Fig. 2. For the purposes of radiation protection dosimetry, the ICRP recommends (ICRP, 1977) that the tracheobronchial region (TB, compartments C and D of Fig. 2), pulmonary region (P, compartments E to I of Fig. 2), and pulmonary lymph tissue (PLT, compartments J and K of Fig. 2), be treated as a single composite organ with mass of 1 kg for Reference Man. The nasopharynx (NP) region is not normally considered an "organ" in the ICRP scheme. However, Bailey and James (1979) have shown that the dose to this region can be significant in accident conditions. The recommended weighting factor (see Section II,B) for the composite (TB, P, and PLT) organ is 0.12. It should be noted, however, that the ICRP (1981) has recently recommended using a "regional lung concept" for dosimetry of radon and thoron daughters. The regions of the lung consist of the TB region (mass = 0.04 kg, weighting factor = 0.06) and the P region plus the pulmonary lymph tissue (mass = 0.96 kg, weighting factor = 0.06).

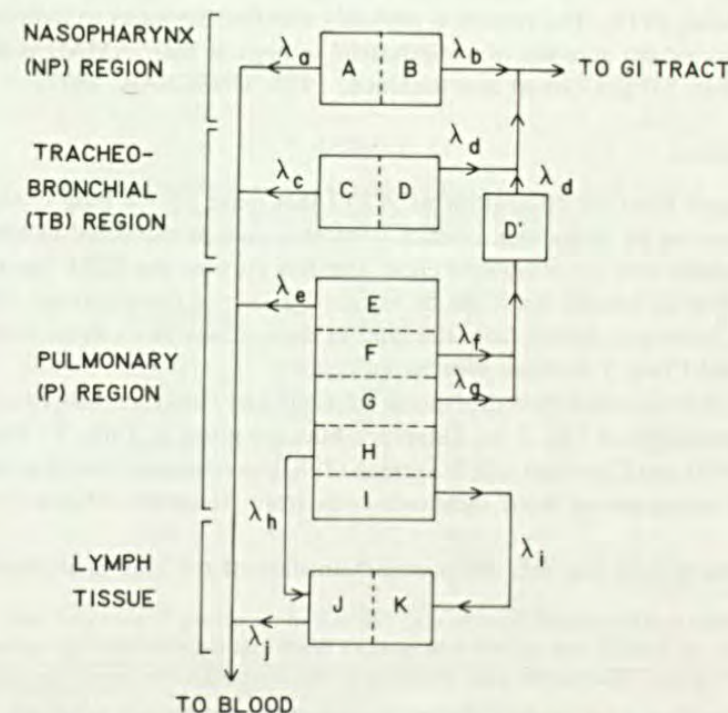


Fig. 2. The compartment model recommended by ICRP (1966) for the lung.

The activity in these regions of the lung will depend on the amount of activity deposited in the various compartments and the rate of clearance from these compartments.

1. Deposition

Fractional deposition of inhaled activity for occupational exposure in the three regions (NP, TB, and P) of the lung is assumed to be a function only of the activity median aerodynamic diameter* (AMAD) of the aerosol. Values recommended by ICRP are given in Fig. 3 (ICRP, 1979). Actual deposition can vary by about a factor of 2 and depends on the dispersion† σ_g of the aerosol about the median diameter, and on the breathing rate, tidal volume, and lung dimensions of the person inhaling the aerosol (Crawford, 1982; Crawford and Eckerman, 1982; James *et al.*, 1981). The fractional depositions, usually assumed when estimating occupational exposures in the absence of measured values for the aerosol (AMAD and σ_g), are $f_{NP} = 0.30$, $f_{TB} = 0.08$, and $f_P = 0.25$ (ICRP, 1979). These values approximate the deposition fractions given in Fig. 3 for AMAD = 1.0 μm , and probably overestimate deposition in the P region of the lung and underestimate the deposition in the NP region, as most measured AMAD's in industrial settings are greater than 1.0 μm (Stuart and Richland, 1973). The reverse is probably true for exposures to radioactivity attached to natural aerosols in nonindustrial settings as their AMAD is usually smaller than 1.0 μm (Stuart and Richland, 1973; UNSCEAR, 1977).

2. Clearance

Clearance from the compartments A to I that make up the lung is assumed to be governed by first-order kinetics (clearance rate proportional to contents) and to depend only on compound class. For this purpose the ICRP has recommended that all inhaled materials be assigned to one of three classes: Class D refers to materials cleared from the lung in days, Class W to those cleared in weeks, and Class Y to those cleared in Years.

The ICRP-recommended deposition fractions and clearance half-times from the compartments of Fig. 2 for Reference Man are given in Table V. Hofmann *et al.* (1979) and Crawford and Eckerman (1982) have recommended a methodology of extrapolating these clearance rates from Reference Man to younger ages.

It is recognized that this compartmentalization of the lung is artificial, and

*The AMAD is the median diameter (i.e., one-half of the activity is associated with particles smaller than the AMAD, and one-half with particles larger than the AMAD) of an aerosol with a density of 1 g/cm³ that has the same aerodynamic characteristics as the aerosol in question.

† The quantity σ_g is the geometric standard deviation of the aerosol, which is normally assumed to be log-normal (Raabe, 1979).

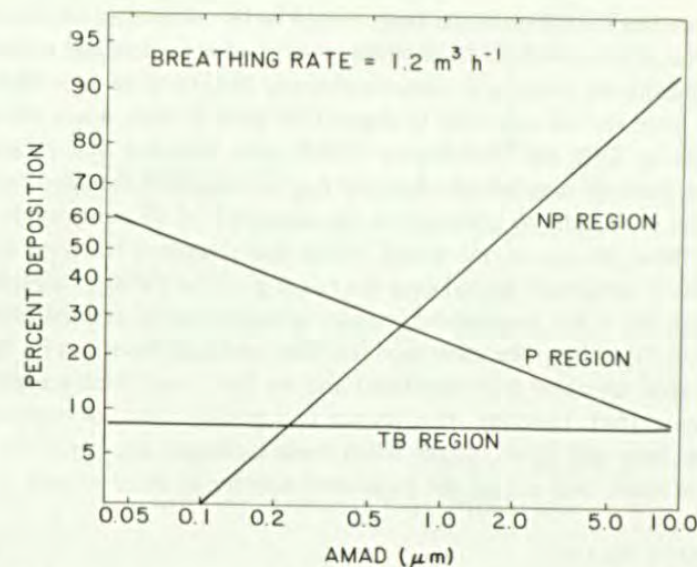


Fig. 3. The deposition probability as a function of AMAD recommended by ICRP (1979) for occupational exposure.

TABLE V

DEPOSITION FRACTIONS D_i AND CLEARANCE HALF-TIMES ($t_{1/2} = \ln 2 / \lambda_i$) FOR THE ICRP LUNG MODEL SHOWN IN FIG. 2

Region	f_i	Compartment	Class D		Class W		Class Y	
			D_i	$t_{1/2}, d$	D_i	$t_{1/2}, d$	D_i	$t_{1/2}, d$
NP	0.30	A(B) ^a	0.5	0.01	0.1	0.01	0.01	0.01
		B(G)	0.5	0.01	0.9	0.4	0.99	0.4
TB	0.08	C(B)	0.95	0.01	0.5	0.01	0.01	0.01
		D(G)	0.05	0.2	0.5	0.2	0.99	0.2
P	0.25	E(B)	0.8	0.5	0.15	50	0.05	500
		F(G)	0	—	0.4	1	0.4	1
		G(G)	0	—	0.4	50	0.4	500
		H(L)	0.2	0.5	0.05	50	0.135	500
		I(L)	0	—	0	—	0.015	500
Lymph tissue	—	J(B)	—	0.5	—	50	—	1000
		K	—	—	—	—	—	∞

^aDeposition probability in a compartment is $f_i D_i$.

^b(B) = transferred to blood, (G) = transferred to GI tract, and (L) = transferred to lymph tissue.

that clearance from a region of the lung should be by competing processes [see, e.g., Johnson *et al.* (1983c)]. In addition, studies of retention and excretion of fused aluminosilicate particles in animals (Snipes, 1983) and humans (Bailey and Fry, 1983) have shown clearance to depend on particle size, a not unexpected result. Work by Diel and Mewhinney (1983) also indicates that clearance of alpha-active material may be enhanced by fragmentation of the deposited aerosol. It is also unrealistic to assume that the clearance of all compounds will fit into one of three classes, as one would expect that clearance fractions and rates should be fairly uniformly spread over the range given in Table V. Despite these shortcomings the ICRP lung model remains a useful tool in internal dosimetry because of its flexibility (the clearance fractions and half-lives can be modified to fit measured retention and excretion) and its fairly universal acceptance. It must be recognized, however, that its use can greatly over- or underestimate doses to the lung and other organs when these estimates are based on intakes or excretion rates, and not on the measured activity in these organs.

B. GI TRACT MODEL

The model recommended by ICRP (1979) for transport of material through, and for dosimetry in, the GI tract is shown in Fig. 4. It is based on the work of Eve (1966); however, Eve considered that radioactivity moves through the various compartments as a "slug" irradiating only the adjacent portion of the walls as it passes. Dolphin and Eve (1966) showed that the average dose to the walls could be adequately approximated by assuming that there was instantaneous mixing in each compartment and that clearance was governed by first-order kinetics. They recommended mean residence times for clearance from these compartments in Reference Man of 1, 4, 13, and 24 h for the stomach (S), small intestine (SI), upper large intestine (ULI), and lower large intestine (LLI), respectively. These mean residence times and first-order kinetics are also currently recommended by ICRP (1979). Crawford-Brown (1983) has suggested modifications to make this model applicable to all ages.

The ICRP recommends that the wall of each compartment of the GI tract be considered a separate organ for the purposes of dosimetry calculations for radiation protection. Each of these organs is to be considered with the other unnamed organs in assigning a weighting factor of 0.06 to each of the five unnamed organs receiving the highest dose (see Section II,B).

According to the model, ingested radioactivity can be absorbed from either the stomach or the small intestine into the blood. If complete absorption is assumed (fraction absorbed, $f_i = 1$) then absorption is assumed to take place directly from the stomach. If f_i is less than one, radioactivity is assumed to be transferred to the small intestine before absorption. Since the dose to organs other than the walls of the GI tract is directly proportional to f_i for ingestion exposures, any uncertainty in f_i will cause an equal uncertainty in the calculated

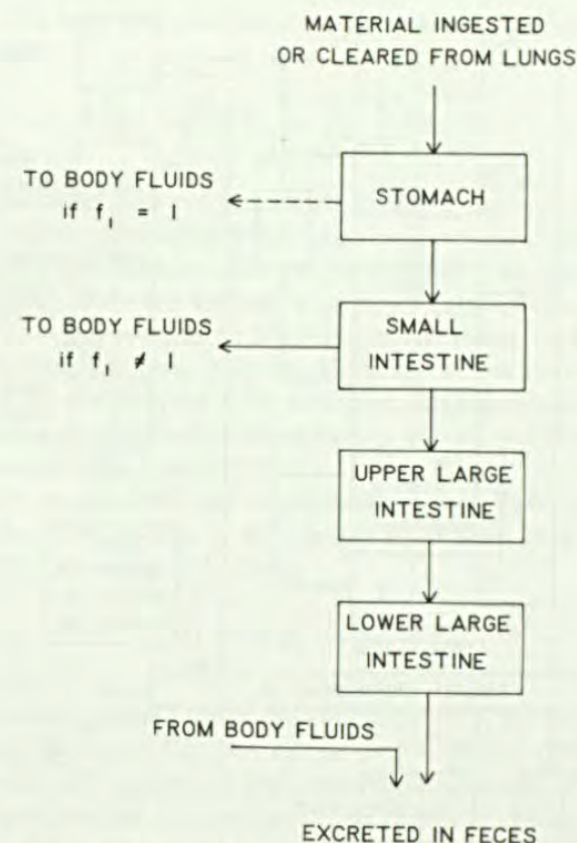


Fig. 4. The compartment model recommended by ICRP (1979) for the GI tract.

dose. Unfortunately, f_i is one of the parameters used in internal dosimetry calculations that has shown the most variability. It is known to depend not only on the chemical form of the radionuclide, but on other contents of the GI tract at the time of ingestion. Variations as large as three orders of magnitude have been noted for actinide radionuclides in animal experiments (Harrison, 1982; Thompson, 1982).

C. METABOLIC MODELS

1. General Model

The general metabolic model recommended by the ICRP (1979) for most radionuclides is shown in Fig. 5 (compartments X_{ij}) along with the interconnected lung and GI tract models. It is described in more detail elsewhere (Adams *et al.*, 1978; ICRP, 1979; Johnson and Myers, 1981) and general characteristics of these types of models have been described in detail by Skrable *et al.* (1980).

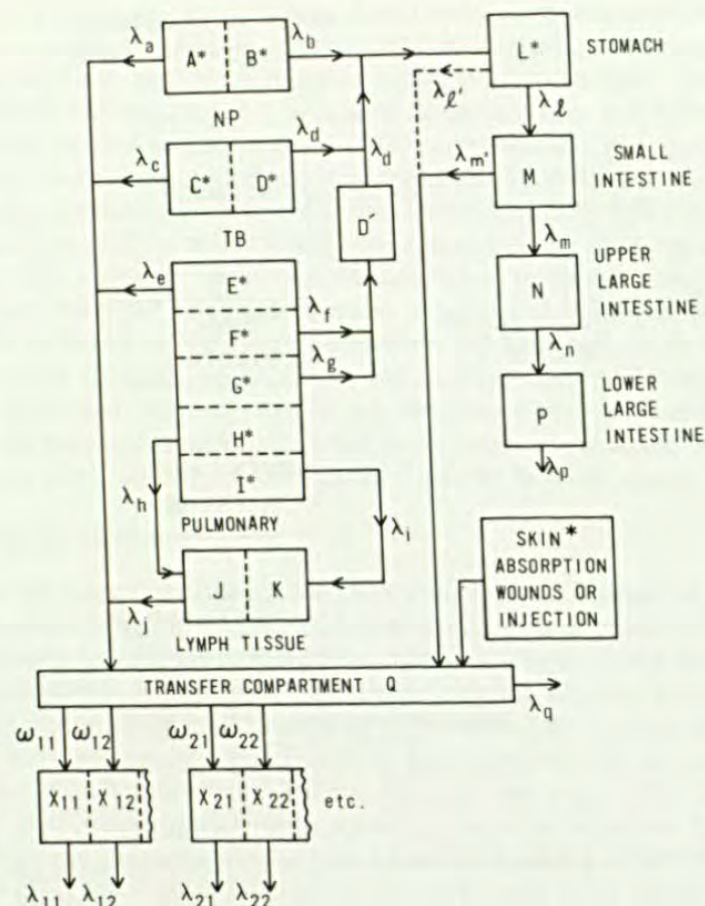


Fig. 5. The general model used by ICRP (1979) to calculate doses from occupational exposure to most radionuclides. Compartments with asterisks indicate those through which radioactivity can enter the body.

Activity entering the blood or transfer compartment (Q) is taken up by the various organs or excreted. Generally, the data available to describe uptake and retention by the organs are in the form of a fractional retention function

$$R_i(t) = \sum_k a_{ik} e^{-\lambda_{ik} t} \quad (15)$$

where the λ_{ik} are identified in Fig. 5, and a_{ik} are the fractions of radioactivity entering the transfer compartment that are taken up by compartment k of organ j . The a_{ik} are related to the rate constants ω_{ik} of Fig. 5 by

$$a_{ik} = \omega_{ik} / \lambda_{TC} \quad (16)$$

The λ_{TC} is the total rate constant for activity leaving the transfer compartment, given by

$$\lambda_{TC} = \lambda_q + \sum_i \sum_k \omega_{ik} \quad (17)$$

This general model does not include recycling of radioactivity from organs back into the transfer compartment for redistribution to the organs or for excretion. Except for a few elements, data are not available from which such models can be developed. The error introduced in the calculation of committed doses from intakes or organ contents by not including recycling explicitly is usually small, as recycling has been included implicitly in the development of the parameters for the equation for $R_i(t)$. However, large errors in doses may occur if calculations are based on excretion rates or blood concentrations and recycling is not considered (Wooten, 1983).

It should also be noted that the committed dose depends on the integral [Eq. (2)] of $R_i(t)$, not on the value of $R_i(t)$ at any fixed time. If $R_i(t)$ is given the form of Eq. (15), the integral becomes

$$\int_0^\infty R_i(t) dt = \sum_k \frac{a_{ik}}{\lambda_{ik}} \quad (18)$$

which is proportional to the equilibrium concentration in the organ from a constant intake (see Section II,C). This can be used to decide if the values of a_{ik} and λ_{ik} give reasonable agreement with organ concentrations of stable isotopes of (or stable isotopes that are assumed to be chemical analogs to) the radionuclide being considered.

Information from which values of a_{ik} and λ_{ik} [Eq. (18)] appropriate to Reference Man can be derived is given for most radionuclides in ICRP Publication 30 (ICRP, 1979). That publication recommends fractional uptake from the transfer compartment (see Fig. 5) by the organs, and the fractional retention and retention half-time of each component (or compartment) in the organs. The a_{ik} [Eq. (18)] are the product of the fractional uptake and fractional retention of each organ compartment to which the rate constants λ_{ik} also apply. These values are usually considered adequate for radiation protection purposes for all adults, but could introduce large errors if used for younger subjects. Since doses are sometimes required for these age groups for environmental assessment and other studies, Johnson and Dunford (1983) and others [see, e.g., Adams (1981)] have extrapolated these adult values to other age groups. These extrapolations will usually be as accurate as the adult values for calculating committed dose if the calculated concentrations in the organ or tissue being considered agree with measured ones [see Eq. (18)].

2. Specific Models

The main drawback of the general model described above is its lack of recycling of radioactivity from organs back to blood for further redistribution to organs or for excretion. Thus, while the committed dose calculated using this model is usually adequate for radiation protection purposes, the relationship between blood and urine concentrations and organ burdens given by these models can be misleading. It is, therefore, preferable to use more realistic models that include recycling when they are available, in conjunction with monitoring programs.

Two models that include recycling explicitly, which have gained wide acceptance in internal dosimetry calculations for radiation protection purposes, and have been extended to all age groups, are the iodine model (Riggs, 1952; Johnson, 1981) and the alkaline-earth model (ICRP, 1973; Johnson and Myers, 1981; Johnson, 1983b). These models will be described in more detail in the next section. Other recycling models, less widely used, include a cerium model (NCRP, 1978), a tritium model (Sanders and Reinig, 1968), an iron model (Robertson *et al.*, 1983) and a technetium model (Hays and Berman, 1977). Detailed descriptions of these models are beyond the scope of this chapter, and the reader is referred to the quoted references if more detail is required.

V. Individual Monitoring and Dose Assessment

The development of large computer data bases containing the SEE or S_{ij} factors described in Section III, and standard metabolic parameters for the models described in Section IV, allows the efficient calculation of such quantities as committed doses per unit intake and ALI's for a large number of radionuclides (ICRP, 1979; Killough *et al.*, 1978; Johnson and Dunford, 1983). The results of these calculations are used primarily in the evaluation and limitation of doses to critical groups, in environmental assessment of releases of radionuclides into the environment, and in defining individual monitoring programs for potentially exposed workers. This section deals exclusively with the latter use.

A. CONDITIONS REQUIRING INDIVIDUAL MONITORING

The ICRP (1968, 1977) has recommended that the committed dose be used in internal dosimetry for comparison to annual dose limits. It is therefore appropriate to use the secondary limit of the ALI (ICRP, 1977) for planning individual monitoring programs. The ICRP (1982) has also recommended that persons working in Working Condition A be subjected to an individual monitoring program. Working Condition A refers to conditions where the annual dose might exceed three-tenths of the annual limit on dose. It is normal to interpret this recommendation conservatively in the sense that "might" refers to a small probability of annual doses of this magnitude occurring (NRPB, 1982). For in-

ternal contamination, Working Condition A refers to those working conditions under which all intakes of a particular radionuclide in a year by any individual might exceed three-tenths of the ALI for that radionuclide (ICRP, 1982). A decision on whether three-tenths of the ALI might be exceeded must be based on considerations of factors such as the amount and form of the radionuclide being handled, the type of containment used, the type of operations being performed, and results of workplace monitoring.

Individuals working in Working Condition A should be monitored to ensure that doses from any significant internal contaminations are appropriately estimated and recorded. Individual monitoring of workers not in Working Condition A (i.e., in Working Condition B) is not required, but may be desirable to demonstrate that no significant contamination has occurred or that doses are as low as reasonably achievable, ALARA (ICRP, 1982).

Even though a workplace might be classified as Working Condition A, it is unusual for the results of individual monitoring to reveal significant contamination, as the objective in the design of most facilities or experimental arrangements is to eliminate internal contamination. In these situations individual monitoring is done to ensure that the facility or experiment is operating as planned. Examples of exceptions to this are exposure to tritium in heavy water nuclear reactors and to uranium in uranium mills and fuel processing facilities. In these cases monitoring is done to routinely assess doses [e.g., tritium; Johnson (1982c)] or to ensure that there is no significant buildup of radioactivity in the body with time [e.g., uranium facilities; Fisher *et al.*, (1983)].

B. DERIVED INVESTIGATION LEVELS

When monitoring is done to ensure that significant internal contamination has not occurred, it is customary to derive a value for any monitoring result, called a *derived investigation level* (DIL), below which no further action is required, as any dose associated with the contamination will be insignificant. If a result is above this level, an investigation to determine the magnitude of the contamination and of the resulting committed dose would be required.

The ICRP (1977) has recommended that these DIL's be set with reference to the ALI for the radionuclide being considered. In routine monitoring, where the time of monitoring is not related to the work being performed in the laboratory, they recommend that the DIL be set with reference to an intake of $0.3 \text{ ALI}/f$, where f is the annual monitoring frequency. In special monitoring, related to a particular job that has a high risk of causing internal contamination or is related to measured contamination in the workplace, they recommend that the DIL be set with reference to an intake of one-twentieth of the ALI (ICRP, 1968, 1977). The calculation of DIL's is straightforward for the special monitoring case, once a metabolic model is selected, because the time of any intake is known. For routine monitoring, the time of intake is not known, and DIL's will vary considera-

bly depending on the selection of this parameter. The ICRP (1982) has recommended that the intake be assumed to occur at the midpoint of the monitoring period. This procedure has been shown to give a reasonably unbiased estimate of dose during intermittent exposures (ICRP, 1971), but it can allow significant doses to occur without exceeding the DIL in some monitoring programs. For this reason others (Health and Welfare Canada, 1980; Brodsky, 1983) have conservatively assumed for the calculation of DIL's that a single intake occurred at the beginning of the monitoring period (i.e., immediately after the last monitoring result was obtained).

The difference between these two approaches to selecting DILs can be illustrated as follows. If the retention of a radionuclide in the body or an organ can be described by a single exponential function of the form

$$R(t) = \exp(-\ln 2 \cdot t/T_{\text{eff}})$$

where T_{eff} is the effective half-life of radionuclide, then the DIL will be equal to

$$\begin{aligned} \text{DIL}_1 &= (0.3 \text{ ALI}/f)R(T/2) \\ &= (0.3 \text{ ALI} \cdot T/365) \exp(-\ln 2 \cdot T/2T_{\text{eff}}) \end{aligned} \quad (19)$$

when the intake is assumed to have occurred at the midpoint of the monitoring period T ($T = 365/f$). If the intake were assumed to occur at the beginning of the monitoring period, the DIL would be given by

$$\begin{aligned} \text{DIL}_2 &= (0.3 \text{ ALI}/f)R(T) \\ &= (0.3 \text{ ALI} \cdot T/365) \exp(-\ln 2 \cdot T/T_{\text{eff}}) \end{aligned} \quad (20)$$

The maximum doses that could accumulate without exceeding these DILs occur for intake at the beginning of the monitoring period, and are given in Fig. 6. It is seen that the maximum committed dose that could occur without exceeding DIL_2 is limited to $15/f$ mSv, or $0.3/f$ of the annual limit on dose. The maximum committed dose that could occur without DIL_1 being exceeded increases exponentially when the monitoring period exceeds the effective half-life for the radionuclide being monitored. It has therefore been recommended (Health and Welfare Canada, 1980) that DIL_2 be used in routine monitoring to ensure that a monitoring result that could indicate that very large doses may have occurred will not be dismissed as being below $15/f$ mSv. Values for DIL_2 are given below for some of the important radionuclides encountered in occupational exposure.

1. Tritium Oxide

Tritium oxide (HTO) is quickly equilibrated with body water upon intake by inhalation, ingestion, or skin absorption (NCRP, 1979). Most of the HTO will

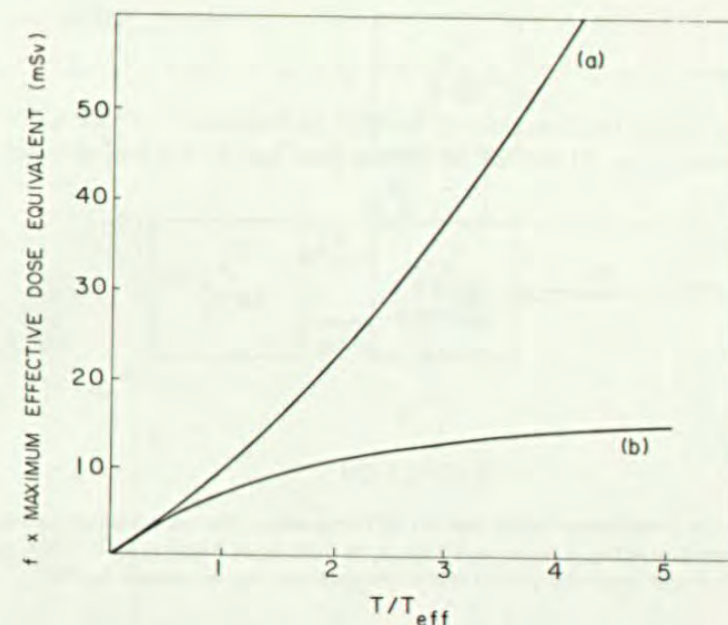


Fig. 6. The maximum committed effective dose equivalent that can occur without DIL_1 [Eq. (19)] being exceeded is given by curve (a), and that which can occur without DIL_2 [Eq. (20)] being exceeded by curve (b), as a function of the ratio of the monitoring period T to the effective half-life T_{eff} . The annual monitoring frequency is given by $f = 365/T$.

be excreted in urine, breath, and sweat with a half-life of about 10 d, and a small fraction will become incorporated into organic molecules. Sanders and Reinig (1968) have suggested a three-compartment model shown in Fig. 7. Applying the appropriate parameters to the model given in Fig. 7 will result in the fractional urinary excretion of tritium, following a single intake, of

$$E(t) = e^{-0.069t} + (4 \times 10^{-3})e^{-0.015t} + (4 \times 10^{-5})e^{-0.0020t}$$

which approximates the average values reported in the literature (NCRP, 1979). The excretion rate following unit input into the body water compartment of Fig. 7 was used to calculate the DIL for urinary analysis following HTO exposures using Eq. (20). This DIL is given in Fig. 8.

2. Cobalt-60

The model used for cobalt retention is shown in Fig. 5. Cobalt can exist as Class D, W, or Y in the workplace and half of the cobalt entering the blood (transfer compartment) is excreted directly (ICRP, 1979). Of the remainder, 5% is transferred to the liver and 45% is equally distributed throughout all other soft tissue. In the latter compartment, the fractional retention is governed by the

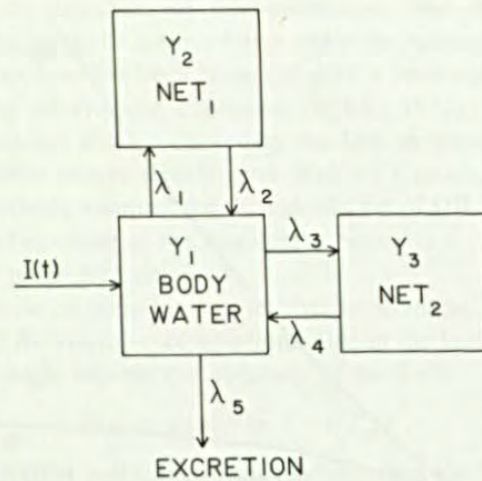


Fig. 7. The compartment model used for HTO exposures. The rate constants used to get the urine concentration of Fig. 8 are $\lambda_1 = 0.0011$, $\lambda_2 = 0.02$, $\lambda_3 = 0.00011$, $\lambda_4 = 0.002$, and $\lambda_5 = 0.0693$. The bound (nonexchangeable) tritium compartments are represented by $\text{NET}_{1,2}$.

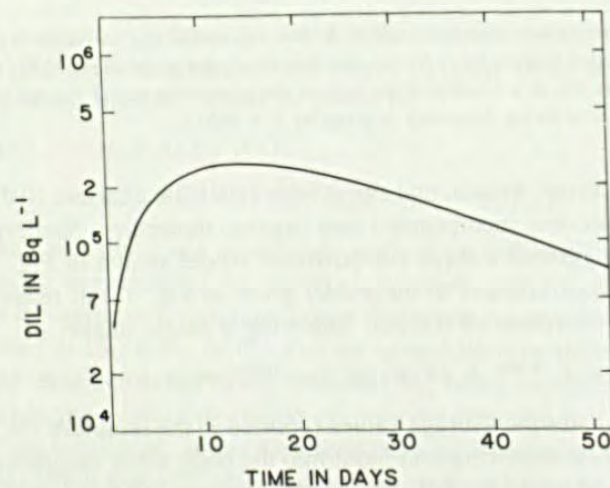


Fig. 8. The DIL of tritium in urine as a function of the monitoring period.

function (ICRP, 1979)

$$R'(t) = 0.5e^{-1.4t} + 0.3e^{-0.12t} + 0.1e^{-0.012t} + 0.1e^{-0.00081t}$$

The DILs for ^{60}Co calculated by GENMOD (Johnson and Myers, 1981), using this function and the GI and lung models of Section IV are given in Fig. 9.

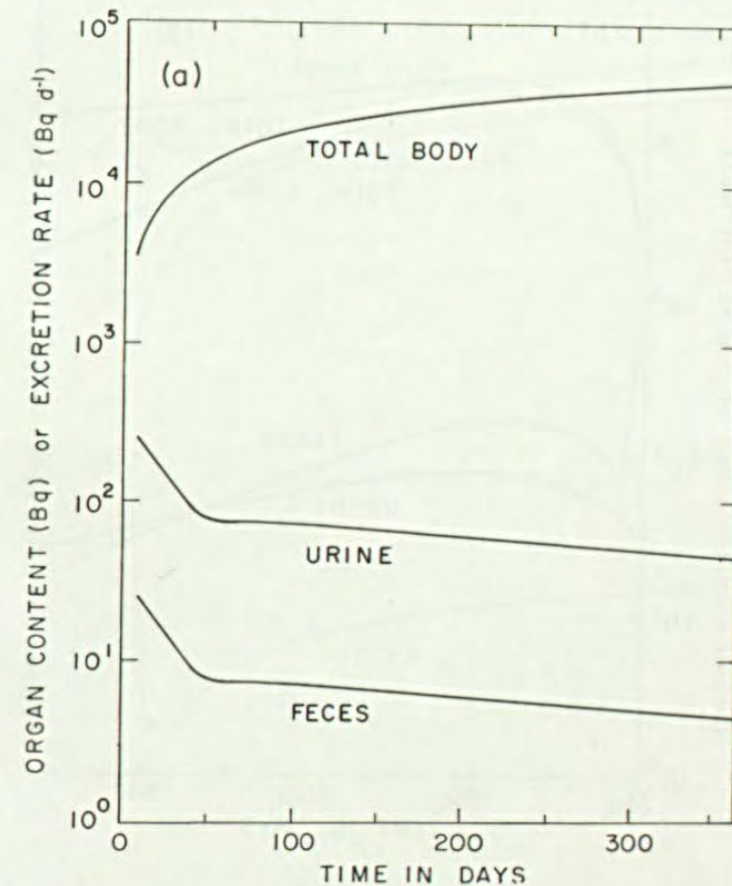


Fig. 9. The DILs for organ contents and excretion rates for inhalation exposures to ^{60}Co as a function of the monitoring period for (a) Class D, (b) Class W, and (c) Class Y compounds (*continues*).

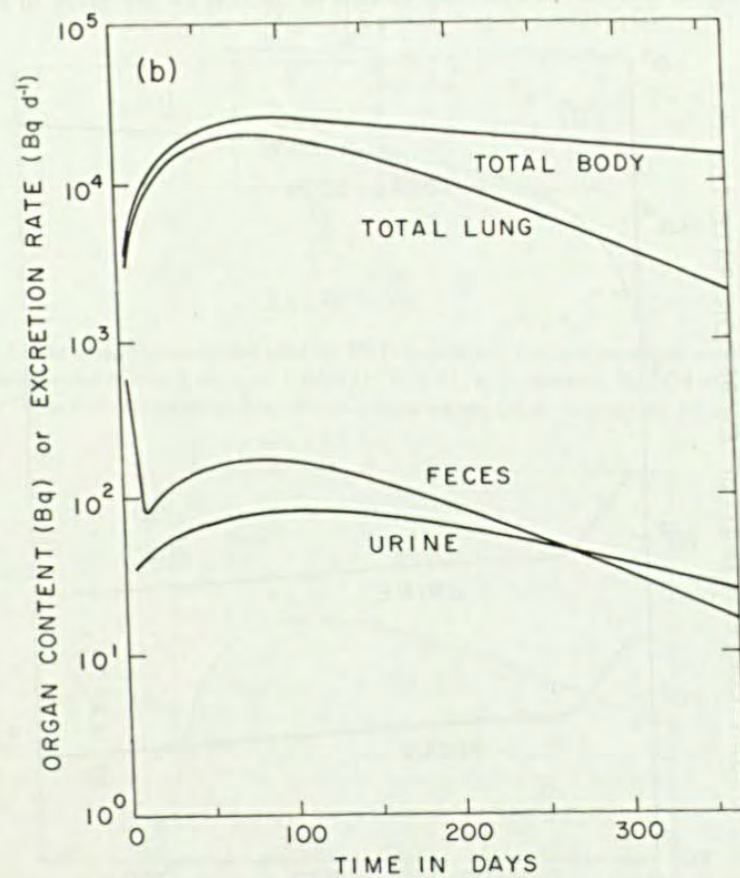


Fig. 9. (continued)

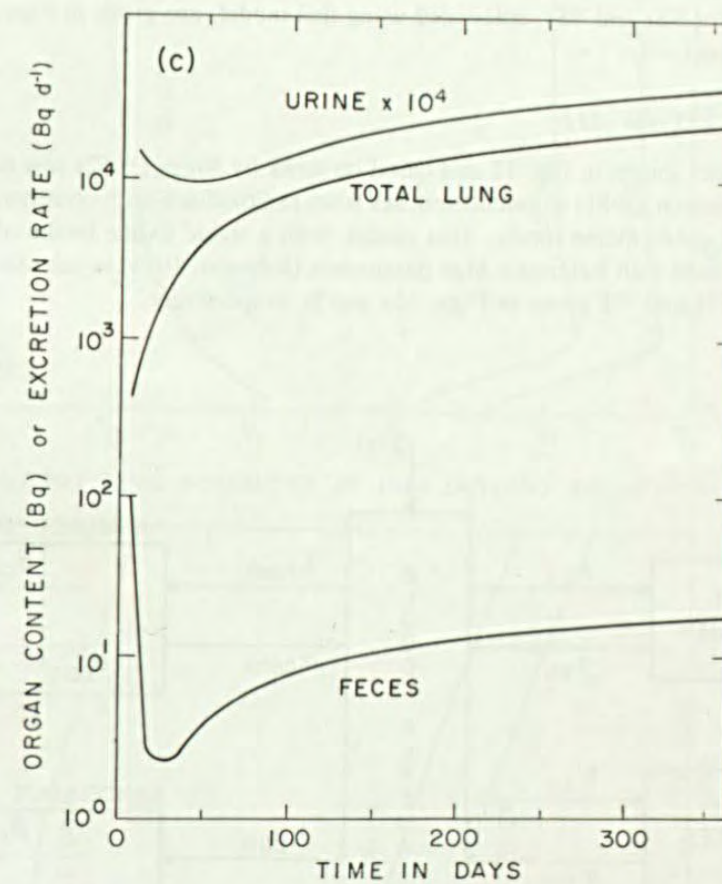


Fig. 9. (continued)

3. Strontium-85 and -90

The ICRP (1973) model for alkaline-earth metabolism has been modified as shown in Fig. 10, and the output of the ICRP lung and GI tract models given in Figs. 2 and 4 form the input $I(T)$ to this model (Johnson and Myers, 1981). The DILs for ^{85}Sr and ^{90}Sr , calculated using this model, are given in Figs. 11a and b, respectively.

4. Iodine-125 and -131

The model shown in Fig. 12 and based on work by Riggs (1952) was developed by Johnson (1981) to calculate doses from radioiodines with consideration of age and stable iodine intake. This model, with a stable iodine intake of 200 $\mu\text{g/d}$, was used with Reference Man parameters (Johnson, 1981) to calculate the DILs for ^{125}I and ^{131}I given in Figs. 13a and b, respectively.

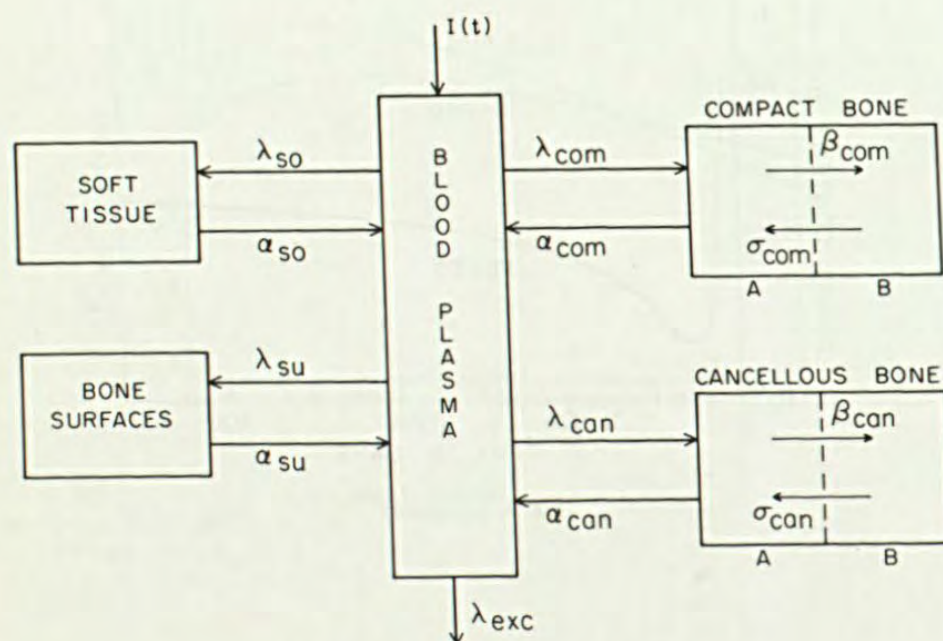


Fig. 10. The compartment model of alkaline-earth metabolism used with the lung and GI tract models of Fig. 2 and 4 to calculate the DILs for ^{85}Sr and ^{90}Sr , as given in Fig. 11.

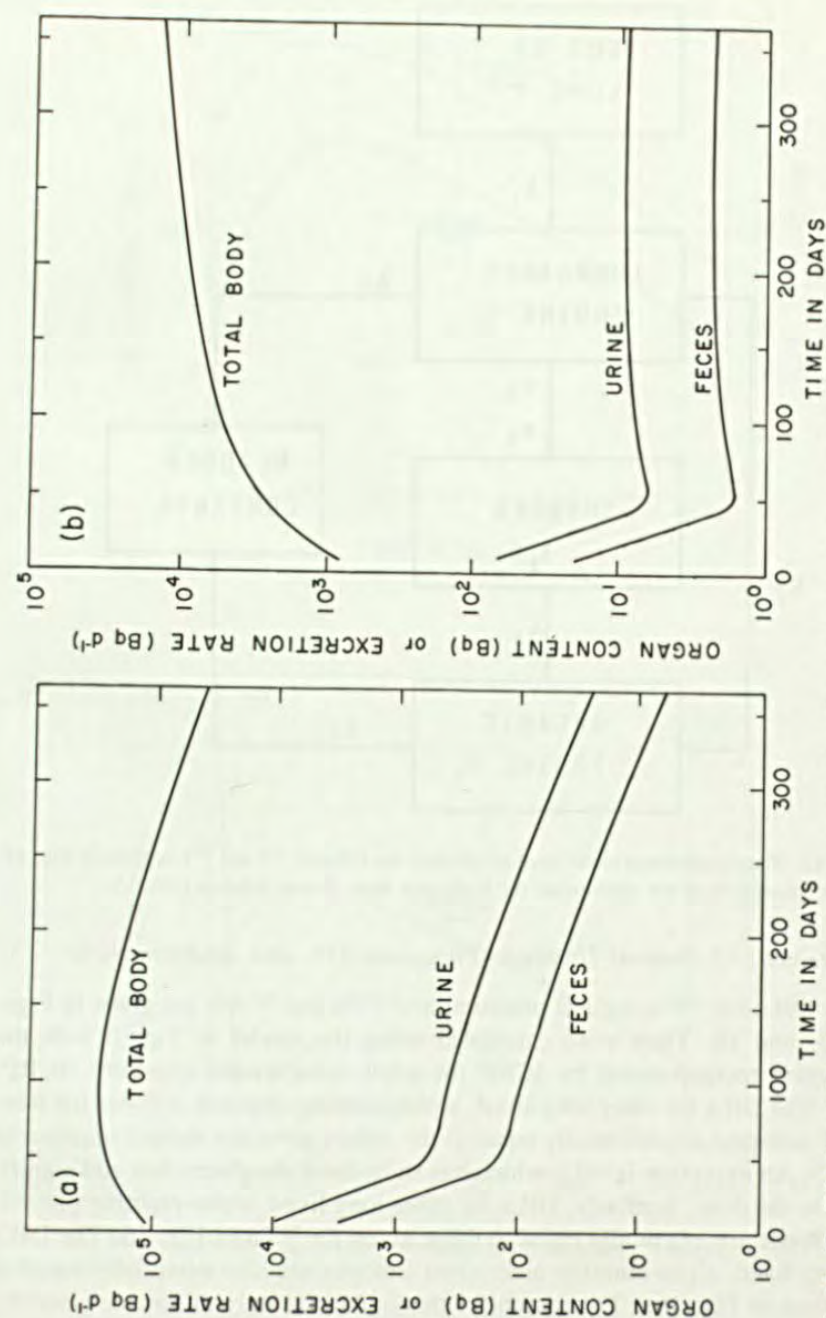


Fig. 11. The DILs for whole-body content and excretion rate of (a) ^{85}Sr and (b) ^{90}Sr for Class D (Classes W or Y rarely occur) inhalation exposure as a function of the monitoring period.

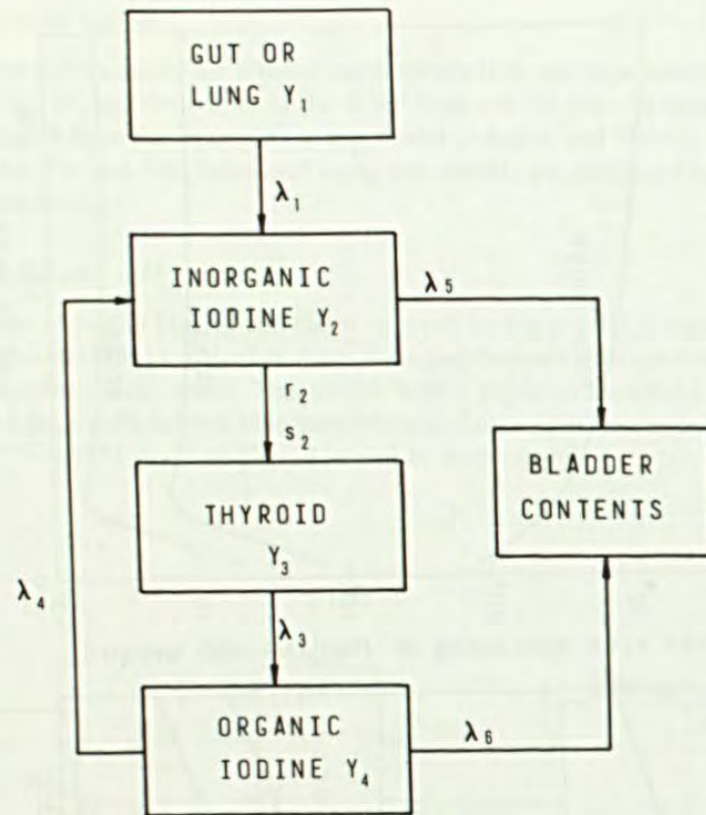


Fig. 12. The compartment model used to calculate the DILs of ^{125}I and ^{131}I as given in Fig. 13. The rate constants used are appropriate for Reference Man. [From Johnson (1981).]

5. Cesium-137, Natural Uranium, Plutonium-239, and Americium-241

The DILs for ^{137}Cs , natural uranium, and ^{239}Pu and ^{241}Am are given in Figs. 14, 15, and 16. They were calculated using the model in Fig. 5 with the parameters recommended by ICRP for adult occupational exposure (ICRP, 1979). The DILs for other long-lived, alpha-emitting uranium isotopes (or mixture of isotopes) are essentially equal to the values given for natural uranium in Fig. 15. An exception is ^{232}U , which has short-lived daughters that add significantly to the dose. Similarly, DILs for other long-lived, alpha-emitting plutonium isotopes are essentially equal to those given for ^{239}Pu in Fig. 16. The DILs for long-lived, alpha-emitting americium isotopes are also essentially equal to the values in Fig. 16a. The reason for this is that the only difference given by ICRP (1979) between Class W plutonium and Class W americium (there are no Class Y americium compounds) is the f_1 factor (see Fig. 4), which is 10^{-4} and

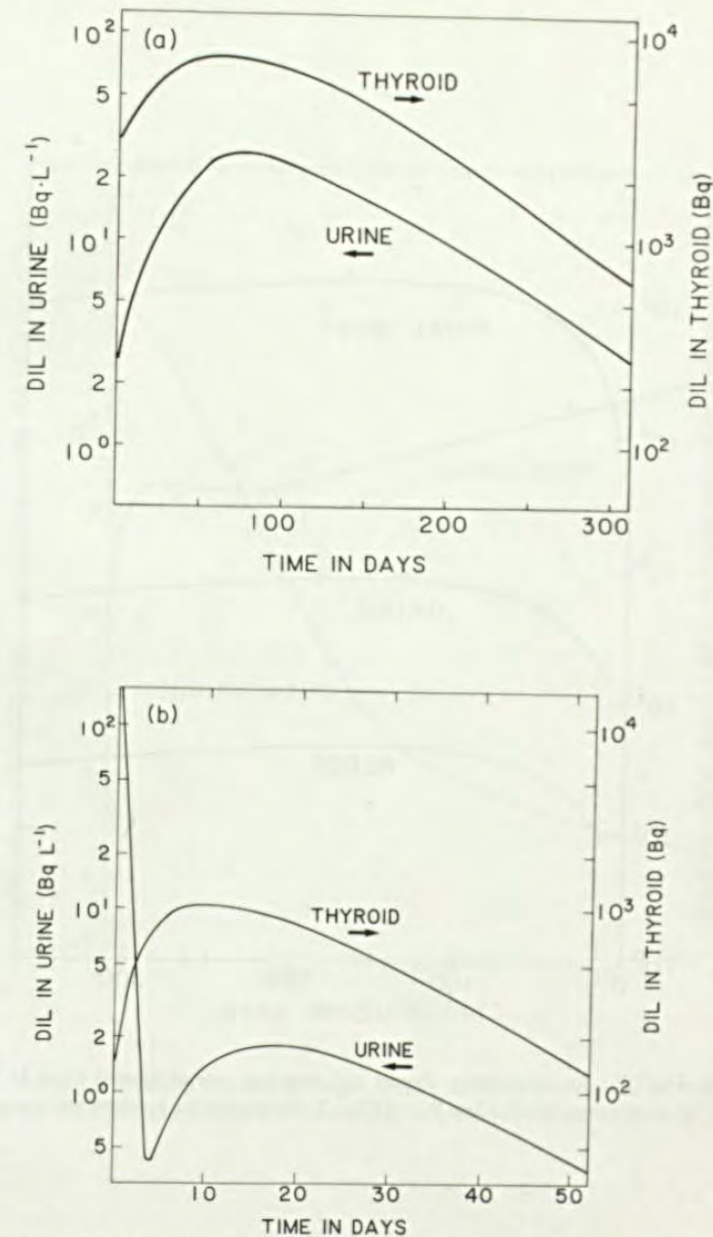


Fig. 13. The DILs for thyroid content and urinary excretion of (a) ^{125}I and (b) ^{131}I as a function of the monitoring period.

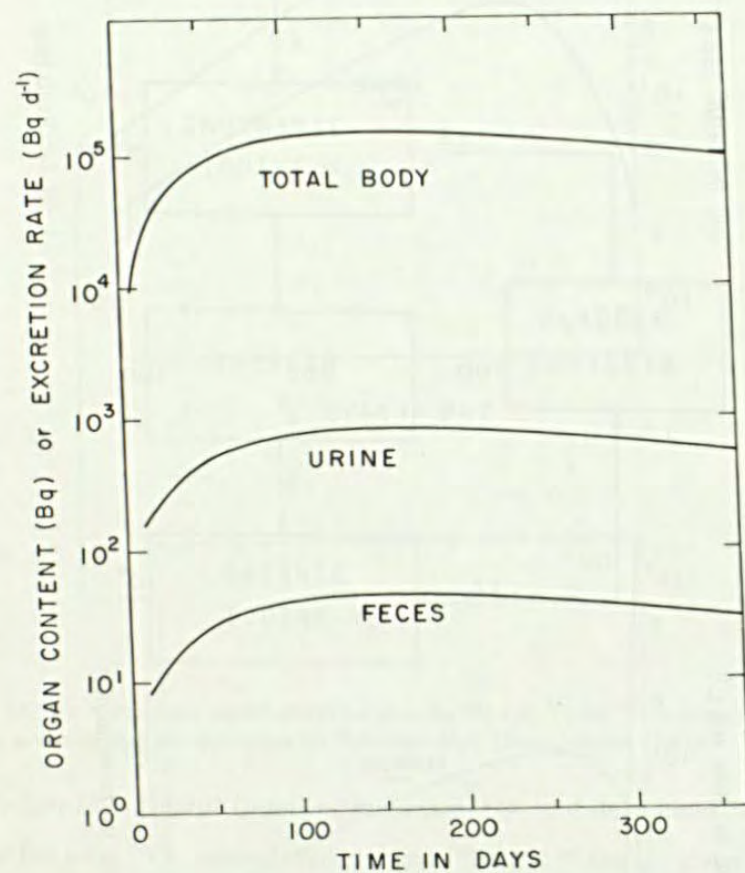


Fig. 14. The DILs for whole-body content and excretion rate of inhaled Class D ^{137}Cs as a function of the monitoring period. Class W and Class Y compounds are assumed not to occur (ICRP, 1979).

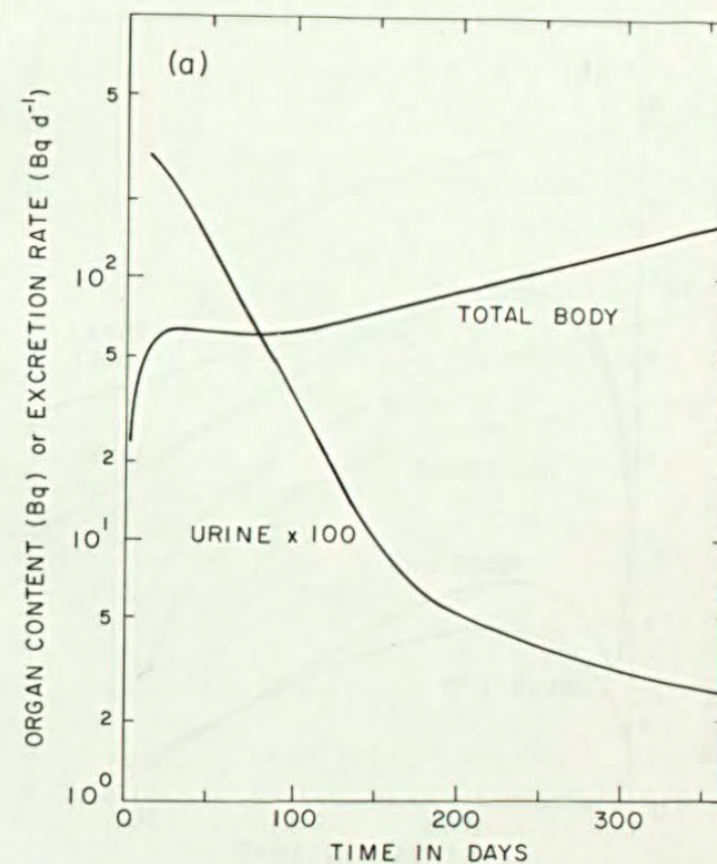


Fig. 15. The DILs of natural uranium for organ content and excretion rate for the inhalation of (a) Class D, (b) Class W, and (c) Class Y compounds as a function of the monitoring period.

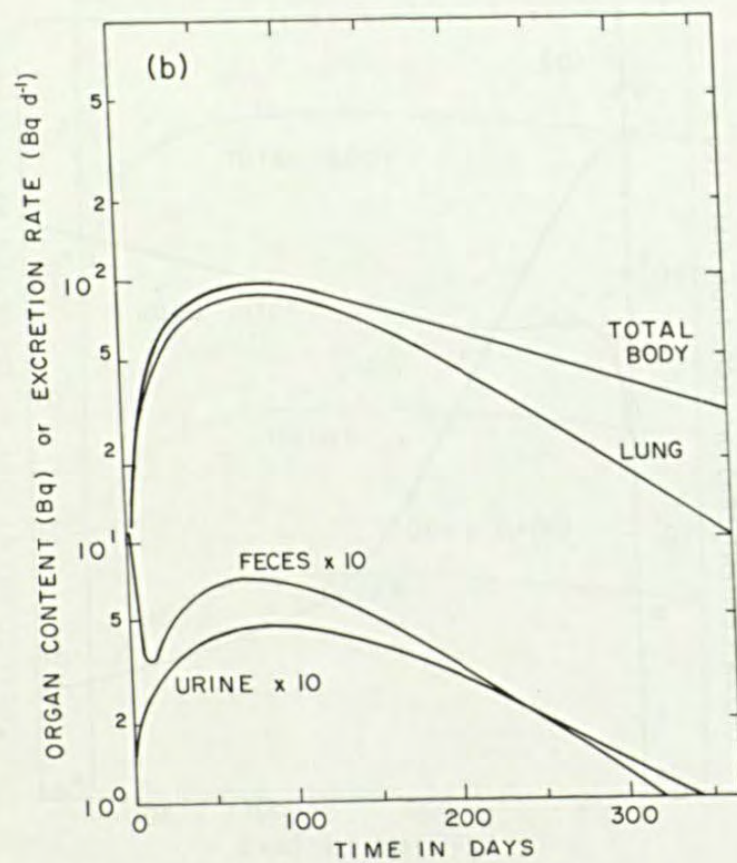


Fig. 15. (continued)

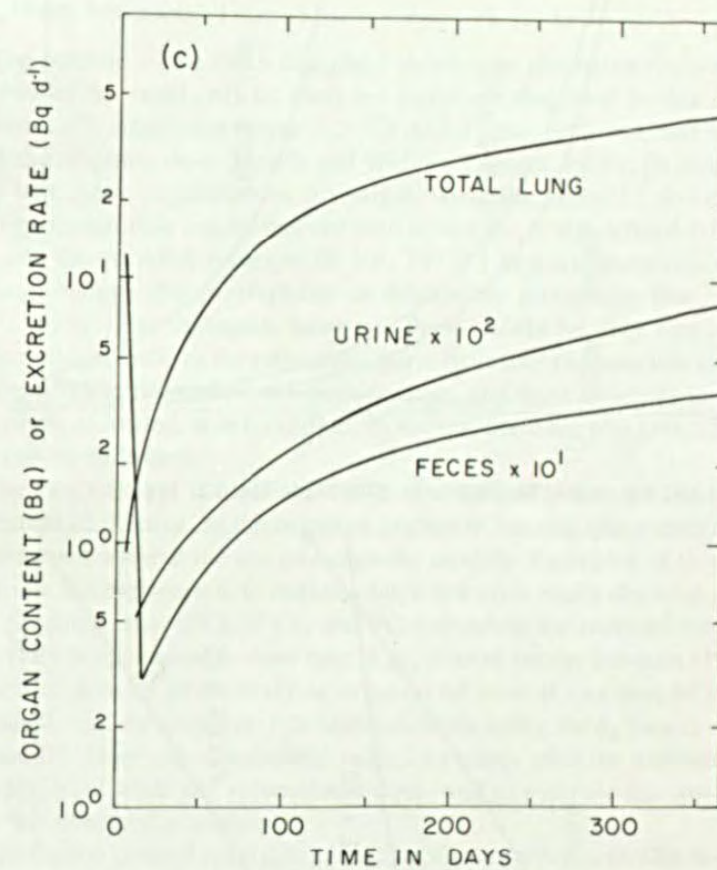


Fig. 15. (continued)

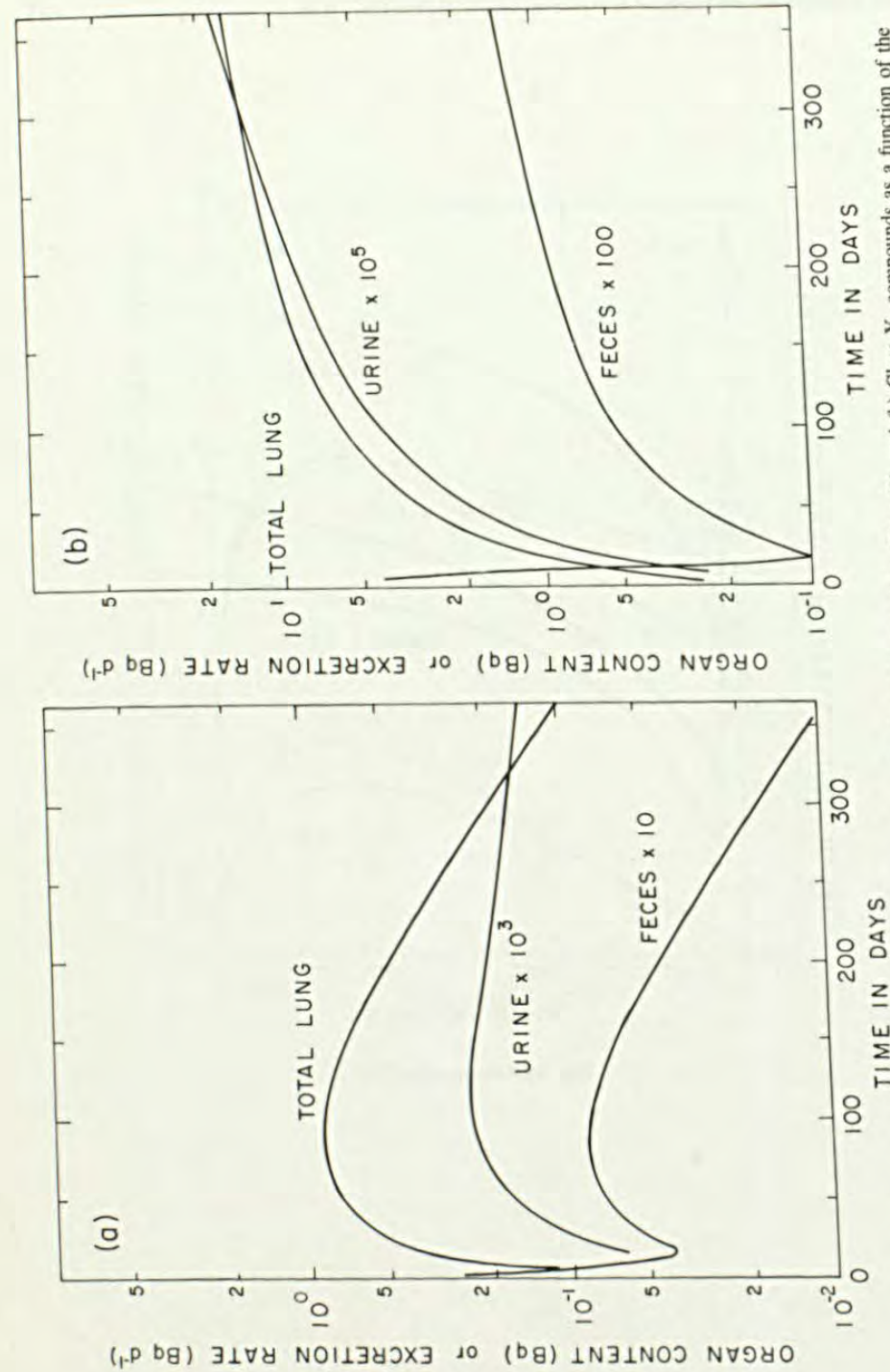


Fig. 16. The DILs of ^{239}Pu for lung content and excretion rate for the inhalation of (a) Class W, and (b) Class Y, compounds as a function of the monitoring period. Class D compounds are assumed not to occur (ICRP, 1979).

5×10^{-4} for plutonium and americium, respectively. Hence very little activity enters the blood from the GI tract following inhalation exposures, and the difference in f_1 factors has no influence on the ALIs or excretion rates.

C. DOSE ASSESSMENT

The purpose of the DILs described above is to distinguish between monitoring results that need only be recorded and those that need further investigation to assess if a significant intake ($>0.3 \text{ ALI}/f$) has occurred, and if so, to estimate the resulting dose (Health and Welfare Canada, 1980). [It should be noted here that many organizations investigate and take remedial actions following positive monitoring results that are well below the ICRP-defined DILs, in keeping with the ALARA principle (ICRP, 1977).] In some instances, intakes may be large enough that prophylactic or therapeutic procedures [see, e.g., NCRP (1977, 1980)] are indicated; however, these should be very rare events.

Specific procedures for estimating doses from any radionuclide exposure cannot be developed because each radionuclide, and most incidents that result in a DIL being exceeded, will be unique. However, there are two general approaches that can be followed.

The first of these is most applicable to situations where the activity, or concentration of activity, in the organ or organs of interest (the source organs) can be estimated without the use of metabolic models. Examples of these situations are when the exposure is to radionuclides that emit easily detected gamma rays (e.g., radioiodines, ^{134}Cs , ^{137}Cs , and ^{60}Co) or when the concentration in excreta is directly proportional to dose rate [e.g., tritium oxide; Johnson (1982c)]. The amount of activity in the body or organ(s) of interest can then be measured or estimated, and the dose rate calculated directly using the S_{ij} factors described in Section III. Dose rates, estimated at various times after the exposure, can then be used to calculate the accumulated dose, and to estimate the committed dose from the observed retention.

The second general approach is most applicable when activities in organs cannot be measured directly, but must be inferred from measurements of excreta. Examples of radionuclides for which this is the case are ^{14}C , ^{35}S , ^{32}P , ^{89}Sr , and ^{239}Pu . In these cases, the analyses of dose must rely on those metabolic models correlating excretion rates and organ burdens that are thought to best represent the radionuclide and the chemical form involved. Because of the uncertainties in these models, individual variability, and because the relationship between dose rate and excretion rate varies with time after the exposure, a good estimate of the dose rate based on a single measurement is virtually impossible. This last point is well illustrated by ^{131}I . Urinary excretion falls very rapidly over the first few days, and the thyroid burden increases during this period. Hence, the ratio of thyroid dose rate to ^{131}I excretion rate varies by orders of magnitude

over the first 10 d following exposure, and a series of measurements of urinary excretion rate would be required to estimate the thyroid dose. It is obviously better to estimate the thyroid dose rate from direct measurements made over the thyroid in this situation, using ^{131}I in urine only as a screening procedure to identify those individuals with possibly significant thyroid burdens. However, direct monitoring of organ burdens is not always possible. For example, the sensitivity for measuring ^{239}Pu in lungs is poor, and inhaled ^{239}Pu that is transferred to other organs (mainly liver and skeleton) is virtually undetectable, even though this transferred radioactivity is a major contributor to the effective dose equivalent. However, air sampling, lung monitoring, and urinary and fecal excretion measurements can be used to refine the ICRP lung deposition and clearance model (ICRP, 1979) to better estimate the amount of transferred ^{239}Pu , and hence dose to organs other than the lung [see, e.g., Ramsden (1983)].

VI. Summary

The basic concepts currently recommended by the ICRP for internal dosimetry calculations in occupational radiation protection have been reviewed, as have their recommended models and model parameters. The major sources of uncertainties in the doses to individuals calculated with these models have been noted, as has the fact that these uncertainties do not necessarily invalidate the use of these models and model parameters in radiation protection programs. Recommended methods of calculating derived investigation levels (DILs) have been reviewed and DILs for use in monitoring for some radionuclides commonly encountered in the workplace are given. General guidance is also provided for assessing doses following an intake of a radionuclide that results in a DIL being exceeded.

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Index

A

- Absorbed dose, 13-15, 19, 23, 26-31, 36-66, 78, 101-105
 - biological effect, 123-124
 - calculations
 - equilibrium, 53-59
 - nonequilibrium, 39-44
 - cavity theory, 67-70
 - for charged particles in phantom, 185, 214-216, 255-264
 - electron fluence, 185-186
 - Fano's theorem, 66-67
 - at interfaces, 39-44
 - measurement, 192-202, 208-216, 252-264
 - for neutrons in phantom, 255-264
 - rate, from internally deposited activity, 370
 - scattering, influence of, 32-36
 - spatial mean, 64, 68
 - Spencer-Attix theory, 42, 63-64, 69
 - Spencer-Fano model, 41-42
 - uncertainties in determination, 216-222
 - for x and γ radiations in phantom, 199-202
- Absorbed fraction, 376-380
 - bone, 377-379
 - GI tract, 379-380
- Actinium decay series, 298
- Activation dosimetry of neutrons, 276-278
- Activity median aerodynamic diameter, 382
- ALI, *see* Annual limit on intake
- Alpha particle
 - counting, 310-314, 345
 - source preparation, 310-314, 346-347, 349
 - standard sources of, 349
- Aluminum-activation dosimetry, of neutrons, 276

- Annual limit on intake, 372-375, 388-389
- Associated volume, for linear track, formulas, 141-143

B

- Background correction, in counting radioactive samples, 350
- Beam monitors, 249-250
- Beta-ray
 - counting 332-340
 - source preparation, 310-314, 332-333
 - spectrometry, 337-338, 340
 - standard sources, 340
- Biological dosimeters, 268-275
- Biological effects, models, 120-125, 151-154
- Bragg peak, 234-236
- Buildup corrections, in counting radioactive samples, 350-351

C

- C_A , 174, 199
- Calorimeters, 165, 168, 169, 244-245, 264
- Cavity theory, 39-40, 67-70
 - Bragg-Gray detector, 39, 64, 69
 - Spencer-Attix, 64
- Cerenkov counting, 338
- Charged-particle beam
 - fluence measurement of, 246-247
 - heavy, depth dose, 234-236
- Charged particle equilibrium, 45-46, 50-59, 62-63, 183-188
- Chemical dosimeters, 165, 168, 169, 248

Chromosome aberrations, dosimetry by
radiation induction of, 270-272
Cobalt-60 gamma-ray beam, depth dose, 237
Coincidence counting, 338-340
Compound Poisson process
algorithm, 154-158
in energy loss, 127-129
in microdosimetry, 97-113
moments, 110-113
Continuous slowing down approximation,
40-42
Convolution, of distribution, 105-107
Counting, precision and accuracy of, 353-356
CSDA, *see* Continuous slowing down
approximation

D

Dead-time correction, 349-350
Decay corrections, in counting radioactive
samples, 350
Delta particle, 41, 60-65
equilibrium, 45-46, 60-65
Derived air concentration, 373-374
Derived investigation level, 389-406
americium, 241, 398, 404
cesium, 137, 398, 400
cobalt, 60, 391, 393-395
iodine, 125, 131, 396, 398-399
plutonium, 239, 398, 404
strontium, 85, 90, 396-397
tritium oxide, 390-392
uranium, 398, 401-403
Detector
Bragg-Gray, 39, 64, 69
efficiency, 16-19
for electrons and photons, 163-181
Fano's theorem, 66, 69
measurement, of absorbed dose, 67-70
response, 15-19
Displacement correction factor, 194-196,
260-264
Dose, concept, 2
Dose assessment, 388-406
Dose equivalent, 370-375
commitment, H_c , 358-359
committed, 372-375, 387-388, 391
effective, 371
rate,

during intake, 374
to target organ, 371, 376
Dosimeters, characteristics of, 168
Dosimetry
electron beam, 202-216
of internal emitters 370-406
in-vivo, 275-278
nonstochastic, 16
neutrons, *see* Dosimetry, nuclear particles
nuclear particles, 238-250, 276-278
photon beam, 181-202
stochastic, 15
Dual radiation action model, 123-125

E

Electron
absorbed dose, 32-36
CSDA range of, 210
scattering of, in phantoms, 209-210
stopping power, 211-213
Electron beam,
absorbed dose
calibration of, 208-214
measurement of, 214-216
depth-dose curve, in low-Z materials,
205-206, 215
depth-ionization curve, 215
dosimetry of, 202-216
energy spectrum of, 203-205
modal energy of, in phantoms, 206-207
Electronic equilibrium, 183-188
lateral, 187-188
transient, 183-184
Energy cutoff, Δ , 40, 42-44, 63-64, 69
Energy deposition, microscopic pattern, 79
Energy fluence, 19, 23, 30-31
absorbed dose, 52
exposure, 59
kerma, 52-53
radiation equilibrium, 45-47
vectorial, 19, 23, 26-27, 38, 45
Energy imparted, 4-15, 23, 78, 90-93, *see also*
Specific energy
biological effectiveness, 87
energy loss, 130-132
equations, 11-12
by primary particle, 48, 51, 61
spatial variation, 13, 87

INDEX

INDEX

Energy loss, by charged particles, 126-132
Energy-loss straggling, 86, 100, 125-137
distribution, 100
effect on lineal energy, 118
Energy transfer, *see* Imparted energy
Environment, naturally occurring radionuclides
in, 293, 294-301
Environmental monitoring programs, 309
Environmental radiation, average annual dose
from, 361, 363
Environmental radioactivity, data reporting on,
356-357
Environmental sampling, 307-309
Errors in dosimetry, 216-222
Event frequency, 95-97
associated volume, 138-143
biological effects, 119
Event size, *see* Lineal energy
Exposure, 2, 59, 176
Extracameral effect, 173

F

Fano's theorem, 65-67
Faraday cup, 246-247
Ferrous sulfate dosimetry, 165, 168, 169
Fluence, 20-26
absorbed dose, 36-40, 42, 62, 69
measurement, 25-26, 246-247
planar, 24
plane, 24, 26
track length, 49
variation, with scattering, 31-36
vectorial, 22-24, 33, 38
Fly-ash, radioactivity in, 300
Fossil fuels, radionuclides in, 300
Fricke dosimeter, 165, 168, 169

G

Gamma-ray
coincidence counting, 327-329
reference sources, 311, 329
whole-body counting of, 320-323
GI model, *see* Radionuclide transport, model,
in humans
Geiger-Müller counters, 244, 336-337, 346

Germanium detectors, for x-ray spectrometry,
314-316, 318-320, 323, 327-332

H

Heavy particles, *see* Nuclear particles
High-LET radiotherapy beams, comparison of,
278-280

I

Imparted energy, 5, 44, 89, 153
absorbed dose, 36-39
equations, 6-10
inchoate distribution, 88-89
mean, 14, 17, 19, 23, 26-31
probability of interaction, 153-154
specific, 17
statistical fluctuations, 15
Inchoate distribution, of energy transfers, 88-89
geometric aspects, 137-154
Inscattering, 36
Interaction coefficient
absorbed dose, 36
exposure, 59
kerma, 52-53
Interaction probability, 38
Interfaces, dose near, 184-187
In-vivo dosimetry, *see* Dosimetry, *in-vivo*
Ionic recombination, 171-172, 239-241
Ionization chambers, 166, 168, 239-243
atmospheric corrections for, 170-171
calibration of, 173-182
design of, 240
displacement correction factor for, 194-196,
260-264
effective center of, 196-199, 209
extracameral effect in, 173
gases for, 242-243
for heavy-particle dosimetry, 239-243, 264
measurement uncertainty, 180-182
perturbation correction factors for, 193-199,
211, 213-214
polarity effects in, 172-173
radial displacement correction for, 196-199
recombination of ions in, 171-172, 239-241
tissue-equivalent, 239-243, 255-264
use of, 169-182, 192-202, 255-264
Ionizing particle, 88

K

- Kerala, high natural radioactivity in, 300-301
 Kerma, 50, 52-59, 176, 183
 Kerma factor, 53, 259-261
 Kinetic energy, of ionizing particles, 6-10, *see also* Radiant energy

L

- Lineal energy, 93, 117-118
 associated volume, 141-143
 average
 frequency, 96, 113
 dose, 96-97, 113, 147-148
 distribution, single event, 94, 113
 Linear energy transfer,
 distribution, 84-86
 quality factor, 117-118
 Lung model, *see* Radionuclide transport,
 model, in humans
 Lyoluminescence, 247

M

- Mass collision stopping power
 absorbed dose, 39-40, 42, 62-63, 69
 restricted, 42, 69
 Mass energy absorption coefficient, 53-54, 59
 Mass spectrometry, for isotope measurement,
 352-353
 Mean free path, of uncharged particles, 82-84
 Microdosimetry, 16, 78-158, 250-254
 biological effect, 79-81, 120-125
 measurements, 87, 113-117
 proximity function, 88, 146-151
 purpose, 79
 in radiation protection, 117-120
 random variables, 132-137
 spatial distribution of ionization, 80-81
 Milk, radioactivity in, 304-305
 Monazite, high natural background in areas rich
 in, 300-301
 Monte Carlo
 calculation for absorbed dose, 44
 simulation of particle track, 81
 Multihit model, 120-122
 Multitarget model, 120-122

N

- Nanodosimetry, 125
 Natural background, *see* Radionuclides,
 naturally occurring
 Neutron activation, for isotope measurement,
 351-352
 Neutron beam, depth dose, 234, 237-238
 Nuclear particles,
 beam characteristics of, 232-238
 beam monitors for, 249-250
 beam quality of, 250-252
 ion chamber dosimetry of, 255-264
 properties of, 231
 protocols for the dosimetry of, 232-233,
 255-264
 therapy treatment planning for, 264-268
 Nuclear track detectors, 249
 Nuclear weapons, release of radionuclides
 from, 302

O

- Outscattering, 36

P

- Particle radiance, 19
 Particle track, 89
 associated volume, 138-143
 computational simulation, 137-143
 proximity function, 146-148
 Perturbation, corrections, 68-70
 for ion chambers, 193-199, 211, 213-214
 Phantoms, nonwater, 201-202, 209-211,
 264-265
 Phoswich x-ray detector, 329
 Photographic film dosimeters, 167, 168, 249
 Photon, absorbed dose, 33-36, 185, 192-202
 Photon beams, calibration of, in phantoms,
 192-199
 Pi-meson therapy, *in-vivo* dosimetry for, 278
 Primary particle, 47-48, 50, 61
 energy imparted, 48, 51, 61
 Proportional counter, 243, 336-337, 342, 345
 Proximity function, 88, 143-151

Q

- Quality factor, 117-118

INDEX

INDEX

- Quality index, 191
 Quality of radiation, *see* Radiation quality

R

- Radial displacement, correction of ion
 chambers, 196-199
 Radiation background, natural, 292
 Radiation chemical yield, 17-19
 Radiation equilibrium, 45-64
 complete, 47-50
 in detector, 68
 Fano's theorem, 66
 transient, 54-59
 Radiation quality, 79, 188-192, 199, 250-252,
 272-273
 Radiant energy, 8, 12, 26-31, 38, 44, *see also*
 Kinetic energy, of ionizing particles
 Radioactivity
 environmental, measurement of, 305-306,
 310-351
 natural, *see* Radionuclides, naturally
 occurring
 Radiochemical separation, of Pu, Am and Th,
 342-343
 Radiochemical yield, determinants, 344
 Radiographic dosimeters, 167, 168
 Radiometric quantities, 19-36, 79
 absorbed dose, 36-39
 exposure, 59
 fluence, 81-82
 kerma, 52-53
 scalar, 19-22
 vectorial, 22-23
 Radionuclide transport
 chains to humans, 357-360
 model, in humans, 380-388
 GI, 384-385
 lung, 381-384
 metabolic, 385-388
 Radionuclides
 in air, 298-299, 306
 artificial, in environment, 293-294, 301-305
 circulation of, by air and water, 302-305
 created by cosmic rays, 299, 303
 in earth's crust, 295-296, 305
 environmental, source preparation of,
 318-327, 332-333
 γ -ray sources, 311
 natural, redistribution of by humans, 299-300

- naturally occurring, 293, 294-301
 primordial, 295-299, 303
 release of
 in medical applications, 302
 from nuclear power stations, 301-302, 304,
 306
 by nuclear weapons, 302, 303-304
 in water, 296-298, 303, 306-307
 Radon (^{222}Rn and ^{220}Rn) in air, 298-299, 362
 Range, of charged particles, 82-84, 86
 Reciprocity, in energy imparted, 49
 Recombination of ions, 171-172
 Relative biological effectiveness, 269, 273-275
 Retention function, 386-387
 Rossi counter, 87, 113-114, 116-117

S

- Scintillation counting, liquid, 332-336, 345
 Scintillators, 247, 314-316, 318-319, 321-337,
 341, 345-346
 Secondary emission monitors, 250
 Secondary particle, 47-48, 50, 61
 SEM, *see* Secondary emission monitors
 Semiconductor detectors, 167, 168, 169, 248,
 276-278, 314-316, 318-320, 323, 327-332,
 337-338, 348
 Silicon detectors, surface barrier, 348
 Silicon diodes, for neutron dosimetry, 276-277
 Single-event
 distribution, 93-97, 100, 133-134
 straggling, 125-137
 variance, 133-137, 151
 weighted average, 143-146, 151
 Sodium iodide crystals, γ -ray response of
 324-326
 Specific effective energy, 376-380
 Specific energy, 90-108
 absorbed dose, 92, 101-106, 108, 122-123
 average
 frequency, 94-95, 100, 113, 138-141
 dose, 96-97, 113
 biological effect, 123-124
 distribution, 90-97, 107-110
 dose dependent, 100-102, 106-107
 simulation, 157-158
 single event, 93-97, 100, 107, 108, 113
 measurement, 114-117
 variance, 108-110, 113-117
 Spectral distribution, 19-22

Spectrometry

- alpha particles, 312-314, 346-347
- beta particles, 312-314
- gamma rays, 310-332
- Stopping power, 39-40, 42, 62-63, 69, 211-213
- Student's *t* factor, 355

T

- Thermoluminescent dosimeters, 167, 247, 277, 278
- Thorium, 232, decay series from, 297
- Threshold model, in microdosimetry, 121-123
- Track length
 - charged particle, distribution, 40-44
 - density, 21
 - fluence, 49
- Transfer point, 88-89
- Transition-zone dosimetry, 184-187
- Transport theory, 37
 - in absorbed dose, 52
 - of radiant energy, 38

INDEX

U

- Uranium, 235, decay series from, 298
- Uranium, 238, decay series from, 296

V

- Vector formalism, examples, 30-36

W

- W*, 17-18
 - for air, 176
 - for TE gas, 258
- Water,
 - as a dosimetry reference medium, 170
 - radioactivity in, 296-298, 300-301, 303

X

- X-ray energy specification, 189-192