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ABSORBED DOSE DETERMINATION IN PHOTON AND ELECTRON BEAMS

An International Code of Practice

PREPARED BY P. ANDREO, J.R. CUNNINGHAM,
K. HOHLFELD AND H. SVENSSON
ON THE BASIS OF A DRAFT
DEVELOPED AT AN ADVISORY GROUP MEETING
ORGANIZED BY THE
INTERNATIONAL ATOMIC ENERGY AGENCY
AND HELD IN VIENNA, 18-22 MARCH 1985



INTERNATIONAL ATOMIC ENERGY AGENCY
VIENNA, 1987

FOREWORD

EDITORIAL NOTE

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Recognizing the importance of accurate dosage in radiation therapy, the International Atomic Energy Agency initiated a programme in dosimetry of ionizing radiations soon after its foundation. At that time neither national nor international absorbed dose standards existed, so a transportable absorbed dose calorimeter was designed and built in the Agency's Laboratory as early as 1960. Being one of the first instruments of its type, this calorimeter was used for making absolute dose measurements and dose intercomparisons with other dose reference instruments. A few years later, the proposal was made to organize a dose intercomparison service by mail for radiation therapy departments. This service was initiated in 1966 for ^{60}Co radiation using TLD. In 1968, WHO joined the programme and since then the service has been run on a regular basis, covering some 200 hospitals per year. There is evidence that, through this service, wide awareness of dosimetric accuracy has been created and, consequently, accuracy has markedly increased. Results from this dose intercomparison service, obtained by statistical evaluations, have been repeatedly used by scientists and national commissions.

In order to further improve the reliability and accuracy of dosimetric measurements in radiation therapy, the IAEA and WHO supported the establishment of Secondary Standard Dosimetry Laboratories (SSDLs) in developing Member States and established a Network of SSDLs in 1976. These SSDLs bridge the gap between the primary measurement standards and the user of ionizing radiation by providing the latter with calibrations against the SSDLs' secondary dose standards and by giving technical advice and assistance. However, a properly calibrated dosimeter is just a necessary first condition for determining a dose. Two more steps are involved, namely the measurement of a (possibly different) quantity and the computation, via some equation, of the absorbed dose from the measurements. The last (computational) step is not simple and experience has shown that it gives rise to most uncertainties. It is not surprising, therefore, that many Codes or Protocols exist, national and international, advising users how to obtain the absorbed dose from a measurement of exposure or another appropriate quantity.

Each of these protocols has its specific merits and peculiarities; no one of them is perfect, and the present one cannot be perfect either. But it is believed that it does represent a step towards a universal Code that can be recommended for use at all radiation therapy centres and SSDLs throughout the world. In fact, it was this idea which motivated the IAEA Secretary of the SSDL Network to call together the co-authors of some of the well received national Codes, and to ask them to write one addressed to radiation physicists in Member States of the IAEA and WHO, where a variety of dosimeters are being used. Such an undertaking seemed worthwhile at a time when new data had become available for pertinent physical parameters, and clearer procedural concepts had been obtained through critical discussions and

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testing of the existing Codes. The Agency gratefully acknowledges the authors' enthusiasm in co-operating and completing the manuscript in short time.

The numerical result of a physical measurement must be complemented by an assessment of its uncertainty. Until recently, different rules have been applied by laboratories in stating the uncertainty of a measurement, often reflecting divergent opinions about the concepts used. This unsatisfactory situation caused the Bureau International des Poids et Mesures (BIPM) to ask a group of experts to consider the subject and to propose a procedure which might lead to harmonization of the diverse practices. The Agency wishes to express its gratitude to A. Allisy and J.W. Müller from the BIPM for writing a brief summary of the uncertainty concept recently recommended by the BIPM. Their contribution is added to this Code as Appendix A.

The first outline of the Code was drafted by W.A. Boyd and H.H. Eisenlohr in consultation with J.R. Cunningham, K. Hohlfield and H. Svensson.

This outline was completed by the members of an Advisory Group who met in Vienna in March 1985. The members of this group were: C.E. de Almeida (Brazil), P. Andreo (Spain), J.R. Cunningham (Canada), A. Dutreix (France), K. Hohlfield (Federal Republic of Germany), A. Kannan (India), N. Suntharalingam (USA), H. Svensson (Sweden) and W. Szymczyk (Poland). From this group, the four principal authors, P. Andreo, J.R. Cunningham, K. Hohlfield and H. Svensson, were selected. Their final draft was reviewed by the other members of the Group.

It is hoped that the present publication will be of use to all those involved with accurate dosimetry in radiation therapy. In particular, it is addressed to the member laboratories of the IAEA/WHO Network of SSDs. Every user is invited to test the code critically and to submit comments to:

Head, Dosimetry Section, RILS,
International Atomic Energy Agency,
P.O. Box 100, A-1400 Vienna, Austria

IAEA sponsor: H.H. Eisenlohr
IAEA scientific secretary: M. Gustafsson

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1. INTRODUCTION

1.1. SCOPE

Present day practices in radiation therapy require high doses of radiation to be delivered with increased accuracy. With the improved anatomical information obtainable from sophisticated diagnostic imaging procedures, the data required to achieve better accuracy in the determination of the radiation dose delivered to the patient are available. It has been demonstrated that the success or failure of radiation treatment depends on the absorbed dose delivered to the tumour and that this should not vary by more than a few per cent from the prescribed values. Thus, the ICRU [1] concludes that there is need for an accuracy of $\pm 5\%$ in the delivery of absorbed dose in some treatment situations. Recent investigations indicate that an even better accuracy is needed [2]. This requires that the overall uncertainties in radiation dosimetry be minimized and that improved accuracy be sought in the determination of absorbed dose from the radiation beams used in cancer therapy. With the current state of theoretical knowledge of radiation dosimetry concepts and with the improvement in the accuracy of the numerical values of the physical data required it is possible to reduce the uncertainty in the calibration of the radiation beam. This is important since this is an essential component of the total uncertainty in the delivery of treatment.

The calibration of a radiation beam used for the treatment of cancer in patients is based on rather complicated measurements and the application of several conversion and correction factors. It is therefore rather important that all the steps in this procedure be unambiguously specified and be capable of execution worldwide with minimum difficulty. The purpose of this code of practice is to outline in detail the methods of procedure and to provide the current best values for physical interaction coefficients (e.g. W/e , and $S_{w,air}$, see Section 2.2) and correction factors (e.g. P_{d, k_m} and k_{at} , see Section 2.2) that will allow the determination of absorbed dose from a radiation beam to the accuracy that is expected.

During the past few years national organizations have introduced a number of codes of practice, protocols and documents (see, for example, Refs [3-10]) that provide a systematic approach to the calibration and determination of absorbed dose from radiation beams used in the treatment of cancer. In general, however, they are somewhat too specific in that they serve the requirements of the facilities in the countries where the documents originated. The situation worldwide is rather different, and the current protocols have certain limitations in meeting the needs of the broader radiation oncology community. There is an urgent need for a unified and easily executable code of practice to be used worldwide for the measurement and determination of absorbed dose from radiation beams. This new international code has drawn on useful information available within the current national protocols and is

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intended to provide the user with a document that is clear and understandable and is easy to follow in rather different environments.

This new international code of practice will necessarily affect all hospitals providing radiation treatment for cancer patients. Even though the nature of these facilities may be widely different, this protocol will serve as a useful document to the medical physics and radiotherapy community and help achieve an acceptable degree of uniformity of radiation dose delivery the world over. This protocol should also be of great value to the network of Secondary Standard Dosimetry Laboratories (SSDLs) to facilitate the accuracy and consistency of their absorbed dose determinations and thereby the standardization of radiation dosimetry in the countries they serve.

1.2. HISTORICAL NOTE ON SSDLS AND THE SSDL NETWORK

There is a general agreement that physical measurements must be traceable to primary standards. Such standards are developed and maintained at national standardizing laboratories and the International Office of Weights and Measures (Bureau International des Poids et Mesures, BIPM). In the case of the measurement of ionizing radiation, primary standards for the quantities exposure and absorbed dose exist only in some twenty laboratories. The lack of calibration facilities in some countries led the IAEA and WHO to consider supporting the establishment of 'Secondary Standard Dosimetry Laboratories (SSDLs)' [11-13].

In a working arrangement between the IAEA and WHO it was further agreed to set up an 'IAEA/WHO Network of Secondary Standard Dosimetry Laboratories' as a means of co-ordinating the work of the individual SSDLS and to facilitate their link to the primary measurement system. Criteria for the establishment of a Secondary Standard Dosimetry Laboratory [14] were set forth by the network secretariat, and dose intercomparisons for SSDLS are organized by the Dosimetry Laboratory of the IAEA. The main objectives of the SSDL Network have been defined as follows:

- To improve dosimetric accuracy and reliability, particularly in radiation therapy, by supporting centres and laboratories for the creation and distribution of knowledge in applied dosimetry;
- to further the exchange of experience between members and affiliated members and to provide support to each other where necessary;
- to facilitate links between members and the international system of radiation measurements through PSDLS (Primary Standard Dosimetry Laboratories) and establish measurement traceability;
- to promote the compatibility of methods applied for calibration and performance of dosimetry in order to achieve uniformity of measurements throughout the world.

At present (1987) the IAEA/WHO Network of SSDLS comprises about 60 laboratories, 46 of which are located in the developing world. The Network is supported by 12 national primary standardizing laboratories holding affiliated membership, and 5 collaborating international organizations. The Network Secretariat is shared between the IAEA and WHO.

1.3. RANGE OF RADIATION BEAM ENERGIES

The present code of practice provides the methodology necessary for the accurate determination of the absorbed dose to water from radiation beams used for radiation therapy. The range of radiation qualities covered by the present publication is: low energy or soft X-rays generated at tube potentials below 100 kV, medium energy or conventional energy X-rays generated at tube potentials above 100 kV, high energy photon beams with mean energies ≥ 0.66 MeV and high energy electron beams (in the energy range $5 \text{ MeV} \leq E_0 \leq 50 \text{ MeV}$, but some data and references are also given which can be used for energies below 5 MeV).

1.4. PRACTICAL USE OF THE PROTOCOL

Much emphasis has been given to making the dosimetry procedure as simple as possible. It has, however, been necessary to include a large amount of information in order to supply a variety of users with the data they may need for absorbed dose determinations. The idea is that after the first study and use of this protocol it should be possible for the user to merely follow the procedures in the flow charts and to use the worksheets (see Appendix B).

2. RADIATION QUANTITIES AND UNITS

2.1. QUANTITIES AND UNITS

The ICRU [15] has provided a clear and unambiguous set of definitions for the quantities and units to be used for radiation dose measurements, which should serve as a useful reference. Only those quantities directly involved with dose calibrations will be discussed in this document.

It is essential that one distinguishes between a quantity and a unit. A *physical quantity* characterizes a physical phenomenon in a form that lends itself to numerical specification. A *unit* is a selected reference sample of a quantity, and is a numerical scale for the specification.

For this document, the most important quantity is absorbed dose, which is a statement of the amount of energy absorbed per unit mass of an irradiated material. Its formal definition is the quotient

$$D = d\bar{\epsilon}/dm$$

where $d\bar{\epsilon}$ is the mean energy imparted by ionizing radiation to matter of mass dm . The mass dm can be taken to be very small but not so small that the mean energy imparted $d\bar{\epsilon}$ is subject to statistical fluctuations. Absorbed dose is therefore a point function and is continuous and differentiable and one may refer to its gradient and its rate. Absorbed dose may be specified in any medium for any type of ionizing radiation.

The unit by which absorbed dose is specified is $\text{J} \cdot \text{kg}^{-1}$ and the special name for this is gray (Gy).

$$1 \text{ Gy} = 1 \text{ J} \cdot \text{kg}^{-1}$$

The absorbed dose rate is defined by

$$\dot{D} = dD/dt$$

The unit for absorbed dose rate is $\text{J} \cdot \text{kg}^{-1} \cdot \text{s}^{-1}$

$$1 \text{ Gy} \cdot \text{s}^{-1} = 1 \text{ J} \cdot \text{kg}^{-1} \cdot \text{s}^{-1}$$

When considering indirectly ionizing particles such as photons it is convenient also to describe the energy released by them as they interact with matter. This quantity *kerma* (kinetic energy released in material, sometimes per unit mass) is defined by the quotient

$$K = dE_{tr}/dm$$

where dE_{tr} is the sum of the kinetic energies of all the charged particles liberated by the interactions of uncharged ionizing particles in a mass dm .

The unit of the quantity *kerma* is $\text{J} \cdot \text{kg}^{-1}$ with the special name gray (Gy). The quantity *kerma*, which may be applied to any material, is closely related to the quantity *exposure*, which applies only to photons interacting with air and is the quotient

$$X = dQ/dm$$

where dQ is the absolute value of the total charge of the ions of one sign produced in air when all the electrons liberated by photons in air of mass dm are completely

stopped in air. The unit is $\text{C} \cdot \text{kg}^{-1}$. However, the special unit of exposure, röntgen (R) is still in fairly extensive use [15]:

$$1 \text{ R} = 2.58 \times 10^{-4} \text{ C} \cdot \text{kg}^{-1}$$

Exposure, as defined above, is the ionization equivalent of air kerma, except that the ionization arising from the absorption of bremsstrahlung emitted by the electrons is not to be included in dQ , i.e.

$$X = K_{\text{air}} (1 - g)/(W/e)$$

where K_{air} is air kerma, g is the fraction of the energy released that is dissipated as bremsstrahlung and the quotient W/e is the mean energy expended per ion pair produced in air and per electron charge.

The use of exposure will in due course be discontinued. It is used in this protocol for two reasons: a number of standardization laboratories continue to calibrate dosimeters with respect to exposure and it plays an important role in many existing reference documents.

2.2. LIST OF SYMBOLS

a_0, a_1, a_2	Coefficients used to derive P_a
B	Backscatter factor
$\bar{D}_{\text{air}}$	Mean absorbed dose to the air in the cavity of an ionization chamber
D_m	Absorbed dose to material m at a point of measurement in a phantom
D_{20}/D_{10}	Ratio of absorbed dose at 20 cm depth to that at 10 cm using a constant source-surface distance
$\bar{E}_a, \bar{E}_0, \bar{E}_z$	Mean electron energy in an electron beam at the accelerator exit window (the vacuum side), phantom surface and at depth z
$E_{\text{max},a}, E_{\text{max},0}, E_{\text{max},z}$	Maximum energy in an electron beam at the accelerator exit window, phantom surface and at depth z

$E_{p,a}, E_{p,0}, E_{p,z}$	Most probable energy in an electron beam the acceleration exit window, phantom surface and at depth z	N_D	Absorbed dose to air chamber factor (this is N_{gas} of AAPM (American Association of Physicists in Medicine) protocol)
FSD	Focus-surface distance	$N_{D,w}$	Absorbed dose to water calibration factor
g	Fraction of energy of secondary charged particles that is lost to bremsstrahlung	N_K	Air kerma calibration factor of an ionization chamber
h_ν	Energy of photon radiation	N_X	Exposure calibration factor of an ionization chamber
h_m	Correction factor to be applied when measurements are carried out at the ionization maximum in material m instead of in water	P_{cel}	Factor to take account of non-air equivalence of the material in the central electrode of an ionization chamber
HVL	Half value layer	P_u	Factor to allow for non-water equivalence of the ionization chamber (i.e. chamber wall material and air cavity), when in the user's beam. An effective point of measurement is to be used when applying this correction factor.
$\bar{J}_{air}$	Mean specific ionization inside the ionization chamber air cavity	P_s	Factor to allow for lack of saturation of charge collection (due to ion recombination) in an ionization chamber
J_{20}/J_{10}	Ratio of ionization at 20 cm depth to that at 10 cm using a constant source-surface distance	P_{TP}	Factor to allow for effects of non-reference temperature and pressure
k_{at}	Factor to allow for attenuation (absorption and scattering) in the walls of an ionization chamber irradiated for calibration purposes	P	Atmospheric pressure during measurement (P_0 is the reference pressure)
k_h	Correction factor to account for the humidity in the chamber cavity air	P_{eff}	The effective point of measurement of the ionization chamber (see Fig. 11 in Section 6)
k_m	Factor to take account of non-air equivalence (at the calibration quality) of the ionization chamber wall and buildup cap material(s)	Q_1/Q_2	Ratio of the charges collected by an ionization chamber using two different collecting voltages (V_1 and V_2)
k_ω	Factor to account for variations in spectral distribution of X-rays used for the ionization chamber calibration free in air and that used by the user in water	r	Inner radius of ionization chamber
K_{air}	Kerma to air	r_0	Continuous slowing down range
K_m	Kerma to material m at a point in an irradiated phantom	R_p	Practical range for electrons
M	Reading of an ionization chamber system	R_t	Therapeutic range, usually corresponding to the 85% level, R_{85}

R_{50}	Half-value depth for electrons	α	Fraction of ionization inside an ion chamber due to electrons arising in the chamber wall
R_{100}	Depth of dose maximum for electrons	ρ	Density
$S_{m,air}$	Ratios of averaged stopping powers for material m to air. The averaging is by the Spencer-Attix theory over the total electron energy spectrum at the point of measurement. (The symbol $S_{m,air}^A$ is used by ICRU [16]. The superscript is here omitted as all stopping-powers below are of the Spencer-Attix type.)	μ_{en}/ρ	Mass energy absorption coefficient
		μ_{tr}/ρ	Mass energy transfer coefficient
SCD	Source to chamber distance	$(\mu_{en}/\rho)_{m_1,m_2}$	Ratios of averaged mass energy absorption coefficients for material m_1 to material m_2 . The averaging is over the photon spectrum at the point of measurement
SSD	Source to surface distance		
T	Temperature of the chamber cavity air during measurement (T_0 is the reference pressure)		In order to keep the overall consistency along the dosimetric chain, all the numerical values for constants used in the present report follow the recommendations of CCEMRI(I) [17]:
TPR_{10}^{20}	Ratio of absorbed dose at 20 cm depth to that at 10 cm using a constant source-detector distance (TPR is short for 'tissue-phantom ratio', see also Ref. [1]).	(a)	stopping power for electrons from ICRU [18];
		(b)	$W/e = 33.97 \pm 0.06 \text{ J}\cdot\text{C}^{-1}$ [19-21];
		(c)	g values from Boutillon [22];
		(d)	energy absorption coefficients from Hubbell [23].
V	Polarizing or collecting voltage of an ionization chamber		
W/e	The mean energy expended in air per ion pair formed and per electron charge		
X	Exposure		
z	Depth in medium in beam direction		
$\Gamma_a, \Gamma_0, \Gamma_z$	Energy spread (width at half of the maximum of the energy spectrum) in an electron beam at the accelerator exit window, phantom surface and at depth z		
ϕ_E	Differential distribution of fluence in energy		
σ	Standard deviation		
Δ	Cut-off energy used in Spencer-Attix stopping-power ratios		
3. EQUIPMENT			
3.1. DOSE MEASURING SYSTEM			
The ionization dosimeter is the only dose measuring system discussed in this report. It consists generally of an electrometer (often separately calibrated in charge or current per scale division) and one or more ionization chambers each intended for different purposes (e.g. different radiation qualities).			
3.1.1. Ionization chamber			
It should be possible to use the same ionization chamber for 100 kV to 300 kV X-rays (medium energy X-rays), ^{137}Cs and ^{60}Co gamma radiation, high energy X-rays, and electron radiation with $\bar{E}_0 \geq 10 \text{ MeV}$ (see Table I). If the chamber is to be used for medium energy X-rays (HVL from about 2 mm Al to 3 mm Cu) then its response in unit scale division or unit charge (depending on the quantity measured with the electrometer) per air kerma should not vary by more than $\pm 2\%$ over this range of half-value layers. This might be attained if the wall thickness of the chamber			

TABLE I. RECOMMENDATIONS FOR IONIZATION CHAMBER DESIGN

Beam quality	Chamber shape	Chamber dimension	— front window or wall	— collecting electrode	— guard width	Internal cavity diameter	Internal cavity length	Material	Response, limit	(M/K _{air}) of variation	Buildup cap + wall	Polarity effect ^b		
Low energy X-rays	Plane parallel	< 0.1 g·cm ⁻²	preferably	dia. 15–20 mm	> 3 mm	15–20 mm	< 25 mm	±2%	±2%			< 0.2%	< 1%	
Medium energy X-rays	Cylindrical	< 0.1 g·cm ⁻²	preferably	dia. 15–20 mm		15–20 mm	< 25 mm	sec ^d			0.4–0.6 g·cm ⁻² preferably	< 0.2%	< 0.2%	
Cs-137 γ-rays	Cylindrical	< 0.1 g·cm ⁻²	preferably	dia. 15–20 mm		15–20 mm	< 25 mm	sec ^d				< 0.2%	< 0.2%	
High energy X-rays	Cylindrical	< 0.1 g·cm ⁻²	preferably	dia. 15–20 mm		15–20 mm	< 25 mm	sec ^d				< 0.2%	< 0.2%	
E ₀ < 5 MeV	Plane parallel	< 0.1 g·cm ⁻²	preferably	dia. 15–20 mm	> 3 mm	15–20 mm	< 25 mm	sec ^d				< 0.2%	< 0.2%	
5 MeV < E ₀ < 10 MeV	Plane parallel ^a or cylindrical	< 0.1 g·cm ⁻²	preferably	dia. 15–20 mm		15–20 mm	< 25 mm	sec ^d				< 0.2%	< 0.2%	
E ₀ > 10 MeV	Cylindrical	< 0.1 g·cm ⁻²	preferably	dia. 15–20 mm		15–20 mm	< 25 mm	sec ^d				< 0.2%	< 0.2%	

^a The plane parallel chamber should be designed as for E₀ < 5 MeV.
^b Defined as the percentage difference in response in making identical irradiations of the chamber with and without reversed polarity (time for stabilization of the chamber is to be allowed for).
^c It would be desirable to have as thin a front window as possible. However, if graphite is used in the window it is not practical to reduce the thickness below about 1 mm (0.17 g·cm⁻²) as the chamber then would be very fragile.
^d Preferably by one single low atomic number material in wall, windows and central electrode.

is less than $0.1 \text{ g} \cdot \text{cm}^{-2}$ ($\approx 1 \text{ mm}$), and if the chamber wall is made of 'air equivalent' materials. In the calibration of such a chamber, free in air using ^{60}Co gamma radiation, a buildup cap has to be added. The thickness for the wall plus buildup cap should then be between $0.4 \text{ g} \cdot \text{cm}^{-2}$ and $0.6 \text{ g} \cdot \text{cm}^{-2}$.

It is very convenient to use a cylindrical chamber (thimble type) for measurements at these radiation qualities as it is robust and simple to use for measurements in a water phantom. The chamber volume should be between about 0.1 cm^3 and 1 cm^3 . This size range is a compromise between the need for sufficient sensitivity and the ability to measure dose at a point. These requirements are met in cylindrical chambers with an internal diameter of preferably not greater than 7 mm and an internal length of not greater than 25 mm. In use, the chamber must be aligned in such a way that the radiation fluence is uniform over the cross-section of the chamber. This chamber length would therefore set a lower limit to the size of the field in which measurements may be made.

The construction of the chamber should be as homogeneous as possible but it is recognized that for technical purposes the central electrode will most likely have to be of a material different from that of the walls and indeed the choice of all these materials may play an important role in ensuring that the energy response is flat. It is not necessary that the chamber wall and the buildup cap be made of the same material but it is essential that the composition of both be known.

It is also necessary for the air cavity not to be sealed. It should be designed so that it will equilibrate rapidly with the exterior pressure condition. Characteristics of a number of ionization chambers are given in Table II.

For measurements in electron beams of energies below $\bar{E}_0 = 10 \text{ MeV}$, plane parallel chambers are recommended and must be used below $\bar{E}_0 = 5 \text{ MeV}$. Cylindrical chambers having a radius less than 2 mm may be used for $5 \text{ MeV} \leq \bar{E}_0 < 10 \text{ MeV}$ instead of plane parallel chambers but the uncertainty may then increase by 1–2%. Plane parallel chambers can, however, also be used above $\bar{E}_0 = 10 \text{ MeV}$. The front window should be thin (preferably < 1 mm) and the material surrounding the cavity should be as 'water-equivalent' as possible (i.e. mass collision stopping powers and linear scattering powers similar to those of water, see Table 6.1 in Ref. [16]). The distance between the plates of the chamber should be less than about 2 mm and the diameter of the collecting electrode should be less than about 20 mm. The collecting electrode should be surrounded by a guard electrode. It is necessary that the chamber should not exhibit a polarity effect (see Section 7.1) and it is also desirable that the chamber be water immiscible. These recommendations are compiled in Table I.

Ionization chambers for measuring low energy X-rays usually possess an entrance window consisting of either a thin membrane or a mesh through which the radiation enters the measuring volume (thin-window chambers). For these chambers the limit of variation of response will normally be within $\pm 2\%$ in a range of half-value layers from 0.05 mm to 2 mm Al, corresponding to X-rays generated by tube

TABLE II. CHARACTERISTICS OF CERTAIN IONIZATION CHAMBERS USED IN RADIOTHERAPY DOSIMETRY

Internal length (mm)	Internal radius (mm)	Wall material	t_{wall} g·cm ⁻²	Cap material	t_{cap} g·cm ⁻²
6.0	2.0	C-552	0.210	Polystyrene	0.568
12.0	2.0	C-552	0.210	Polystyrene	0.568
22.0	3.2	C-552	0.050	C-552	0.924
22.0	3.2	C-552	0.050	Polystyrene	0.539
22.0	3.2	C-552	0.050	PMMA	0.547
23.8	3.25	Graphite	0.046	PMMA	0.625
9.7	4.7	C-552	0.182	C-552	0.712
9.7	4.7	C-552	0.182	C-552	0.356
9.7	4.7	C-552	0.182	C-552	0.712
9.7	4.7	A-150	0.115	A-150	0.450
7.2	2.0	A-150	0.117	A-150	0.450
12.0	3.5	Delrin	0.071	Delrin	0.430
9.5	2.3	A-150	0.183	A-150	0.386
7.0	2.95	Tufoal	0.074	PMMA	0.543
7.0	3.15	Graphite	0.0655	PMMA	0.545
8.3	3.15	Graphite	0.0655	Delrin	0.5515
24.0	2.95	Tufoal	0.075	PMMA	0.415
24.0	2.95	Tufoal	0.075	PMMA	0.545
24.0	2.95	Tufoal	0.075	PMMA	0.545
24.0	2.95	Nylon 66	0.063	PMMA	0.545

TABLE II (cont.)

Internal length (mm)	Internal radius (mm)	Wall material	t_{wall} g·cm ⁻²	Cap material	t_{cap} g·cm ⁻²
24.0	3.15	Graphite	0.065	PMMA	0.57
24.0	3.15	Nylon 66	0.041	PMMA	0.551
24.0	3.15	Graphite	0.065	Delrin	0.551
24.0	3.15	A-150	0.040	PMMA	0.584
24.0	3.15	A-150	0.0407	Polystyrene	0.584
9.2	3.7	Graphite	0.090	Delrin	0.600
21.9	3.05	PMMA	0.059	PMMA	0.356
21.9	3.05	PMMA	0.053	PMMA	0.551
21.9	2.35	PMMA	0.142	PMMA	0.551
18.0	2.5	PMMA	0.059	PMMA	0.356
22.0	4.0	PMMA	0.059	PMMA	0.356
15.1	2.75	PMMA	0.089	PMMA	0.356
4.3	2.5	Delrin	0.529		
23.0	2.4	Polystyrene	0.117	PMMA	0.481
18.0	2.5	PMMA	0.06	PMMA	0.36
23.0	3.05	PMMA	0.06	PMMA	0.36
22.0	4.0	PMMA	0.06	PMMA	0.36
17.9	4.0	Graphite	0.084	Graphite	0.384
17.9	4.0	A-150	0.056	A-150	0.373

Year of manufacture.

potentials ranging from 12 kV to about 70 kV. It is generally the practice to calibrate such chambers together with a phantom as corrections for backscatter are then avoided.

3.1.2. Electrometers

The charge (or current) induced in an ionization chamber is extremely small and must be measured by a very sensitive charge measuring device having a high input impedance ($> 10^{14} \Omega$) known as an electrometer. Such instruments are delicate and must be handled with care. Ideally, the electrometer should be provided with a digital display and should be capable of four digit resolution or 0.1%. The change in response due to lack of long term stability should not exceed $\pm 0.5\%$ in a year. The electrometer and the ionization chamber can be calibrated separately. However, in some cases the electrometer is an integral part of the dosimetry system and the ionization chamber and electrometer must be calibrated as a unit.

It should be possible (at least if the chamber is used in pulsed beams) to vary the voltage applied to the chamber so as to determine ion collection efficiency and to reverse the polarity so that the polarity effect of the ionization chamber may be determined.

3.2. PHANTOMS

Water is recommended as the reference medium for absorbed dose measurements both for photon and electron beams. Plastic phantoms in slab form such as polystyrene, acrylics and 'solid water' may be used for solid phantoms but the dose determination must always be referenced to water. However, great care is needed with non-conductive plastic phantoms (in particular if these are made as solid blocks) because several investigations have shown [24-26] that charge buildup can take place, mainly in electron beams. This causes a very large electrical field around the chamber that will direct the flow of electrons towards the cavity, producing an increased signal. This effect may cause an error of several per cent; therefore, measurements for absorbed dose determination should, at least for electron beams, always be performed in water or conductive plastics.

The dimensions of the phantom should be such that, with the largest field size to be employed, a margin of 5 cm is provided on all four sides at the depth of measurement. There should also be a margin of 5 cm beyond the maximum depth to be measured, see Fig. 1.

3.3. RADIATION BEAM GEOMETRY

The position of all of the components of the experimental arrangement must be known, capable of description and reproducible. Included are the beam and its

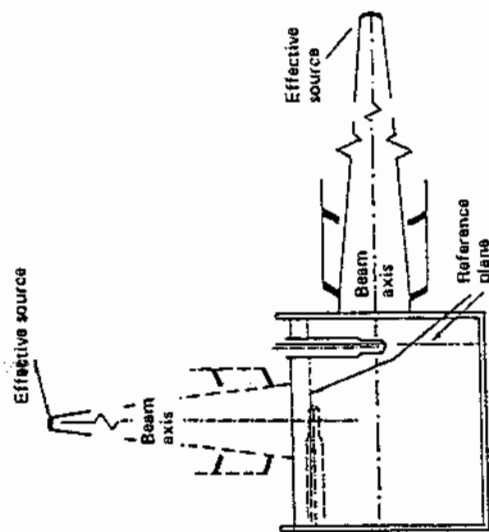


FIG. 1. Reference water phantom for absorbed dose determination. The use of a thimble ionization chamber for absorbed dose determination at the reference depth in the phantom is shown for two positions of the beam. The effective depth of measurement of the chamber (see Fig. 11) should be placed at the reference point in the phantom.

collimating system, the phantom and the location of the dosimeter within the phantom. The relevant parameters are shown in Fig. 2.

3.3.1. Beam alignment and position

The alignment of the different beam axes must be checked before an absorbed dose distribution is determined. There are four axes to be considered. They are:

- The collimator rotation axis, or the axis about which the collimator rotates (Fig. 3). This is the basic mechanical axis to which the others are referred.
- The geometric beam axis. This is the line passing through the centre of the effective source (X-ray focus or effective electron source) [16] and the centre of the final beam limiting diaphragm.
- The radiation beam axis. This is the line passing through the centre of the source and the centre of gravity of the area in the reference plane¹ within which the absorbed dose exceeds 50% of the maximum absorbed dose at this plane. (Generally, the centre of gravity coincides with the point of intersection P_i defined by $AP_i = BP_i$ and $CP_i = DP_i$ for a rectangular field as in Fig. 4.)

¹ The reference plane is perpendicular to the radiation beam axis at the reference depth, see Table VII in Section 7.

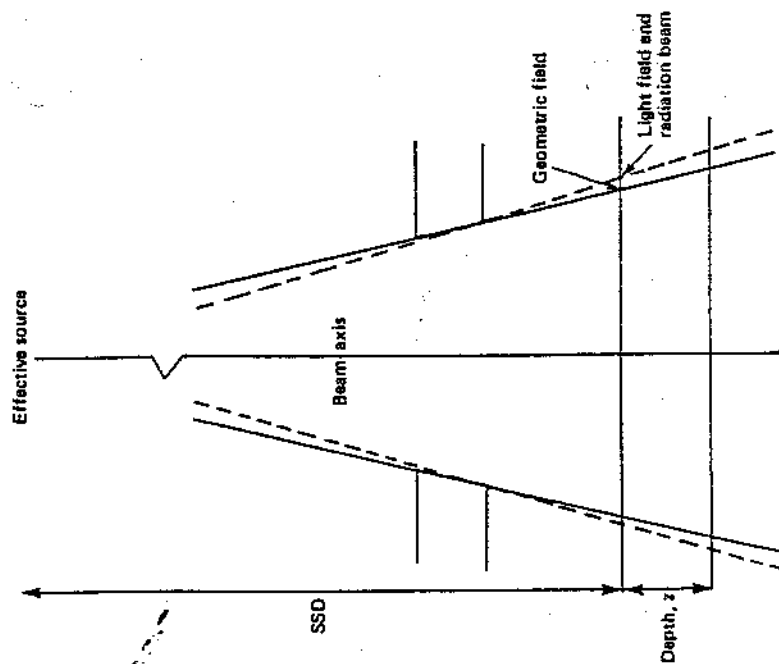


FIG. 2. Beam geometry. The collimator rotation axis, geometric beam axis and radiation beam axis coincide in the figure though in practice there may be small differences. Also, the geometric field and light field may differ. The radiation beam should coincide with the light beam. The tolerances are given in the text.

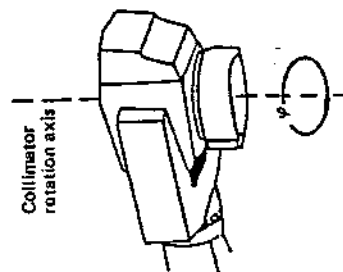


FIG. 3. Definition of the collimator rotation axis.

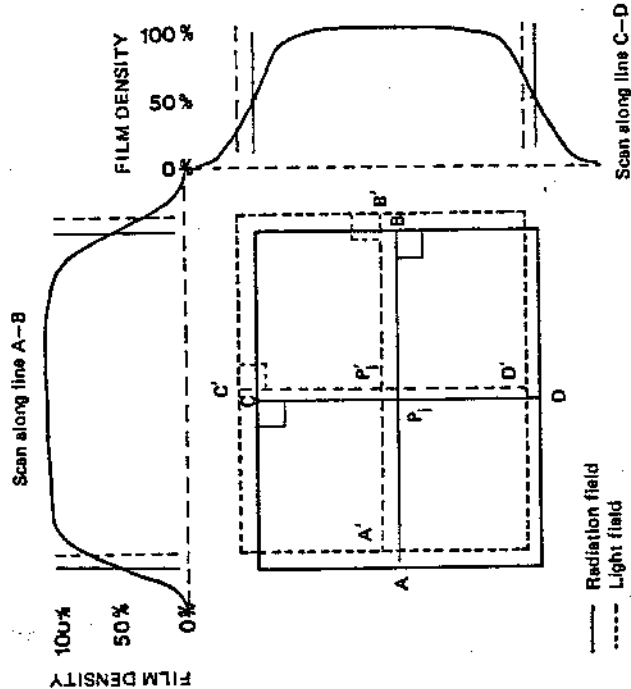


FIG. 4. Field symmetry and alignment of the light beam to the radiation beam. A film is exposed perpendicularly to the central axis of the radiation beam with the edges of the light field marked by metal or pressure. The angular rotation of the collimator should be noted for each irradiation and related to the orientation of the film. Electron buildup material is placed over the film if necessary. From densitometric scans across the field, the location of the edges of the therapy beam is taken to be where the film density is 50% of that at the centre of the field; the edges of the light field are at the markings (from Ref. [27]).

- (d) The light beam axis. This is defined as the line from the light source focus and the centre of gravity of the area within which the light intensity exceeds 50% of the maximum light intensity at the phantom surface (the centre of gravity coincides, generally, with the point of intersection P'_1 defined by $AP'_1 = BP'_1$ and $CP'_1 = DP'_1$ for a rectangular field as in Fig. 4).

The alignment of these axes may be checked by the following procedures.

- (a) Collimator axis. Figure 5 shows a simple alignment device that can be clamped in the jaws of the collimator. The centre of this device must be accurately known and provided with a projecting pointer. When the collimator is rotated on its axis the point should describe a circle of diameter ≤ 2 mm at the normal SSD. The axis of this circle is the centre of the collimator. The centre of the effective radiation source should be on the collimator axis. Its position can be checked as part of the radiation beam check.

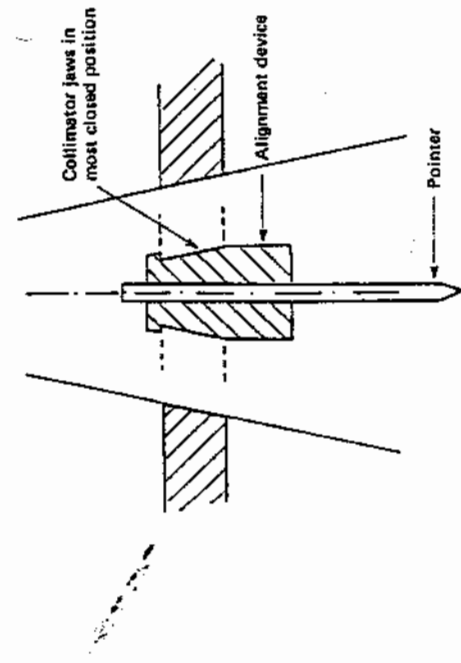


FIG. 5. Alignment device used for the determination of the collimator (rotation) axis.

- (b) It is not necessary to check the alignment of the geometrical beam separately as this is included in (a).
- (c) The radiation beam axis can be checked by using an ionization chamber in a water tank or using radiographic film between slabs of a plastic phantom and obtaining two dose profiles at right angles to each other as shown in Fig. 4. A special investigation should be carried out to find out if the relative absorbed dose profile measured in a water phantom can be approximated by relative film density curves using a plastic phantom for the beam qualities in use. The profiles should be scanned to determine the 50% points² on each profile and then the mid-point between them. The position of these mid-points defines the centre of the radiation beam for a given position of the collimator. The collimator should then be rotated through 180° and the process repeated. If the locations of the mid-points are within about 2 mm of each other at the normal SSD the source can be taken to be on the axis of rotation of the collimator and so also is the axis of the radiation beam.
- (d) The coincidence of the light field with the radiation field is most easily checked by using radiographic film. With the film in place as for (c) above, the position of the light field may be indicated on the film by means of suitably located lead markers; alternatively a ballpoint pen may be used to mark the light field directly through the envelope of the film. Again, the corners of the light field

² In some therapy units the radiation beam is defined by the 80% or 90% isodose level. This criterion can be used but it has then to be uniformly used for all the equipment at a department.

may be checked by using a needle to pierce the envelope and film base. Inspection of the differences between the radiation beam image at its 50% (isodose) level and the light beam marks will determine whether the light field and radiation field are symmetrical and whether this symmetry is retained on rotation of the collimator. The congruence between radiation and light beam should be within 2 mm on each side at the normal SSD. This process should also be repeated at 0°, 90°, 180° and 270° angulations of the gantry of the treatment unit.

4. RADIATION QUALITY SPECIFICATION AND DETERMINATION

The need for specification of radiation quality comes from the fact that several of the parameters required for absorbed dose determination depend on the photon or electron energy. Examples of such parameters are photon absorption coefficients, electron stopping powers, and various perturbation factors. A complete characterization of a radiation beam would involve specification of the type of particles and their energy and angular distribution. This type of information is in general very difficult to obtain but it is possible to perform accurate dosimetry with much less detail. A number of practical schemes for specifying radiation beam quality have been devised. Those chosen for use in this code of practice are described in the following sections.

4.1. HIGH ENERGY ELECTRONS

The electron beam already has a certain energy spread before meeting the accelerator exit window. This spread depends on a large number of factors such as type of accelerating mechanism, method of injection and extraction, and beam handling system (see Ref. [16]). The energy spectrum at the accelerator exit window (vacuum side), shown on the right side of Fig. 6 and labelled (a), may be characterized by a relatively small number of parameters such as those indicated: $E_{\text{max},a}$ is the maximum energy, $\bar{E}_a$ is the average electron energy, $E_{p,a}$ is the most probable electron energy and Γ_a is the energy spread, i.e. the spread at half maximum of the energy spectrum.

As the electron beam passes through the different materials between the accelerator exit window and the phantom surface, energy losses displace the spectrum to lower energies and at the same time fluctuations in such energy losses widen the spectrum. This also happens when the electrons pass into the phantom. Thus, the energy spread at half the maximum of the spectrum Γ increases so that $\Gamma_a < \Gamma_0 < \Gamma_z$ where a , 0 , and z indicate accelerator window (vacuum side), phantom surface and a certain phantom depth respectively, see Fig. 6. Because the spectrum is

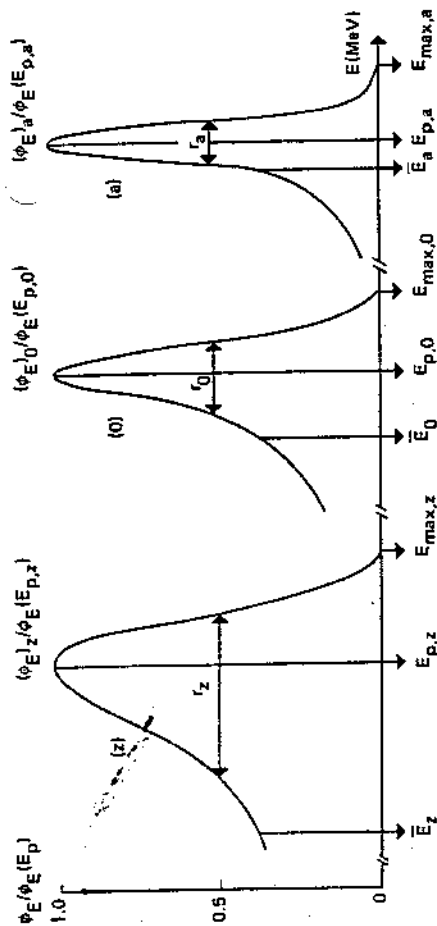


FIG. 6. Electron energy spectra and their parameters: (a) indicates the energy distribution of the electron beam as it exits before meeting the accelerator window; (b) indicates the distribution at the phantom surface and (c) indicates the distribution at a depth z in the phantom.

somewhat skewed towards low energies, $\bar{E}_0$ and $E_{p,0}$ do not coincide and $\bar{E}_0$ will always be less than $E_{p,0}$. The difference between $E_{p,0}$ and $\bar{E}_0$ increases with the energy spread Γ_0 .

The electron beam quality can be specified in a practical way by two of these parameters, $\bar{E}_0$ and $E_{p,0}$. Procedures for determining them are given in Section 4.1.1.

The most probable energy $E_{p,0}$ has been found to be a convenient parameter for characterizing isodose curves or depth dose distributions. This is because it can be related to an experimentally accessible quantity, the practical range R_p .

Different electron beams with the same most probable energy $E_{p,0}$ and the same field size and SSD, however, may produce different absorbed dose distributions. This is because the energy and angular spread of the electrons incident on the phantom surface [16] may be different. For instance, the so called therapeutic range, R_t (useful depth of treatment), which is related to a certain absorbed dose level, usually 85%, may differ considerably for different accelerators at a certain $E_{p,0}$ [28].

The most probable energy at the vacuum exit window $E_{p,a}$ is an intrinsic property of the beam and it may be convenient to use this to characterize the beam at the operator's console. This parameter is often used as it is independent of flattening filters, monitor chambers, etc. Its determination, basically theoretical, can be made from $E_{p,0}$ [6] but a detailed knowledge of the structure and materials of the accelerator treatment head is needed, and the method will not be considered here.

As the electron spectrum at the depth of interest is not known, the mean energy at the phantom surface $\bar{E}_0$ together with the depth of measurement z are used as input parameters for stopping power ratios. This method gives only approximate values for these parameters as the data used below (i.e. Table X) are strictly valid for monoenergetic and monodirectional broad electron beams perpendicularly incident on the phantom surface³.

The mean energy $\bar{E}_z$ of the electrons at the depth of interest is the essential parameter for determining some dosimetric factors (i.e. p_u , see Section 6.1) that allow for the calculation of absorbed dose. Strictly, these factors should be evaluated by considering the whole spectrum, but their evaluation by considering only the mean energy is a reasonable approximation.

4.1.1. Range-energy relationships

The determination of $E_{p,0}$ and $\bar{E}_0$ is based on the use of empirical relationships between electron energy and range parameters.

Range parameters are determined from depth absorbed dose distributions as shown in Fig. 7. The practical range R_p and the half-value depth R_{50} are of special importance for range-energy measurements: R_p is defined as the depth where the tangent to the descendent part of the curve intersects the prolongation of the bremsstrahlung tail⁴; R_{50} is defined as the depth where the absorbed dose is 50% of the maximum.

The empirical energy-range relations that will be given here are strictly valid for broad and parallel electron beams only, perpendicularly incident on a semi-infinite water phantom. It is difficult to set a limit beyond which a clinical beam can be considered as a broad beam, i.e. when central axis depth dose distributions become independent of the field size, as it will depend on the energy and angular spread of the beam. On the basis of experimental results [30] the use of field sizes of 12 cm x 12 cm or more are recommended for energies up to $\bar{E}_0 = 15$ MeV, and sizes of 20 cm x 20 cm or more are recommended above that energy.

This report does not consider absorbed dose determinations to a medium different from water. Nevertheless, for electron beams with $\bar{E}_0 \leq 10$ MeV, it may

³ It has been suggested by Johansson and Svensson [29] that these two parameters, $\bar{E}_0$ and z , are not sufficient for the selection of stopping power ratios in beams of large spectral width. They used the difference $(E_{p,0} - \bar{E}_0)$ as an indication of the spectral width, which is often wider in accelerators with a flattening system composed of a single scattering foil and applicators or cones. It was suggested that the value of $(s_{w,air})_d$ (at the reference depth for electron beams with $\bar{E}_0$ between about 15 and 25 MeV) should be increased by 2-3% for a difference of $E_{p,0} - \bar{E}_0$ of about 2 MeV.

⁴ Some authors consider the intersection with the depth axis. This produces practical ranges different from those considered here.

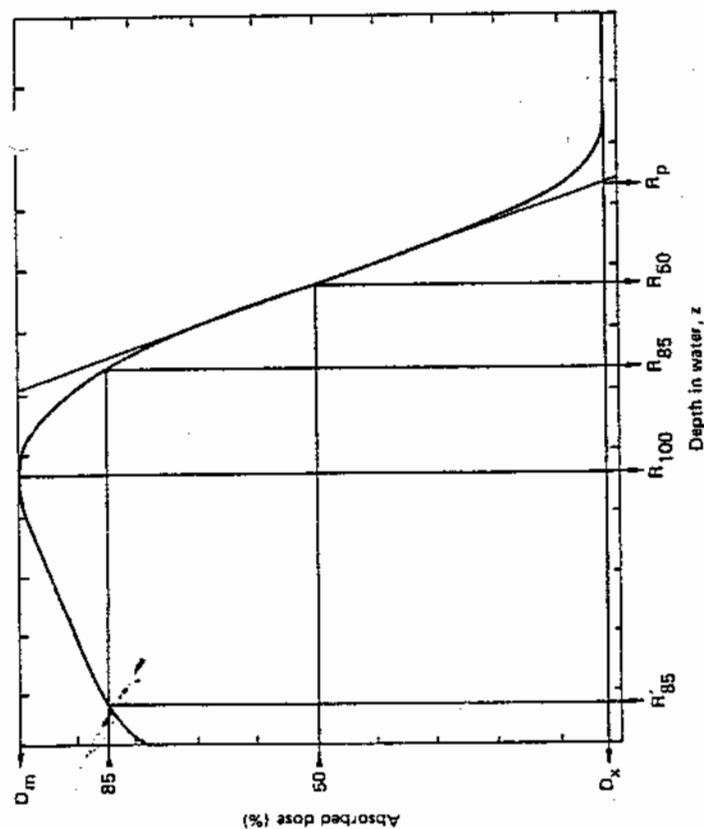


FIG. 7. An electron beam absorbed dose distribution in a water phantom showing the significance of various parameters: D_m is the maximum absorbed dose, D_x is the absorbed dose due to bremsstrahlung, R_{100} is the depth of dose maximum, R_{85} is the therapeutic range (it is here assumed that $R_t = R_{85}$; the depth at which the therapeutic interval intersects the depth dose curve near the skin entrance is designated by R_{85}); R_{50} is the depth for 50% absorbed dose; R_p is the practical range. (From Ref. [28])

be more practical to use slabs of plastic to determine range parameters. In this case, it is necessary to scale R_p and R_{50} to water equivalent ranges by using the relationship

$$\frac{R_{pl}}{R_w} = \frac{(r_0/\rho)_{pl}}{(r_0/\rho)_w} \quad (1)$$

where R_w and R_{pl} are ranges (R_t , R_{50} , R_p , etc.) in water and plastic respectively, r_0/ρ is the linear continuous-slowing-down range tabulated as a function of energy by ICRU [16]. Values of the r_0/ρ ratios for a few materials are reported in Table III. The mean energy at the phantom surface $\bar{E}_0$ can be used as an input parameter as the ratio in Eq. (1) is not very sensitive to the variation of the energy and therefore also not to the energy parameter chosen.

TABLE III. r_0/ρ VALUES OF THE LINEAR CONTINUOUS-SLOWING-DOWN RANGE r_0/ρ^a AS A FUNCTION OF ELECTRON ENERGY FOR DIFFERENT PLASTICS TO THE CORRESPONDING VALUES FOR WATER FROM REF. [16]

The densities for A-150, polystyrene and PMMA were 1.127, 1.06, and 1.19 g·cm⁻³, respectively^b

$\bar{E}_0$ (MeV)	A-150	Polystyrene	PMMA
0.1	0.879	0.961	0.863
0.2	0.880	0.963	0.864
0.5	0.882	0.964	0.864
1.0	0.886	0.967	0.865
2	0.891	0.970	0.867
5	0.897	0.974	0.870
10	0.902	0.978	0.872
15	0.906	0.983	0.874
20	0.909	0.987	0.876
30	0.914	0.993	0.878
40	0.919	0.999	0.882
50	0.922	1.003	0.883

^a Note that r_0 conventionally is given in units of g·cm⁻² while R_p , R_{50} and R_p generally are given in the unit cm.

^b The density of plastic may vary from one sample to another. It is therefore recommended that the density is measured and corrections are applied if necessary to the figures given in the table.

The range-energy relationships are strictly valid only for depth absorbed dose distributions (using broad and parallel beams, see above). However, in practical dosimetry, depth ionization distributions are generally used instead of convenience. This is adequate if R_p is measured to determine $E_{p,0}$ (Eq. (2)), but not for determination of $\bar{E}_0$ by Eq. (3), in particular not at high energies. A correction has then to be performed (this is already included in the values given in Table IV).

In the measurement of depth ionization distributions it is necessary to use the effective point of measurement of the cylindrical chamber (see Fig. 11) and to correct the readings for the ion recombination and polarity effects (Section 7.2). For energies $\bar{E}_0 \leq 10$ MeV, or whenever the inherent uncertainty associated with the

TABLE IV. RELATIONSHIP BETWEEN THE MEAN ENERGY AT WATER PHANTOM SURFACE OF AN ELECTRON BEAM ($\bar{E}_0$) AND THE HALF-VALUE DEPTH MEASURED FROM ABSORBED DOSE AND IONIZATION CURVES AT SSD = 1 m AND BROAD BEAMS (R_{50}^D AND R_{50}^I , RESPECTIVELY), FROM REF. [6]
(Equation (3) is only valid for absorbed dose curves in a broad plane parallel beam)

$\bar{E}_0$ (MeV)	R_{50}^D (cm)	R_{50}^I (cm)
1	0.3	0.3
2	0.7	0.7
3	1.2	1.2
4	1.6	1.6
5	2.1	2.1
6	2.5	2.5
7	3.0	3.0
8	3.4	3.4
9	3.8	3.8
10	4.3	4.3
12	5.1	5.1
14	6.0	5.9
16	6.8	6.7
18	7.8	7.6
20	8.6	8.4
22	9.4	9.2
25	10.7	10.4
30	12.8	12.3
35	14.6	14.0
40	16.3	15.4
45	18.1	16.9
50	19.7	18.2

determination of the effective point of measurement and (or) the perturbation correction effect (section 6.1) has to be reduced, the use of plane parallel chambers is recommended [6, 7].

(a) Determination of $E_{p,0}$

The most probable energy at the surface is related to the practical range by the following empirical relation [6]

$$E_{p,0} = C_1 + C_2 R_p + C_3 R_p^2 \quad (2)$$

with $C_1 = 0.22$ MeV, $C_2 = 1.98$ MeV·cm⁻¹, $C_3 = 0.0025$ MeV·cm⁻². This equation is valid for large field sizes and for most of the existing accelerators: R_p can be determined from depth ionization or absorbed dose distributions measured at SSD ≥ 1 m.

(b) Determination of $\bar{E}_0$

The mean energy at the phantom surface is related to the half-value depth R_{50} by means of the expression [28]

$$\bar{E}_0 = C_4 R_{50} \quad (3)$$

where $C_4 = 2.33$ MeV·cm⁻¹.

Strictly speaking, Eq. (3) is valid only for large field sizes (following the criterion given above), in the energy range between 5 and 35 MeV, and for R_{50} determined from absorbed dose distributions at a constant source-chamber distance (i.e. plane parallel beam). However, it is also possible to determine $\bar{E}_0$ if the half-value depth has been obtained from depth dose or ionization curves at SSD = 1 m using Table IV.

(c) Determination of $\bar{E}_z$

The mean energy as a function of depth is normally determined by means of approximate relations. For monoenergetic electron beams with the electron energy E_0 at the phantom surface, Harder [31] suggested that the depth dependence of the mean energy could be approximated by the simple relation

$$\bar{E}_z \approx E_0 (1 - z/R_p) \quad (4)$$

where z is the depth and R_p is the practical range in a water phantom for an initial energy E_0 . Usually, E_0 is replaced by $\bar{E}_0$ in this equation. This approximation is

recommended or for use with mean energy $\bar{E}_0$ less than 10 MeV or for small depths at higher energies.

In this report Monte Carlo calculations (Andreo and Brahme [32] and Andreo [33]) are recommended for the evaluation of $\bar{E}_z$. Table V gives the ratio of the mean energy $\bar{E}_z$ at a depth z in water (as the fraction z/R_p), to the mean energy $\bar{E}_0$ at the phantom surface, for electron beams between 5 and 50 MeV. These values are to a first approximation valid for clinical beams independently of their energy and angular spread.

4.2. HIGH ENERGY PHOTONS

4.2.1. ^{137}Cs and ^{60}Co gamma rays

The radiation from ^{137}Cs is monoenergetic at 0.662 MeV and that from ^{60}Co is nearly monoenergetic, being made up of two photon energies, 1.17 MeV and 1.33 MeV, giving an average of 1.25 MeV. The beams from SSDL or therapy sources, however, include a certain amount of lower energy scattered photons, thus somewhat reducing the effective energies.

4.2.2. X-rays from accelerators

Photon beams from accelerators are produced by electrons striking a target. The resulting bremsstrahlung spectrum depends on the electron energy, the target material and the target thickness. It will also depend on the thickness and material used and the design of the beam flattening filter. This, and the shape of an X-ray (bremsstrahlung) spectrum, makes it very difficult to satisfactorily characterize the photon beam by a single parameter (such as maximum energy, most probable energy or mean energy) as was done for electron beams. In this sense quality specification for high energy photon beams is more complicated than for electron beams but in another sense it is less complicated because the photon spectrum does not change appreciably with depth. This, and the idea that high energy photon beams are 'somewhat like gamma rays', has led people to believe that a photon beam energy can be used as a practical beam quality specifier.

The parameter that has been most commonly used is a 'nominal maximum energy' (or 'nominal accelerating potential') which approximates the energy of the electrons striking the target. This is not good enough [34, 35] to specify either beam parameters such as depth dose data and isodose charts or dosimetric parameters such as stopping power ratios and perturbation corrections.

For proper specification of these parameters the absorbed dose is to be determined in the user's beam at depths of 20 cm and 10 cm for a constant source-detector distance and a 10 cm x 10 cm field at the plane of the chamber, see Fig. 8 [8]. The ratio of these absorbed doses is designated as TPR_{10}^{20} (TPR is short for 'tissue-

TABLE V. RATIO OF THE MEAN ENERGY AT DEPTH z ($\bar{E}_z$) AND THE MEAN ENERGY AT PHANTOM SURFACE $\bar{E}_0$ FOR ELECTRON BEAMS IN WATER.

Depths are given as a fraction of R_p [32, 33]

z/R_p	5 MeV	10 MeV	20 MeV	30 MeV	40 MeV	50 MeV
0.00	1.000	1.000	1.000	1.000	1.000	1.000
0.05	0.943	0.941	0.936	0.929	0.922	0.915
0.10	0.888	0.884	0.875	0.863	0.849	0.835
0.15	0.831	0.826	0.815	0.797	0.779	0.761
0.20	0.772	0.766	0.754	0.732	0.712	0.692
0.25	0.712	0.705	0.692	0.669	0.648	0.627
0.30	0.651	0.645	0.633	0.607	0.584	0.561
0.35	0.587	0.583	0.574	0.547	0.525	0.503
0.40	0.527	0.523	0.514	0.488	0.466	0.444
0.45	0.465	0.462	0.456	0.432	0.411	0.390
0.50	0.411	0.407	0.399	0.379	0.362	0.345
0.55	0.359	0.355	0.348	0.329	0.314	0.299
0.60	0.313	0.309	0.300	0.282	0.269	0.256
0.65	0.270	0.265	0.255	0.239	0.228	0.217
0.70	0.231	0.226	0.216	0.202	0.192	0.182
0.75	0.197	0.191	0.180	0.168	0.159	0.150
0.80	0.164	0.159	0.149	0.138	0.131	0.124
0.85	0.137	0.131	0.120	0.111	0.105	0.099
0.90	0.114	0.108	0.096	0.089	0.084	0.079
0.95	0.091	0.086	0.076	0.069	0.065	0.061
1.00	0.077	0.071	0.059	0.053	0.049	0.045

phantom ratio' ⁵. Stopping power ratios are given as a function of this parameter in Table XIII.

4.3. LOW AND MEDIUM ENERGY X-RAYS

The radiation quality of an X-ray beam is normally characterized by tube potential, total filtration and first half-value layer.

The half-value layer, HVL, of a radiation beam is the thickness of an absorber which reduces the air kerma rate of a narrow unidirectional X-ray beam at a point distant from the absorbing layer to 50% compared with the non-attenuated beam.

The dependence of calibration factors as well as correction, conversion and backscatter factors on the radiation quality is expressed by their dependence on the half-value layer in aluminium or copper. Aluminium is used for X-rays with tube potentials up to 100 kV. Copper is used above 100 kV but also as an overlap down to 50 kV.

For very accurate determinations of the half-value thickness the experimental set-up according to the Fig. 9 is recommended. The purity of the aluminium absorber material should be better than 99.99% for HVL (Al) < 0.2 mm and 99.8% in the case HVL (Al) ≥ 0.2 mm. The thickness of the absorber should be known to better than ± 5 μm or ± 1%, whichever is the stronger requirement.

For hospital practice a simple experimental set-up may be applied and also less restricted purity of the absorber may be used.

The energy dependence of the response of the ionization chamber should be within the limits given in Table I, so as not to introduce errors in measuring at different qualities with and without an absorber in the X-ray beam.

5. MEASUREMENT CHAIN

The first step in the measurement chain, namely, the transfer of a calibration expressed in terms of calibration factors, has been described by OIML [36]. The same quantity (i.e. exposure, air kerma, water kerma or absorbed dose) is measured at each stage of this first step. The irradiation conditions, such as the photon energy,

⁵ Another and almost equivalent quantity can be obtained from depth dose data by using the dose ratio for 20 cm and for 10 cm depth D_{20}/D_{10} , measured at SSD = 1 m for a 10 cm x 10 cm field at the phantom surface, which is approximately equal to the depth ionization ratio J_{20}/J_{10} (see Ref. [6]). This quantity is approximately related to TPR_{10}^{20} by an inverse square relationship. More accurate values obtained by fit to a large set of experimental data are given later in this text (Tables XX and XXI).

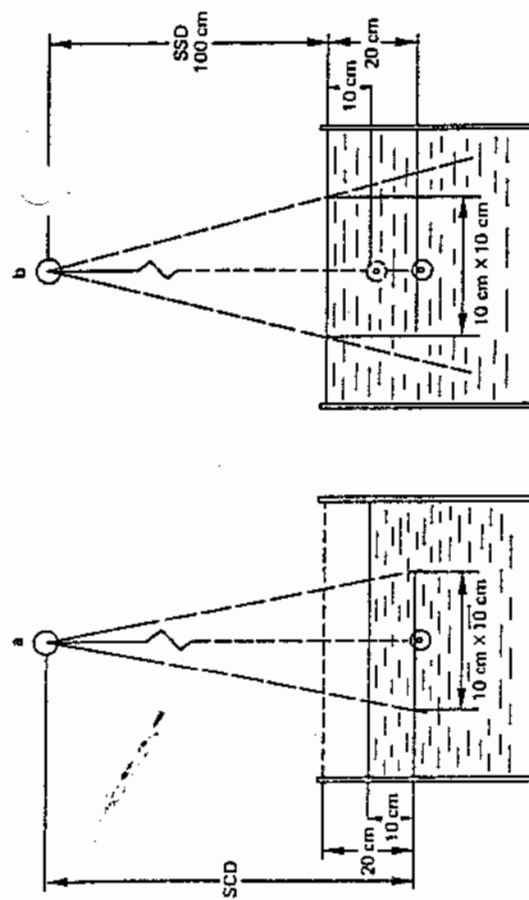


FIG. 8. The two experimental set-ups to determine the quality of photon beams. (a) The source-chamber distance (SCD) is kept constant and depth is changed by varying the amount of phantom material over the detector. TPR_{10}^{20} is measured (see text). (b) The source-surface distance (SSD) is kept constant and the detector is moved to different depths. D_{20}/D_{10} is measured.

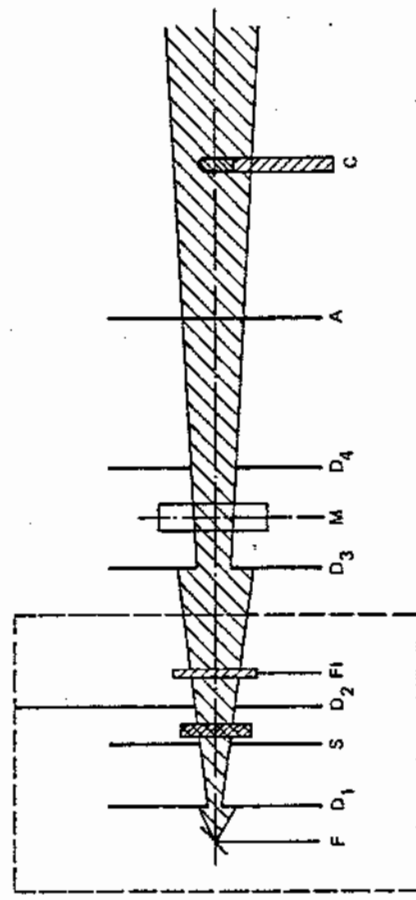


FIG. 9. Experimental set-up for determining half-value thickness: F is the focus of the X-ray tube, $D_1...D_4$ diaphragms to define beam cross-section or to reduce scattered radiation, S the shutter, F1 a filter, M a monitor chamber, A an absorber of variable thickness (Al or Cu), C an ionization chamber. Recommended dimensions: FC about 100 cm, MA about 20 cm for tube potentials greater than 50 kV and about 30 cm for tube potentials less than 50 kV. The components inside diaphragm D_4 should reduce the beam diameter at A to at least 4 cm. The positions are, however, not very critical for HVL determinations.

spectrum and the field size, must be as similar as possible at the SSDL (Primary Standard Dosimetry Laboratory) and the SSDL (Secondary Standard Dosimetry Laboratory), and also at the user's beam in order to avoid errors due to quality dependence of the chamber response. However, in the most usual situation at a hospital neither the same quantity is wanted as is available for calibration nor are the same beam qualities used. Thus, the more general aim is to determine the absorbed dose to water at several photon and electron beam qualities using an ionization chamber calibrated in air kerma or exposure. This protocol covers this more general situation.

It is necessary to use a consistent set of interaction coefficients at all the steps in order not to introduce unnecessary errors. In order to stress this point the complete measurement chain is analysed (see Svensson, Ref. [37]).

5.1. PRIMARY STANDARDS

Primary standards for absorbed dose to graphite (D_{gr}), exposure free in air (X), and air kerma (K_{air}) have been developed by several PSDLs. Intercomparisons of standards of the same type generally give agreement within 0.2-0.3% (see e.g. Refs [38-40]) at ^{60}Co gamma ray quality. Absorbed dose to graphite is also considered to be known within these limits in absolute terms because of calorimetric measurements.

The absolute uncertainty in the exposure standard for ^{60}Co gamma radiation is much larger than the proven consistency between the standards. The reason is that exposure is measured in terms of electric charge and mass of an air ionization chamber. Interaction coefficients must be used to convert these measurements to a statement of absorbed energy. The coefficients, $s_{gr,air}$ and $(\bar{\mu}_{en}/\rho)_{air,gr}$ are needed if a graphite ionization chamber is used for exposure determination, see Eq. (a) in Fig. 10. The interaction coefficients are generally taken from the same sources by the different PSDLs, and so consistency is observed but accuracy is not tested. Recent computations of $s_{gr,air}$ suggest that conventionally accepted values [41] should be decreased by about 1% (see, for example, Ref. [42]). The uncertainty in $(\bar{\mu}_{en}/\rho)_{air,gr}$, however, is about 0.1% (1 σ) for the ^{60}Co gamma quality and is therefore of minor importance.

The determination of air kerma, $K_{air,c}$ at the calibration radiation quality (usually ^{60}Co gamma rays; index c stands for calibration) from measurements of the mean specific ionization inside the ionization chamber air cavity (i.e. liberated charge of one sign through the mass of air inside the sensitive part of the ionization chamber air cavity) J_{air} also includes the values of $(1 - g)$ and $(W/e)_c$.

The value of g (the fraction of the energy of the secondary charged particles lost to bremsstrahlung in air) is close to 0.3% for ^{60}Co gamma (CCEMRI-1 [20])

and lower than the value for medium and low energy X-rays. The uncertainty in the value of g is insignificant compared to other sources of uncertainties for the determination of K_{air} .

The mean energy expended in air per ion pair formed and per electron charge W/e , is very often considered constant for electron radiation and electron beams produced by photons or radioactive sources [43]. A value of $W/e = 33.85 \text{ J/C}$ is recommended for dry air by the ICRU [43]. However, this value is to a large part based on experiments carried out by comparison measurements using a graphite calorimeter and a graphite ionization chamber inside in a graphite phantom. The Bragg-Gray equation $D_{gr} = s_{gr,air} W/e J_{air}$ is applied to determine the product $s_{gr,air} W/e$ (D_{gr} is the absorbed dose to the graphite absorber and J_{air} is the specific ionization measured at the corresponding depth in the graphite phantom in the ionization chamber). Theoretical values of $s_{gr,air}$ are used and a value of W/e is obtained.

Recent recommended changes in stopping-power values [18] will thus directly influence the value of W/e . CCEMRI(1) [20] has therefore recommended the use of $W/e = 33.97 \text{ J/C}$ for dry air (from 1986-01-01), which is therefore recommended here. It is seen from Eq. (a) in Fig. 10 that the air kerma includes the product of $s_{gr,air}$ and W/e , which is better known than are the separate values. Therefore the air kerma has a much smaller uncertainty than the exposure (which only makes use of the numerical value of $s_{gr,air}$).

For low energy and medium X-rays the standards are based on measurements using free-air chambers. Intercomparisons of exposure standards generally give an agreement within $\pm 0.5\%$, see, for example, Ref. [44]. The same value of W/e (i.e. 33.97 J/C) as for ^{60}Co gamma radiation is used.

5.2. SECONDARY STANDARDS

The main task of the SSDLs is to transfer the calibration from the PSDLs to the user. It is assumed that calibrations either in terms of air kerma or exposure have been carried out for the SSDLs at the PSDL (calibration factors N_k and N_x respectively, see Fig. 10). The beam qualities and the irradiation geometries at the PSDLs and at the SSDLs need to be as similar as possible in order not to introduce errors in this transfer.

Another important task for the SSDL is to advise and help the users to calculate chamber factors to be used in the Bragg-Gray equation (Eqs (f) and (g) in Fig. 10) in order to determine absorbed dose at the user's beam. The derived factor to be used for this purpose is the absorbed dose to air chamber factor, N_D (Eqs (d) and (e) in Fig. 10) (see Refs [6, 16]).

Geometry

St
(Eq. No.)

Step
(Eq. No.)

Relationship

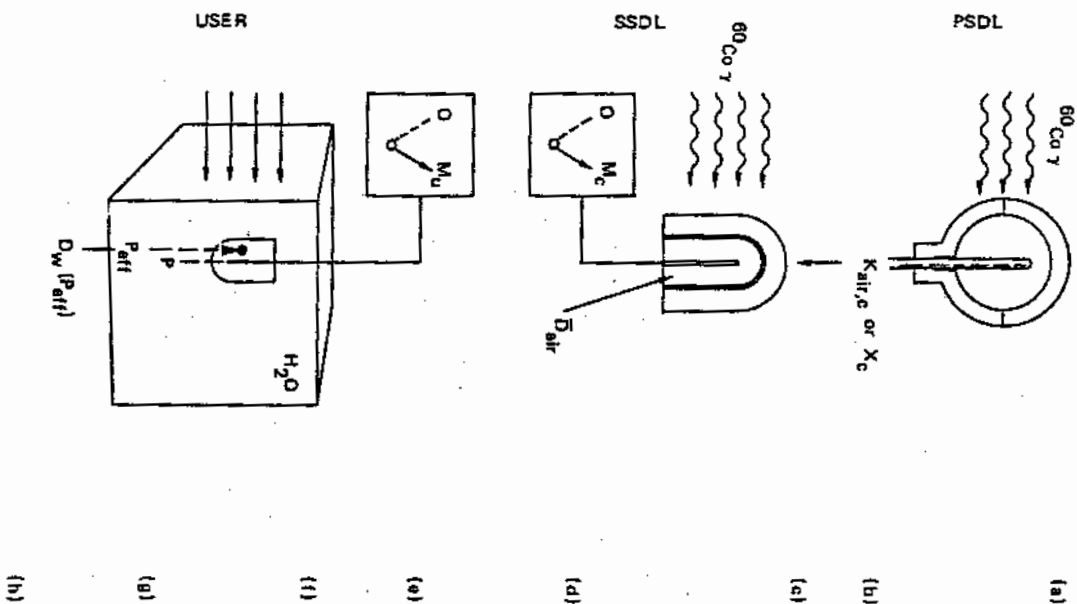
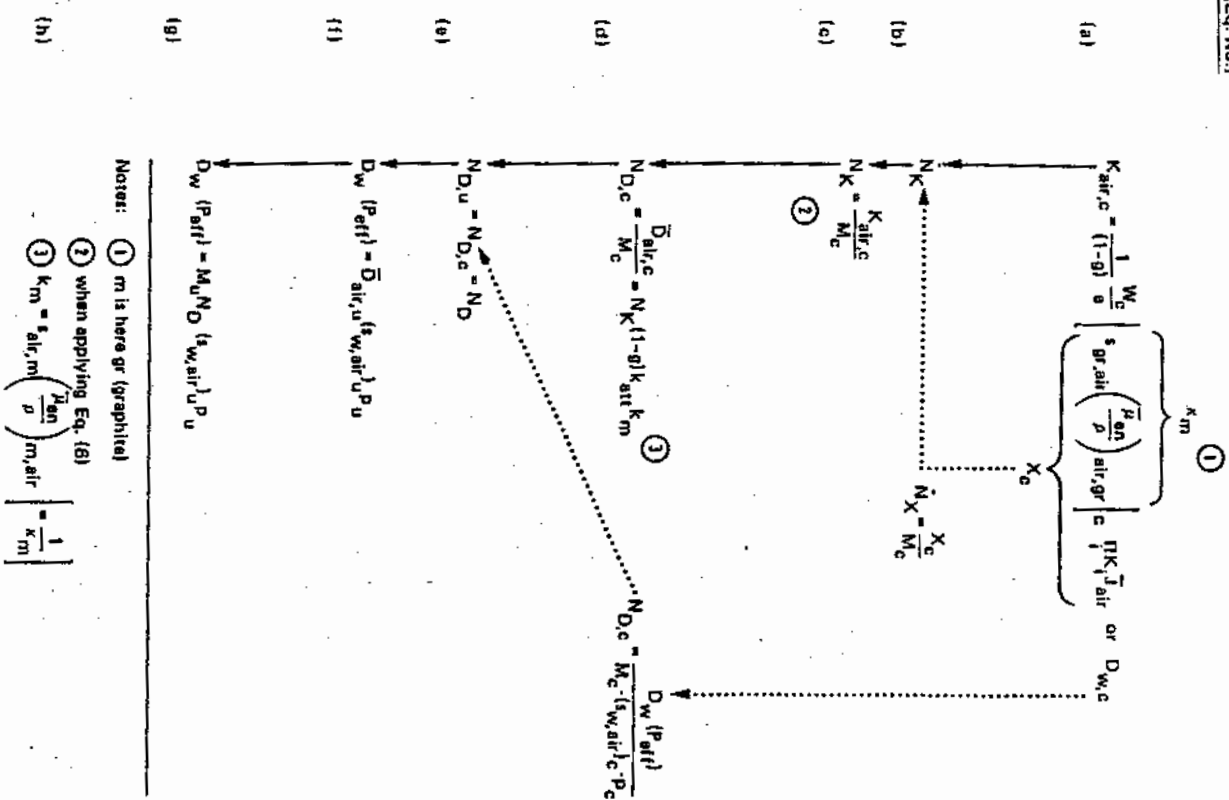


FIG. 10. The calibration chain for electron radiation and high energy photons from PSDL to SSDL to user. It is recommended that the solid line is followed. The air kerma or exposure calibration of the user's chamber can be utilized in this chain; k_m and k_f are correction



Notes: ① m is here gr (graphic)
② when applying Eq. (6)
③ $k_m = \left(\frac{\rho_{air}}{\rho} \right) \left(\frac{P_{air}}{P_u} \right) \left(\frac{1}{k_m} \right)$

factors applied when determining exposure or air kerma in standard laboratories. The rest of the symbols are explained in the text.

5.3. USER'S DOSIMETRY SYSTEM

The user must have an ionization chamber and a measuring assembly (referred to as the local reference instrument) following the performance specifications given in Section 3.2.1. It is essential that this chamber be checked regularly against a stability check source. Suitable for such checks are $^{90}\text{Sr} + ^{90}\text{Y}$ sources or ^{60}Co gamma therapy sources. Constancy checks must always be performed before and after the calibration at the SSDL.

6. FORMALISM

A formalism based on an air kerma calibration of the user's ionization chamber is recommended. It is recommended that the transfer be carried out according to the calibration chain shown in Figure 10. The values of the interaction coefficients to be used in the different steps of the absorbed dose determination are given in Section 7. In this Section the component steps that make up the formalism are discussed and some explanation of each is given.

The calibration quality (subscript c) in Section 6.1 is assumed to be ^{60}Co gamma radiation. The user's quality (subscript u) is assumed to be ^{137}Cs gamma radiation, ^{60}Co gamma radiation, photons or electrons from an accelerator (these radiation qualities are for simplicity named 'high energy photon and electron radiation'). Electrons with energies below 10 MeV are discussed separately because different types of ionization chambers are often used for electron beams below and above about $E_0 = 10$ MeV as the scattering effect is fairly large and special restrictions are needed for the shape of the air cavity. Somewhat modified procedures will also be discussed for the situation where the user wants to determine absorbed dose in X-ray beams with photon energy below 300 keV (Section 6.2).

6.1. ELECTRON AND HIGH ENERGY PHOTON RADIATION

The determination of the exposure and air kerma standard for ^{60}Co gamma rays at PSDL has been discussed in detail elsewhere (see Refs [45-47]). The air kerma or exposure standard at the PSDL is used in order to determine calibration factors for the secondary standard of the SSDL in terms of air kerma per meter reading or charge ($\dot{N}_k$), or exposure per meter reading or charge ($\dot{N}_x$). This standard is then used in the calibration of the user's instrument at the SSDL to yield a factor N_k . An alternative would be to make a calibration of the chamber at a point of interest in a water phantom where the absorbed dose to water could be determined

by the PSDL (Fig. 10). However, this procedure is not discussed in detail as absorbed dose to water calibration is not generally offered by the PSDLs. It is recommended that the calibration of the user's chamber be made free in air in a ^{60}Co gamma beam. If a therapy source is used then a source to chamber distance of 1 m (to the centre of the chamber) and a field size of 10 cm $\times$ 10 cm is preferable as the reference condition. The choice of source distance and field size is not critical, however [48]. The uniformity of the beam should always be investigated and an increase of field size above 10 cm $\times$ 10 cm may prove to be necessary for some radiation therapy units.

For a given field size, measured at the position of the chamber, the scattered photon contribution from the source and collimator to the exposure at central axis will decrease with an increase of the distance to the source. An SCD larger than 1 m is, however, often impractical as the exposure rate may be too low.

The interest of the user at a hospital is how to utilize the air kerma calibration factor of the ionization chamber, N_k ($= K_{\text{air},c}/M_c$), valid only at the calibration quality (here ^{60}Co gamma rays, index c) to determine the absorbed dose at the radiation beams used for radiation treatment. The procedure can be divided into two steps:

(1) The air kerma $K_{\text{air},c}$ (giving a meter reading M_c) corresponds to a certain mean absorbed dose to air inside the cavity of the chamber, $\bar{D}_{\text{air},c}$, which is calculated. The relation between $K_{\text{air},c}$ and $\bar{D}_{\text{air},c}$ depends on the construction of the ionization chamber and buildup cap (Eqs (6) and (7) below). The absorbed dose to air chamber factor defined by

$$N_{D,c} = \frac{\bar{D}_{\text{air},c}}{M_c} \quad (5a)$$

can be determined. It is assumed that this factor also is valid at the user's beam quality (subscript u), i.e.

$$N_{D,c} = N_{D,u} = \frac{\bar{D}_{\text{air},u}}{M_u} \quad (5b)$$

(2) The Bragg-Gray equation (Eq. (4) in Fig. 10) is used to determine the absorbed dose to water at the point of interest in the water phantom.

In the first step, air kerma (free in air) $K_{\text{air},c}$ can be related to the (mean) absorbed dose $\bar{D}_{\text{air},c}$ inside the air cavity of the user's ionization chamber by

$$\bar{D}_{\text{air},c} = K_{\text{air},c} (1 - g) k_{\text{air}} k_m \quad (6)$$

i.e.

$$N_{D,c} = N_k (1 - g) k_{\text{air}} k_m \quad (7)$$

$$D_w(P_{eff}) = M_u N_d (S_{w,air})_u P_u \quad (9)$$

It is thus assumed in Eq. (9) that the $N_{d,c}$ determined at the calibration quality (^{60}Co gamma) is also valid at the user's beam quality, i.e. $N_{d,c} = N_{d,u}$. This assumption holds only if $W_e = W_u$. Recently, Svensson and Brahme [42] showed that this may not be the case in the energy region covered in this report. More experimental data on W_u/W_e are needed. However, as the energy dependence of W_u/W_e is probably within 1.5% this ratio has been taken to be unity in this report. No subscripts are therefore used for the absorbed dose to air calibration factor in Eq. (9), i.e. $N_{d,c} = N_{d,u} = N_d$.

The photon and electron fluence in the uniform water phantom is *perturbed* in the volume which will be occupied by the ionization chamber when the measurements are performed. This effect is corrected for by:

- (1) Applying a *perturbation correction factor* P_u , that corrects: (a) for the different properties in electron production and scattering in the chamber wall and corresponding volume of water; and (b) for the difference in electron scattering in the air cavity and in the water which is 'replaced' by the air cavity.
- (2) The use of an *effective point of measurement* that takes into account the spatial extent of the air cavity by locating the point of interest P_{eff} in front of the chamber centre P (see Fig. 11) to correct for the gradient of fluence within the chamber cavity.

The Bragg-Gray equation has been corrected in Eq. (8) to take these two effects into account. Generally, it is not possible to separate them. However, the magnitude of the perturbation correction factor P_u can be obtained experimentally from a comparison of data obtained with a cylindrical and a flat coin-shaped chamber at or near the broad dose maximum of the depth-dose curve (see Ref. [16] for electron radiation). In such experiments for photon beams the atomic composition of the wall material of the plane parallel chamber must be similar to water. The material is less critical for electron radiation (see Johansson et al. [54]). Values of P_u are given in Sections 7.2 and 8. These factors are thus valid only if the 'effective point of measurement' of the ionization chamber is located at the point of interest during the measurements.

Recommended values for the displacement of the 'effective point of measurement' in relation to the chamber centre are given in Fig. 11 and in Section 7.2.

The scattering power of electrons increases rapidly with a decrease in electron energy. The perturbation correction factor for electron beams will therefore also

The factor k_m takes into account the lack of air equivalence of the ionization chamber material at the calibration in the ^{60}Co gamma ray beam. The theoretical and experimental evaluation of k_m is discussed in Section 8, and values for a large number of chambers are given in Table XVIII (Section 8).

The factor k_{air} takes into account the attenuation and scatter of the photons in the ionization chamber material (including the buildup cap [16]). The influence of the photons absorbed and scattered in the chamber is thus included in this factor. Most electrons crossing the cavity and producing ionization in the air are generated upstream in the cavity wall. The correction factor, therefore, only includes attenuation in the material from the front surface of the buildup cap to the 'centre of electron production'. Theoretical values, based on Monte Carlo calculations, have been computed by Nah and Schulz [49] and Rogers et al. [50]. Values for a large number of chambers are given in Table XVIII.

It is preferable that the thickness of the buildup cap together with the chamber wall be standardized and a value of $(0.45 \pm 0.05) \text{ g} \cdot \text{cm}^{-2}$ has been recommended by NACP [6] and ICRU [16] for cylindrical chambers. A value of 0.990 ± 0.005 for k_{air} can be used for cylindrical chambers of the size and material recommended in Section 3.1.1. Chambers constructed of only one type of material in the walls, buildup cap and collecting electrode are preferable as the calculation of k_m and k_{air} is then simplified (see Ref. [7]). A value of $k_m k_{air} = 0.989$ (if $S_{air,c}$ from Table XVII, $\rho = 1.7 \text{ g} \cdot \text{cm}^{-3}$ is used; see also Table XVIII for the SS-graphite chamber) is recommended for a cylindrical chamber of pure graphite (i.e. graphite in the central electrode, wall and buildup cap). However, commercial chambers used at most hospitals today are composed of several materials. Experimental values of the product $k_m k_{air}$ (the separate values of k_m and k_{air} are not needed, see Eq. (d) of Fig. 10) are, however, available for a large number of chambers (Mattsson and Johansson [51]). These experiments are based on the assumption that $S_{air,gr}$ is known. In Table XVIII, calculated values of the product $k_m k_{air}$ are given instead as the agreement between the experiments and calculations are fairly good (within about 0.5% if the same $S_{air,gr}$ are used for the experiments and the calculations). Corrections for the 'non-air equivalence' of the central electrode have been disregarded in these calculations (see Section 8).

In the second step, the Bragg-Gray equation can be used to determine the absorbed dose to water at the point of interest (i.e. at the effective point of measurement of the chamber, P_{eff} ; see below):

$$D_w(P_{eff}) = \bar{D}_{air,u} (S_{w,air})_u P_u \quad (8)$$

where $(S_{w,air})_u$ is the stopping-power ratio water to air at the user's quality at the point of interest (see numerical values in Section 7.2), and P_u the perturbation correction factor (see below).

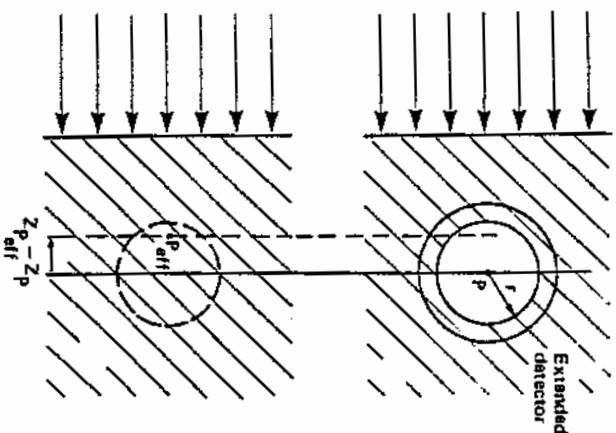


FIG. 11. Displacement of the effective point of measurement P_{eff} (depth $z_{p_{\text{eff}}}$) from the centre P (depth z_p) of an ionization chamber. Recommended values of $z_{p_{\text{eff}}} - z_p$ are:

- (a) for electron beams 0.5r (see ICRU 35 [16]);
- (b) for 'high' energy photon beams 0.75r [52-54];
- (c) for ^{60}Co gamma radiation 0.5r [55];
- (d) for ^{137}Cs gamma radiation 0.35r [55];
- (e) for medium X-rays zero [55].

increase with a decrease in electron energy (see Table XI). The uncertainty in this factor will increase for low energy electrons. One way to reduce this problem down to about $E_z \approx 4$ MeV (which corresponds to $E_0 \approx 5$ MeV for $z = 0.5$ cm, the approximate depth of dose maximum) is to use a cylindrical chamber with a very small cavity radius, i.e. r being less than 2 mm. However, the recommended method is to use a plane parallel chamber. The NACP [7] describes a procedure for determining N_D for such a chamber by experimentally comparing, in an electron beam, the reading of the plane parallel chamber with unknown N_D to the cylindrical chamber with known N_D (the P_u correction for the cylindrical chamber must be taken into account). The effective point of measurement will be situated at the front surface of the air cavity. The perturbation correction is very close to unity if an adequate guard is used [7]. The purpose of the guard, besides producing a suitable electrical field at the border of the collecting electrode, is to minimize the in-scattering of electrons from solid material at the lateral sides of the sensitive volume of the chamber air cavity [16].

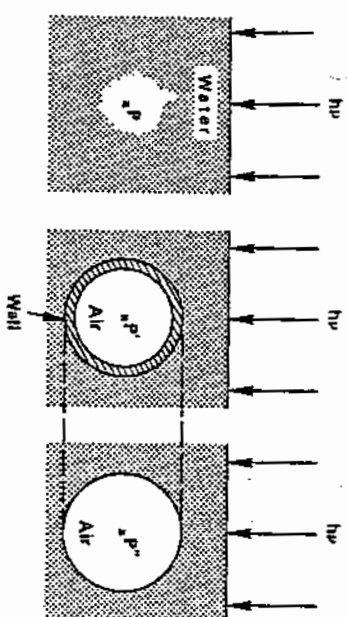


FIG. 12. The absorbed dose at a point P in an 'undisturbed' medium (water phantom) is to be determined. An exposure or air kerma calibrated free in air is placed with its centre P' at that point. The chamber has been calibrated free in air. Included in the calibration factor are any disturbances due to the chamber material. In the phantom measurement, the calibrated chamber will therefore give the exposure or air kerma value in the centre P'' of an air cavity equal to the external size of the chamber.

6.2. MEDIUM AND LOW ENERGY X-RAYS

6.2.1 Medium energy X-rays: 100 kV to 300 kV

The PSDs generally have air kerma and exposure standards for several X-ray energy regions. Therefore it is possible at the SSDL to transfer calibrations to the user's chamber in terms of exposure $N_X (= X_0/M_0)$ and air kerma $N_K (= K_{\text{air},0}/M_0)$. The relationship between these two calibration factors is (see Section 2.1):

$$N_K = N_X \frac{W}{e} \frac{1}{(1-g)} \quad (10)$$

The method for determination of absorbed dose in a water phantom using an exposure or air kerma calibrated ionization chamber will differ somewhat for medium energy X-rays from that at high energy radiation. The main reason is that none of the electrons generated in the water phantom will reach the air cavity. (A 200 keV electron will have a maximum range in graphite less than 0.3 mm, i.e. less than the thickness of most chamber walls.) Electrons producing ionization inside the air-cavity are therefore mainly generated in the chamber walls. (A few electrons are also generated in the air of the cavity.) The formalism below follows mainly that described by Johns and Cunningham [56] (p. 246).

The absorbed dose is to be determined by the user in a uniform water phantom at the point P (Fig. 12). An ionization chamber is placed with its centre P' at that point. A measurement is carried out giving the meter reading M_u . The air kerma

$$K_{\text{air}} = M_u N_K k_u \quad (11)$$

is obtained at a point P" in the centre of an air cavity inside the phantom as the chamber is calibrated to indicate the air kerma free in air, i.e. as if the chamber was not present. This means that we obtain the air kerma at the centre (P") of an air cavity of a size defined by the outer wall surfaces. Correction factors for attenuation in the chamber walls and the non-air equivalence of the walls are thus already included in the calibration factor N_K . The radiation quality correction factor k_u is required because the ratio K_{air}/M_u may be sensitive to the difference in spectral distribution of the radiation field used for the calibration free in air and that in the phantom at the position of the detector. For most practical situations k_u can be taken as unity, as it is recommended to use reference instruments for which the change in response (i.e. meter reading to air kerma) with energy is small (less than $\pm 2\%$ in the range of half-value layers from 2 mm Al to 3 mm Cu, i.e. from approximately 70 kV to 250 kV X-ray tube potentials [57]).

The water kerma is then obtained from

$$\frac{K_w}{K_{air}} = \left(\frac{\bar{\mu}_{tr}}{\rho} \right)_{w,air} \quad (12)$$

For conventional X-rays the energy transfer to bremsstrahlung from generated electrons is almost negligible, i.e. $g = 0$ and therefore $(\bar{\mu}_{tr}/\rho) = (\bar{\mu}_{en}/\rho)$, and furthermore $K_w \approx D_w$. Equations (11) and (12) give

$$D_w = M_u N_K k_u (\bar{\mu}_{en}/\rho)_{w,air} P_u \quad (13)$$

The perturbation correction factor P_u is here introduced explicitly because the equation without this factor would give the absorbed dose to a small mass of water at the centre P" of the cavity. In the formalism recommended by ICRU in its Report 23 [58] a 'displacement correction' is incorporated in the conversion coefficient F but in the medium energy X-ray region no detailed information was available at the time. The factor P_u corrects for the replacement of water by air and the chamber wall material. The values of P_u given in Section 7.3 are evaluated assuming the centre of the ionization chamber to be the effective point of measurement (see Fig. 11). The values of P_u were derived from measurements of absorbed dose in a phantom by means of an extrapolation chamber [59] and by means of thimble ionization chambers calibrated to indicate air kerma. Monte Carlo calculations by Schneider and Grosswendt [60] support the values given in Section 7.3. Recent calorimetric work [61, 62] leads to similar results, suggesting that much previous and present dosimetry with medium energy X-rays in phantom is in error (see also Ref. [63]). Therefore, use of the formalism described here may produce deviations from the results obtained by procedures neglecting perturbation corrections. Absorbed dose values derived by the present formalism will therefore be a few

per cent higher in the upper energy range with a maximum of up to about 10% at 100 kV.

6.2.2 Low energy X-rays: 10 kV to 100 kV

The main dosimetric task in this photon energy range is the determination of the absorbed dose at the surface of a phantom. The determination, using a plane-parallel chamber, can either be based on a calibration at the surface of a phantom, method (a), or (when the former method is not available) on a calibration carried out free in air, method (b).

(a) Primary standards based on extrapolation ionization chamber techniques have been established recently [64, 65]. These allow direct calibration of the user's ionization chamber indicating absorbed dose to water at the surface of a water phantom.

As the photon fluence decreases very rapidly within the first few millimetres of depth in the phantom, only plane-parallel ionization chambers with small electrode distances and very thin entrance foils are suitable (see Table I). These design features do not allow the use of the chamber inside a water phantom and therefore a solid phantom must be used. The calibration factor $N_{D,w}$ is defined as

$$D_w = M_u N_{D,w} \quad (14)$$

where D_w is the water absorbed dose at the surface of a water phantom as measured with a primary standard, and M_u is the reading of the instrument placed at the surface of a solid phantom, at the same distance from the source. It should be noted that an ionization chamber calibrated free in air should not be used on the surface of a phantom unless a correction factor which takes into account the scattering contribution from the phantom is known and applied.

(b) The chamber may be calibrated in terms of air kerma free in air. A calibration factor N_K is then available. No extra phantom is used for the measurement but the ionization chamber may be embedded in some material which then has to be regarded as part of the chamber. The inner surface of the entrance foil is the effective point of measurement. This is brought to the reference point. The absorbed dose to water at that reference point at the surface of a phantom in the absence of the ionization chamber is then given by

$$D_w = M_u N_K B k_u (\bar{\mu}_{en}/\rho)_{w,air} \quad (15)$$

The backscatter factor B for reference field size can be taken from Table XVI. The term $(\bar{\mu}_{en}/\rho)_{w,air}$ is the ratio of the averaged mass absorption coefficient of water to

that of air averaged over the spectral energy fluence distribution surface of the phantom. Values of this ratio are given in Table XIV, as a function of the beam quality expressed as half value thickness. The factor k_d corrects for the difference in spectral distribution of the radiation field used for the calibration and that at the surface of the phantom. For most practical situations this factor should be very close to unity.

7. DETERMINATION OF ABSORBED DOSE TO WATER

7.1. REFERENCE CONDITIONS AND CORRECTION FACTORS

The calibration factor for an ionization chamber is the ratio of the true value of the quantity to be measured to the indicated value under reference conditions. The reference condition is described by a set of influence quantities for which the calibration factor is valid without further correction factors. Influence quantities are defined as quantities not being the subject of the measurement but yet influencing the quantity under measurement. They may be of different nature as, for example, temperature, humidity and mains voltage; they may arise from the dosimeter (e.g. ageing, zero drift, warm-up) or may be quantities related to the radiation field (e.g. radiation quality, dose rate, field size, depth in a phantom). In calibrating a dosimeter as many influence quantities as possible are kept at reference conditions. However, some influence quantities, e.g. temperature, pressure, humidity, dose rate and polarity effect, must generally be corrected for to yield the 'influence' corresponding to the reference conditions. Reference conditions for both calibration of ionization chambers and measurement of absorbed dose, related to the radiation field, are recommended in Tables VI and VII.

As a general recommendation, when using therapy accelerators and X-ray units, it is strongly recommended that an additional monitoring dosimetry system be used during the experimental procedure to account for fluctuations in the radiation output.

Warm-up effect. Before measurements are made with an ionization chamber system, enough time should be allowed for the chamber to reach thermal equilibrium and for the measuring system to warm up.

Leakage current. The leakage current should be measured and should be insignificant compared to that current obtained for the real measurements. It should be observed that the leakage may be much larger during an irradiation than the leakage without irradiation [66, 67].

The polarity effect. The polarity effect of the chamber must be checked, particularly for plane parallel ionization chambers, and found to be within the limits

TABLE VI. REFERENCE CONDITIONS OF THE IRRADIATION GEOMETRY FOR THE CALIBRATION OF AN IONIZATION CHAMBER 'FREE IN AIR'

Radiation beam	Radiation quality	Distance from source to the centre of chamber cavity	Field size in cm × cm
Low energy X-rays ^a (10 kV to 100 kV)	0.03 mm Al < HVL ≤ 2 mm Al (calibration)	Usual treatment distance	3 × 3 or ø 3 cm
Medium energy X-rays (100 kV to 300 kV)	2 mm Al < HVL < 3 mm Cu	same	10 × 10
Co-60 gamma radiation ^b	(hv) = 1.25 MeV	1 m	10 × 10

^a It is generally preferred to calibrate the ionization chamber together with a phantom (see Section 6.2.2).

^b A buildup cap should be used (see Section 6.1).

given in Table I. This effect may be large for electrons of low energy. The mean value of the currents (or charges) measured at both polarities of the chamber voltage is considered to be the best approximation of the true ionization current. However, in practical situations, only one polarity is generally used as the correction using the chambers discussed here should be small.

Temperature and pressure. With ionization chambers open to the ambient air the mass of air in the ionization chamber volume during the measurement may be different from that to which the calibration is related. The correction factor to be applied to the measured current is

$$P_{TP} = \frac{P_0 (273.2 + T)}{P (273.2 + T_0)} \quad (16)$$

where P and T are the air pressure and temperature during measurement and P_0 and T_0 are the reference values (generally 101.3 kPa and 20°C) as stated in the calibration certificate.

Humidity. The humidity of the ambient air has — except when hygroscopic wall materials are used such as A-150 plastic or nylon [68] — only a minor influence on the charge produced in the chamber cavity. If the calibration factor is related to

TABLE VII. REFERENCE CONDITIONS OF THE IRRADIATION GEOMETRY FOR ABSORBED DOSE MEASUREMENTS USING AN IONIZATION CHAMBER IN A PHANTOM

Radiation beam	Beam quality	Reference depth in phantom in cm ^a	Effective point of measurement of the ionization chamber	SSD	Field size in cm × cm
Low energy X-rays	0.03 mm Al < HVL ≤ 2 mm Al	Surface	Front surface of the plane parallel chamber	Usual treatment distance	3 × 3 or ø 3 cm
Medium energy X-rays	2 mm Al < HVL ≤ 3 mm Cu	5	Centre of cylindrical chamber	"	10 × 10
Cs-137 gamma rays		5	0.35r ^b	"	10 × 10
Co-60 gamma rays		5	0.5r ^b	"	10 × 10
High energy X-rays	TPR ₁₀ ²⁰ ≤ 0.70	5	0.75r ^b	"	10 × 10
	TPR ₁₀ ²⁰ > 0.70	10	0.75r ^b	"	10 × 10
Electrons	$\bar{E}_0/\text{MeV} < 5$	R ₁₀₀	0.5r ^b	"	10 × 10 ^c
	5 ≤ $\bar{E}_0/\text{MeV} < 10$	R ₁₀₀ or 1 ^d	0.5r ^b	"	10 × 10 ^c
	10 ≤ $\bar{E}_0/\text{MeV} < 20$	R ₁₀₀ or 2 ^d	0.5r ^b	"	10 × 10 ^c
	20 ≤ $\bar{E}_0/\text{MeV} < 50$	R ₁₀₀ or 3 ^d	0.5r ^b	"	15 × 15 ^c

^a Depth of P_{eff}.^b The distance $z_{p,eff} - z_p$ is given, see Fig. 11.^c Field sizes for energy range measurements are different and discussed in Section 4.1.1.^d The larger depth should always be chosen.

a relative humidity of 50% then in a range of 20% to 70% relative humidity no correction is needed for temperatures ranging between 15°C and 25°C. If, on the other hand, the calibration factor is related to dry air and measurements are taken between 20% and 70% relative humidity and in a temperature range of 15°C to 25°C, a correction factor $k_h = 0.997$ is adequate for ⁶⁰Co gamma radiation [69].

Recombination. The incomplete efficiency in collecting charge in the cavity volume due to ion recombination requires the use of a correction factor. The effect depends on the geometry of the chamber, the applied collection voltage, and the rate of charge production by the radiation. In the case of pulsed radiation, especially with scanned beams, the correction factor for recombination becomes rather important. On the other hand, for continuous radiation (gamma ray beams) the effect is usually very small.

It is possible to derive a correction factor P_3 theoretically, using the theory of Boag [70-73] but as the magnitude of the correction depends on the position of the central electrode and this could accidentally be changed without an external change in the chamber [9], it is recommended here that an experimental procedure be used to determine the recombination factor.

The so called 'two-voltage' method [71] is, owing to its simplicity, the method to be used. It is based on two measurements of the collected charge, Q_1 and Q_2 , using different voltages V_1 (the normal operating bias voltage) and V_2 for the same irradiation conditions. The ratio V_1/V_2 should be equal to or greater than 3. Many of the commercially available ionization chamber instruments provide multiple bias voltages while some others can easily be modified to do so.

The recombination correction factor P_3 at the normal operating bias voltage V_1 can then be obtained from a quadratic equation that has been fitted to the numerical solutions of the expressions for pulsed and pulsed scanned radiation

$$P_3 = a_0 + a_1 \left(\frac{Q_1}{Q_2} \right) + a_2 \left(\frac{Q_1}{Q_2} \right)^2 \quad (17)$$

where constants a_i are given in Tables VIII and IX [74]. For continuous radiation P_3 is given in Fig. 13 [9].

7.2. DETERMINATION OF ABSORBED DOSE UNDER REFERENCE CONDITIONS

The irradiation geometry is set up as shown in Fig. 1. Geometrical parameters should be checked according to Section 3.3. Plastic phantoms are generally used for electron radiation with $\bar{E}_0 \leq 5$ MeV (see Section 4.1.1) and also for low energy X-rays (see Section 6.2.2). A water phantom is the best choice for other qualities

TABLE VIII. QUADRATIC FIT COEFFICIENTS FOR PULSED RADIATION AS A FUNCTION OF THE VOLTAGE RATIO V_1/V_2

These coefficients are used to calculate P_s by the two-voltage technique using $P_s = a_0 + a_1(Q_1/Q_2) + a_2(Q_1/Q_2)^2$. (From Weinhaus and Meli [74])

Voltage ratio	a_0	a_1	a_2
2.0	2.337	-3.636	2.299
2.5	1.474	-1.587	1.114
3.0	1.198	-0.8753	0.6773
3.5	1.080	-0.5421	0.4627
4.0	1.022	-0.3632	0.3413
5.0	0.9745	-0.1875	0.2135
6.0	0.9584	-0.1075	0.1495
8.0	0.9502	-0.03732	0.08750
10.0	0.9516	-0.01041	0.05909

TABLE IX. QUADRATIC FIT COEFFICIENTS FOR PULSED SCANNED RADIATION AS A FUNCTION OF VOLTAGE RATIO V_1/V_2

These coefficients are used to calculate P_s by the two-voltage technique using $P_s = a_0 + a_1(Q_1/Q_2) + a_2(Q_1/Q_2)^2$. (From Weinhaus and Meli [74])

Voltage ratio	a_0	a_1	a_2
2.0	4.711	-8.242	4.533
2.5	2.719	-3.977	2.261
3.0	2.001	-2.402	1.404
3.5	1.665	-1.647	0.9841
4.0	1.468	-1.200	0.7340
5.0	1.279	-0.7500	0.4741
6.0	1.177	-0.5081	0.3342
8.0	1.089	-0.2890	0.2020
10.0	1.052	-0.1896	0.1398

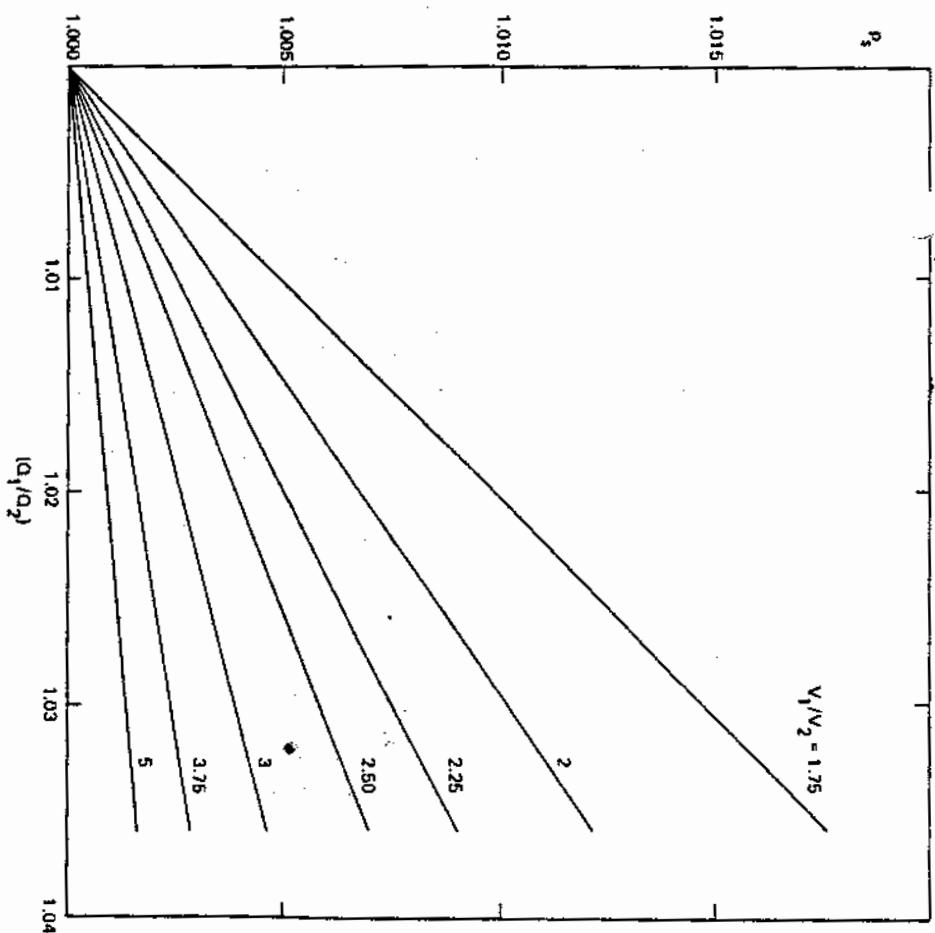


FIG. 13. The correction factor for recombination P_s in continuous radiation beams (from SEFM [9]). The 'two-voltage' method is used to determine the ratio of collected charges (or electrometer readings) Q_1/Q_2 (see text).

although for electrons with energies below 10 MeV, plastics may be used (see Section 3.2). The ionization chamber is irradiated without a buildup cap inside the water phantom. A thin, well fitted, PMMA tube (wall thickness about 0.5 mm) may however be used to protect the chamber.

The chamber must have a known absorbed dose to air chamber factor $N_{D,c}$. (As already pointed out, this report assumes no energy dependence for W (see Section 6.1, Eq. (7)) and consequently no subscript will be used for N_D):

$$N_D = N_K (1-g) k_{at} k_m$$

TABLE X. (cont.)

Depth in water (cm)	Electron beam energy $\bar{E}_0$ (MeV)										
	R_p	10.0 5.02	9.0 4.52	8.0 4.02	7.0 3.52	6.0 3.02	5.0 2.52	4.0 2.02	3.0 1.51	2.0 1.01	1.0 0.505
0.0		0.997	1.003	1.011	1.019	1.029	1.040	1.059	1.078	1.097	1.116
0.1		0.998	1.005	1.012	1.020	1.030	1.042	1.061	1.081	1.101	1.124
0.2		0.999	1.006	1.013	1.022	1.032	1.044	1.064	1.084	1.106	1.131
0.3		1.000	1.007	1.015	1.024	1.034	1.046	1.067	1.089	1.112	1.135
0.4		1.002	1.009	1.017	1.026	1.036	1.050	1.071	1.093	1.117	1.136
0.5		1.003	1.010	1.019	1.028	1.039	1.054	1.076	1.098	1.122	
0.6		1.005	1.012	1.021	1.031	1.043	1.058	1.080	1.103	1.126	
0.8		1.009	1.016	1.026	1.037	1.050	1.067	1.090	1.113	1.133	
1.0		1.013	1.021	1.031	1.043	1.058	1.075	1.099	1.121		
1.2		1.017	1.026	1.037	1.050	1.066	1.085	1.108	1.129		
1.4		1.022	1.032	1.044	1.058	1.075	1.095	1.117	1.133		
1.6		1.027	1.038	1.050	1.066	1.084	1.104	1.124			
1.8		1.032	1.044	1.057	1.074	1.093	1.112	1.130			
2.0		1.038	1.050	1.065	1.082	1.101	1.120	1.133			
2.5		1.053	1.067	1.083	1.102	1.120	1.131				
3.0		1.069	1.084	1.102	1.119	1.129					
3.5		1.085	1.102	1.118	1.128						
4.0		1.101	1.116	1.126							
4.5		1.115	1.125	1.127							
5.0		1.123	1.126								
5.5		1.125									

where $g = 0.0$. Values of k_{at} can be found in Table XVIII (in Section 8) for commercial ionization chambers. Values of k_m in this table do not include additional corrections for those chambers having the central electrode of a material different from the wall, but the lack of data for all the commercially available chambers makes this correction uncertain. Corrections for 'Farmer-type' chambers are small (about 0.8%), see Section 8. It is also possible to experimentally determine N_D according to the procedure given by NACP [7].

The absorbed dose should be determined using the reference conditions shown in Table VII. Corrections to the meter reading M_u should be carried out as described in Section 7.1 (e.g. for recombination, polarity, temperature, and pressure).

7.2.1. High energy electrons

The absorbed dose to water D_w at the effective point of measurement is given by

$$D_w (P_{eff}) = M_u \cdot N_D (s_{w,air})_u P_u \quad (\text{see Eq. (9)})$$

Stopping-power ratio ($s_{w,air})_u$. The Spencer-Attix cavity theory [75] is used. The cut-off energy Δ used for the restricted stopping powers was chosen to be 10 keV. (Strictly, the energy Δ should be equal to the energy of an electron whose range in air is equal to the mean chord length across the cavity. The exact value chosen for Δ is not very critical.) Stopping-power ratios are from Ref. [8], i.e. Monte Carlo calculations of electron spectra from Berger [76]. Data to be used are listed in Table X. Input parameters for ($s_{w,air})_u$ are the mean energy at the phantom surface $\bar{E}_0$ and the depth to the effective point of measurement z ; $\bar{E}_0$ is to be determined from R_{50} , see Section 4.1.

The perturbation correction factor P_u . Values from Johansson et al. [54] are given in Table XI. The mean electron energy at the effective depth of measurement $\bar{E}_z$ is needed as an input parameter for P_u . No correction for the central electrode is included in P_u . This correction is generally not well known but is probably small, see Section 8. For the determination of $\bar{E}_z$, see Section 4.1 and Table V.

Effective point of measurement P_{eff} . The distance between P_{eff} and chamber centre is equal to 0.5r (i.e. $z_{P_{eff}} - z_p = 0.5r$) for a cylindrical chamber where r is the internal chamber radius (see Fig. 11); the front surface of the air cavity is to be used for a plane parallel chamber.

When measurements are performed in plastic phantoms (alternative for $\bar{E}_0 \leq 10$ MeV) depths in plastic and water have to be related according to

$$\frac{z_w}{z_{pl}} = \frac{(r_0/\rho)_w}{(r_0/\rho)_{pl}} \quad (19)$$

TABLE XI. PERTURBATION CORRECTION FACTORS FOR ELECTRON BEAMS

Values of P_u for cylindrical thin-plate ionization chambers of inner radius r and a cavity length of 15 mm for different mean energies, $\bar{E}_e$ at the reference depths inside a water phantom (Johansson et al. [54])

$\bar{E}_e$ (MeV)	$r = 1.5$ mm	$r = 2.5$ mm	$r = 3.5$ mm
4	0.981	0.967	0.955
6	0.984	0.974	0.963
8	0.988	0.980	0.971
10	0.991	0.984	0.978
12	0.993	0.988	0.984
15	0.995	0.992	0.989
20	0.997	0.995	0.994

where r_0 is the continuous slowing down range of electrons having energy $\bar{E}_0$ (see Table III). An additional factor is required to transfer the electron fluence between plastic and water. The electrometer reading in plastic is then converted to an equivalent reading in water (see Ref. [7]) by means of

$$M_u(\text{water}) = M_u(\text{plastic}) h_m \quad (20)$$

where h_m is given in Table XII for some common plastic phantoms.

7.2.2. High energy photons

The absorbed dose to water, D_w , at the effective point of measurement is given by

$$D_w(P_{\text{eff}}) = M_u N_D (s_{w,\text{air}})_u P_u \quad (\text{see Eq. (9)})$$

which is the same as the equation that was used for the absorbed dose determination for high energy electrons (see Sections 6 and 7.2.1). For the meanings of the symbols, see Section 7.2.1.

Stopping-power ratio ($s_{w,\text{air}})_u$. The Spencer-Attix cavity theory is used with the cut-off energy for restricted stopping powers $\Delta = 10$ keV. Stopping-power data are taken from Berger and Seltzer [80]. Monte Carlo calculations by Andreo and

TABLE XII. VALUES OF h_m

The factor h_m is defined as the ratio of the signal measured using a plane parallel chamber at the ionization maximum on the central axis in a water phantom to that measured (at the same source to phantom surface distance) for the same accelerator monitor setting at the ionization maximum in other phantom materials. It is independent of $\bar{E}_0$ (between 2 and 10 MeV) within the limits of the stated uncertainty (see Mattsson et al. [67]; NACP [7]; Bruinvis et al. [79])

Phantom material	h_m
PMMA	1.000 ± 0.003
'White' polystyrene	1.006 ± 0.004
'Clear' polystyrene	1.016 ± 0.006
A-150	1.006 ± 0.003

Brahme [81] were used for generating electron energy spectra and determining average stopping power values. Data to be used at the reference depths are given in Table XIII. The parameters for specifying beam quality are TPR₁₀²⁰ and/or D₂₀/D₁₀ (see Section 4.2.2).

The perturbation correction factor P_u .⁷ Values from Andreo et al. [82] are given in Fig. 14. These are valid for a chamber wall thickness of about 0.5 mm. The P_u values for other wall thicknesses or composite materials have to be calculated following Section 8 if good accuracy is to be obtained. However, it can be seen from Fig. 14 that P_u for most wall materials is very close to unity. The error introduced by using the values from this figure also for walls of a slightly different thickness is therefore generally small but should be estimated by the user. No correction for the central electrode is included in P_u . This correction is generally not well known but is probably small (see Section 8).

The effective point of measurement P_{eff} . The following values are recommended for the reference depth:

¹³⁷ Cs gamma radiation	$Z_{p,\text{eff}} - Z_p = 0.35r$
⁶⁰ Co gamma radiation	$Z_{p,\text{eff}} - Z_p = 0.5r$
High energy X-rays	$Z_{p,\text{eff}} - Z_p = 0.75r$

⁷ It should be observed that P_u is defined in connection with the use of an effective point of measurement. This is different from the procedure recommended in some other protocols, e.g. in the Nordic protocol [6], AAPM protocol [8] or SEFM protocol [9].

TABLE XIII. THE STOPPING POWER RATIO WATER TO AIR ($S_{w,air}$) AT THE REFERENCE DEPTHS AS A FUNCTION OF THE PHOTON BEAM QUALITY (Andreo et al. [82])

Beam quality TPR_{10}^{20}	D_{20}/D_{10}^*	$S_{w,air}$	Ref. depth (cm)
0.50	0.44	1.135	5
0.53	0.47	1.134	5
0.56	0.49	1.132	5
0.59	0.52	1.130	5
0.62	0.54	1.127	5
0.65	0.56	1.123	5
0.68	0.58	1.119	5
0.70	0.60	1.116	5
0.72	0.61	1.111	10
0.74	0.63	1.105	10
0.76	0.65	1.099	10
0.78	0.66	1.090	10
0.80	0.68	1.080	10
0.82	0.69	1.069	10
0.84	0.71	1.059	10
Cs-137		1.136	5
Co-60		1.133	5

* At SSD = 100 cm: obtained from TPR_{10}^{20} by a fit to experimental data.

7.2.3. Medium energy X-rays: 100 to 300 kV

In the X-ray energy region below 300 kV tube potential the centre of a thimble-type ionization chamber is considered to be the effective point of measurement. This point is placed at the reference depth in a water phantom. As derived in Section 6.2.1, the absorbed dose to water is given by the equation

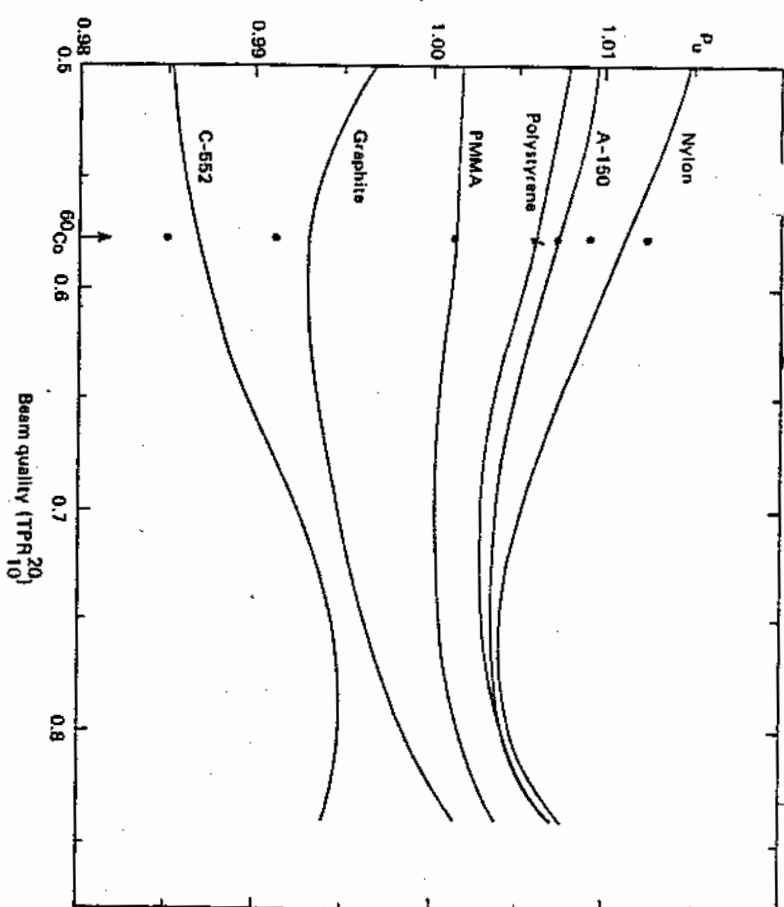


FIG. 14. The perturbation factor P_u as a function of the quality of photon beams for different chamber wall materials. It is assumed that the chamber is of thimble type having a wall thickness equal to 0.5 mm. (From Andreo et al. [82])

$$D_w = M_u N_k k_u (\bar{\mu}_{en}/\rho)_{w,air} P_u \quad (\text{see Eq. (13)})$$

Here, N_k , the air kerma calibration factor, is to be taken for the radiation quality (expressed in half-value thickness) of the primary beam (i.e. not influenced by the penetration into the phantom) and is obtained for the free in air calibrations at the SSD; k_u is a correction factor taking into account the change in response due to the change of the spectral distribution of the beam penetrating into the phantom to the reference depth. This factor should be close to 1 for the chambers recommended in this report (see Sections 3.1.1 and 6.2.1). (The influence of different beam qualities should be investigated by making calibrations for at least two beam qualities.) The term $(\bar{\mu}_{en}/\rho)_{w,air}$ is the ratio of the mass energy absorption coefficients of water and air averaged over the spectral energy fluence in the phantom at

TABLE XIV. ENERGY ABSORPTION COEFFICIENT RATIO FOR X-RAYS AS A FUNCTION OF THE HALF-VALUE THICKNESS FOR A CIRCULAR FIELD WITH A DIAMETER OF 11.3 cm; FSD IS 100 cm (Grosswendt [83]).

Tube potential (kV)	Beam quality		$(\bar{\mu}_{en}/\rho)_{w,air}$		
	Half value thickness in mm		$z = 0 \text{ cm}^*$	$z = 2 \text{ cm}$	$z = 5 \text{ cm}$
50	2.24	0.072	1.013	1.012	1.012
70	2.94	0.09	1.017	1.017	1.017
100	4.28	0.17	1.027	1.027	1.028
120		0.30	1.036	1.036	1.036
140		0.49	1.047	1.045	1.044
150		0.83	1.061	1.058	1.057
200		1.60	1.080	1.075	1.073
250		2.47	1.092	1.089	1.085
280		3.37	1.100	1.097	1.094

* Depth in water.

the reference depth. Values for this conversion factor are given in Table XIV as a function of the X-ray radiation quality. The perturbation correction factor P_u correcting the replacement of water by an air volume in the shape of the ionization chamber (see Fig. 12, P^u) can be found in Table XV. Values are available only for a typical chamber design.

7.2.4. Low energy X-rays: 10 to 100 kV

If calibration with the chamber placed in the phantom has been made directly in terms of absorbed dose to water per meter reading (see Section 6.2.2, method (a)) then only corrections for different influence quantities (see Section 7.1) need to be carried out. The absorbed dose to water is given by

$$D_w = M_u N_{D,w}$$

(see Eq. (14))

TABLE XV. PERTURBATION CORRECTION FACTOR P_u FOR THIMBLE IONIZATION CHAMBERS FOR X-RAYS AT 5 cm DEPTH INSIDE A WATER PHANTOM

The X-ray qualities are expressed as Al or Cu half-value layers (Schneider [65]). The values are valid for ionization chambers with volumes between 0.3 cm³ and 1 cm³, outer diameters between 5 and 9 mm and wall thicknesses of about 0.5 mm. The uncertainty of the values is estimated to be 2% (1 σ).

Tube potential (kV)	HVL in		Perturbation correction factor P_u
	mm Al	mm Cu	
100	4.28	0.17	1.10
120	6.31	0.30	1.09
140		0.49	1.08
150		0.83	1.06
200		1.70	1.04
250		2.47	1.02
280		3.37	1.01

If the calibration has been made free in air the absorbed dose is calculated from

$$D_w = M_u N_k B k_u (\bar{\mu}_{en}/\rho)_{w,air} \quad (\text{see Eq. (15)})$$

where the values for B and $(\bar{\mu}_{en}/\rho)_{w,air}$ should be taken from Table XVI and Table XIV, respectively. Corrections for different influence quantities (see Section 7.1) must be carried out.

7.3. DETERMINATION OF ABSORBED DOSE UNDER CONDITIONS OTHER THAN REFERENCE CONDITIONS

Once the absorbed dose to water has been determined under the reference conditions, it is possible to determine the absorbed dose for any other conditions by means of relative distributions of absorbed dose (percentage depth doses or isodose curves). Such distributions should be obtained for all the possible combinations of energy, field size, SSD, etc., used for radiotherapy treatments.

Owing to the intrinsic differences in radiation generators, as a general rule, the above mentioned distributions should be obtained experimentally for each treatment

TABLE XVI. BACKSCATTER FACTOR B FOR LOW ENERGY X-RAYS AS A FUNCTION OF THE HALF-VALUE LAYER FOR DIFFERENT FIELD SIZES

The backscatter factor B is defined as the ratio of the air kerma at the surface of a semi-infinite water phantom to the air kerma at that point in the absence of the phantom. The values can be applied to moderately filtered spectra. The values are derived from Grosswendt [84]

Beam quality Aluminium HVL (mm)	Field diameter (mm)			
	3 cm	5 cm	10 cm	20 cm
0.1	1.007	1.007	1.007	1.007
0.15	1.012	1.013	1.013	1.013
0.2	1.021	1.023	1.023	1.023
0.3	1.036	1.040	1.041	1.041
0.4	1.048	1.054	1.060	1.061
0.5	1.061	1.071	1.081	1.086
0.6	1.071	1.085	1.100	1.110
0.8	1.088	1.109	1.133	1.148
1.0	1.099	1.126	1.155	1.176
1.5	1.122	1.164	1.204	1.245
2.0	1.138	1.190	1.241	1.296
3.0	1.157	1.219	1.291	1.363
4.0	1.169	1.230	1.330	1.412

unit. A lack of adequate instrumentation could justify the use of dose distributions published either in the open literature or by the manufacturer, provided that the relative dose at several points on the beam axis and off-axis is experimentally checked.

7.3.1. High energy electrons

In the case of electron beams, relative absorbed dose distributions are different from relative ionization distributions because of the continuous decrease in energy and increased scattering with depth of the electrons. Both the stopping power ratio

and the perturbation factor vary significantly with energy and the ratio D_w/M_u cannot be considered to be constant with depth. Consequently, it will be necessary to determine the absorbed dose at each point of interest (where the effective point of measurement is located) by repeating the procedure described in Section 7.2.1. Water/air stopping power ratios and perturbation correction factors are given in Tables X and XI. Because the energy and angular spread at the surface are not negligible in some accelerators, the use of $\bar{E}_0$ and the depth of interest z to select stopping power ratios can introduce large uncertainties at depths beyond a few centimetres. It is then possible to use $\bar{E}_0$ and a scaled depth

$$z' = z R_p^{\text{calc}}/R_p^{\text{meas}} \quad (21)$$

where R_p^{calc} is the practical range for monoenergetic electrons of energy $\bar{E}_0$ given in Table X and R_p^{meas} is the experimentally determined practical range for the actual conditions. (The value of the electron energy $\bar{E}_0$ is here equal to the value of the mean energy $\bar{E}_0$ of the actual beam.)

The perturbation correction factor at each depth can be estimated by using data given in Table XI after the mean energy at the location of the effective point of measurement has been determined (see Table IX). (The data in Table XI were experimentally determined at the depth for dose maximum. However, Johansson and Svensson [29] showed that they can be used at least for depths corresponding to 0.2 to $0.5 R_p$.)

Measurements with ionization chambers using different correction factors are rather laborious, especially when iso-absorbed dose distributions have to be obtained. Film dosimetry in a solid plastic phantom is often preferred for the determination of isodose distributions because it is rapid and it presents the great advantage of a high spatial resolution. In this case, depths in plastic have to be scaled to depths in water according to Eq. (19).

For improved accuracy, the dose distribution along the beam axis should be measured with an ionization chamber and used to assign a depth absorbed dose value to the points on the central axis in the film.

Other types of detectors may also be used, such as TLD or silicon diodes. All of them require certain precautions and systematic errors can be introduced if they are not handled in a proper way. It is therefore strongly recommended that some of the measured distributions be checked against ionization chamber measurements; otherwise large errors may be introduced.

7.3.2. X-rays and high energy photons

With medium energy X-rays the spectral and angular distribution of the radiation changes with depth and field size [85]. The response of ionization chambers and the factors to be applied for the determination of absorbed dose will then be different

for conditions other than those used as reference. There are no comparable sets of data available, but $(\mu_{en}/\rho)_{w,air}$ and P_u [60] are expected to vary within a few per cent. Values of the $(\mu_{en}/\rho)_{w,air}$ ratio with respect to depth in the phantom are given in Table XIV. Additional information can be found in Ref. [63].

In the case of high energy photons the factors to be applied to the readings of the measurement system can be assumed to be independent of depth for a given quality. Only at very large depths does this assumption introduce small uncertainties and they can be disregarded for practical dosimetry. This means that relative ionization distributions can be used as relative absorbed dose distributions and procedures are then simpler than for electron beams.

As was stated in the formalism (see Fig. 12), the effective point of measurement of the ionization chamber has to be situated at the depth of interest within the phantom.

Film dosimetry can introduce large uncertainties, especially in the penumbra region, owing to the enhanced sensitivity of the film to low energy photons [29]. The photographic method should not be used to determine absorbed dose profiles, but only for certain relative checks or comparisons of the flatness and uniformity of photon beams.

Similar recommendations to those given for electron beams can be considered here regarding alternative measuring systems. It should be noticed that for ^{60}Co gamma radiation and X-ray beams below about 10 MV large errors have been reported with some semiconductor detectors [29]. However, different types of diodes may have different characteristics regarding quality dependence.

8. DETAILS ON CORRECTION FACTORS

The formalism has been discussed in Section 6 and values of interaction coefficients have been given in Section 7. It is the purpose in this section to give more detailed information to enable, if required, the determination of values of k_m , k_{air} and P_u for various types of ionization chambers.

Value of k_m

The relationship

$$k_m = S_{air,m} (\mu_{en}/\rho)_{m,air} \quad (22)$$

can be used in those cases where the wall material and the buildup cap are made of the same material m . Values are given in Table XVII. If instead the chamber wall

TABLE XVII. VALUES OF THE FACTOR k_m ($= S_{air,m} (\mu_{en}/\rho)_{m,air}$)
This factor takes into account the non-air equivalence of the chamber wall and cap material at the calibration in ^{60}Co gamma ray beams

Material	$S_{air,m}$ (a)	$(\mu_{en}/\rho)_{m,air}$ (b)	k_m
A-150 (T.E. plastic)	0.876	1.101	0.965
C-552 (A.E. plastic)	1.005	1.001	1.006
Delrin (CH_2O) _n	0.926	1.058	0.989
Graphite ($\rho = 1.7 \text{ g/cm}^3$)	0.998	1.001	0.999
Graphite ($\rho = 2.265 \text{ g/cm}^3$)	1.000	1.001	1.001
Nylon 66 ($\text{C}_6\text{H}_{11}\text{ON}$) _n	0.875	1.098	0.961
PMMA (perspex, lucite) ($\text{C}_5\text{H}_8\text{O}_2$) _n	0.908	1.081	0.982
Polystyrene (C_8H_8) _n	0.901	1.078	0.971
Tufnol	—	—	0.979 ^c

^a From Andreo and Brahme [81].

^b From Cunningham [86].

^c From Henry [87].

(or inner part of the wall) and the buildup cap (and in some cases in addition part of the chamber wall) are made of different materials then k_m can be calculated from:

$$k_m = \alpha S_{air,wall} (\mu_{en}/\rho)_{wall,air} + (1 - \alpha) S_{air,cap} (\mu_{en}/\rho)_{cap,air} \quad (23)$$

where α is the fraction of ionization inside the air cavity due to electrons from the chamber wall, and $(1 - \alpha)$ is the fraction of ionization due to electrons generated in the buildup cap (from Almond and Svensson [88]). Values of α are given in Figs 15 and 16. Table XVIII includes k_m values for a large number of commercially available chambers [82].

Effect of the central electrode

Equation (23) only takes account of the non-air equivalence of the chamber wall and cap materials and therefore does not include the non-air equivalence of the central electrode material. The correction for this effect is very small for electrodes of graphite or plastics and can be disregarded. However, central electrodes are often

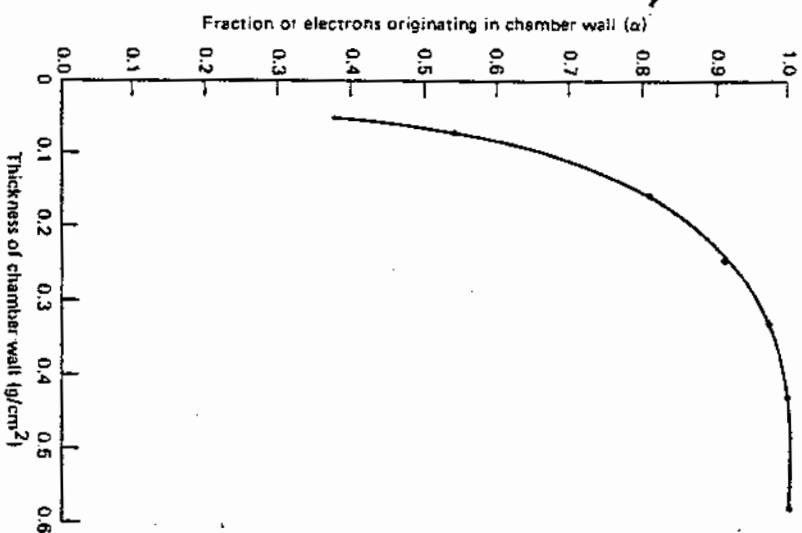


FIG. 15. The fraction α of ionization due to electrons arising in the chamber wall as a function of wall thickness for ^{60}Co gamma rays. (From Lempert et al. [89])

made of aluminium. In a ^{60}Co gamma beam the response of the ionization chamber increases significantly compared to an ionization chamber with 'air equivalent' central electrode and k_m should be corrected for this effect. For example, an aluminium electrode with a diameter of 1 mm inside a chamber of the 'Farmer size' (i.e. inner length of 25 mm and inner radius of 3.15 mm) would increase the response at ^{60}Co gamma rays by 0.8% (see Refs [50, 91, 92]).

The increase in response will, however, also occur for the absorbed dose measurement at the user's radiation quality and a corresponding correction to P_a will be needed. The corrections to k_m and P_a will cancel out at the ^{60}Co quality. For higher photon energies the correction to P_a due to the use of an aluminium electrode decreases when the photon energy increases and the corrections to k_m and P_a will not fully cancel out. For electron radiation the correction to P_a is negligible and the correction carried out in k_m results in a maximum deviation (see Ref. [92]).

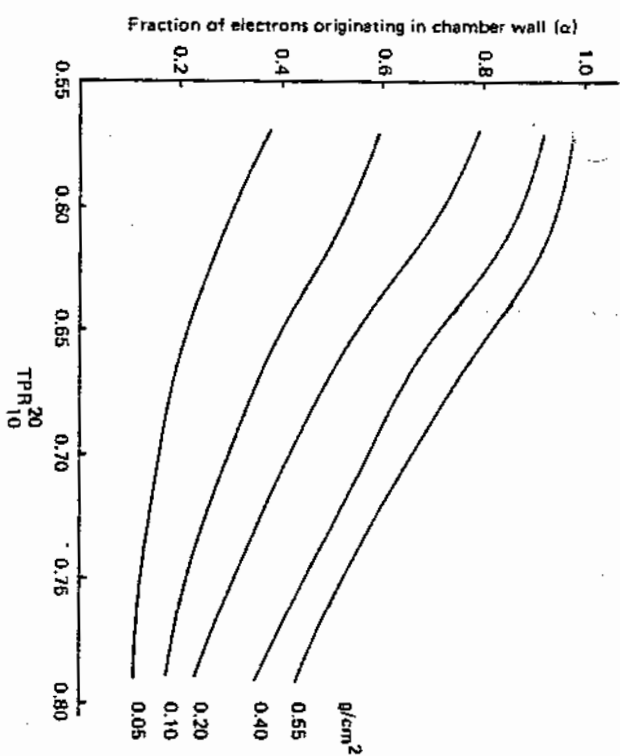


FIG. 16. The fraction α of ionization due to electrons arising in the chamber wall as a function of wall thickness for high energy X-rays. The quality of the photon beam is given by TPR_{10}^{20} (Data taken from Lempert et al. [89])

TABLE XVIII. VALUES OF k_m AND $k_{m, \text{air}}$ AND THE PRODUCT $k_m \cdot k_{m, \text{air}}$ FOR THE IONIZATION CHAMBERS IN TABLE II

The values of k_m were computed by Andreo et al. [82] using Eq. (23).

Corrections for the non-air equivalence of the central electrode are not included; $k_{m, \text{air}}$ was computed from data in Nath and Schultz [49]

	k_m	$k_{m, \text{air}}$	$k_m \cdot k_{m, \text{air}}$
CAPINTEC 0.07 cm ³ PR-05P minichamber	1.002	0.988	0.990
CAPINTEC 0.14 cm ³ PR-05 minichamber	1.002	0.989	0.991
CAPINTEC 0.65 cm ³ PR-06C Farmer type (AB cap)	1.006	0.984	0.990
CAPINTEC 0.65 cm ³ PR-06C Farmer type (polystyrene cap)	0.987	0.990	0.977
CAPINTEC 0.65 cm ³ PR-06C Farmer type (PMMA cap)	0.993	0.990	0.983
CAPINTEC 0.60 cm ³ (AAPM)	0.989	0.989	0.978
EXRADIN 0.5 cm ³ Al (2 mm cap)	1.006	0.985	0.991
EXRADIN 0.5 cm ³ Al (4 mm cap)	1.006	0.976	0.981

TABLE XVIII. (cont.)

	k_m	k_{at}	$k_m \cdot k_{at}$
EXRADIN 0.5 cm ³ T2	0.965	0.985	0.950
EXRADIN 0.05 cm ³ T1 min Shonka	0.965	0.991	0.957
FAR WEST TECH 0.1 cm ³ IC-18	0.965	0.991	0.956
FZB, 0.4 cm ³ TK 01 waterproof	0.989	0.989 ^c	0.978
NE 0.20 cm ³ 2515	0.980	0.988	0.968
NE 0.20 cm ³ 2515/3	0.991	0.987	0.978
NE 0.20 cm ³ 2577	0.994	0.987	0.982
NE 0.60 cm ³ Farmer 2505 '54-'59 ^a	0.980	0.992	0.973
NE 0.60 cm ³ Farmer 2505 '59-'67 ^a	0.980	0.990	0.971
NE 0.60 cm ³ Farmer 2505/A '67-'74 ^a	0.971	0.991	0.962
NE 0.60 cm ³ Farmer 2505/3, 3A '71-'79 ^a	0.991	0.990	0.981
NE 0.60 cm ³ Farmer 2505/3, 3B '74-present ^a	0.974	0.991	0.965
NE 0.60 cm ³ Guarded Farmer 2571	0.994	0.990	0.985
NE 0.60 cm ³ Robust Farmer 2581 (PMMA cap)	0.975	0.990	0.966
NE 0.60 cm ³ Robust Farmer 2581 (polystyrene cap)	0.969	0.990	0.959
NE 0.325 cm ³ NPL Sec Sid 2561	0.995	0.984	0.979
PTW 0.6 cm ³ 2333 (3 mm cap)	0.982	0.993	0.975
PTW 0.6 cm ³ 2333 (4.6 mm cap)	0.982	0.990	0.972
PTW 0.4 cm ³ 23331	0.982	0.990	0.971
PTW 0.3 cm ³ Normal M233332	0.982	0.993	0.975
PTW 0.1 cm ³ Transit M233331	0.982	0.992	0.974
PTW 0.3 cm ³ Waterpr M2333641	0.982	0.992	0.974
VICTOREEN 0.1 cm ³ Radcon II 555	0.989	0.990	0.979
VICTOREEN 0.3 cm ³ Radcon III 550	0.974	0.991	0.965
VICTOREEN 0.30 cm ³ 30-348	0.982	0.993	0.975
VICTOREEN 0.60 cm ³ 30-351	0.982	0.993	0.975
VICTOREEN 1.00 cm ³ 30-349	0.982	0.992	0.974
SSI GRAPHITE	0.999	0.990 ^b	0.989
SSI A-150	0.965	0.990 ^b	0.955

^a Year of manufacture.^b Experimental, Johansson et al. [54].^c Experimental, Dufschmid [90].

In order to simplify the use of the central electrode correction, the use of a global correction factor P_{cel} is recommended to account for the non-air equivalence both at the calibration quality and the user's radiation quality instead of performing corrections to both k_m and P_u . For 'Farmer type' chambers having an aluminium central electrode values of P_{cel} are given in Table XIX for different radiation beams

TABLE XIX. CORRECTION FACTOR P_{cel}
To be used if the chamber has a central electrode of aluminium (data for a Farmer type of chamber (Rogers et al. [50], Mattson and Johansson [51] and Mattson [92]))

Electrode radius (mm)	Electrons	Photons ($(h\nu)_{max} > 25$ MeV)	⁶⁰ Co and photons ($(h\nu)_{max} \leq 25$ MeV)
0.5	1.008	1.004	1.000
1	1.015	1.008	1.000
1.5	1.020	1.010	1.000
2.5	1.032	1.016	1.000

as a function of the radius of the central electrode. The equation to determine absorbed dose to water will then be:

$$D_w (P_{el}) = M_u N_D (S_{w,air})_u P_u P_{cel} \quad (24)$$

Value of k_{at}

The values of k_{at} for different types of chambers have been computed using the Monte Carlo method by Nath and Schulz [49] and Rogers et al. [50]. Values for a large number of chambers are given in Table XVIII. It can be seen that $k_{at} = 0.990 \pm 0.005$ for cylindrical, thin type, chambers with the dimensions recommended in Section 3, having a total thickness of chamber wall together with buildup cap (made of graphite or plastics) between 0.45 and 0.6 g·cm⁻². The product $k_m k_{at}$ from Table XVIII can be used in Eqs (5) and (6) to determine N_D for an ionization chamber calibrated in air kerma (see also Eq. (d) in Fig. 10).

The perturbation correction factor p_u

For electron radiation the perturbation correction factor takes into account the difference in scattering in the phantom (and also chamber wall) material and in the air cavity. Data are given in Section 7.2.

For photon beams it must be considered that electrons are produced and stopped differently in the chamber wall material and in water. In the computation it is always assumed that the measurement in the water phantom is carried out

TABLE XX. VALUES OF $s_{\text{wall,air}}$ AS A FUNCTION OF PHOTON BEAM QUALITY, SPENCER-ATTIX THEORY ($\Delta = 10$ keV) (from Andreo et al. [82])

Beam quality TPR ₁₀ ²⁰ in water	D_{20}/D_{10} ^a	A-150	C-552	Delrin	Graphite	Nylon	PMMA	Polystyrene
0.50	0.44	1.147	0.997	1.083	1.008	1.148	1.105	1.114
0.53	0.47	1.145	0.997	1.082	1.007	1.146	1.104	1.113
0.56	0.49	1.142	0.995	1.080	1.003	1.143	1.102	1.110
0.59	0.52	1.138	0.992	1.077	1.000	1.139	1.099	1.107
0.62	0.54	1.135	0.990	1.074	0.996	1.135	1.096	1.104
0.65	0.56	1.130	0.986	1.070	0.992	1.130	1.091	1.100
0.68	0.58	1.124	0.983	1.065	0.987	1.125	1.087	1.095
0.70	0.60	1.121	0.980	1.062	0.984	1.121	1.084	1.092
0.72	0.61	1.115	0.976	1.057	0.979	1.115	1.079	1.087
0.74	0.63	1.108	0.970	1.051	0.973	1.108	1.073	1.080
0.76	0.65	1.101	0.964	1.044	0.967	1.100	1.066	1.074
0.78	0.66	1.091	0.956	1.035	0.959	1.090	1.057	1.065
0.80	0.68	1.080	0.948	1.026	0.950	1.080	1.047	1.055
0.82	0.69	1.069	0.938	1.015	0.941	1.068	1.037	1.044
0.84	0.71	1.058	0.929	1.005	0.932	1.057	1.027	1.034
¹³⁷ Cs		1.149	0.999	1.085	1.011	1.150	1.107	1.116
⁶⁰ Co		1.142	0.995	1.080	1.002	1.142	1.102	1.110

^a At SSD = 100 cm. Obtained from TPR₁₀²⁰ by a fit to experimental data.

without a built-in cap. The perturbation correction effect p_a can then be approximately calculated from the equation by Almond and Svensson [88]:

$$p_a = \frac{\alpha s_{\text{wall,air}} (\bar{\mu}_{\text{en}}/\rho)_{w,\text{wall}} + (1-\alpha) s_{w,\text{air}}}{s_{w,\text{air}}} \quad (25)$$

where α is the fraction of the total ionization produced in the air in the cavity by electrons arising in the chamber wall. Values of α for different thicknesses of the wall as a function of beam quality are given in Fig. 16.

Stopping power ratios $s_{\text{wall,air}}$ for different chamber materials as a function of the beam quality to be used in Eq. (25) are given in Table XX; $s_{w,\text{air}}$ is given in Table XIII and mass energy absorption coefficients can be found in Table XXI.

9. THE UNCERTAINTY IN ABSORBED DOSE DETERMINATION AT THE REFERENCE POINT

9.1. GENERAL

The various steps necessary for establishing traceability in a measurement chain from the primary standard to the user's instrument have been discussed in Section 5. The practical realization of each step in this chain brings in a number of uncertainties. Frequently these are not very well known but often simple and reasonable estimates of their magnitude can be made. In this chapter the uncertainties in the various steps are evaluated and added according to the procedure described in Appendix A.

9.2. DETERMINATION OF UNCERTAINTIES

In the present report the total uncertainty in the calibration chain is evaluated from the determination of air kerma at the standards laboratory to the measurements of absorbed dose in the reference point at the user's beam at the hospital.

Loevinger and Loftus [93] have analysed a similar calibration chain. Svensson [37] adjusted the different steps in the chain to be in agreement with the calibration procedures used by NACP [6]. This procedure is very similar to that of the present document and is therefore followed. However, it has been noted that some of the data involved have been recently re-evaluated ($s_{m,1}$ and W/e). The

TABLE XXI. VALUES OF $(\mu_{en}/\rho)_{w,wall}$ AS A FUNCTION OF PHOTON BEAM QUALITY (from Cunningham [86])

Beam quality TPR ₁₀ ²⁰ in water	D ₂₀ /D ₁₀ ^a	A-150	C-552	Delrin	Graphite	Nylon	PMMA	Polystyrene
0.53	0.47	1.011	1.110	1.043	1.114	1.015	1.031	1.035
0.56	0.49	1.011	1.110	1.043	1.113	1.015	1.030	1.034
0.59	0.52	1.012	1.110	1.042	1.113	1.015	1.030	1.034
0.62	0.54	1.012	1.110	1.042	1.113	1.015	1.031	1.035
0.65	0.56	1.013	1.110	1.043	1.114	1.016	1.031	1.036
0.68	0.58	1.015	1.109	1.043	1.115	1.018	1.032	1.038
0.70	0.60	1.016	1.108	1.043	1.115	1.019	1.033	1.040
0.72	0.61	1.019	1.107	1.043	1.117	1.021	1.035	1.042
0.74	0.63	1.023	1.105	1.044	1.119	1.026	1.038	1.048
0.76	0.65	1.028	1.103	1.045	1.121	1.030	1.041	1.053
0.78	0.66	1.035	1.100	1.046	1.125	1.037	1.045	1.061
0.80	0.68	1.043	1.096	1.048	1.130	1.045	1.051	1.071
0.82	0.69	1.051	1.093	1.049	1.134	1.054	1.056	1.081
0.84	0.71	1.059	1.089	1.051	1.139	1.062	1.062	1.090
¹³⁷ Cs		1.010	1.111	1.042	1.111	1.013	1.029	1.032
⁶⁰ Co		1.011	1.110	1.042	1.113	1.015	1.030	1.034

^a At SSD = 100 cm. Obtained from TPR₁₀²⁰ by a fit to experimental data.

TABLE XXII. COMBINED UNCERTAINTIES IN THE DIFFERENT PHYSICAL QUANTITIES OR PROCEDURES
The values of the uncertainties correspond to one standard deviation.
Step 1 in the calibration procedure (see text)

Type of physical quantity or procedure	Uncertainty (%)
Interaction coefficients ($W/e \cdot s \cdot g^{-1} (R_{w,air}/\rho)_{air,c}$)	0.4
Measurement of air kerma rate with the standard chamber (belonging to PSDL)	0.3
Calibration of secondary instrument (belonging to SSDL)	0.3
Calibration of local reference instrument (belonging to the hospital)	0.3
Transfer of air kerma rate calibration to a reference point in the ⁶⁰ Co gamma beam (at the hospital)	0.5
Calibration of the field instrument (at the hospital)	0.5
Combined uncertainty	1.0

figures below are to be considered as guidelines. They will, however, give an indication of the accuracy that might be achieved by using a careful measurement technique and applying this protocol.

Step 1 in the measurement chain starts with the realization of air kerma at the standards laboratory in the ⁶⁰Co gamma beam and ends with the air kerma calibration of the field instrument to be used by the hospital (see Table XXII). In the example it is assumed that absorbed dose determinations are to be carried out in a ⁶⁰Co gamma beam, high energy photon beam, or electron beam at the hospital.

Step 2 is the absorbed dose determination at the reference point in a water phantom using the air kerma calibrated ionization chamber (see Table XXIII).

Combining the uncertainties in Steps 1 and 2 will give the overall uncertainty for the determination of the absorbed dose at the reference points. This will give 2.7% (i.e. $\sqrt{1.0^2 + 2.5^2}$) for ⁶⁰Co gamma radiation, 3.4% for high energy X-rays, and 3.8% for electron radiation.

TABLE XXIII. COMBINED UNCERTAINTIES IN THE DIFFERENT PHYSICAL QUANTITIES OR PROCEDURES

The values of the uncertainties correspond to one standard deviation. Step 2 in the calibration procedure

Type of physical quantity or procedure	Uncertainty (%)		
	⁶⁰ Co gamma rays	High energy X-rays	Electrons
Interaction coefficients (k_{co} , k_{air} , $S_{\text{w,air}}$, P_{air} ; also uncertainties due to input parameters are included, i.e. E_0 , TPR_{10}^{20})	2.4	2.6	3.2
Field instrument measurements in the therapy beam	0.5	1	1
Dose monitor of the therapy unit	0.5	1.5	1.5
Combined uncertainty	2.5	3.2	3.7

10. EVALUATION OF PERFORMANCE

10.1. QUALITY CONTROL

The traditional way to test the actual performance of radiotherapy procedures is by inspection. In its most formal form this has been by analysis of results of treatment. Now the procedure is more sophisticated and the term 'quality control' is used. Quality control is the process by which the actual performance is measured and compared with performance standards, and the actions necessary to maintain or regain conformance with the standard are identified.

10.2. QUALITY ASSURANCE

Quality assurance for radiation dosimetry consists of the formalization of those procedures that must be taken in order to enable accurate and consistent dose measurements to be carried out. Some examples only can be given.⁸

⁸ They will be discussed in an ICRU report (in preparation) on accuracy requirements and quality assurance of external beam therapy with photons and electrons.

(1) A 'local' reference dosimeter must be available. This dosimeter must have a calibration that is traceable to a Primary Standards Laboratory by no more than one exchange or have been calibrated by a Secondary Standard Dosimetry Laboratory (SSDL). It should be accompanied by a calibration certificate which is dated and includes calibration factors for one or more beam qualities. This should include a specification of linearity of response, electrical leakage, stem effects if any and presence of openings to the atmosphere.

(2) This 'local' reference dosimeter should, ideally, be reserved for this role. Under certain circumstances, particularly if an isotopic source is available for constancy check, it may be used for the relatively infrequent initial primary reference calibration of a machine, but another dosimeter, similar to the local reference, should be available. This second, or 'field dosimeter' may be less restrictive in design than the local standard and should be used for on-going calibration and checking of equipment.

(3) The local reference dosimeter should be calibrated against a standard at certain intervals. In some countries these are given by regulations. As an example two-year periods have often been chosen [6]. The field dosimeter should be checked more frequently against the local reference standard.

(4) It is necessary that the response of the local reference be checked before shipment to a standardization laboratory (SSDL, for example) and after its return to ensure that no change has occurred in its sensitivity as a result of handling and shipment.

(5) A record must be kept of all dosimeter tests and calibrations carried out.

(6) Certain general procedures should be followed for the care and maintenance of the dosimeters. These may include:

- Instructions given in the manufacturer's handbook should be followed.
- When not in use, the dosimeter should be kept in a protective and sealed carrying case. In moist environments this should include an active desiccant (except when chambers are used having hygroscopic wall materials, see Section 7.1 under 'humidity').
- The linearity of response between the instrument reading and a ⁶⁰Co gamma ray exposure should be within $\pm 1\%$. This linearity should be checked from time to time.
- The collection efficiency of the chamber should be assessed for all the conditions under which the chamber will be used. Stem effects and charge leakage should be assessed and be known and taken into account as needed. All ionization readings must be corrected for deviation of ambient temperature and pressure from those pertaining to the calibration. The rate of attaining pressure and temperature equilibrium should be known.

(7) In calibration measurements all parameters such as distances, depths and field sizes should be checked repeatedly and several readings should always be taken to confirm precision.

(8) When and where possible, calibrations of new equipment could be done independently by at least two different persons and preferably by different dosimeter instruments. The results should be compared only after the completion of both measurements. If they differ by more than 2% they should be repeated and the reasons for the difference should be investigated.

10.3. QUALITY AUDIT

Quality audit (assessment) is defined as an independent review of the quality assurance programme and the results achieved by it. The essential point is that it be performed by an independent person who is not responsible for the results achieved. In calibration procedures this can take the form of intercomparisons of the results of dose measurements e.g. through the TLD postal service of the IAEA.

When differences are found the reasons for them must be investigated as part of the general quality control programme.

Appendix A

EVALUATION OF UNCERTAINTIES

According to a widespread procedure, uncertainties in measurements are subdivided into two classes — denoted as random and systematic. This traditional view, however, has no objective basis [94]. Thus, since by 'systematic' uncertainty one usually means a contribution which cannot be reduced by taking more measurements in which this component is involved, this is a concept which depends on the way a certain result (with its uncertainty) is subsequently used. As a consequence, the type of an uncertainty can change (from random to systematic, or vice versa), depending on the ensuing data handling. This shows that the classification, although it may sometimes be practical, is not a very profound one.

It may be noted in passing that the concepts correction and uncertainty should not be confused; whenever there exists a bias and its sign and magnitude are known, it should be corrected for by the introduction of an appropriate correction. Once applied, the known bias is eliminated, but any insufficient knowledge of this shift will have to be included in the uncertainty of the corrected value. Modern usage is to distinguish between 'error' and 'uncertainty' in such a way that an error can (and therefore should) be corrected, whereas for an uncertainty this cannot be done since its sign is unknown. In practice, however, one should not rely too heavily on the exact use of the vocabulary as this is still somewhat fluctuating.

A.1. CLASSIFICATION OF DIFFERENT UNCERTAINTIES

A simple and stable classification of the various uncertainty components should be based on the way the numerical values have been obtained. Then it is normally possible to decide whether the value results from some statistical procedure (no matter how simple or elaborate), or whether it relies essentially on personal opinion. This does not automatically imply that 'educated' guesses made by an experienced researcher are necessarily less reliable than values which result from some kind of statistical estimation, but they are inevitably subjective.

In the BPM recommendation on the statement of uncertainties (Giacomo [95], Kaaris [96]) this has led to the subdivision of uncertainties into two groups, called class A (statistical estimates, objective) and class B (guesses, subjective).

In fact, this classification is often not very important in practical applications, as both types are subsequently treated in a completely similar manner. The distinction may, however, have some pedagogical justification as it compels the experimenter to give some thought to the origin of his estimates. In addition, it may be useful as a warning, for whenever uncertainties of class B play an important role

among the various contributions, one should refrain from a precise evaluation of confidence intervals, because no density function can be attributed with any reliability to a class B value, which would be a prerequisite for such a calculation.

Uncertainties of class A. Since the numerical values of these uncertainties are based on some statistical reasoning, they are in most cases already expressed in terms of their standard deviation. As an example we may consider a dose determination made with an ionization chamber. This involves a factor which allows for the fact that the temperature of the chamber during the measurements may be different from that for which it was calibrated. In addition, the temperature may be fluctuating. The factor is known to be inversely proportional to the absolute temperature and its uncertainty is proportional to the uncertainty in the temperature. One way to assess this would be to make several measurements of temperature and to determine the standard deviation of their average. This would yield

$$s(\bar{T}) = \left\{ \frac{1}{N(N-1)} \sum_{i=1}^N (T_i - \bar{T})^2 \right\}^{1/2} \quad (\text{A.1})$$

where T_i is the result of an individual determination and $\bar{T}$ is the average of the N measurements performed.

The standard deviations of other relevant quantities may be determined in an analogous manner; their combination and effect on the final result will be discussed later. For some specific applications (e.g. in time and frequency measurements) it may be necessary to use a somewhat modified form of variance, called Allan variance.

Uncertainties of class B. It is not always possible to assess the uncertainty of a quantity by repeated measurements. Thus, it may be necessary to use values from the literature. If uncertainties are stated, possibly in the form of estimated standard deviations, they may be considered as components of class A and treated as such. Examples are listed in Table XXII. However, such information may be lacking. It then is the duty of the experimenter to assign to them, to the best of his knowledge, uncertainties which may be considered as estimated standard deviations. This cannot be done in a completely objective way as it involves personal experience and judgement; in spite of the difficulties, such uncertainties should be estimated in the form of standard deviations (or variances) because these are the quantities which are actually needed for further propagation, and the assignment should be made by the experimenter since he alone has a comprehensive knowledge of the method and the experimental procedures.

Some people prefer to state in these cases uncertainties in the form of 'maximum limits', in the sense that it is very unlikely that the 'true' value will be outside this interval. While it cannot be denied that such values are easy to estimate,

one has also to realize that they are ill defined and of limited practical usefulness. In particular, they do not lend themselves to the formulation of a suitable propagation law. Since in most practical cases various components will contribute to the total uncertainty, it is of prime importance to have at one's disposal a general expression for doing this. By maintaining this requirement, however, one is inevitably led to use variances (or their positive square root, called standard deviation) as the relevant quantities. All other 'measures' will usually have to be 'translated' to variances before being applicable.

Covariances. Practical measurements often involve quantities which are not independent of each other, either because one of them directly influences another one, or also because they are functions of another quantity (for instance temperature or humidity). In this case they are said to be correlated and the degree of correlation can formally be expressed by a quantity called covariance. Experimenters sometimes think that covariances are quantities which are rather enigmatic and difficult to handle, but happily of so little practical impact that they can nearly always be neglected. Unfortunately, this is not quite a realistic view. For a clear account of the basic concepts see, for instance, Allisy [97].

A close formal similarity exists between the quantities variance and covariance. If for two quantities x and y there are N measurements available, we are not only in a position to form the variance, which, for the measurements x_j , is given by

$$\text{Var}(x) = \frac{1}{N-1} \sum_{j=1}^N (x_j - \bar{x})^2 \quad (\text{A.2})$$

with $\bar{x}$ the arithmetical mean value, and similarly for y , but also the covariance between the quantities, defined as

$$\text{Cov}(x, y) = \frac{1}{N-1} \sum_{j=1}^N (x_j - \bar{x})(y_j - \bar{y}) \quad (\text{A.3})$$

From this it is obvious that $\text{Cov}(x, x) = \text{Var}(x)$. If x and y are independent of each other, we expect their covariance to vanish. Note that the concept of covariance always involves a specific pair of variables (here x and y).

A.2. PROPAGATION LAW FOR UNCERTAINTIES

The quantity we want to determine by means of an experiment can rarely be obtained by a direct measurement. In most cases the desired result is obtained in an

indirect way, namely by the measurement of several quantities $x_1, x_2, \dots, x_n$ which, by the application of some mathematical relation, it is then possible to arrive at the quantity actually looked for. Hence, if y is the final result and if the measured quantities are denoted by $x_1, x_2, \dots, x_n$, we can describe the situation by writing $y = F(x_1, x_2, \dots, x_n)$, which means that y is a (known) function of the n quantities $x_1, x_2, \dots, x_n$, which are supposed to be directly accessible to measurement. (Note that the x_j now stand for different variables in what follows.)

A basic result of mathematical statistics ensures us that the variance of y is always given to a good approximation by the equation

$$s_y^2 = \sum_{j=1}^n \left(\frac{\partial F}{\partial x_j} \right)^2 s_j^2 + 2 \sum_{j < k} \left(\frac{\partial F}{\partial x_j} \frac{\partial F}{\partial x_k} \right) s_{jk}^2 \quad (\text{A.4})$$

where $s_j^2 = \text{Var}(x_j)$ and $s_{jk} = \text{Cov}(x_j, x_k)$ are in practice the sample estimates of these quantities. This is the general propagation law for uncertainties. Note that no distinction is made between the uncertainty components of class A or class B.

Two important special cases of simple functional relations may be distinguished. Thus, if

$$y = F(x_1, x_2, \dots, x_n) = c_0 + \sum_{j=1}^n c_j x_j \quad (\text{A.5})$$

where the coefficients c_j are constants, one readily obtains for the variance of x the expression

$$s_y^2 = \sum_{j=1}^n c_j^2 \cdot s_j^2 + 2 \sum_{j < k} c_j \cdot c_k \cdot s_{jk} \quad (\text{A.6})$$

The other special case occurs if the variables are factors of a product (multiplications or divisions, thus for

$$y = F(x_1, x_2, \dots, x_n) = c_0 \prod_{j=1}^n x_j^{p_j} \quad (\text{A.7})$$

In this case it is convenient to use the concept of relative uncertainty which is defined for a quantity x_j by

$$r_j \equiv \frac{s_j}{x_j}, \quad \text{likewise } r_{jk} \equiv \frac{s_{jk}}{x_j x_k} \quad (\text{A.8})$$

for the relative covariance.

Some simple algebra then leads to

$$r_y^2 = \sum_{j=1}^n c_j^2 \cdot r_j^2 + 2 \sum_{j < k} c_j \cdot c_k \cdot r_{jk} \quad (\text{A.9})$$

These two special cases already cover a large fraction of the practical situations.

The proper use of the notion of correlation thus allows one to arrive at a real unification of uncertainty statements. A possible general bias in a set of measurements is now seen to correspond to a common positive covariance, and the effect this has on the resulting variance is given by the propagation law mentioned above. As a consequence, the previous and often somewhat arbitrary subdivision of uncertainties into systematic and random components, with separate rules applicable to their propagation, is now recognized as an unnecessary complication. The consistent application of the propagation law in its general form not only allows one to simplify the practical evaluations, but also clarifies the concepts used. Further details can be found in Müller [98] or DIN [99].

INTRODUCTION TO WORKSHEETS AND FLOW DIAGRAMMS

The following worksheet examples show the different steps to be followed in the present Code of Practice.

First, the absorbed dose to air chamber factor N_D of a cylindrical chamber is determined. It has been assumed that the SSDL has provided an air kerma calibration factor N_K (or exposure calibration factor N_X) for a given dosimetry assembly, i.e. electrometer plus ionization chamber. The calculation of the factor k_m has been considered in detail, although it will most often not be needed as many chambers have been included in Table XVIII.

Second, the absorbed dose to water is calculated for the reference conditions in electron and photon beams covering the whole range of application of the present publication. For high energy photon beams calculation for ^{60}Co has been chosen since this is the most widely used radiation therapy unit.

Blank worksheets can be obtained free of charge from the

International Atomic Energy Agency
Dosimetry Section
Division of Life Sciences
P.O. Box 100
A-1400 Vienna
Austria

who can also supply copies of the flow diagrams.

The entries in italics in the worksheets are those assumed to have been filled in by the user.

WORKSHEET 1 FOR CALCULATING THE ABSORBED DOSE TO AIR CHAMBER FACTOR N_D

Name: *User*

Date:

1. Ionization chamber

Chamber model and serial number: *NE 2505/34, No. 4075*Cavity inner radius: *3.14 mm*Wall material: *Graphite* ($\rho = 1.82 \text{ g/cm}^3$), thickness: *0.0655 g/cm²*Buildup cap material: *PMMA* ($\rho = 1.18 \text{ g/cm}^3$), thickness: *0.543 g/cm²*
total thickness: *0.6085 g/cm²*

2. Calibration laboratory data

Calibration laboratory and date: *SSDL, 860808*Calibration factor (kerma in air)¹, $N_K = 9.08 \times 10^{-3} \text{ Gy/scale div}$
given at $P_0 = 101.3 \text{ kPa}$, $T_0 = 20^\circ \text{C}$ and $50\% \text{ R.H.}$ Polarizing voltage: *-250 V*, field size: *$10 \times 10 \text{ cm}^2$* Source chamber distance: *100 cm*

3. Constants

 $W/e = 33.97 \text{ J/C}$, and $g = 0.003$ (for ^{60}Co gamma radiation).

4. Determination of k_{air} k_m ²

Fraction of ionization due to electrons from chamber wall
(Fig. 15),

$$\alpha = 0.53$$

Stopping power ratio air/wall (Table XVII),

$$S_{\text{air,wall}} = 0.998$$

Energy absorption coefficient ratio wall/air
(Table XVII),

$$(\bar{\mu}/\rho)_{\text{wall,air}} = 1.001$$

Fraction of ionization due to electrons from
buildup cap,

$$(1-\alpha) = 0.47$$

Stopping power air/cap (Table XVII),

$$S_{\text{air,cap}} = 0.908$$

Energy absorption coefficient ratio cap/air
(Table XVII),

$$(\bar{\mu}/\rho)_{\text{cap,air}} = 1.081$$

$$k_m = \alpha S_{\text{air,wall}} (\bar{\mu}/\rho)_{\text{wall,air}} + (1-\alpha) S_{\text{air,cap}} (\bar{\mu}/\rho)_{\text{cap,air}}$$

$$= 0.991$$

$$k_{\text{air}} = 0.990$$

$$k_{\text{air}} k_m = 0.981$$

5. Absorbed dose to air calibration factor

 $N_D = N_K (1-g) k_{\text{air}} k_m = 0.888 \times 10^{-2} \text{ Gy/div}$,
obtained at *101.3 kPa*, *20 °C*, *50% R.H.*

¹ If N_K instead of N_K is known, N_K is given by

$$N_K = N_x \frac{W}{e} \frac{1}{1-g}$$

N_x should be in C/kg per division. If N_x is in R/div:
 $N_x (\text{C/kg div}) = N_x (\text{R/div}) 2.58 \times 10^{-4} (\text{C/kg} \cdot \text{R})$

² If the chamber is included in Table XVIII it is recommended to use the product $k_{\text{air}} k_m$ given there.

WORKSHEET 2 FOR CALCULATING THE ABSORBED DOSE TO WATER UNDER REFERENCE CONDITIONS USING ELECTRON BEAMS

Name: *User*

Date:

1. Radiation treatment unit: *Sagitaire*

Nominal energy: *19 MeV*

Depth of the effective point of measurement: *3 cm*, ($z_{pr} - z_p = 0.5r$)¹

Field size: *10 × 10 cm²* at SSD = *100 cm*

Nominal dose rate of the accelerator: *200* monitor units/min

2. Ionization chamber

Model and serial number: *NE 2505/3A, No. 4075*

Inner radius¹: *3.14 mm*, wall material and thickness: *Graphite, 0.0655 g/cm²*

Absorbed dose to air chamber factor:

$N_D = 0.888 \times 10^{-2}$ Gy/div given at $P_0 = 101.3$ kPa, $T_0 = 20^\circ\text{C}$, 50% R.H.

Polarizing voltage: *-250 V*

Response change as compared to calibration date derived from checking against a radioactive source: *O.K. within 0.2%*

3. Electrometer reading correction

Reading², $M_u^0 = 0.9976$ div/m.u., monitor setting: *200 m.u.*

Pressure, $P = 100.3$ kPa

Temperature, $T = 24.3^\circ\text{C}$

$$P_{TP} = \frac{P_0 (273.2 + T)}{P (273.2 + T_0)} = 1.025$$

Humidity correction, $k_h = 1.000$

Recombination correction (Table VIII or IX)

$V_1 = 250$ V, $V_2 = 83.33$ V, $M_1/M_2 = 1.095$,

$P_s = 1.054$

$$M_u = M_u^0 P_{TP} k_h P_s = 1.078 \text{ div/m.u.}$$

Electron fluence³ correction plastic to water (Table XII), $h_m = 1.0$

$$M_{u,w} = M_u h_m = 1.078 \text{ div/m.u.}$$

4. Absorbed dose to water

Ranges obtained by measurement at SSD = *1 m* with absorbed dose curves

$R_{50} = 7.4$ cm, $R_p = 9.0$ cm

Phantom material (plastics can only be used if $\bar{E}_0 < 10$ MeV)

a) *water* b) *plastic*

Ranges converted to ranges in water (Eq. (1), Table III)

$R_{50} =$ cm, $R_p =$ cm

Most probable energy at the surface

$$E_{p,0} = 0.22 + 1.98 R_p + 0.0025 R_p^2$$

$$E_{p,0} = 18.24 \text{ MeV}$$

Mean energy at the surface (Table IV),

$$\bar{E}_0 = 17.56 \text{ MeV}$$

$$z/R_p = 0.333, \bar{E}_z/\bar{E}_0 \text{ (Table V)} = 0.596$$

Mean energy at depth ($z = 3$ cm),

$$\bar{E}_z = 10.47 \text{ MeV}$$

Stopping power ratio water/air (Table X),

$$S_{w,air} = 0.999$$

Perturbation factor (Table XV)⁴,

$$P_a = 0.981$$

$$D_w (P_{eff}) = M_u N_D S_{w,air} P_a = 0.938 \times 10^{-2} \text{ Gy/m.u.}$$

¹ For plane parallel chambers, P_{eff} is situated in the front surface. Data regarding radius are therefore not needed.

² Averaged value of readings corrected for leakage and polarity.

³ Equal to 1 if a water phantom is used.

⁴ Equal to 1 for plane parallel chambers.

WORKSHEET 3 FOR CALCULATING THE ABSORBED DOSE TO WATER UNDER REFERENCE CONDITIONS USING HIGH ENERGY PHOTON BEAMS

Name: User Date:

1. Radiation treatment unit: *Theratron 80, ⁶⁰Co*

Nominal accelerating potential: MV

Depth in water of the effective point of measurement: 5 cm,
($z_{p,eff} - z_p = 0.5$ cm, Fig. 11)

Field size: 10×10 cm² at SSD = 80 cm

Nominal dose rate of the accelerator: monitor units/min

2. Ionization chamber

Model and serial number: *NE 2505/3A, No. 4075*

Inner radius: 3.14 mm, wall material and thickness: *Graphite, 0.0655 g/cm²*

Absorbed dose to air chamber factor:

$N_D = 0.888 \times 10^{-2}$ Gy/div given at $P_0 = 101.3$ kPa, $T_0 = 20^\circ\text{C}$, 50 % R.H.

Polarizing voltage: ~ 250 V

Response change as compared to calibration date derived from checking against a radioactive source: *O.K. within 0.1 %*

3. Electrometer reading correction

Reading¹, $M_u^0 = 76.43$ div/min², monitor setting: — m.u.

Pressure, $P = 100.7$ kPa

Temperature, $T = 23.8^\circ\text{C}$

Recombination correction (Table VII or IX, or Fig. 13)

$V_1 = 250$ V, $V_2 = 83.3$ V, $M_1/M_2 = 1.001$,

$$M_u = M_u^0 P_{TP} k_h P_s = 77.88 \text{ div/min}^2$$

4. Absorbed dose to water

Quality of the beam, TPR_{10}^{20} (or D_{20}/D_{10}) = — for 10×10 cm²
at SCD = 1 m (SSD = 1 m)

Stopping power ratio water/air (Table XIII),

$$S_{w,air} = 1.133$$

Perturbation factor (Fig. 14)³,

$$P_u = 0.991$$

$$D_w (P_{eff}) = M_u N_D S_{w,air} P_u = 77.65 \times 10^{-2} \text{ Gy/min}^2$$

¹ Average value of readings corrected for leakage and polarity.

² Corrections for the effective irradiation time in gamma therapy beams should be accounted for.

³ The perturbation factor can also be obtained from Eq. (23). Fraction of ionization due to electrons from chamber wall (Figs 15 and 16),

Stopping power ratio wall/air (Table XX),

$$S_{wall,air} = 1.002$$

Energy absorption coefficient ratio water/wall (Table XXI),

$$(\bar{\mu}_{en}/\rho)_{w,wall} = 1.113$$

Fraction of ionization due to electrons from water,

$$(1-\alpha) = 0.47$$

Stopping power ratio water/air (Table XIII),

$$S_{w,air} = 1.133$$

$$P_u = \frac{\alpha S_{wall,air} (\bar{\mu}_{en}/\rho)_{w,wall} + (1-\alpha) S_{w,air}}{S_{w,air}} = 0.992$$

WORKSHEET 4 FOR CALCULATING THE ABSORBED DOSE TO WATER UNDER REFERENCE CONDITIONS USING X-RAYS

Name: *User* Date: _____

1. Radiation treatment unit:

Nominal tube potential 200 kV, filtration: 1 mm Cu, 3.8 mm Al

Nominal tube current: 10 mA

Applicator/field size: 10 cm × 10 cm at FSD = 80 cm

Quality of the beam to be used, HVL = 1.6 mm of Cu

2. Ionization chamber

2a. Plane parallel (low energy X-rays, 10–100 kV)

Model and serial number:

Calibration factor at HVL = mm of

(Nominal kV = , filtration:)

$N_K =$ Gy/div, $N_{D,w} =$ Gy/div

given at $P_0 =$ kPa, $T_0 =$ °C, % R.H.

Polarizing potential: V, field size: cm²

Source–chamber distance: cm

Response change as compared to calibration date derived from checking against a radioactive source:

2b. Cylindrical (medium energy X-rays, 100–300 kV)

Model and serial number: PTW M23331 No. 307

Calibration factor at HVL = 1.6 mm of Cu

(Nominal kV = 200, Filtration: 3.8 mm Al 1 mm Cu)

$N_K = 28.78 \times 10^6$ Gy/C,

given at $P_0 = 101.325$ kPa, $T_0 = 20$ °C, 60 % R.H.

Polarizing potential: +500 V, Field size: 10 cm × 10 cm

Source–chamber distance 100 cm

Response change as compared to calibration date derived from checking against a radioactive source: O.K. within %

3. Electrometer reading correction

Reading, $M_u^0 = 6.072 \times 10^{-11}$ C/min

Pressure, $P = 101.84$ kPa

Temperature, $T = 21.3$ °C

$$P_{TP} = \frac{P_0 (273.2 + T)}{P (273.2 + T_0)} = 0.9994$$

Humidity correction, $k_h = 1.000$

Recombination correction (Fig. 13):

$V_1 =$ V, $V_2 =$ V, $M_1/M_2 =$,

$P_2 = 1.000$

$$M_u = M_u^0 P_{TP} k_h P_2 = 6.068 \times 10^{-11} \text{ C/min}$$

4. Absorbed dose to water

4a. Low energy X-rays (surface)

(i) $N_{D,w}$ available

$$D_w = M_u N_{D,w} = \text{Gy/min}$$

(ii) N_K available

(a) Energy absorption coefficient ratio water/air (Table XIV),

$$(\bar{\mu}_{en}/\rho)_{w,air} =$$

(b) Backscatter factor (Table XVI),

$$B =$$

(c) Correction for spectral dependence¹,

$$k_u =$$

$$D_w = M_u N_K B k_u (\bar{\mu}_{en}/\rho)_{w,air} = \text{Gy/min}$$

4b. Medium energy X-rays (5 cm depth)

(a) Energy absorption coefficient ratio water/air (Table XIV),

$$(\bar{\mu}_{en}/\rho)_{w,air} = 1.074$$

(b) Perturbation factor (Table XV),

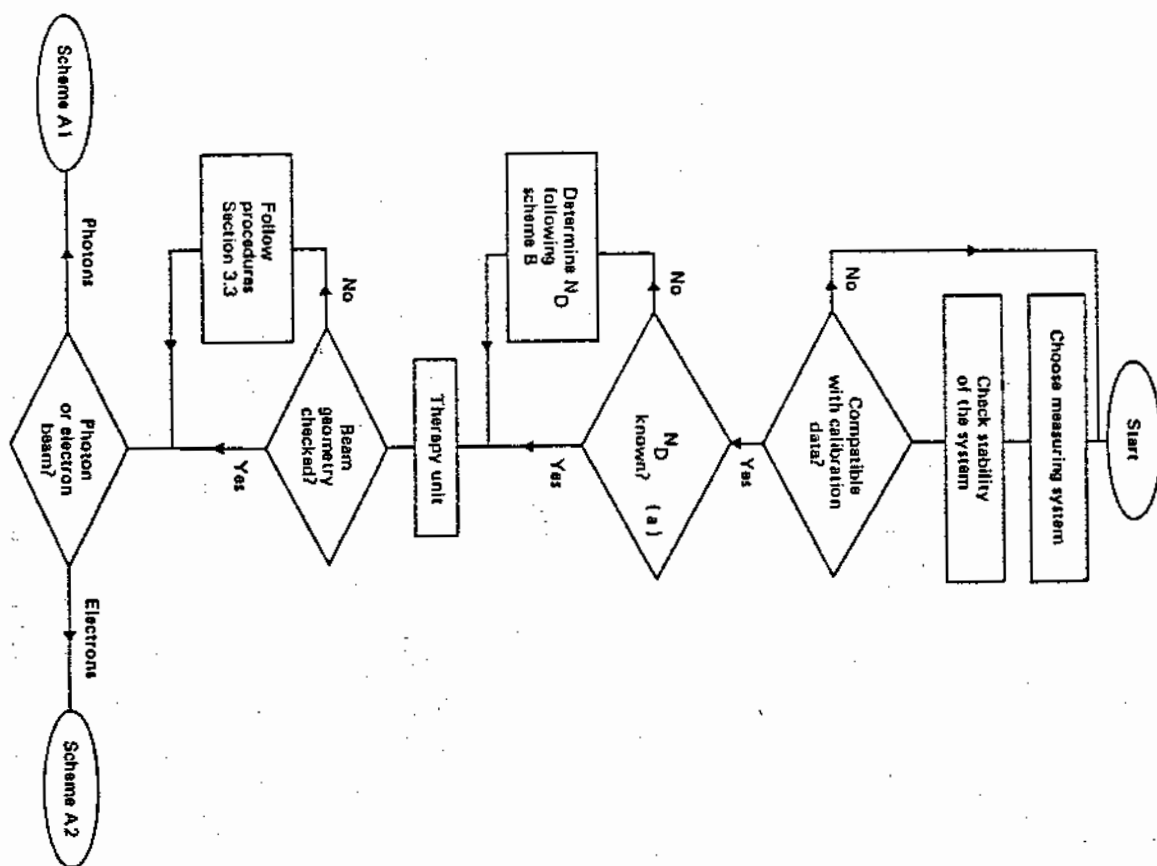
$$P_u = 1.04$$

(c) Correction for spectral dependence¹,

$$k_u = 1.000$$

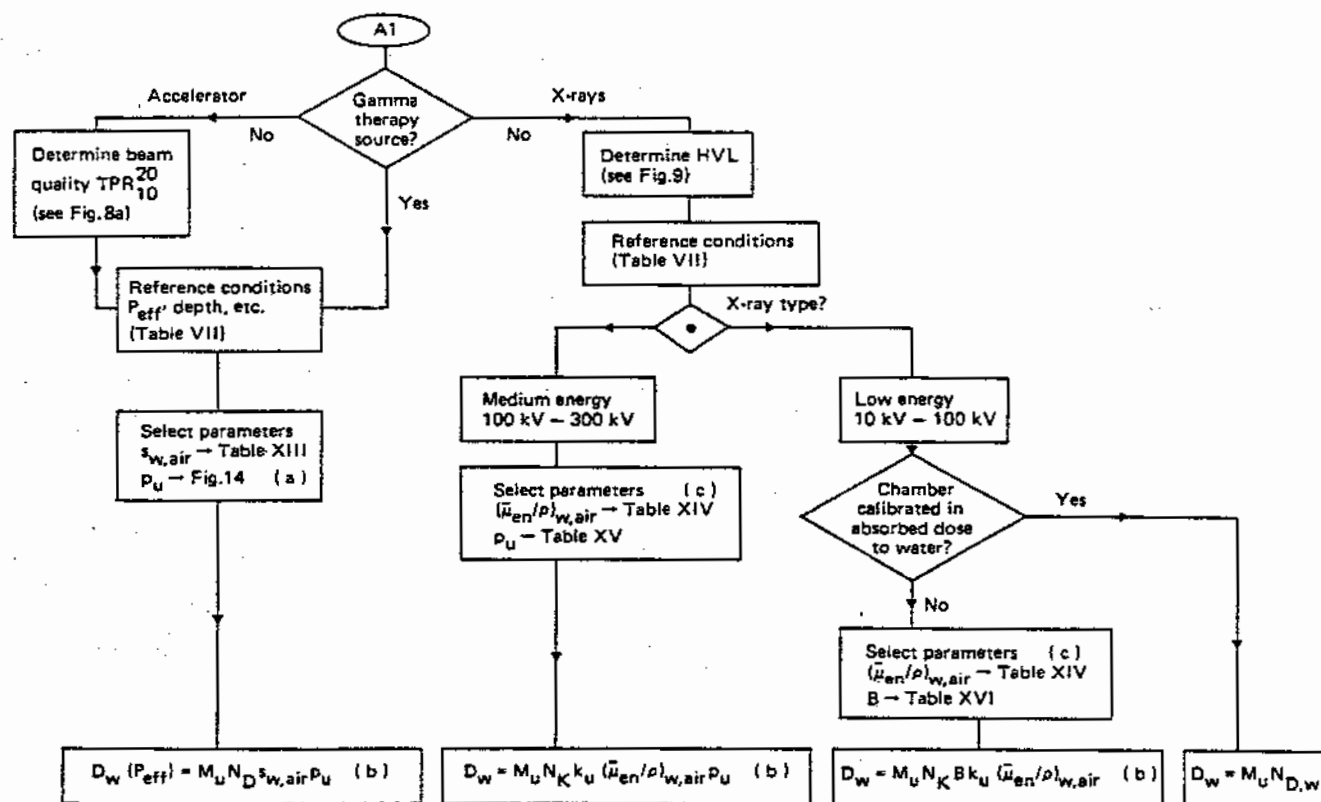
$$D_w = M_u N_K (\bar{\mu}_{en}/\rho)_{w,air} P_u k_u = 1.95 \times 10^{-3} \text{ Gy/min}$$

¹ Equal to 1 for most practical situations.



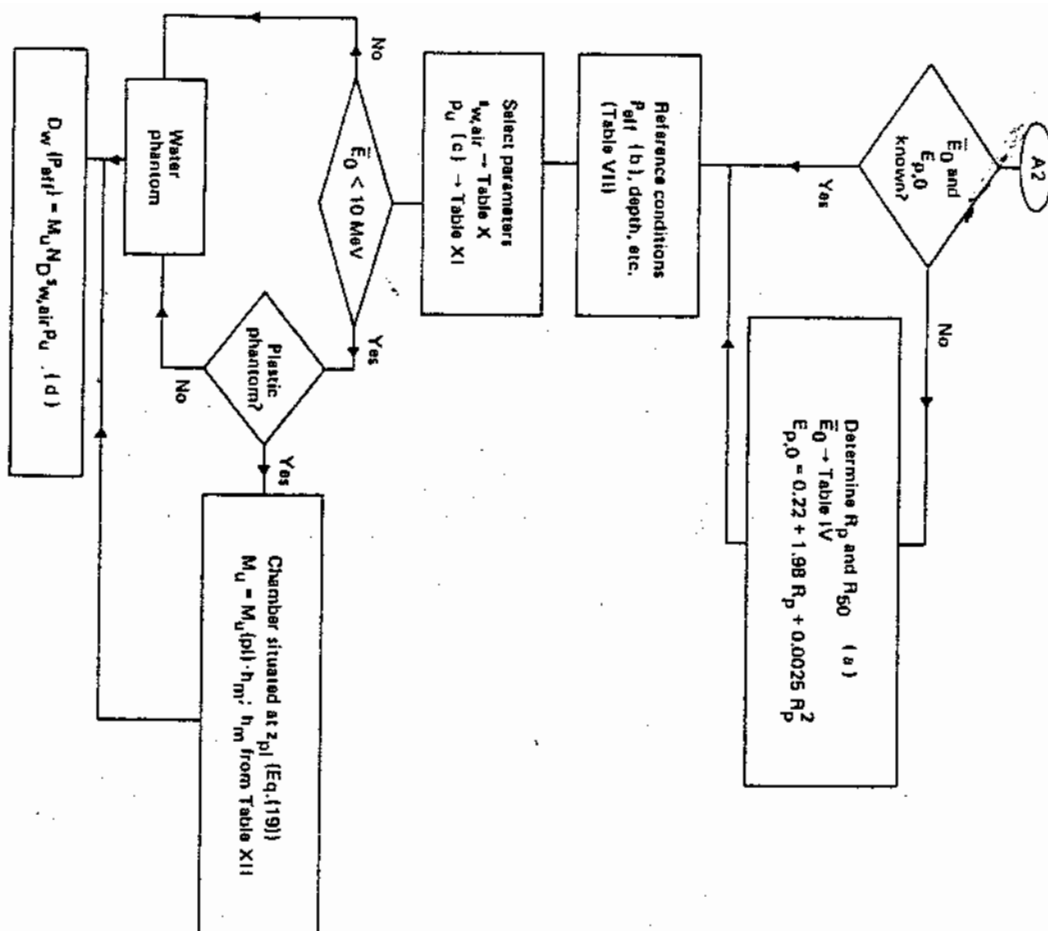
(a) For X-rays only N_K (or $N_{D,w}$ for low energy X-rays - see Section 6.2.2) is needed.

Diagram A. Determination of absorbed dose to water under reference conditions: general.



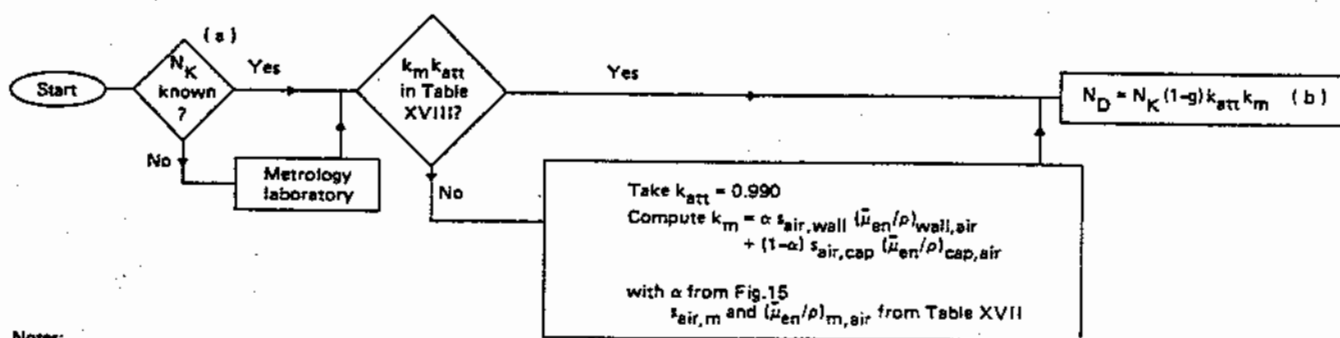
Notes: (a) For greater accuracy use Eq.(23) with α from Fig.16; $s_{w,air}$ and $(\mu_{en}/\rho)_{w,m}$ from Tables XX and XXI.
 (b) M_u should be corrected for temperature, pressure, recombination, etc. (see Section 7.1).
 (c) $k_u = 1$ for most practical situations.

Diagram A1. Absorbed dose to water in photon beams.



- Notes:
- (a) Broad beams always have to be used. If ranges are obtained in plastics, scale R_p and R_{50} according to Eq. (1).
- (b) If a plane chamber is used, P_{aff} is situated in the front surface.
- (c) For a plane chamber, $P_u = 1.0$. For cylindrical chambers, compute z/R_p and obtain $\bar{E}_z/\bar{E}_0$ from Table V. Using E_2 and the inner radius of the chamber, obtain P_u from Table XI.
- (d) M_u should be corrected for temperature, pressure, recombination, etc. (see Section 7.1).

Diagram A2. Absorbed dose to water in electron beams.



Notes:

- (a) If N_X instead of N_K is known, $N_K = N_X \frac{W}{e} \frac{1}{1-g}$ (Eq. (10)).
 $W/e = 33.97 \text{ J/C}$.
 N_X should be in C/kg per division. If N_X is in R/div, $N_X (\text{C/kg} \cdot \text{div}) = N_X (\text{R/div}) 2.58 \times 10^{-4} (\text{C/kg} \cdot \text{R})$.
- (b) $g = 0.003$ for ^{60}Co gamma rays.

Diagram B. Determination of the absorbed dose to air calibration factor N_D of a cylindrical ionization chamber.

REFERENCES

- [1] INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, Determination of Absorbed Dose in a Patient Irradiated by Beams of X or Gamma Rays in Radiotherapy Procedures, ICRU Rep. 24, ICRU Publications, Bethesda, MD (1976).
- [2] BRAHME, A., Dosimetric precision requirements in radiation therapy, *Acta Radiol., Oncol.* **23** (1984) 379.
- [3] HOSPITAL PHYSICISTS ASSOCIATION, A Practical Guide to Electron Dosimetry 5-35 MeV, HPA Rep. Series No. 4, London (1971).
- [4] HOSPITAL PHYSICISTS ASSOCIATION, A Practical Guide to Electron Dosimetry below 5 MeV for Radiotherapy Purposes, HPA Rep. Series No. 13, London (1975).
- [5] HOSPITAL PHYSICISTS ASSOCIATION, Revised code of practice for the dosimetry of 2 to 35 MV X-ray, and of caesium-137 and cobalt-60 gamma-ray beams, *Phys. Med. Biol.* **28** (1983) 1097-1104.
- [6] NORDIC ASSOCIATION OF CLINICAL PHYSICS, Procedures in external radiation therapy dosimetry with electron and photon beams with maximum energies between 1 and 50 MeV, *Acta Radiol., Oncol.* **19** (1980) 55.
- [7] NORDIC ASSOCIATION OF CLINICAL PHYSICS, Electron beams with mean energies at the phantom surface below 15 MeV, *Acta Radiol., Oncol.* **20** (1981) 403.
- [8] AMERICAN ASSOCIATION OF PHYSICISTS IN MEDICINE, Code of practice of X-ray therapy linear accelerators, *Med. Phys.* **2** (1975) 110; A protocol for the determination of absorbed dose from high-energy photon and electron beams, *Med. Phys.* **10** (1983) 741.
- [9] SOCIEDAD ESPAÑOLA DE FÍSICA MÉDICA, Procedimientos Recomendados para la Dosimetría de Fotones y Electrones de Energías Comprendidas entre 1 MeV y 50 MeV en Radioterapia de Haces Externos, SEFM Publ. No. 1/1984, Comité de Dosimetría en Radioterapia, Madrid (1984).
- [10] EUROPEAN SOCIETY FOR THERAPEUTIC RADIOLOGY AND ONCOLOGY, Proc. Symp. on New Protocols for the Dosimetry of High-Energy Photon and Electron Beams, *Radiother. Oncol.* **4** (1985) 289-328.
- [11] EISENLOHR, H.H., BOYD, A.W., NAM, J.W., The dosimetry programme of the International Atomic Energy Agency in Vienna, *Radiat. Prot. Dos.* **14** (1981) 245-248.
- [12] EISENLOHR, H.H., The IAEA/WHO Network of Secondary Standard Dosimetry Laboratories — a novel approach to modern radiation metrology, *Med. Phys. World* **1** (1984) 9.
- [13] INTERNATIONAL ATOMIC ENERGY AGENCY, Secondary Standard Dosimetry Laboratories: Developments and Trends, IAEA, Vienna (1985).
- [14] INTERNATIONAL ATOMIC ENERGY AGENCY, WORLD HEALTH ORGANIZATION, Criteria for the Establishment of a Secondary Standard Dosimetry Laboratory, IAEA/WHO (1984).
- [15] INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, Radiation quantities and units, ICRU Rep. 33, ICRU Publications, Bethesda, MD (1980).

- [16] INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, Radiation Dosimetry: Electron Beams with Energies Between 1 and 50 MeV, ICRU Rep. 35, ICRU Publications, Bethesda, MD (1984).
- [17] COMITÉ CONSULTATIF POUR LES ETALONS DE MESURE DES RAYONNEMENTS IONISANTS (Section I), Report to the Comité International des Poids et Mesures (ELLIS, S.C., Reporter), 8th Mtg (1985).
- [18] INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, Stopping Powers for Electrons and Positrons, ICRU Rep. 37, ICRU Publications, Bethesda, MD (1984).
- [19] NIATEL, M.-T., PERROCHE-ROUX, A.M., BOUTILLON, M., Two determinations of W for electrons in dry air, Phys. Med. Biol. 30 (1985) 67.
- [20] COMITÉ CONSULTATIF POUR LES ETALONS DE MESURE DES RAYONNEMENTS IONISANTS (Section I), Effect of a change of stopping-power values on the W value recommended by ICRU for electrons in dry air, Rep. CCEMRI(85-8 Bureau International des Poids et Mesures, Sèvres (1985).
- [21] BOUTILLON, M., PERROCHE-ROUX, A.M., Re-evaluation of the W value for electrons in dry air, Phys. Med. Biol. 32 2 (1987) 213.
- [22] BOUTILLON, M., Values of g in air for BIPM Radiation Qualities, Comité consultatif pour les étalons de mesure des rayonnements ionisants (Section I), Rep. CCEMRI(85-18, Bureau International des Poids et Mesures, Sèvres (1985).
- [23] HUBBELL, J.H., Photon mass attenuation and energy-absorption coefficients from 1 keV to 20 MeV, Int. J. Appl. Radiat. Isot. 33 (1982) 1269.
- [24] GALBRAITH, D.M., RAWLINSON, J.A., MUNRO, P., Dose errors due to charge storage in electron irradiated plastic phantoms, Med. Phys. 11 (1984) 197.
- [25] MATTSOON, L.O., SVENSSON, H., Charge build-up effects in insulating phantom materials, Acta Radiol., Oncol. 23 (1984) 393.
- [26] ANTHONY, K., BLUNDAIT, H., PALIWAL, R., ATTIX, F.H., Charge storage in electron irradiated phantom materials, Med. Phys. 11 (1986) 99.
- [27] NATIONAL COUNCIL ON RADIATION PROTECTION AND MEASUREMENTS, Dosimetry of X-ray and Gamma Ray Beams for Radiation Therapy in the Energy Range 10 keV to 50 MeV, NCRP Rep. 69, Washington, DC (1981).
- [28] BRAHME, A., SVENSSON, H., Specification of electron beam quality from the central axis depth absorbed dose distribution, Med. Phys. 3 (1976) 95.
- [29] JOHANSSON, K.-A., SVENSSON, H., Dosimetric Intercomparison at the Nordic Radiation Therapy Centres. Part II. Comparison between Different Detectors and Methods, Thesis, Univ. of Göteborg (1982).
- [30] LAX, I., BRAHME, A., Collimation of high energy electron beams, Acta Radiol., Oncol. 19 (1979) 199.
- [31] HARDER, D., "Energiespektren schneller Elektronen in verschiedenen Tiefen", Symp. on High Energy Electrons, Montreux, 1964 (ZUPPINGER, A., PORETTI, G., Eds), Springer-Verlag, Berlin (West) (1965) 260.
- [32] ANDREO, P., BRAHME, A., Mean energy in electron beams, Med. Phys. 8 (1981) 6.
- [33] ANDREO, P., Dosis: A computer program for the calculation of absorbed dose in photon and electron beams from ionization measurements in a phantom, Nucl. Instrum. Methods 211 (1983) 481.
- [34] BRAHME, A., SVENSSON, H., Radiation beam characteristics of a 22 MeV microtron, Acta Radiol., Oncol. 18 (1979) 244.
- [35] RAWLINSON, J.A., JOHNS, H.E., Percentage depth dose for high energy X-ray beams in radiotherapy, Am. J. Roentgenol. 118 (1973) 919.
- [36] ORGANISATION INTERNATIONALE DE METROLOGIE LEGALE, Secondary Standard Dosimetry Laboratories for the Calibration of Dosimeters Used in Radiotherapy, OIML, Paris (1983).
- [37] SVENSSON, H., Quality assurance in radiation therapy: physical aspects, Int. J. Radiat. Oncol., Biol. Phys. 10, Suppl. 1 (1984) 59-65.
- [38] NIATEL, M.-T., LOFTUS, T.P., OETZMANN, W., Comparison of exposure standards for ^{60}Co gamma rays, Metrologia 11 (1975) 17.
- [39] ZSDÁNSZKY, K., "Primary and secondary standards of dosimetry. Calibration methods in Hungary", National and International Standardization of Radiation Dosimetry (Proc. Symp. Atlanta, 1977), Vol. 1, IAEA, Vienna (1978) 107.
- [40] HOFMEESTER, G.H., "Calorimetric determination of absorbed dose in water for 1-25 MeV X-rays", Biomedical Dosimetry: Physical Aspects, Instrumentation, Calibration (Proc. Symp. Paris, 1980), IAEA, Vienna (1981) 235.
- [41] NIATEL, M.-T., "Détermination de W_{air} (énergie moyenne nécessaire pour produire une paire d'ions dans l'air) basée sur des comparaisons d'étalons de dose absorbée effectuées au BIPM", CCEMRI(77-113 (1977) R(0) 56.
- [42] SVENSSON, H., BRAHME, A., "Recent advances in electron and photon dosimetry", Radiation Dosimetry, Physical and Biological Aspects (ORTON, C.G., Ed.), Plenum Press, New York (1986) 87.
- [43] INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, Average Energy Required to Produce an Ion Pair, ICRU Rep. 31, ICRU Publications, Bethesda, MD (1979).
- [44] COMITÉ CONSULTATIF POUR LES ETALONS DE MESURE DES RAYONNEMENTS IONISANTS (Section I), Comparison of Standards of Exposure, CCEMRI(80)30 Bureau International des Poids et Mesures, Sèvres (1981).
- [45] ALLISY, A., Contribution à la mesure de l'exposition produite par les photons émis par le Co-60, Metrologia 3 (1967) 41.
- [46] BOUTILLON, M., NIATEL, M.-T., A study of a graphite cavity chamber for absolute exposure measurements of Co-60 gamma rays, Metrologia 9 (1973) 139.
- [47] LOFTUS, T.P., WEAVER, J.T., Standardization of Co-60 and Cs-137 gamma-ray beams in terms of exposure, J. Res. Natl. Bur. Stand. (U.S.) 78A (1974) 465.
- [48] LÖFROTH, P.-O., WESTMAN, S., HETTINGER, G., Spectral distributions from Co-60 therapy units, Acta Radiol., Ther., Phys., Biol. 12 (1973) 552.
- [49] NATH, R., SCHULZ, R.J., Calculated response and wall correction factors for ionization chambers exposed to Co-60 gamma rays, Med. Phys. 8 (1981) 85.
- [50] ROGERS, D.W.O., BIELAJEW, A.F., NAHUM, A.E., Ion chamber response and A_{wall} correction factors in a Co-60 beam by Monte Carlo simulation, Phys. Med. Biol. 30 (1985) 429.
- [51] MATTSOON, L.O., JOHANSSON, K.-A., Experimentally Determined Wall Correction Factors k_m and k_{air} for Cylindrical Ionization Chambers Used in High Energy Photon and Electron Beam Dosimetry, Internal Rep., Radiation Physics Department, University of Göteborg, 1984.

- [52] HETTINGER, G., PETERSSON, C., SVENSSON, H., Dispersion effect of thin-ble chambers exposed to a photon or electron beam from a betatron, *Acta Radiol. Ther., Phys. Biol.* 6 (1967) 61.
- [53] DUTREIX, A., "Problems of high-energy X-ray beam dosimetry" (KRAMER, S., Ed.), Wiley, New York (1976).
- [54] JOHANSSON, K.-A., MATTHESSON, L.O., LINDBORG, L., SVENSSON, H., "Absorbed-dose determination with ionization chambers in electron and photon beams having energies between 1 and 50 MeV", National and International Standardization of Radiation Dosimetry (Proc. Symp. Athens, 1977), Vol. 2, IAEA, Vienna (1978) 243.
- [55] ZOETELIEF, J., ENGELS, A.C., BROERSE, J.J., "Displacement corrections for spherical ion chambers in phantoms irradiated with neutron and photon beams", *Bio-medical Dosimetry: Physical Aspects, Instrumentation, Calibration* (Proc. Symp. Paris, 1980), IAEA, Vienna (1981) 125.
- [56] JOHNS, H.E., CUNNINGHAM, J.R., *The Physics of Radiology*, Charles C. Thomas, Springfield (1983).
- [57] INTERNATIONAL ELECTROTECHNICAL COMMISSION, *Medical Electrical Equipment, Dosimeters with Ionization Chamber as Used in Radiotherapy*, IEC Standard Publication 731, IEC, Geneva (1986).
- [58] INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, *Measurement of Absorbed Dose in a Phantom Irradiated by a Single Beam of X- or Gamma-Rays*, ICRU Rep. 23, ICRU Publications, Bethesda (1973).
- [59] SCHNEIDER, U., "A new method for deriving the absorbed dose in phantom material from measured ion dose for X-rays generated at voltages up to 300 kV", *Biomedical Dosimetry: Physical Aspects, Instrumentation, Calibration* (Proc. Symp. Paris, 1980), IAEA, Vienna (1981) 223.
- [60] SCHNEIDER, U., GROSSWENDT, B., "Dosismessung in Phantom bei konventioneller Röntgenstrahlung (70 kV bis 300 kV) mit Ionisationskammern unter Bedingungen, die von den Bezugsbedingungen abweichen", *Medizinische Physik*, Hsrg. v. Klitzing, Laibach (1986).
- [61] KUBO, H., *Water calorimetric determination of absorbed dose by 280 kVp orthovoltage X-rays*, *Radiother. Oncol.* 4 (1985) 275.
- [62] MATTHESSON, O., Comparison of water calorimetry and ionization chamber dosimetry in 100 and 200 kV X-ray beams, Preliminary Report for the CCEMRI meeting in Paris 1985, Severs (1985).
- [63] CUNNINGHAM, J.R., WOO, M., ROGERS, D.W.O., BIELAJEW, A., The dependence of mass energy absorption coefficient ratios on beam size and depth in a phantom, *Med. Phys.* 13 (1986) 496.
- [64] BÖHM, J., HOHLFELD, K., REICH, H., "A primary standard for determination of absorbed dose in water for X-rays generated at potentials of 7.5 to 30 kV", National and International Standardization of Radiation Dosimetry (Proc. Symp. Atlanta, 1977), Vol. 1, IAEA, Vienna (1978) 53.
- [65] SCHNEIDER, U., Bestimmung der Wasser-Energiedosis im Wasserphantom mit frei in Luft kalibrierten Ionisationskammern, Jahresbericht 1985, Physikalisch-Technische Bundesanstalt, Braunschweig (1986).
- [66] BOAG, J.W., "Ionization chambers", in: *Radiation Dosimetry* (ATTIX, F.H., ROESCH, W.C., Eds), Academic Press, New York (1966) 1.
- [67] MATTHESSON, L.O., JOHANSSON, K.-A., SVENSSON, H., Calibration and use of plane-parallel ionization chambers for the determination of absorbed dose in electron beams, *Acta Radiol., Oncol.* 20 (1981) 385-399.
- [68] MUNHEER, B.J., Variation of response to radiation of a nylon-walled ionization chamber induced by humidity changes, *Med. Phys.* 11 (1986) 625.
- [69] COMITE CONSULTATIF POUR LES ETALONS DE MESURE DES RAYONNEMENTS IONISANTS (Section I), *Correction d'humidité, Recommandation CCEMRI(80) 30*, Bureau International des Poids et Mesures, Severs (1977).
- [70] BOAG, J.W., *Ionization measurements at very high intensities. I. Pulsed radiation beams*, *Br. J. Radiol.* 23 (1950) 601.
- [71] BOAG, J.W., CURRANT, J., Current collection and ionic recombination in small cylindrical ionization chambers exposed to pulsed radiation, *Br. J. Radiol.* 53 (1980) 471.
- [72] BOAG, J.W., The recombination correction for an ionization chamber exposed to pulsed radiation in a 'swept beam' technique. I. Theory, *Phys. Med. Biol.* 27 (1982) 201.
- [73] INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, *The Dosimetry of Pulsed Radiation*, ICRU Rep. 34, ICRU Publications, Bethesda, MD (1982).
- [74] WEINHOUS, M.S., MELI, J.A., Determining P_{ion} , the correction factor for recombination losses in an ionization chamber, *Med. Phys.* 11 (1984) 846.
- [75] SPENCER, L.V., ATTIX, F.H., A theory of cavity ionization, *Radiat. Res.* 3 (1955) 239.
- [76] BERGER, M.J., Personal communication to the American Association of Physicists in Medicine, 1983.
- [77] STERNHEIMER, R.M., The density effect for the ionization loss in various materials, *Phys. Rev.* 88 (1952) 851.
- [78] CARLSSON, T.A., Atomic binding energies for each subshell for elements $Z = 1 - 106$, *Photoelectron and Auger Spectroscopy*, Plenum Press, New York (1975) 337.
- [79] BRUNVIS, I.A.D., HEUKLOM, S., MUNHEER, B.J., Comparison of ionization measurements in water and polystyrene for electron beam dosimetry, *Phys. Med. Biol.* 30 (1985) 1043-1053.
- [80] BERGER, M.J., SELTZER, S.M., Stopping Powers and Ranges of Electrons and Positrons, Rep. NBSIR-82-2550, Natl Bur. Stand., Gaithersburg, MD (1982).
- [81] ANDREO, P., BRAHME, A., Stopping power data for high-energy photon beams, *Phys. Med. Biol.* 31 (1986) 839.
- [82] ANDREO, P., NAHUM, A.E., BRAHME, A., Chamber-dependent wall correction factors in dosimetry, *Phys. Med. Biol.* 31 (1986) 1189.
- [83] GROSSWENDT, B., Physikalisch-Technische Bundesanstalt, Braunschweig, Federal Republic of Germany, private communication, 1986.
- [84] GROSSWENDT, B., Backscatter factors for X-rays generated at voltages between 10 and 100 kV, *Phys. Med. Biol.* 29 (1984) 579.

- [85] SKARSGARD, L.O., JOHNS, H.E., Spectral flux density altered and primary radiation generated at 250 kV, *Radiat. Res.* 14 (1961) 231.
- [86] GUNNINGHAM, J.R., Work carried out for this working group, 1986.
- [87] HENRY, W.H., On C_{α} , C_{β} , and the effective wall material of a Baldwin-Farmer chamber, *Phys. Med. Biol.* 24 (1979) 37.
- [88] ALMOND, P.R., SVENSSON, H., Ionization chamber dosimetry for photon and electron beams. Theoretical considerations, *Acta Radiol., Ther., Phys., Biol.* 16 (1977) 177.
- [89] LEMPET, G.D., NATH, R., SCHULTZ, R.J., Fraction of ionization from electrons arising in the wall of an ionization chamber, *Med. Phys.* 10 (1983) 1.
- [90] DUFTSCHMID, K., Austrian Research Centre, Seibersdorf, A-1082 Vienna, Private communication (1986).
- [91] KRISTENSEN, M., Measured influence of the central electrode diameter and material on the response of a graphic ionization chamber to cobalt-60 gamma rays, *Phys. Med. Biol.* 28 (1983) 1269.
- [92] MATTSSON, L.O., Application of the Water Calorimeter, Fricke Dosimeter and Ionization Chamber in Clinical Dosimetry, An Evaluation of Correction Factors and Interaction Coefficients, Thesis, Univ. Gothenburg, Sweden (1984).
- [93] LOEVINGER, R., LOFTUS, T.P., "Uncertainty in the delivery of absorbed dose", Ionizing Radiation Metrology, International Course, Varese, Italy, 1974, Editrice Compositori Bologna (1977).
- [94] MÜLLER, J.W., Some second thoughts on error statements, *Nucl. Instrum. Methods* 163 (1979) 241-251.
- [95] GIACOMO, P., New from the BIPM, *Metrologia* 17 (1981) 69.
- [96] KAARLS, R., "Rapport du groupe de travail sur l'expression des incertitudes au CIPM", *BIPM Proc. Verb. Com. Int. Poids et Mesures* 49 Annexe A (1981).
- [97] ALLSY, A., in Proc. Int. School of Phys. Enrico Fermi, Course LXVIII, Metrology and Fundamental Constants (FERRO MILONE, A., GIACOMO, P., LESCHULT-TRA, S., Eds), North-Holland, Amsterdam (1980) 20, also Bulletin BNM No. 53 (1983) 5.
- [98] MÜLLER, J.W., in IMEKO Interregional Training Course, Seibersdorf, 1984.
- [99] DIN Norm 1319, Teil 4, Grundbegriffe der Messtechnik - Behandlung von Unsicherheiten bei der Auswertung von Messungen, Beuth Verlag, Berlin (West) (1985).



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