

Chapter 3

Recent Advances in Electron and Photon Dosimetry

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1. INTRODUCTION

The possibilities to improve radiation therapy have increased during recent years, not only because of the use of new or improved tools such as computed tomography and dose planning, and high-quality electron and photon beams from therapy accelerators, but also because of increased knowledge in fields like clinical radiation biology about dose fractionation and dose-response relations. These new developments increase the demand for accurate dosimetry as illustrated by the following examples.

1. One department may today use several different radiation qualities to treat a special type of cancer, and the choice of radiation quality is individually optimized to the anatomy of each patient. To give consistent treatments it is therefore necessary to be able to give the same absorbed dose independent of the beam quality;
2. The CT information together with accurate computed dose planning systems facilitates the correction of the dose distribution for the density and composition of the various tissues in the body. Accurate input dose distribution data are needed if the advantages of a precision computer system are to be exploited fully;

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3. Finally, a more accurate analysis of dose-effect curves will be possible as corrections for various fractionation schedules are better known today (e.g., from calculations of the NSD, TDF, or CRE). The accuracy in the dose determination and dose delivery is in many cases the limiting factor in such an analysis.

The ICRU 24⁷⁵ has concluded that there is evidence, for certain types of tumors, that the accuracy in the determination of absorbed dose to the target volume should be within $\pm 5\%$ or even lower if an eradication of the tumor is sought. Recent intercomparisons in the Scandinavian countries⁸² show that this accuracy is not even obtained at the reference depth in a water phantom. Even larger uncertainties have been shown in other surveys (e.g., Refs. 48, 124, 148).

There is thus a great need for improving the dosimetry. However, today this is a fairly complicated task as no single dominant reason for the large uncertainty can be found. Instead, Johansson and Svensson⁸² showed that the total uncertainty is due to a combination of a number of different systematic errors and random uncertainties. In this analysis it was assumed that the ionization chamber method was used for the absorbed dose determination. An improvement of the dosimetry can therefore only be achieved if most of the uncertainties in the calibration procedure can be reduced or if a completely different method is applied. The aim in the present report is to analyze the complete dosimetrical procedure for the determination of absorbed dose in a phantom.

The report contains both original material and a review of recent publications. It starts by describing the basic dosimetric quantities and their physical relations (Section 2). The special problems of spatially extended detectors including such problems as fluence perturbation and effective point of measurement are dealt with in Section 3. The basic physical data and the implications of the most recent stopping power values¹¹ for the determination of absorbed dose are discussed in Section 4. Finally, more practical problems in the calibration of a medical accelerator are analyzed in Section 5.

2. BASIC DOSIMETRIC QUANTITIES AND THEIR PHYSICAL RELATIONS

2.1. Energy Deposition

A principal problem in the description of the interaction of electron and photon beams with matter is that all real beams consist of a finite number of discrete particles, whereas they are most conveniently de-

scribed by continuously variable probability distributions. For this reason the most fundamental concepts in radiation dosimetry are stochastic quantities. This is the case with the energy imparted ε , defined as the difference in energy of particles incident on and emerging from the volume under consideration plus the decrease in rest mass energy that might occur inside that volume.⁷⁷ When the energy imparted is due to a very large number of energy depositions it is often sufficient to use the expectation value of the stochastic quantity ε , i.e., the mean energy imparted $\bar{\varepsilon}$.

The response of most radiation dosimeters is proportional to the energy imparted. Such a detector, when appropriately calibrated, can be used to determine the energy imparted in the radiation sensitive volume according to

$$\varepsilon = N_{\varepsilon} M \quad (1a)$$

where M is the value of the detector signal and N_{ε} is the energy imparted calibration factor. For the most common case in dosimetry, the gas-filled ionization chamber, $M = Q_g$, that is the recombination corrected charge collected in the gas volume, and $N_{\varepsilon} = W_g/e$ is the mean energy needed to generate an ion pair in the gas. For this simple case Eq. (1a) becomes

$$\varepsilon_g = \frac{W_g}{e} Q_g \quad (1b)$$

The stochastic quantity for the energy absorbed per unit mass is called the specific energy imparted, z , and is obtained by dividing the energy imparted by the mass of the volume under consideration $z = \varepsilon/m$. The expectation value of the specific energy imparted, $\bar{z}$, is by definition equal to the mean value of the absorbed dose in the volume considered:

$$\bar{z} = \frac{\bar{\varepsilon}}{m} = \bar{D} \quad (2)$$

The absorbed dose at a point is thus defined as the differential quotient:

$$D = \frac{d\bar{\varepsilon}}{dm} = \lim_{m \rightarrow 0} \bar{z} \quad (3)$$

which thus coincides with the limiting value of the specific energy when the volume of interest or the mass approaches zero (cf. Ref. 77).

By using Eq. (2), Eq. (1) may be rewritten in the following form:

$$\bar{D} = N_D \bar{M} \quad (4a)$$

where $\bar{M}$ is the expectation value of the detector signal, $\bar{D}$ the mean absorbed dose to the detector, and N_D the absorbed dose calibration factor, which is related to N_ϵ by $N_D = N_\epsilon / m$.

When we apply these equations on a normal air ionization chamber Eq. (4a) takes the form

$$\bar{D}_{\text{air}} = \frac{W_{\text{air}}}{em_{\text{air}}} Q_{\text{air}} \quad (4b)$$

which may be rewritten

$$\bar{D}_{\text{air}} = \frac{W_{\text{air}}}{e} J_{\text{air}} \quad (4c)$$

where J_{air} is the specific ionization or mass ionization in the air volume ($Q_{\text{air}}/m_{\text{air}}$) assuming the detector signal is measured in this unit.

For uncharged particles like photons it is useful to define an auxiliary field quantity K , the kerma, defined as the initial kinetic energy of all charged particles liberated per unit mass.⁷⁷ Because part of this kinetic energy will be converted back to photons through bremsstrahlung and annihilation in flight processes, it is of interest to distinguish that part of the kerma that remains as kinetic energy of charged particles, the collision kerma, K_{col} .⁸ This latter quantity is of special interest as the collision kerma in air is directly related to the traditional unit exposure, X , through the relation

$$K_{\text{col,air}} = X \frac{W_{\text{air}}}{e} \quad (5)$$

where W_{air} is the mean energy expended in air per ion pair formed and e is the electron charge. Because the collision kerma is an expression of the energy in a photon beam that is permanently converted to kinetic energy of charged particles, the three quantities mean energy imparted, collision kerma, and absorbed dose are related through the equality

$$\bar{\epsilon}_V = \iiint_V K_{\text{col}}(\mathbf{r}) \rho(\mathbf{r}) dV = \iiint_V D(\mathbf{r}) \rho(\mathbf{r}) dV \quad (6)$$

where $\rho(\mathbf{r})$ is the density distribution in the irradiated volume V . This relation can be used to determine the kerma distribution in a medium where the absorbed dose distribution is known.¹¹¹

2.2. Fluence and Absorbed Dose

The physical quantity which describes the radiation field is the fluence, Φ , defined by the ICRU as the quotient of the number of particles dN incident on a sphere and its cross-sectional area da :

$$\Phi = \frac{dN}{da} \quad (7)$$

This definition can be generalized by multiplying both dN and da by the mean cord length $\bar{l} = 4dV/dA$ of the sphere; see Fig. 1. Because $\bar{l}dN$ is just equal to the total path length, $d\bar{s}$, and $\bar{l}da$ is the volume dV , dA being the surface area, the fluence can also be defined as

$$\Phi = \frac{d\bar{s}}{dV} \quad (8)$$

This definition is more general than Eq. (7) as it is independent of the shape of the volume considered and avoids the problem of whether the incident particles just enter the volume or really cross it. By the latter definition the fluence is easily visualized as the total path length per unit volume.

It is important that the definitions (7) and (8) contain all particles independent of energy and direction. However, it is easy to generalize the fluence concept and define a fluence differential in energy and angle

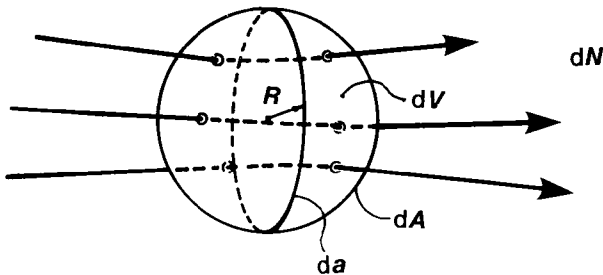


Figure 1. Definition of the fluence of particles as dN/da the number crossing a sphere of cross section da or $d\bar{s}/dV$, the path length per unit volume. When the radius of the sphere is R , then $da = \pi R^2$, $dV = \frac{4}{3}\pi R^3$, and $dA = 4\pi R^2$.

$$\bar{l} = \frac{4R}{3}$$

$\Phi_{E,\Omega}$ such that $\Phi_{E,\Omega} dE d\Omega$ contains all the particles of energy between E and $E + dE$ with direction in the interval Ω to $\Omega + d\Omega$ of solid angles. Based on this differential fluence all other fluence concepts can be defined, e.g., the fluence differential in energy:

$$\Phi_E = \int \Phi_{E,\Omega} d\Omega \quad (9)$$

and the vectorial energy fluence:

$$\Psi = \iint \Omega E \Phi_{E,\Omega} dE d\Omega \quad (10)$$

This latter expression is of special interest as it can be used to calculate the mean energy imparted and the absorbed dose from first principles.

When, for simplicity, the changes in rest mass energy can be disregarded, the mean energy imparted inside a closed surface A is given by the net energy flow across this surface:

$$\bar{\varepsilon}_V = \oiint_A \Psi \mathbf{n} dA = - \iiint_V \text{div } \Psi dV \quad (11)$$

where Gauss' law has been used in the last equality, $\mathbf{n}$ is a unit vector normal to A , and V is the volume enclosed by A . By allowing A and V to decrease to zero, Eq. (3) can be used to express the absorbed dose at a point $\mathbf{r}$:

$$D(\mathbf{r}) = - \frac{1}{\rho(\mathbf{r})} \text{div } \Psi(\mathbf{r}) \quad (12)$$

The absorbed dose distribution can now be calculated from the spatial distribution of the vectorial energy fluence of all contributing particles. By using Eq. (10), assuming Φ includes the total fluence of electrons, and by changing the order of the integration and the divergence operations, the absorbed dose becomes

$$D = \frac{-1}{\rho} \iint (\Omega \Phi_{E,\Omega} \text{grad}(E) + \Omega E \text{grad}(\Phi_{E,\Omega})) dE d\Omega \quad (13)$$

because $\text{div } \Omega = 0$ as Ω by definition is an independent variable. When E is regarded as a variable depending on the space coordinates along the

particle path l and after the Ω integration is performed, Eq. (13) becomes

$$D = - \int \left(\frac{1}{\rho} \frac{dE}{dl} \Phi_E + \frac{E}{\rho} \frac{d\Phi_E}{dl} \right) dE \quad (14)$$

This relation shows that two types of energy depositions by the electrons can be distinguished: quasicontinuous energy losses described by the first term, or track ends described by the second term. In the Spencer–Attix formulation Eq. (14) is closely approximated by

$$D = \int_{\Delta}^{\infty} \frac{L_{\Delta}(E)}{\rho} \Phi_E dE + \frac{\Delta}{\rho} S_{\text{col}}(\Delta) \Phi_E(\Delta) \quad (15)$$

where the last term gives the track end contribution (cf. Refs. 31 and 103 and also the discussion by Svensson and Brahme¹³⁵).

Under assumption of an equilibrium in the transport of secondary electrons, such that the outflow through each volume element is compensated by an equal inflow, Eqs. (14) and (15) can be simplified considerably

$$D_{\text{eq}} = \int \frac{S_{\text{col}}(E)}{\rho} \Phi_E^p dE \quad (16)$$

where Φ_E^p is the fluence of primary electrons. An alternative expression, when the electrons are set in motion by photons, is obtained if the energy fluence differential in energy of the photons, Ψ_E , is used in Eq. (13):

$$D_{\text{eq}} = \frac{-1}{\rho} \iint \Omega \text{grad } \Psi_E dE d\Omega \quad (17)$$

which in analogy with Eq. (14) becomes

$$D_{\text{eq}} = - \int \frac{1}{\rho} \frac{d\Psi_E}{dl} dE = \int \frac{\mu_{\text{en}}(E)}{\rho} \Psi_E dE \quad (18)$$

where $\mu_{\text{en}}(E)$ is energy absorption coefficient of photons of energy E . Under equilibrium conditions both Eqs. (16) and (18) can be used, whereas in absence of complete equilibrium in the electron transport the basic expressions Eqs. (14) and (15) have to be applied.

3. GENERAL THEORY OF RADIATION DOSIMETERS

3.1. Background

Since the earliest theories about the response of radiation detectors (cf, e.g., Ref. 19) it has been well known that the material surrounding the radiation sensitive part of a small detector is of critical importance. This is so because, independent of the energy of the beam, about 30% of the absorbed dose is due to low-energy secondary electrons, the range of which is only about 0.5 mm or less in unit density material. Over such distances there will be a transition zone near the interface between two materials, where the slowing down spectrum of electrons changes from that characteristic of one material to that of the next (cf. Refs. 77 and 14). This implies that the response of a small detector will depend on the atomic number of the surrounding medium.

There are at least three different approaches to dosimetry depending on how this central problem is handled. The traditional Bragg-Gray type approach is to use such a small detector that the electron spectrum existing in the medium will penetrate the detector and not be affected by its presence. Strictly speaking, this is theoretically impossible, as no interactions can then be allowed in the detector and thus no signal will be obtained. In particular, it is a very poor assumption when the elemental composition of the detector differs from that of the medium, as interface phenomena will then have a strong influence and the important low-energy part of the spectrum will be affected (cf. Ref. 150).

The second approach is to make the dosimeter of finite size with only the radiation sensitive part different from the medium. This simplifies the theoretical treatment, and, for example, a Spencer-Attix type theory could be applied.^{13,31,103,126} A typical example is the use of a water equivalent ionization chamber in water with only the air volume as the nonwater equivalent part (see Fig. 2a).

The third possibility is to use a detector which is surrounded by a wall of the same material as used in the radiation sensitive part of the detector. This approach will ensure that the interface effects will mainly disturb the dose distribution in the wall, and the low-energy spectrum in the radiation sensitive part of the detector will be fully built up. In the first approximation, the simple theory of Harder covers this situation (Ref. 62; cf. also Refs. 20 and 35). A typical example here is an air ionization chamber with air equivalent walls when used in water (see Fig. 2b).

In many cases the most accurate dosimetry is performed when one of the above approaches is applied to a detector which is as similar as possible in elemental composition to the medium. This is the case with

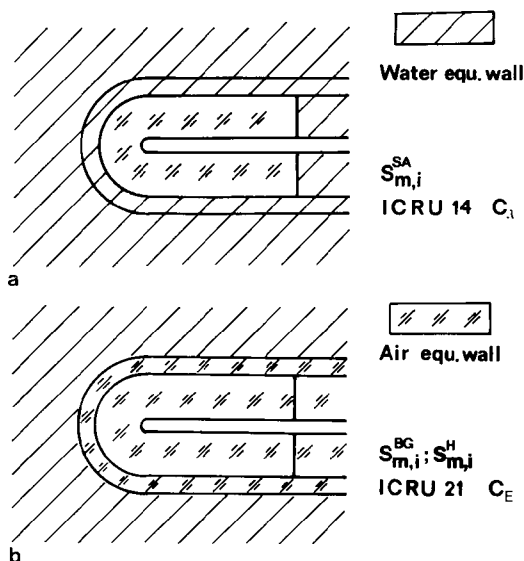


Figure 2. Illustration of the two principal types of ionization chambers used in electron and photon dosimetry. The two-media detector (a), with phantom equivalent walls (m) around the radiation sensitive material (i), has generally been used in photon beams (cf. ICRU 14) and should strictly be used with a Spencer-Attix type stopping power ratio [cf. Eq. (40)]. The matched detector (b), with walls of the same material as the radiation sensitive part of the detector, is often assumed in electron beams (cf. ICRU 21) and could be used with a Bragg-Gray stopping power ratio [Eq. (38)], or often equally well with the approximation due to Harder [eq. (39)]. In photon beams a difference in mass ionization of about 3% may result between these two detector types (cf. Ref. 2).

detector systems like ferrous sulfate, liquid ionization chambers, and tissue equivalent or water calorimeters. These detector systems are of special importance when consistent dosimetry at quite different radiation qualities is required.

One of the most important requirements of a theory of the response of radiation dosimeters is that all manipulations of the detector signal are performed with internal consistency. For example, it must be assured that the stopping power ratio, normally calculated for the same energy spectrum in the detector material and in the medium, really is applicable. This requires that the electron spectrum at the effective point of measurement (see Sections 3.4.2 and 3.4.3) in the medium is the same as that averaged over the radiation sensitive part of the detector. Similarly, it is essential that necessary fluence perturbation factors (see Sections 3.4.2 and 3.4.4) really are based on the fluence in the medium at the effective point of measurement where the determined absorbed dose later is described.

Most published cavity theories have shortcomings in these respects. One of the aims with the present treatment is to give a consistent set of equations defining the most important quantities already identified by ICRU 14⁷³ and 35⁷⁴ in a stringent and consistent way, namely, the stopping power ratio, the perturbation correction factor, and the effective point of measurement.

Because the user of a calibrated dosimeter will make measurements in many different beam qualities, the definitions will sometimes differ from those used by the standard laboratories where mainly ^{60}Co γ photons are used. For example, in the present treatment a more general reference volume concept (see Section 3.4.2) will be used instead of the chamber center which in clinical use necessitates more or less complex gradient corrections to be performed on the detector signal.

3.2. Fano's Theorem

The fundamental properties of the secondary electron spectrum in low- (or high-) density cavities are made use of in many radiation dosimeters as was described in detail already by Bragg.¹⁹ However, a strict formal proof was first given by Fano.⁵² A slightly generalized formulation of "Fano's" theorem, which is approximately applicable to radiation dosimeters, is as follows: In a medium of uniform composition, but of varying density, and with a local source density of generated particles which is a given function of the density, the fluence of these particles is uniform and independent of the density at least one particle range inside the volume where this condition is fulfilled, provided the interaction cross sections for these particles with the medium have the same density dependence as the source density (cf. Refs. 51 and 65).

This implies that the fluence of secondary electrons in photon or electron beams (photoelectric, Compton, and pair electrons and positrons in photon beams, and δ rays in electron beams) is uniform and independent of density variations in a volume V , provided the fluence of primary photons or electrons is uniform in V and at least one maximum secondary electron range outside V . This very general statement relies on the fact that the cross section for secondary electron production is proportional to the electron density N_e and so is the interaction cross section for the secondary electrons. Intuitively this can be understood as follows. When the density is high the number of secondary electrons produced per unit volume is high but their range is short so their fluence or path length per unit volume (cf. Section 2.2) is the same as if the density were low, with a long range of the secondary electrons but a small number produced per unit volume.

A more formal proof is almost as straightforward if the time-

independent Boltzman equation describing the spatial distribution of the differential fluence $\Phi_{E,\Omega}(\mathbf{r})$ is considered (cf. Ref. 44):

$$\begin{aligned} \Omega \text{ grad } \Phi_{E,\Omega}(\mathbf{r}) = & \int_E^\infty N_e(\mathbf{r}) \Phi_{E',\Omega'}(\mathbf{r}) \sigma(E', \Omega', E, \Omega) dE' d\Omega' \\ & - \Phi_{E,\Omega}(\mathbf{r}) \int_0^E \int N_e(\mathbf{r}) \sigma(E, \Omega, E', \Omega') dE' d\Omega' + S_{E,\Omega}(\mathbf{r}) \end{aligned} \quad (19)$$

Here $N_e(\mathbf{r})$ is the electron density, the term $S_{E,\Omega}(\mathbf{r})$ describes the source density, the term $\Omega \text{ grad } \Phi$ describes the collision free motion of the particles, the positive collision integral describes the inflow of particles of higher energies to the energy E and direction Ω , and the corresponding negative integral the collisional loss of particles from this energy and angle to lower energies. In the case of a uniform collision and source density, $N_e(\mathbf{r})\sigma = N_e^u\sigma$ and $S_{E,\Omega}(\mathbf{r}) = S_{E,\Omega}^u$ there is no spatial dependence and the fluence will also be uniform in space, $\Phi_{E,\Omega}(\mathbf{r}) \equiv \Phi_{E,\Omega}^u$ and consequently $\text{grad } \Phi^u \equiv 0$.

However, when the collision density is proportional to the density or some function of the density, $N_e(\mathbf{r})\sigma = f(\rho(\mathbf{r}))N_e^u\sigma$, and the source density is the same function of the density, $S_{E,\Omega}(\mathbf{r}) = f(\rho(\mathbf{r}))S_{E,\Omega}^u$, the term $f(\rho(\mathbf{r}))$ is common to all terms on the right-hand side in the Boltzman equation:

$$\begin{aligned} \Omega \text{ grad } \Phi = f(\rho(\mathbf{r})) \left\{ \int_E^\infty N_e^u \Phi_{E',\Omega'}(\mathbf{r}) \sigma(E', \Omega', E, \Omega) dE' d\Omega' \right. \\ \left. - \Phi_{E,\Omega}(\mathbf{r}) \int_0^E \int N_e^u \sigma(E, \Omega, E', \Omega') dE' d\Omega' + S_{E,\Omega}^u \right\} \end{aligned} \quad (20)$$

When this is observed it is immediately clear that $\Phi_{E,\Omega}^u$ will also make the right-hand side of the nonuniform equation zero, as the expression inside the brackets is just the right-hand side of the uniform equation. Thus $\Phi_{E,\Omega}^u$ is also a solution to the whole nonuniform equation since $\text{grad } \Phi^u$ is already known to be zero. According to the uniqueness theorem this is also the only solution of the nonuniform case.

This very general result relies on at least three conditions that are rarely strictly met in practical dosimetry:

1. The fluence of primary electrons or photons is seldom constant

over a sufficiently large volume to make $N_e(\mathbf{r})\sigma$ and $S_{E,\Omega}(\mathbf{r})$ proportional to each other.

2. The Möller cross section σ is modified by Fermi's density effect at relativistic energies and this makes the density dependence of the collision integrals and the source term somewhat different.
3. The wall around the radiation sensitive material of the detector is often not strictly detector equivalent, which again violates the proportionality between the collision integrals and the source term.

In these cases the fluence of secondary electrons in the dosimeter will differ from that in the surrounding uniform medium and a fluence perturbation correction factor should in principal be introduced to compensate for this. By designing the detector appropriately the perturbation of the secondary electron fluence can often be sufficiently reduced to bring the dosimetric uncertainty below the 1% level without using a perturbation correction factor.

3.3. Bragg–Gray Detectors

For a detector which is small relative to the spatial variation of the electron fluence in the uniform medium and also small enough not to disturb the electron fluence in the medium as it penetrates the detector, the Bragg–Gray relation

$$D_m = D_i s_{m,i} \quad (21a)$$

can be used for calculating the absorbed dose D_m to the medium at the point of interest from the absorbed dose D_i to the detector, where $s_{m,i}$ is the collision mass stopping power ratio which accounts for the differences in energy deposition per unit mass in materials m (medium) and i (detector). For such a small detector it is sufficient to evaluate $s_{m,i}$ for the electron spectrum at the point of interest in the uniform medium, as this, by definition, is identical to the spectrum inside the detector.

For medium equivalent detectors it is in the first approximation only necessary that the detector be small compared to fluence variations in the medium, as the fluence in the detector will be independent of density according to Fano's theorem.

If the slowing down properties of the detector material differ from those of the medium, it has recently been shown by Zheng-Ming¹⁵⁰ that, no matter how small the cavity is, the change in the low-energy part of the electron spectrum can never be neglected because the spectrum is proportional to E^{-2} .

For a small medium equivalent gas ionization chamber the Fano

condition can often be fulfilled with fairly high accuracy. The Bragg–Gray relation for a small gas cavity can thus be written using Eq. (4c)

$$D_m = \frac{W_i}{e} s_{m,i} J_i \quad (21b)$$

When the influence of different basic quantities on the dosimetric accuracy is investigated it is convenient to combine the factors in front of the specific ionization according to

$$D_m = \omega_{m,i} J_i \quad (21c)$$

The ratio of the absorbed dose to the specific ionization is thus given by the quantity $\omega_{m,i}$.

A further advantage of this simplified version of the Bragg–Gray relation is that ω is exactly the quantity determined in experiments where D_m and J_i are known, e.g., by calorimetric and ionometric measurements (cf. Refs. 43 and 108).

3.4. Extended Detectors

3.4.1. Definitions

In practical dosimetry it is generally not possible to construct a sufficiently small detector with adequate sensitivity. Most practical detectors are thus large relative to fluence variations in the medium and they perturb the fluence in the medium. The signal from an extended detector of different slowing-down and/or scattering properties therefore differs from that of a Bragg–Gray detector of negligible spatial extension, and the basic assumptions in the Bragg–Gray relation are violated in two principal respects:

1. The electron fluence will be sampled over an extended volume, which means that the response of the detector is proportional to the mean value of the absorbed dose distribution inside its radiation sensitive volume ($\bar{D}_i$) rather than being a well-defined point value (D_i).
2. The electron fluence averaged over the radiation sensitive part of the detector volume will in general differ from that at the point of interest in the uniform medium in absence of the detector. Three principally different types of fluence perturbations can be identified as discussed in detail in Section 3.4.4 and illustrated schematically in Fig. 5.

These two effects of spatially extended detectors have to be taken into account in the generalization of the Bragg-Gray relation.

3.4.1.1. Reference Volume. The averaging property of an extended detector can be taken into account by relating the mean absorbed dose inside the detector volume ($\bar{D}_i$) to that inside a similarly shaped reference volume in the unperturbed medium ($\bar{D}_m$) (see Fig. 3). This is the natural generalization of an extended medium-equivalent detector, which will necessarily measure a mean value over its radiation sensitive volume.

It is desirable that the basic form of the Bragg-Gray relation is retained for extended detectors. Therefore the dimensions of the reference volume should be chosen such that the shape of the electron spectrum (i.e., the relative electron energy distribution, irrespective of absolute normalization) averaged over the detector volume is as similar as possible to the spectrum averaged over the reference volume in the unperturbed medium. This requirement is due to the fact that most formulations of stopping power ratios are based on the same electron

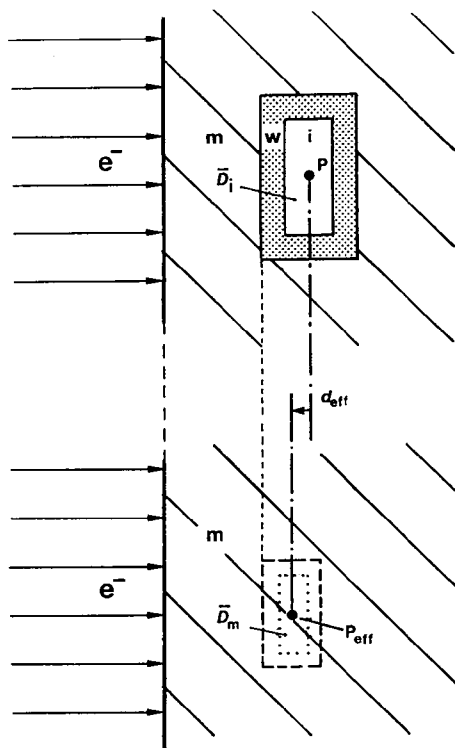


Figure 3. Definition of the reference volume in the uniform medium, the dash-dotted line in the lower half of the figure, and the effective point of measurement, P_{eff} , for solid and liquid detectors. In this figure it is assumed that the radiation sensitive material of the detector (i) is surrounded by a wall (w) of the same material, and that the linear stopping power is lower than for the medium (m). The mean absorbed dose inside the detector is $\bar{D}_i$ and the one inside the dash-dotted reference volume is $\bar{D}_m$. Under the assumption of a linear dose variation over the detector, the effective point of measurement will be the center of the reference volume.

energy spectrum, both for the detector and the medium. The present choice of reference volume will therefore allow the use of existing tabulations of stopping power ratios.

The reference volume is particularly simple to identify when the energy straggling in detector and medium is similar and a negligible disturbance of the energy distribution in the detector is produced by electrons that are scattered laterally into the detector from the medium (e.g., a flat detector, cf. Fig. 5c). The reference volume in the uniform medium should simply be chosen such that the mean energy loss by an electron in crossing the detector is the same as in crossing the reference volume in the medium. This implies that all linear dimensions of the reference volume in the uniform medium, t_m , are obtained by multiplying the corresponding dimension of the detector volume, t_d , by the ratio of the total linear stopping power of detector to medium evaluated at the mean energy of the primary electrons in the detector volume:

$$t_m = \frac{S_{\text{tot},i}}{S_{\text{tot},m}} t_d \quad (22)$$

This scaling is illustrated in Fig. 3 and was used already by Gray^{58,59} and more recently by Brahme and Lax.²⁵ This choice will thus make the first two moments of the energy distribution, the mean energy and the energy spread, equal in the detector and the reference volume. The special importance of the mean energy of the primary electrons for electron and photon dosimetry due to the energy dependence of the stopping power ratio, through the influence of the density effect in gas-filled detectors, is well known.^{6,20,62,135}

When parameters other than the absorbed dose in a broad uniform beam are of prime interest the reference volume must be chosen in a different way. For example, when the relative dose profile across the beam in a certain direction is determined the reference volume should be chosen equal to the true dimension of the detector in the direction of measurement.²²

3.4.1.2. Effective Point of Measurement. It is in practice more convenient to use an effective point of measurement, to which the absorbed dose in the medium may be allocated, rather than a reference volume. When the distribution of absorbed dose in the uniform medium is a continuously varying, monotonic function of the space coordinates there will exist one and only one point inside the reference volume (according to the mean value theorem) where the absorbed dose is exactly equal to the mean dose inside that volume:

$$D_m(P_{\text{eff}}) = \bar{D}_m \quad (23)$$

This point, P_{eff} , is therefore defined as the effective point of measurement and the “upstream” distance between P and P_{eff} is defined as the shift of the effective point of measurement, d_{eff} :

$$d_{\text{eff}} = z_p - z_{p_{\text{eff}}} \quad (24)$$

3.4.1.3. Fluence Perturbation. There will always be a perturbation of the electron fluence of the medium in the presence of an extended detector when the electron slowing down and/or scattering properties of the detector differ from those of the medium. The different types of fluence perturbation are discussed in detail in Section 3.4.4.

The total effect of the fluence perturbations is that even if the shape of the electron energy spectrum in the detector volume is not significantly different from that in the reference volume, the absolute values of the mean electron fluences in these two volumes may differ. The mean fluence inside the reference volume, $\bar{\Phi}_m$, can be related to that inside the detector volume, $\bar{\Phi}_d$, by defining a fluence perturbation factor $p_{m,i}$ according to

$$\bar{\Phi}_m = p_{m,i} \bar{\Phi}_d \quad (25)$$

The perturbation factor is unity for a small “Bragg–Gray” detector or a detector of the same composition and density as the medium but departs from unity with increasing size and difference in linear electron stopping and scattering properties between detector and medium.

3.4.1.4. Generalized Absorbed Dose Relation. By the above choice of the reference volume the basic Bragg–Gray relation, Eq. (21), can be generalized to apply to extended detectors by introducing the mean dose values $\bar{D}_m$ and $\bar{D}_d$, averaged over the reference and detector volumes, respectively, and by multiplication by the fluence perturbation correction factor $p_{m,i}$ which accounts for the difference in mean fluence in the two volumes:

$$\bar{D}_m = \bar{D}_d s_{m,i} p_{m,i} \quad (26)$$

Using the concept of an effective point of measurement the Bragg–Gray relation for extended detectors can also be written

$$D_m(P_{\text{eff}}) = \bar{D}_d s_{m,i} p_{m,i} \quad (27)$$

It should be pointed out here that the above procedure of defining a perturbation correction factor and an effective point of measurement in

connection with a reference volume is not the only way of generalizing the Bragg–Gray relation to extended detectors. An important advantage of this method, however, is that it allows a strict separation of the definition of the effective point of measurement from that of the perturbation correction factor in a straightforward way and further that it is a general approach which is applicable both for low-density gaseous detectors and solid or liquid detectors. It is also of fundamental importance that the shape of the energy distribution spectrum of the electrons in the detector volume and that at the effective point of measurement in the uniform medium should be as similar as possible to validate the use of existing theoretical expressions and tabulations of stopping power ratios.

3.4.2. Effective Point of Measurement for Gaseous Detectors

3.4.2.1. Theory. The effective point of measurement, as defined in Section 3.4.1, may be located at an appreciable distance from the detector center when the linear stopping power of the detector differs substantially from that of the medium. This is the case particularly for gaseous detectors.

As an illustrative example the distance from the effective point of measurement to the central axis of a cylindrical gas-filled detector of radius r , with the axis perpendicular to the direction of the electron beam (Fig. 4), will be evaluated under two different sets of simplifying assumptions.

In the first set it is assumed that the electron fluence varies linearly in the direction of the beam over the diameter of the gas volume but that the mean electron energy does not vary appreciably over the diameter of the gas volume. These two conditions are generally met in photon beams and approximately also over the steep dose fall-off section of the electron depth–dose curve.

In the second set it is assumed that the variation of the electron fluence or absorbed dose over the diameter of the gas volume is fairly small and that the mean electron energy in the medium decreases linearly over the diameter of the gas volume. These latter conditions are complementary to the first set of conditions and are generally met over the therapeutically useful plateau part of the electron depth dose curve.

The following four assumptions are common to both sets:

1. all electrons entering the gas volume move parallel to the beam direction;
2. the number of ions generated by an electron crossing the gas volume is proportional to the chord length;

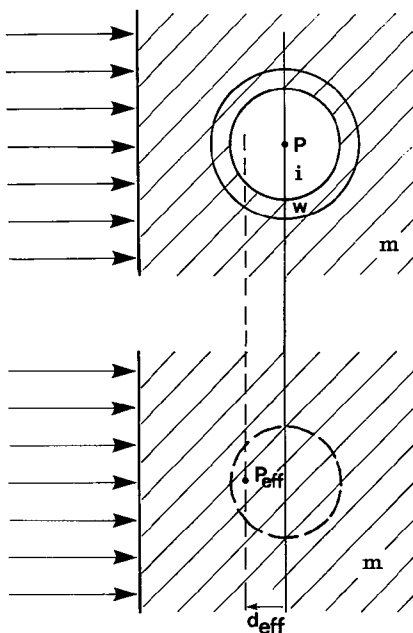


Figure 4. The shift of the effective point of measurement P_{eff} , from the center P , of a cylindrical gas-filled detector. The shift d_{eff} is defined as the upstream distance from the detector center P to P_{eff} . Owing to the low density of the gas relative to the medium, the size of the reference volume (cf. Section 3.4.2) will be negligible for gaseous detectors and it will be located at P_{eff} .

3. the wall of the detector is medium equivalent;
4. the influence of the central electrode can be neglected.

For the first set of conditions it can be calculated that the average mass ionization in the cylindrical probe corresponds to an effective point of measurement located upstream at a distance $d_{\text{eff}} = 8r/3\pi \sim 0.85r$ from the cylinder axis.^{46,66,125} This result holds for a decreasing as well as an increasing fluence.

For the second set of conditions the effective point of measurement will be located at that point in the unperturbed medium where the mean electron energy is equal to that in the gas volume. This is so since the density of the gas is only 10^{-3} of that in the medium and thus the reference volume will be very small and in practice coincide with the effective point of measurement. Based on the second set of conditions, it can be shown that the mean electron energy in the gas volume will be equal to that at a point in the uniform medium, P_{eff} , located a distance $d_{\text{eff}} = 8r/3\pi$ upstream of the cylinder axis, P (Fig. 4).

For both sets of assumptions the same effective point of measurement is thus obtained. The reason for this is that in both cases a linear variation of the variable under consideration is assumed. This indicates that, in the transition region between the first and second set of assumptions, the same effective point of measurement could also be used.

Indeed, a simultaneous assumption of linear variation in energy and fluence results in a small additional shift of P_{eff} depending on the steepness and sign of the fluence variation relative to the energy variation.

However, several other factors will in practice influence the location of the effective point of measurement. The most important factor is that the electrons generally have a wide angular distribution. To make the above analysis applicable this has to be taken into account. For example, if a fraction f of the electrons enters the gas volume parallel to the beam direction, whereas a backscattered fraction g enters the volume in an antiparallel direction, and assuming the fluences at both the front and the back surfaces have constant negative or positive gradients in the beam direction, the location of the effective point of measurement is $(8r/3\pi)(f - g)^{66}$ upstream from the chamber center of a cylindrical chamber.

An even more realistic description is obtained if the true angular distribution of the electrons is taken into account. The shift of the effective point of measurement then becomes $(8r/3\pi)\overline{\cos \Theta} \approx (8r/3\pi)[1 - \overline{\Theta^2}(z)/2]$, where $\overline{\Theta^2}(z)$ is the mean square angular spread of the electrons in the forward hemisphere at the depth z of the chamber. When both the angular spread and the backscattered electrons are included simultaneously the distance may be estimated from

$$d_{\text{eff}} = \frac{8r}{3\pi} \left[1 - \frac{\overline{\Theta^2}(z)}{2} - \eta_B \right] \quad (28)$$

where η_B is the backscattering coefficient. The backscattering coefficient is a lower limit for the fluence backscatter which strictly should be used. However, η_B is only a few percent for electrons with energies above 2 MeV in water.^{47,64} In electron beams $\overline{\Theta^2}(z)$ increases quasilinearly with depth and reaches a saturation value of about 0.6 rad² near the dose maximum.

In low-energy photon beams (⁶⁰Co, 5 MV) the value increases from about 0.45 rad² near the surface to 0.6 rad² at the depth of transient equilibrium. At higher photon energies both values are lower. At 20 MV, for example, the value increases from 0.15 rad² close to the surface and saturates at 0.45 rad² at the depth of transient equilibrium.²⁴

At the dose maximum and beyond, the effective point of measurement will under the above assumptions therefore be located about 0.55 r upstream from the center of a cylindrical chamber in electron beams and low-energy photon beams.

Other factors that have an influence on the location of the effective point of measurement are as follows:

- (i) The actual shape of the gas volume. In the case of a thimble

chamber, the contribution of the hemispherical part of its gas volume to the total ionization will move the effective point of measurement closer to the central axis compared to its value for a cylindrical volume.

(ii) The presence of the central electrode. Since the number of ions produced in the gas in the cavity is reduced by the insertion of a central electrode, the relative weight of the electrons passing through the peripheral parts of the gas volume is increased. Again this results in a smaller shift of the effective point of measurement from the chamber axis.

(iii) The actual weighting factor. The number of ions generated by an electron entering the gas-filled volume need not be exactly proportional to the chord length along the entrance direction. Ionization due to secondary electrons generated in the chamber wall, gas, or central electrode will influence the location of the effective point of measurement. It has been calculated that the shift becomes $(\pi/4)r = 0.79r$, if the weighting factor for electrons entering a cylindrical volume is assumed to be independent of the point of entry instead of being proportional to the chord length.⁶⁶

3.4.2.2. Experimental Results for Electrons. Experimental determinations of the effective point of measurement can be made with cylindrical ionization chambers of varying radii, using an extrapolation to $r = 0$. Alternatively, the effective point of measurement can be found by use of flat detectors like film emulsions, Fricke solution in flat irradiation cells, or plane-parallel ionization chambers.

Dutreix and Dutreix⁴⁶ found a depth-independent upstream shift of $0.65r$ for 20-MeV electrons, whereas the upstream shift for 10-MeV electrons appeared to increase with depth. Harder⁶³ showed that after the application of the perturbation correction, the 10-MeV values showed an almost constant shift of $0.50r$. Hettinger *et al.*⁶⁷ found a shift of $0.75r$ for 13- and 34-MeV electrons when no perturbation correction was performed. The displacement measured by Johansson *et al.*⁸¹ for 19- and 27-MeV electrons was about $0.55r$ independent of depth, whereas the upstream shift appearing in their 4.5- and 9.5-MeV beams increased with depth in a similar manner as the uncorrected 10-MeV values of Dutreix and Dutreix.⁴⁶ After administration of the perturbation correction, the value of Johansson *et al.*⁸¹ is consistent with a constant shift of $0.40r$.

Weatherburn and Stedeford¹⁴¹ recommended a constant value of $0.45r$ to be used from 3 to 10 MeV after the perturbation correction according to Harder⁶³ is made. Similarly they recommended $0.6r$ between 10 and 30 MeV. Up to 10 MeV their recommendations are based on their own data, while at higher energies they are based on their data at 30 MeV together with recalculations of data from Dutreix and Dutreix⁴⁶

and Hettinger *et al.*⁶⁷ This recommendation is also in approximate agreement with the data of Johansson *et al.*⁸¹ and also with the theoretical value after correction for the angular spread.

Based on the experimental data available, a constant value of $(0.5 \pm 0.1)r$ can therefore be used in the whole energy range from 3 MeV up to 30 MeV. When the uncertainty stated is unacceptable a chamber of smaller radius or a flat chamber should be used.

For flat, thin-window chambers with the window perpendicular to the beam and with no in-scattering perturbation, the effective point of measurement lies just inside the plane of the entrance wall of the air cavity [Ref. 125 and Eq. (31)].

3.4.2.3. Experimental Results for Photons. A very extensive set of measurements of the effective point of measurement in photon beams was made by Johansson *et al.*⁸¹ They found, as a general trend, that the surface value was quite high and close to the theoretical value in a monodirectional beam, but the value decreased rapidly with depth over the first few centimeters. This result is in agreement with the present theory as the angular spread of the secondary electrons increases with depth up to the depth of transient equilibrium.²⁴ The same trend should be expected from the increase in backscatter with depth as the complete slowing down spectrum is being built up. Johansson *et al.* also found a slightly larger value at depth in high-energy photons (42 MV) than at ⁶⁰Co and 5 MV, again in reasonable agreement with the lower saturation value in angular spread in high-energy photon beams.²⁴ An additional factor responsible for this effect could be the larger fraction of scattered photons in low-energy photon beams. Experimental results essentially in agreement with the above statements were reported by Hettinger *et al.*,⁶⁷ Casanovas,³⁵ and Dutreix.⁴⁵

3.4.3. Effective Point of Measurement for Nongaseous Detectors

For solid and liquid detectors, the variation of the absorbed dose within the detector, rather than the variation of the electron fluence over the entrance surface as in a gas-filled detector, will determine the effective point of measurement. The dimensions of nongaseous detectors such as ferrous sulfate dosimeters, film dosimeters, TLD rods, or disks should generally be chosen in such a way that any expected displacement of the effective point of measurement from the detector center can be neglected. However, in a practical case a compromise has to be made between the two conflicting aims of a well-defined effective point of measurement and a high detector sensitivity.

3.4.3.1. Medium-Equivalent Detectors. For extended, medium-equivalent detectors such as the ferrous sulfate dosimeter used in a water phantom, the effective point of measurement may be significantly shifted from the center of the detector volume if the dose distribution is varying in a strongly nonlinear fashion over the dosimeter volume. When the thickness, t , of the detector in the direction of measurement is constant, the measured dose profile may be corrected in a simple way for the finite detector size by using the measured data. The corrected dose distribution is given by²²

$$D_{\text{corr}}(x) = \bar{D}_{\text{meas}}(x) - \frac{t^2}{24} \bar{D}_{\text{meas}}''(x) + \dots \quad (29)$$

where $\bar{D}_{\text{meas}}(x)$ is the measured mean absorbed dose with the detector center at a point x in the medium and x is the variable along the direction of measurement e.g., parallel or perpendicular to the direction of the beam. The resultant upstream shift of the effective point of measurement from the detector center at P is therefore given in the first approximation by

$$d_{\text{eff}} = -\frac{t^2}{24} \frac{\bar{D}_{\text{meas}}''(P)}{\bar{D}_{\text{meas}}'(P)} \quad (30)$$

The first and second derivatives of the measured dose distribution, $\bar{D}_{\text{meas}}'(x)$ and $\bar{D}_{\text{meas}}''(x)$, may be obtained by fitting any three consecutive data points by a second-order polynomial.

3.4.3.2. Non-Medium-Equivalent Detectors. When the detector is non-medium-equivalent, the situation is more complicated as fluence perturbation phenomena also have to be considered. For flat detectors (cf. Figs. 5c and 5d), where the influence of the inscattering effect on the energy distribution in the detector is small, the shift of the effective point of measurement is given by

$$d_{\text{eff}} = \left(1 - \frac{S_{\text{tot},i}}{S_{\text{tot},m}}\right) \frac{t}{2} \quad (31)$$

where t is the thickness of the detector. This expression is based on the definition of the reference volume and the effective point of measurement [Eqs. (23) and (24); cf. Fig. 3] and is valid when the dose variation is linear [$\bar{D}_{\text{meas}}'' \approx 0$; cf. Eq. (29)].

An analytic expression for the effective point of measurement in

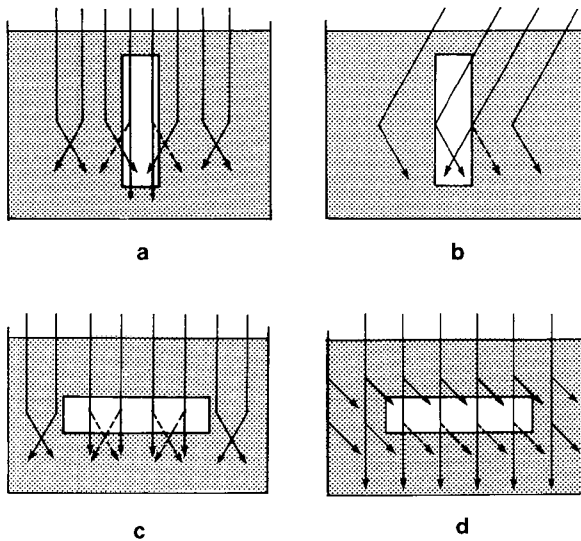


Figure 5. Disturbances of the electron fluence by scattering effects for the case of a low-density detector, such as an air ionization chamber. The multiple scattering of the electrons is schematically illustrated by one large-angle single scattering interaction. Dashed electron paths would occur in the absence of the detector. Panel (a) illustrates the in-scattering, (b) the quasireflection and integrating effects, and (c) the build-up of primary electrons, whereas (d) illustrates the lack of secondary electron equilibrium. (Panels a and c are adapted from Ref. 65.)

electron beams which also takes the fluence perturbation into account was derived by Svensson and Brahme¹³⁵ for a cylindrical detector of radius r and thickness t when irradiated in a region of linearly increasing fluence with the beam axis parallel to the axis of the cylinder. The distance to the effective point of measurement as measured from the cylinder center can be determined from

$$d_{\text{eff}} = \frac{t}{2} \left(1 - 0.43 \frac{t}{r} \right) \left(1 - \frac{T_i}{T_m} \right) \quad (32)$$

where T_i and T_m are linear scattering powers of the detector and the medium, respectively. In the first approximation the shift of P_{eff} from the detector center is negligible if either the linear scattering powers of detector and medium are equal or the detector shape is selected such that $t \approx 2.3r$. Furthermore, for a low-density dosimeter the effective point of measurement is very close to the front surface, as is expected for the limit of a flat gas ionization chamber ($T_i/T_m \approx 0$) provided $t/r \ll 1$.

The effective point of measurement used in deriving Eq. (32) differs from that of Section 3.4.2 and Eq. (31) in that it defines that point in the uniform medium where the electron fluence is equal to the mean fluence inside the detector volume. By this choice, no additional perturbation correction is needed, but instead different mean electron energies should be used for the detector and for the effective point of measurement when calculating mean stopping power values¹³⁵:

$$\bar{E}_i = \bar{E}_m + (S_{\text{tot},m} - S_{\text{tot},i}) \frac{t}{2} - S_{\text{tot},m} d_{\text{eff}} \quad (33)$$

where $\bar{E}_m$ is the mean electron energy at the effective point of measurement in the uniform medium, and $\bar{E}_i$ is the mean energy in the detector.

3.4.4. Fluence Perturbation Correction Factor

3.4.4.1. Perturbation Mechanisms. There will necessarily be a perturbation of the electron fluence of the medium as it penetrates an extended detector of different electron slowing-down and/or scattering properties compared to those of the surrounding medium. This fluence perturbation can be of three principal types as specified below and illustrated schematically in Fig. 5 for the case of a low-density cavity.

(i) *Lack of lateral scattering equilibrium* across the boundaries of the detector volume (cf. Figs. 5a and 5b). This effect is due to different linear scattering powers in the detector and the medium. At least two separate contributions can be recognized. Firstly, owing to the low density there is a lack of scattering out of the side walls of the detector, and hence a net scattering into the cavity from the medium with a resultant increased fluence in the detector volume. This “inscattering effect” is illustrated by Fig. 5a. Secondly, there is an apparent reflection of cavity electrons by the walls of the detector (cf. Ref. 1), which is not compensated by a corresponding apparent reflection of medium electrons by the radiation sensitive material in the detector. This “quasireflection” effect, illustrated by Fig. 5b, will increase the fluence in the detector when the fraction of cavity electrons meeting the walls of the detector at small angles is large. This is the case, for instance, near the surface in electron beams with a small angular spread. A third effect in a low-density cavity is due to the integrating or accumulation of inscattered electrons from different depths in the medium as their absorption in the cavity is negligible. This integrating effect of a low-density cavity will also increase the mean electron fluence in the detector volume.

(ii) *Difference in buildup or attenuation* of the electron fluence in

detector and medium. A difference in fluence buildup is obtained when the angular distribution of primary electrons in the detector volume and in the reference volume of the medium differs (cf. Fig. 5c). The “buildup effect” is thus due to the lack of scattering inside the detector volume compared to that in the medium. This results in a reduced buildup of the primary electron fluence near the surface and a reduced fluence decrease near the end of the electron range in an electron beam. This effect arises when the values of the linear stopping and scattering power ratios for detector to medium differ. The fluence buildup or attenuation in the detector and in the reference volume will generally be very similar when the reference volume is chosen appropriately (see Sections 3.4.2 and 3.4.3.2) and the detector material is not too different from the medium.

(iii) *Lack of secondary electron equilibrium* is due to differences in the slowing down properties of the detector and the medium. In an extended detector a significant part of the low-energy secondary electron spectrum is characteristic of the detector material rather than of the medium. Often, the scattering properties of the two materials will also differ, and a considerable disturbance of the dose distribution near the interface between detector and medium may result (cf. Section 3.2 and Refs. 14, 15, 112, and 120). This perturbation of the fluence of secondary electrons is not discussed further for electron beams, but in principle it could be taken into account either by adjusting the stopping-power ratio or by applying a perturbation factor. For photon beams this effect is very important (cf. text of Fig. 2), and it is generally treated by a generalized type of cavity theory as discussed in Section 3.5.

3.4.4.2. *Gaseous Detectors. (i) Theory.* The inscattering effect is of great importance for gas-filled detectors in a solid or liquid medium. It increases with the surface area of the detector that is parallel to the initial beam direction and is therefore largest in slim detectors elongated in the direction of the beam (Figs. 5a and 5b).

A theoretical treatment of the perturbation correction for plane, *monodirectional electron beams*, based on the small-angle multiple scattering approximation, has been given by Harder.⁶³ For a coin-shaped, gas-filled detector with its axis parallel to the incident beam, the inside plane of the circular entrance window defines the effective point of measurement because the detector essentially samples the electron fluence from the window. Accordingly, inscattering through the side walls of the gas volume (Fig. 5a) disturbs the measured values. The resulting formula⁶³ can be written as

$$p_{m,i} = 1 - 0.13 \frac{t}{r} (tT_m)^{1/2} \quad (34)$$

where t is the thickness and r the radius of the coin-shaped volume, and T_m is the linear scattering power for the medium m . The perturbation correction factor is material and energy dependent through its dependence on T_m . Owing to the change of the electron spectrum with depth the energy dependence can also be described as a depth dependence.

It is important to note that the above expression, Eq. (34), is strictly applicable only at very small depths in beams with negligible angular divergence at the surface. At slightly greater depths or in clinical beams with a significant angular spread the treatment of inscattering from the medium (Fig. 5a) must take account of the angular divergence of the primary electrons, and allowance should also be made for the quasi-reflection effect (Fig. 5b).

The inscattering effect can be taken into account more accurately by generalizing Eq. (34) to take the angular spread of the electrons incident on the cavity into account. Interestingly enough, this changes the size dependence of the perturbation correction factor and it becomes instead

$$p_{m,i} = 1 - 0.11 T_m t \frac{t}{r} \left(1 - \frac{T_i}{T_m} \right) \left\{ \frac{[3\bar{\Theta}^2(z)]^{1/2} + [3\bar{\Theta}^2(z)]^{-1/2}}{2} \right\} \quad (35)$$

where $\bar{\Theta}^2(z)$ is the mean square angular spread of the electrons incident on the front surface of the cavity and T_i is the linear scattering power of the detector gas. This expression should be used instead of Eq. (34) when the depth of the detector is larger than one third of the detector thickness, or rather when $\bar{\Theta}^2(z) > T_m t/3$. In practice this means that the condition is nearly always fulfilled since there is generally a substantial angular spread already in the incident beam.^{21,22} The new term containing the angular spread $[\bar{\Theta}^2(z)]$ differs by less than 10% from unity provided $\bar{\Theta}^2(z)$ is larger than 0.15 rad^2 .

The more general expression taking the angular spread into account shows the direct proportionality between cavity size t and the perturbation factor assuming the shape parameter t/r remains constant. This general result is consistent with the experimental results of Johansson *et al.*⁸¹ for cylindrical thimble ionization chambers when irradiated perpendicular to the chamber axis.

However, for practical dosimetry the perturbation by scattered electrons entering a coin-shaped gas volume from the sides can in practice be reduced to a negligible magnitude when the sensitive volume of the collecting electrode is defined by a surrounding guard electrode of sufficient width (see Fig. 6). An estimate of this width is $2t[\bar{\Theta}^2(z) + T_m t/3]^{1/2}$.

(ii) *Experimental.* In carrying out experimental determinations of the perturbation factor it is desirable to avoid any complications caused

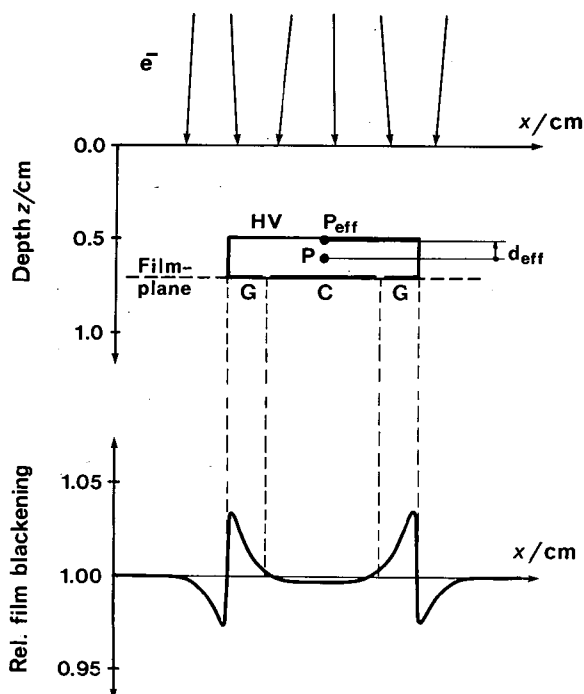


Figure 6. Thin-window, coin-shaped ionization chamber in a weakly scattered electron beam; HV, high-voltage electrode; C, collecting electrode; and G, guard ring. With a well-designed guard ring the scattering effect (cf. Figs. 5a and 5b) mainly influences the electron fluence over the guard. Therefore the chamber can be considered to measure the electron fluence at the entrance window as both the energy loss and scattering in the gas are negligible. This is illustrated by the relative film blackening profile in the lower half of the figure. The measurements were carried out at $\bar{E}_0 = 6 \text{ MeV}$ in PMMA (cf. Refs. 101 and 56).

by the uncertainty in the position of the effective point of measurement. The perturbation effect due to in-scattering is therefore determined from dose measurements made at or near a wide dose maximum, where the exact position of the effective point of measurement is immaterial.

Johansson *et al.*⁸¹ made measurements of the perturbation corrections for cylindrical chambers with their axes perpendicular to the electron beam for mean electron energies between 2.5 and 22 MeV at the depth dose maxima in PMMA. These results (Fig. 7) show the increase of the perturbation with increasing chamber diameter and decreasing electron energy, and demonstrate the fact that the theoretical values given by ICRU⁷⁴ underestimate the effect. The strong energy variation of the perturbation effect implies that relative dose measurements are also affected when made at different depths in a phantom (cf. Ref. 6).

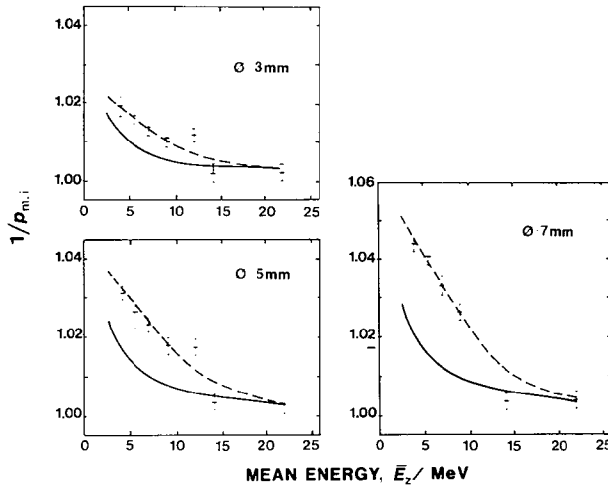


Figure 7. The reciprocal of the perturbation correction factor ($1/p_{m,i}$) for cylindrical ionization chambers with the cylinder axis perpendicular to the beam axis. The chamber diameters were $d = 0.3, 0.5$, and 0.7 mm, respectively. Experimental data: from Ref. 81 measured at the depth of dose maximum (in PMMA); calculated data for water: from Ref. 63 and Ref. 74, solid line). The deviations from the experimental results are mainly caused by the quasireflection and integrating effects (see Fig. 5b). The fairly large spread in the measured points can be explained by the build-up of large electrostatic fields in insulating plastic after irradiations with electron beams. This field disturbs the electron fluence in the phantom (see Section 5.4.1).

For plane-parallel chambers the perturbation factor has been determined by an extrapolation method.^{89,96} The first reference found that the effect was negligible for electron beams with mean energies of 2.2 and 14.3 MeV at the point of measurement; the second reference reported that the perturbation correction was less than 0.5% at electron energies above 0.5 MeV. No fluence perturbation correction is thus needed for plane-parallel ionization chambers that have a thin air cavity and a sufficiently wide guard ring when the entrance window is taken as the effective point of measurement.

However, flat chambers could have an angular dependent response as they do not possess the same symmetry as a spherical chamber or a cylindrical thimble chamber (cf. Ref. 24).

3.4.4.3. Nongaseous Detectors. For solid and liquid detectors the density of the radiation sensitive material can no longer be disregarded and the reference volume will be of finite extension. This implies, for

example, that the effective point of measurement will no longer be at the front surface of a flat detector [cf. Eq. (31)].

Very few theoretical and experimental investigations of the perturbation of the primary electron fluence by solid and liquid detectors have been published. Svensson and Brahme¹³⁵ treated the general case of an arbitrary detector material (*i*) in a uniform medium (*m*), for the case of flat circular detectors with the axis parallel to the beam axis. The buildup effect (cf. Fig. 5c) was treated in detail as it dominated in their application whereas the inscattering effect was only treated in an approximate manner. The total perturbation correction factor derived by Svensson and Brahme has therefore been modified here to take the inscattering effect into account in a more accurate fashion [cf. Eq. (35)] and also to make it applicable to a reference volume reduced in size by the detector to medium stopping power ratio (see Section 3.4.2):

$$p_{m,i} = 1 + \frac{T_m t}{4} \left\{ \frac{S_{\text{tot},i}}{S_{\text{tot},m}} - \frac{T_i}{T_m} - 0.43 \frac{t}{r} \left(1 - \frac{T_i}{T_m} \right) \right. \\ \left. \times \left[\frac{[3\bar{\Theta}^2(z)]^{1/2} + 1/[3\bar{\Theta}^2(z)]^{-1/2}}{2} \right] \right\} \quad (36)$$

where *t* is the thickness of the cylinder in the direction of the beam and *r* the cylinder radius, $\bar{\Theta}^2(z)$ the mean square angular spread of the electrons in a uniform medium at the entrance surface of the cylinder, $S_{\text{tot},m}$ and $S_{\text{tot},i}$ the total linear stopping powers, and T_m and T_i the linear scattering powers of the medium and the detector, respectively. The expression is valid for $\bar{\Theta}^2(z) > T_m t/3$, which is always fulfilled provided that $z > t/3$. The first two terms inside the curly brackets are due to the buildup effect (Fig. 5c), whereas the remainder is due to the inscattering effect (Fig. 5a). The term within the large square brackets differs by less than 10% from unity provided that $\bar{\Theta}^2(z) > 0.15$. It is seen that the perturbation of the primary electron fluence due to the buildup effect is negligible when the linear stopping power ratio equals the linear scattering power ratio, a theoretically interesting result as this is one of the most important conditions for material equivalence. If instead the linear scattering powers are identical ($T_i = T_m$), no perturbation due to in- or outscattering of the primary electrons should be expected. Furthermore, the inscattering effect decreases with increasing angular spread of the electrons $[\bar{\Theta}^2(z)]$ whereas the buildup effect is not affected in the first approximation. These two principal types of fluence perturbation have different signs. Therefore, by appropriate choice of the shape parameter *t/r*, they may compensate each other. It should also be noted that the buildup effect is only valid for depths smaller than the

therapeutic range because the angular distribution then becomes saturated at larger depths (cf. Refs. 119 and 23).

It can be noted that also in this general case the fluence perturbation [Eq. (36)] is directly proportional to the linear dimension of the detector provided the shape parameter t/r has a constant value. This general result agrees phenomenologically well with the experimental results of Johansson *et al.*⁸¹ even though these pertain strictly to thimble chambers with the cylinder axis perpendicular to the beam. From the general expression for the perturbation correction for a flat dosimeter [Eq. (36)] it can also be concluded that thin, disk-shaped detectors such as TLD chips will introduce a negligible in-scattering effect (actually an out-scattering effect since the linear scattering power of the TLD detector is generally larger than for the medium). However, the buildup effect may become quite large since the difference between the linear stopping- and scattering-power ratios is large. For a 0.4-mm-thick LiF-Teflon TLD disk in a polystyrene phantom, the buildup effect introduces a perturbation of the primary electron fluence of about 0.5% in the surface region of a 5-MeV electron beam.

3.5. Review of Recent Dosimetric Theories

The theoretical work for gas detectors in photon beams was reviewed by Burlin.³² Many new ideas have evolved since then.

A very interesting and new approach to dosimetry has been published by Zheng-Ming.^{149,150} By studying the Boltzman transport equation [cf. Eq. (19)] for a uniform medium with a small enclosed gas volume he derived an explicit expression for the equivalent electron source which is responsible for the perturbation of the electron fluence in the region of the gas volume. This allows a more accurate calculation of the energy deposition than the Spencer-Attix theory including second-order effects and taking into account bremsstrahlung and the energy deposited in the cavity by electrons from the equivalent source with energies above Δ .

During the last decade the group in Ghent⁷⁸⁻⁸⁰ have extensively treated the theory of small and large cavities. Their theory could also be regarded as a modified form of the Spencer-Attix theory with a more accurate energy deposition model than the simple two-step approximation of Spencer and Attix.

Interface effects in solid (TLD) dosimeters were measured by Bertilsson.¹⁵ She interpreted the results in the form of a generalized Burlin-type equation where the atomic numbers of the medium and the detector were taken into account more accurately by considering the electron scattering phenomena near the interfaces.

By defining restricted energy absorption coefficients the local energy deposition by photon interactions in a cavity was calculated by Brahme.²¹ The contribution to the direct energy deposition by photons in a detector was obtained by multiplying the traditional mass energy absorption coefficient ratio by the ratio of the restricted to the unrestricted energy absorption coefficients of the detector. The restricted mass energy absorption coefficients were calculated in a manner similar to the energy deposition function also used by Janssens.⁸⁰ By studying the mean path length of insiders (cf. Ref. 36), Horowitz and Dubi⁶⁹ introduced a modification of the detector generated dose contribution.

Based on the method of treatment of electron scatter at an interface suggested by Bernard,¹⁴ Kearsley⁸³ introduced further modifications of the expression for extended cavities in photon beams. His approach is in this respect similar to the method used by Bertilsson.¹⁵

Unfortunately, none of the above theories has treated the problems of the effective point of measurement or the fluence perturbation in a way consistent with their detailed discussion of the ratio of the absorbed dose in the detector to that in the medium.

4. BASIC PHYSICAL DATA

4.1. Background

A determination of the absorbed dose at a point in a phantom is generally based on a measurement of the signal from a detector, the use of one or more calibration factors (e.g., for a detector and an electrometer), the application of one or several physical constants or conversion factors, and corrections of these factors to the situation valid at the point of measurement. The purpose of this section is to analyze the presently available value of the different physical constants, like G , W , and $s_{m,i}$, while the complete absorbed dose calculation procedure is discussed in detail in Section 5.

The experimental determination of $s_{m,i}$ [for definition see Section 3.3 and Eqs. (21), (26), and (27)] is based on certain numerical values of W and G ; see, e.g., Refs. 43 and 136. On the other hand, W and G are often determined using certain assumed values of $s_{m,i}$. For instance, in order to determine W from measurements with a graphite calorimeter and a graphite ionization chamber, values on $s_{gr,air}$ ($m = \text{graphite}$, and $i = \text{air}$) are applied. The W value recommended by the ICRU No. 31⁷⁶ is thus mainly based on such measurements in Co-60 gamma ray beams or in beams at lower energies.

The aim in the present analysis is to avoid such a circular reasoning.

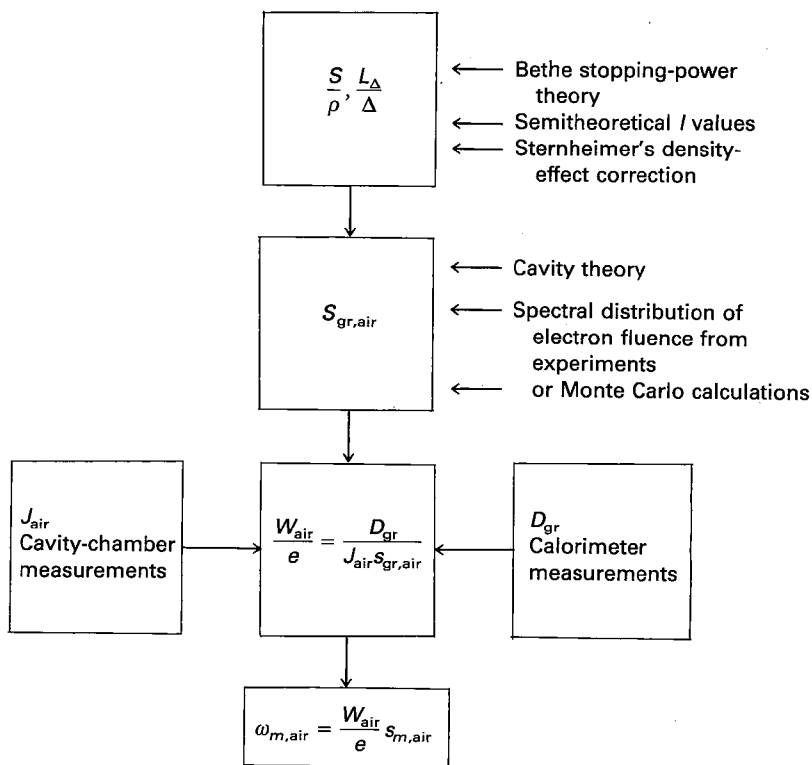


Figure 8. A flow-chart for the determination of W/e and $\omega_{m,air}$. This is the method generally used for Co-60 gamma rays but is here also applied at higher energies. It is seen that W/e depends on the assumption of values for stopping power ratios. For use in dose determinations, numerical values of $\omega_{m,air} = W/e \cdot s_{m,air}$ are needed. These values can be checked by experiments, e.g., from $\omega_{m,air} = D_m/J_{air}$ or $\omega_{m,air} = \omega_{gr,air} s_{m,gr}$.

A flow chart has therefore been followed which has stopping power values as a starting point; see Fig. 8. These values are based on theories and experiments that are independent of assumptions of the values of certain dosimetric constants. The next step is the application of one of the cavity theories (see Section 3) in order to evaluate the stopping power ratios, $s_{m,i}$. Input data for $s_{m,i}$ computations are, besides the stopping powers, the spectral distribution of the electron fluence at the point of measurement, or in a simplified treatment the mean energy of the primary electrons; see Section 4.3. These $s_{m,i}$ values are then used to determine W as shown in the figure (see also Section 4.5) and G as discussed in Section 4.4.

4.2. Stopping Power, S

The collision stopping powers in most widespread use up to 1982 have been those published by Berger and Seltzer in 1964 and 1966.^{9,10} They were adopted by the ICRU in the photon report No. 14⁷³ and electron report No. 21.⁷⁴ These stopping powers have since been reevaluated. Some of these new data from water are shown in Fig. 9. The changes are mainly due to different density effect corrections and use of different mean excitation energies (I values).

The computation by Berger and Seltzer¹¹ was undertaken in connection with a new ICRU report on the subject and represents the most complete data for different elements and compositions. These computations are considered to give the present state-of-the-art values and are therefore used below. It is of great interest to compare these stopping powers with those of Berger and Seltzer^{9,10} as these will directly influence the determined absorbed dose value keeping other factors unchanged. For instance, the decrease in $S_{\text{col},w}$ is 1.8% at an electron energy of

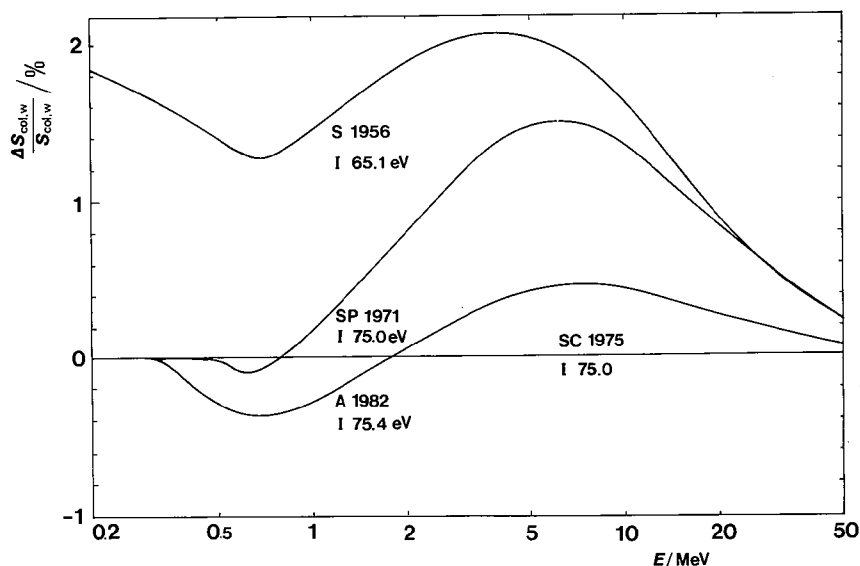


Figure 9. Comparisons of different sets of collision stopping powers for water. The density effect corrections have been performed according to Sternheimer¹²⁸ (S 1956), Sternheimer and Peirels¹²⁹ (SP 1971), Sternheimer¹²⁷ and Carlsson³³ (SC 1975), and Ashley⁷ (A 1982). The computations have been carried out by Berger and Seltzer (see the text). The SC 1975) data have been recommended by Berger and Seltzer¹¹ and are therefore used in the present report. The (S 1956) data were used in Berger and Seltzer.^{9,10}

TABLE I
The Change in I Values for Some Important Materials for Dosimetry

Material	I values by Berger and Seltzer (eV)		ΔI (eV)
	1964	1982	
Air	86.8	85.7	-1.1
Water	65.1	75.0	9.9
Graphite	78.0	78.0	0
Lucite (PMMA)	65.6	74.0	8.4
Polystyrene	63.6	68.7	5.1
Polyethylene	54.6	57.4	2.8

0.3 MeV (Fig. 9), i.e., at an energy of importance for Co-60 gamma dosimetry.

The I values from 1964–1966 and 1982 are compared for some materials of importance for clinical dosimetry in Table I. The change of the I value is particularly large for water, about 10 eV. The change in collision stopping-power values will, however, be moderate as I is included in the logarithmic term in the stopping power formula. Thus

$$S_{\text{col}} \propto \rho \frac{Z}{A} [f(E) - 2 \ln I - \delta] \quad (37)$$

where ρ is the density, Z/A is the atomic number to the atomic weight (in compounds or mixtures weighted by the mass fractions of the different elements), $f(E)$ is a function of the energy of the incident electron, and δ is the density effect correction.

The new density effect corrections differ up to about 2% from earlier calculations, a change large enough also to be of importance in clinical dosimetry. The situation is particularly complicated for graphite as it is not clear if the bulk density (about 1.7 g cm^{-3}) or the crystalline density (2.265 g cm^{-3}) should be used. The reduction in $S_{\text{col,gr}}$ as compared to the data from 1964 is shown in Fig. 10; curves are shown for the two densities. The density problem for graphite represents a special complication, since for many years graphite calorimeters have been used as the standard instrument to determine absorbed dose at national standards laboratories. The flow scheme (Fig. 8) using both these two densities is therefore discussed in Sections 4.3 and 4.4 in order to find arguments in the field of dosimetry for one or the other density.

The I value for air has been almost unchanged (Table I). The density effect correction is below 1% for energies <50 MeV. The stopping power

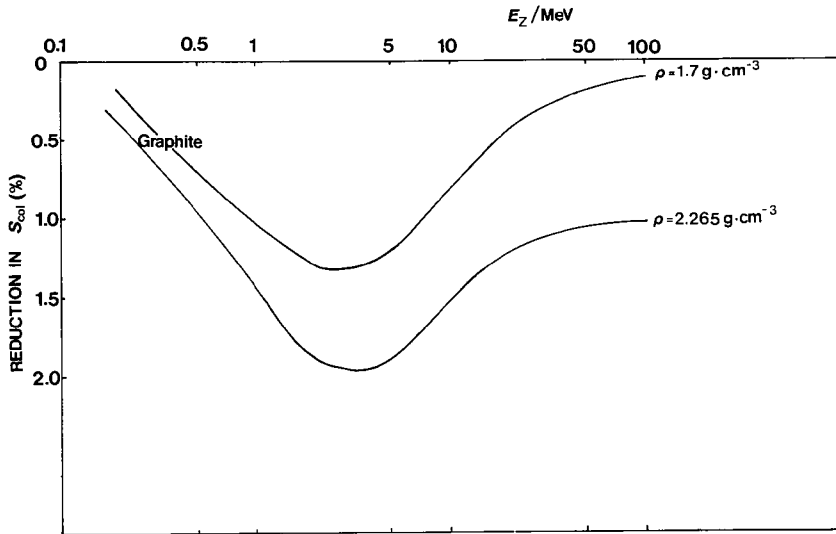


Figure 10. The percentage reduction of the collision stopping power for graphite from Berger and Seltzer 1964 (Ref. 9) to their 1982 values (Ref. 11). The graphite density 1.7 g cm^{-3} was assumed in Ref. 9.

ratios $s_{m,\text{air}}$ will therefore mainly be influenced by changes for the material m .

4.3. The Stopping Power Ratio, $s_{m,i}$

Two major irradiation geometries for the medium and the detector will be discussed in this section as shown in Fig. 2. These lead to two different sets of stopping power ratios, $s_{m,i}$, to be applied in the Bragg-Gray equation [see Eqs. (21), (26), and (27)].

In the *Bragg-Gray* (BG) approach it is assumed (see Section 3.1) that the spectral fluence of the primary electrons in the detector is the same as that in the medium, $\Phi_{E,m}^p$. Also, it is assumed that the energy dissipation by secondary electrons occurs “on the spot,” or alternatively, that there is a secondary electron equilibrium, i.e., that the energy from secondary electrons entering the cavity is equal to the energy from such electrons leaving the cavity. Interface effects (Fig. 5d) between the medium, chamber wall, and the cavity gas are not taken into account. Under this assumption the ratio of the energy dissipation in the uniform

medium and in the air inside a cavity is given by [compare Eq. (16)]

$$s_{m,\text{air}}^{\text{BG}} = \frac{\int_0^{E_{\text{max}}} (\Phi_{E,m}^p) \left(\frac{S}{\rho} \right)_{\text{col},m} dE}{\int_0^{E_{\text{max}}} (\Phi_{E,m}^p) \left(\frac{S}{\rho} \right)_{\text{col},\text{air}} dE} \quad (38)$$

In order to avoid significant interface effects the medium, chamber wall, and cavity should be of very similar composition. This is not possible in many practical situations. However, many workers^{19,68,62} showed that a difference in composition may be overcome by having a fairly thin lining of detector-equivalent material around the radiation sensitive material of the detector. This lining practically eliminates the interface problem. Figure 2b illustrates this geometry for an ionization chamber.

With such a lining Eq. (38) may be simplified, as suggested by Harder,⁶² by using the collision stopping power at the mean energy of the primary electrons:

$$s_{m,\text{air}}^{\text{H}} = \left[\frac{S(E)}{\rho} \right]_{\text{col},m} / \left[\frac{S(E)}{\rho} \right]_{\text{col},\text{air}} \quad (39)$$

This simplification is valid when the collision stopping powers vary linearly with the energy of the primary electrons or when the stopping power ratio is a slowly varying function of energy over the energy range of interest.

The situation for which the Spencer–Attix (SA) stopping power ratio is valid is an air cavity with walls of the same composition as the surrounding medium, m ; see Fig. 2a. This theory accounts in an approximate manner for the finite ranges of secondary electrons or δ rays. Here the spectral fluence of both primary and secondary electrons, $\Phi_{E,m}$, above a cutoff energy, Δ , must be computed. The stopping powers L_{Δ}/ρ are now restricted to losses smaller than the cutoff energy. According to Eq. (15) the Spencer–Attix stopping power ratio, $s_{m,\text{air}}^{\text{SA}}$ can be written

$$s_{m,\text{air}}^{\text{SA}} = \frac{\int_{\Delta}^E \Phi_{E,m} \left(\frac{L_{\Delta}}{\rho} \right)_m dE + \Phi_e(\Delta)_m \left(\frac{S(\Delta)}{\rho} \right)_m \Delta}{\int_{\Delta}^E \Phi_{E,m} \left(\frac{L_{\Delta}}{\rho} \right)_{\text{air}} dE + \Phi_e(\Delta)_m \left(\frac{S(\Delta)}{\rho} \right)_{\text{air}} \Delta} \quad (40)$$

The second term in the numerator and denominator is the energy dissipation by the track ends (see Refs. 102 and 103).

The cutoff energy is often chosen to be equal to the energy just needed for an electron to be able to cross the air cavity. The cutoff energy has thus to do with the size and shape of the chamber, but this cutoff is not very critical for the values of $s_{m,air}^{SA}$. As an example, the practical range (standard pressure and 20°C) for 10-keV electrons is about 2 mm and about 7 mm for 20 keV. The stopping power ratio, $s_{w,air}^{SA}$, at an electron energy of 10 MeV, decreases by only about 0.2% when the cutoff is increased from the lower to the higher value. The exact value is somewhat dependent on the electron energy. Berger *et al.*¹³ gave an empirical formula for the relative variation of $s_{w,air}^{SA}$ with the cavity size with input parameters depending on the value of Δ , electron energy, and the depth in the phantom.

The values of the three different stopping power ratios will now be discussed. In particular, $s_{w,air}$, i.e., the stopping power ratio water to air will be analyzed, first for electron beams and then for photon beams.

Values of $s_{w,air}^{SA}$ for different electron energies at the phantom surface and different depths have been computed for electron radiation, using a Monte Carlo technique, by Berger and Seltzer¹² and by Nahum.¹⁰⁴ These two sets of calculations agree very well, as can be seen in Fig. 11. This agreement shows that the calculated electron spectra are sufficiently similar as the same set of input stopping power data have been used. There is also indirect experimental evidence that at least the computation of $\Phi_{E,m}$ is sufficiently accurate for the dosimetry. It is possible to find the same numerical values of $s_{m,air}^{SA}(E_0, z)$ for various incident energies, E_0 , and phantom depths z . For instance, according to computations by Berger and Seltzer,¹² $s_{gr,air}^{SA}$ for $E_0 = 15$ MeV, and $z = 1$ cm will have the same value as for $E_0 = 50$ MeV and $z = 14$ cm ($s_{gr,air}^{SA} = 0.865$ using $\rho = 1.7 \text{ g cm}^{-3}$ and $\Delta = 10$ keV). In spite of the extreme difference in parameters, experiments based on absolute measurements of D_{gr} and J_{air} ⁴³ also show that D_{gr}/J_{air} at these two sets of energies and depths are the same within the uncertainty of measurement (a few parts of 1%), thus supporting the theoretical $\Phi_{E,m}$. More details will be found in Section 4.5, where W/e has been evaluated from experimental determinations of D_{gr}/J_{air} and the input of theoretical $s_{gr,air}^{SA}$. A consistent W/e value, independent of the combination of E_0 and z , is obtained there. The same type of experiment has been carried out in water, i.e., D_w and J_{air} were determined. Also in this case good agreement is obtained between Monte Carlo calculations and measurements. Computations of the Bragg-Gray stopping power ratios (Eq. (38)] have been carried out by Nahum^{102,103} for various electron energies. These computations have been reevaluated using the stopping power data from Ref. 11. The

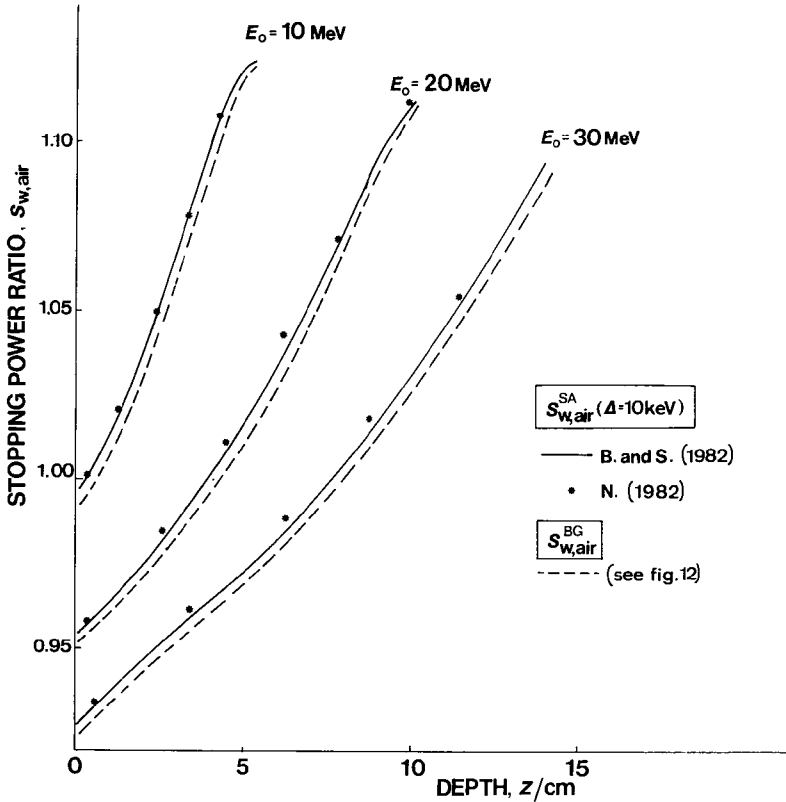


Figure 11. Monte Carlo calculated $s_{w,air}^{SA}$ from Berger and Seltzer¹² (—) and Nahum¹⁰⁴ (*). The $s_{w,air}^{BG}$ are taken from Fig. 12.

calculations, shown as dotted lines in Fig. 12, are made in a straightforward way using Eq. (38). In practice complete secondary electron equilibrium is never achieved. This is particularly so at small depths where the slowing down spectrum of secondary electrons is not yet built up.

A modified BG ratio has therefore also been computed (dash-dot line in Fig. 12) taking into consideration the lack of equilibrium. Here $(L/\rho)_{\Delta}$ replaces $(S/\rho)_{col}$. The restriction value Δ is adjusted so that the range of electrons with energy Δ is equal to z ;²⁰ this modified $s_{w,air}^{BG}$ is included as the dashed line in Fig. 12. It is seen that the difference amounts to about 1.5% and is largest at small depths.

In Fig. 12 $s_{w,air}^H$ is also shown. The mean energy at the point of

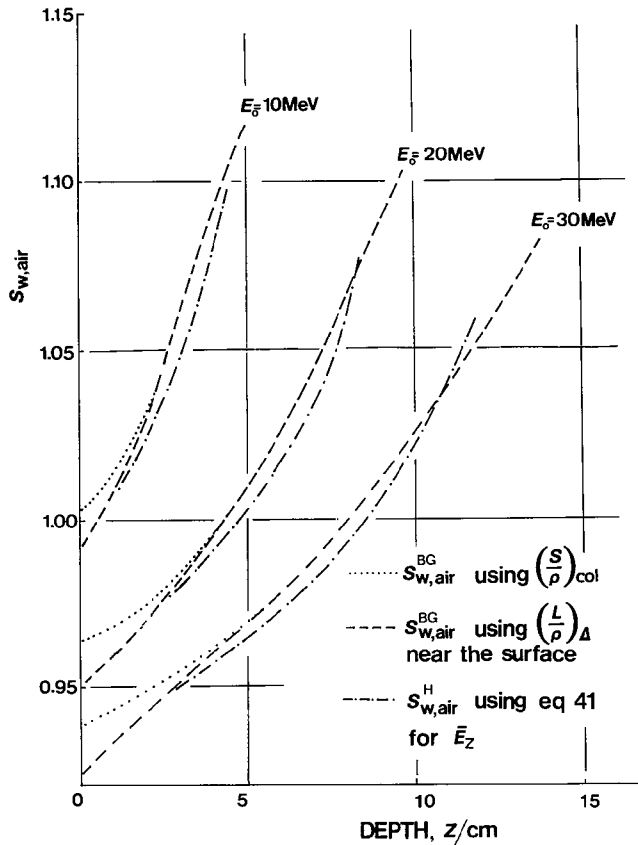


Figure 12. The straightforward Bragg-Gray stopping power ratios for water to air are shown ($\cdots$). Corrections for lack of secondary electron equilibrium near the surface have been performed ($---$). The result from a simplified computation of the Bragg-Gray ratios, indicated $s_{w,air}^H$, are also shown ($- \cdots -$).

measurements was calculated from the relation ^{20,6}

$$\bar{E}(z) = \bar{E}(0) \frac{S_{\text{tot}} \exp[-zS_{\text{rad}}/\bar{E}(0)] - S_{\text{col}}}{S_{\text{rad}}} \quad (41)$$

The $s_{w,air}^{BG}$ value [Eq. (38)] agrees within about 0.5% with the $s_{w,air}^H$ value [Eq. (39)], at least at depths corresponding to about dose maximum.

It is also of interest to compare $s_{w,air}^{BG}$ and $s_{w,air}^{SA}$ (see Fig. 11). It is seen that the BG ratio is 0.3%–0.5% smaller than the SA ratio. Svensson and Nahum¹³⁷ computed the difference $s_{m,air}^{SA} - s_{m,air}^{BG}$ as a function of Δ

$I = I_{\text{air}} - I_m$, where I is the mean excitation energy. They found that the difference in stopping power ratios is very weakly dependent on the quality of the electron and photon beams, but increases with $|\Delta I|$. The difference is below about 0.5% for $|\Delta I| < 10$ eV.

All the stopping power ratio expressions [Eqs. (38), (39), and (40)] are equally applicable for photons and electrons. For photons the complete spectrum of secondary electrons must be known. Very few data have been reported in the literature. However, $s_{w,\text{air}}^{\text{SA}}$ ratios are now available for some qualities from Nahum^{102,103} and Andreo.⁵ These data agree fairly well, except at small phantom depths, i.e., before dose maximum; cf. Section 4.6.

Numerical data of $\omega_{w,\text{air}} = s_{w,\text{air}}^{\text{SA}}(W/e)$ are presented both for electron and photon beams in Section 5. They may thus also be used for electrons in the case where the ionization chamber has an air-equivalent wall, as the difference in $s_{w,\text{air}}^{\text{SA}} - s_{w,\text{air}}^{\text{BG}}$ for most practical situations is unimportant, i.e., below 0.5%. The situation is more complicated for photons because photons generate electrons differently in water and in an air-equivalent wall. In practical dosimetry, it is therefore more straightforward to use the Spencer–Attix values for photons and then correct for the non-water-equivalence of the chamber wall by means of a perturbation factor.¹⁰⁰

4.4. Mass Energy Absorption Coefficient, μ_{en}/ρ

Values of μ_{en}/ρ are needed in order to compute the influence of different phantom and wall materials on the signal of the detector in photon beams. The μ_{en}/ρ for the detector material is also of interest in cases where a significant part of the electron fluence in the detector is generated in the detector material itself; see Sections 3 and 5. Values of μ_{en}/ρ have been computed by Hubbel^{71,72} and they are considered to be known within a fraction of 1%.

4.5. Radiation Chemical Yield, $G(X)$

The radiation chemical yield is defined by $G(X) = n(X)/\bar{\epsilon}$, where $n(X)$ is the mean amount of substance of a specified entity, X , produced, destroyed, or changed by the mean energy imparted, $\bar{\epsilon}$, to the matter (see ICRU No. 33).⁷⁷ Here the radiation chemical yield of the ferrous sulfate dosimeter will be discussed.

The dose measurement technique based on the ferrous sulfate dosimeter is well established (e.g., ICRU report No. 14⁷³). It is assumed here that the standard dosimeter solution is used, i.e., consisting of 1 mol m⁻³ ferrous sulfate, 400 mol m⁻³ sulfuric acid, and 1 mol m⁻³ NaCl.

The reference temperature during the irradiation and spectrophotometric measurements is assumed to be 25°C.

The absorbed dose to the dosimeter solution D_i is determined from the relationship

$$D_i = \frac{\Delta A_t}{\rho l (\epsilon_m)_t G_{t'}} \quad (42)$$

where D_i is the absorbed dose to the dosimeter solution, ΔA_t is the increase in absorbance due to irradiation at temperature t ($=25^\circ\text{C}$) during the spectrophotometric measurements, ρ is the density of the solution, l is the length of the light path in the photometer cell, $(\epsilon_m)_t$ is the difference between the molar linear absorption coefficient for ferric ions and for ferrous ions at temperature t ($=25^\circ\text{C}$), and $G_{t'}$ is the radiation chemical yield of ferric ions at the irradiation temperature t' ($=25^\circ\text{C}$).

Experimental data for $\epsilon_m G$ from different investigators will be compared rather than G values. The reason is that the experimental determination of ϵ_m needs great care and large systematic errors are often introduced. However, in several intercomparisons it has been shown by using a reference solution that the difference in measured absorbance with different spectrophotometers generally is very small. This seems to be the reason why a smaller spread is obtained if experimental values of $\epsilon_m G$ are compared instead of G (see Ref. 136).

Values of $\epsilon_m G$ determined from calorimeter measurements for electron radiation are given in Table II. The $\epsilon_m G$ values for different types of calorimeters are given in different columns. The authors' values have been corrected in two ways:

(a) The calorimeter absorber is in some experiments replaced by a ferrous sulfate dosimeter. This dosimeter has different scattering properties than the absorber. The absorber and the dosimeter will therefore "see" a somewhat different fluence of primary electrons. The applied correction is dependent on the shape and the material of the dosimeter and on the spectral and angular distribution of the electrons [see Section 3.4.4, in particular Eq. (36)].

(b) The stopping-power data by Berger and Seltzer¹¹ have been applied. For the graphite calorimeter stopping powers based both the bulk density ($\rho = 1.7 \text{ g cm}^{-3}$) and the crystalline density ($\rho = 2.265 \text{ g cm}^{-3}$) are used. These stopping powers are considered to represent the extreme values within which the true effective density is located.

Table II includes the corrected values for electron radiation. The mean values are calculated giving the same weight to the result from each investigator. If $\rho = 1.7 \text{ g m}^{-3}$ is used when selecting stopping powers, a

TABLE II
 ϵ_m and G Values Determined by Different Investigators^a

Graphite calorimeter				
Reference	$\bar{E}_z$ (MeV)	$\epsilon_m G$ ($\text{m}^2 \text{kg}^{-1} \text{Gy}^{-1}$)		ϵ_m ($\text{m}^2 \text{mol}^{-1}$)
		$\rho = 1.7 \text{ g cm}^{-3}$ ($\times 10^{-6}$)	$\rho = 2.265 \text{ g cm}^{-3}$ ($\times 10^{-6}$)	
37	3.5	348.2	345.8	217.34
	7	351.9	349.2	
	9	352.3	349.4	
	13	352.6	349.2	
	14.5	352.8	349.4	
		351.6	348.6	
55	6	352.2	349.5	220.0
	16	348.4	245.1	
		350.3	347.3	
115	16.5	355.7	352.4	220.0
97	24	349.1	345.2	216.4
Mean:		351.7	348.4	218.4

Nongraphite calorimeter				
Reference	$\bar{E}_z$ (MeV)	$\epsilon_m G$ ($\text{m}^2 \text{kg}^{-1} \text{Gy}^{-1}$)		ϵ_m ($\text{m}^2 \text{mol}^{-1}$)
		($\times 10^{-6}$)		
122	1	351.1		219.6
53	3	349.0		217.0
3	7	350.4		222.5
95	5	351.9		
	15	354.1		
		353.0		
114	20	354.2		219.6
Mean:		351.5		

^a The $\epsilon_m G$ values are corrected as described in the text.

value of $\epsilon_m G$ is obtained ($351.7 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$), in close agreement with the determination from nongraphite calorimeters ($351.5 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$). This is one indication that the bulk density should be used at least with the most recent formulation of the density effect. However, the value using the higher density ($348.4 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$), is 0.9% lower but still within the systematic errors. The calorimetric experiments alone are therefore not sufficiently conclusive for the use of the lower density.

For Co-60 gamma-ray beams a G value of 15.5 ± 0.2 per 100 eV (i.e., $1.607 \times 10^{-6} \text{ mol kg}^{-1} \text{ Gy}^{-1}$) has been recommended by the ICRU in report No. 14.⁷³ The uncertainties of the individual determinations referred to in report No. 14 are fairly large due to uncertainties in spectrophotometer calibrations (ϵ_m value) and also in physical constants. A reevaluation of earlier measurements is complicated as the exact values of some constants used by the authors are not known. This is so in particular for G values based on the ionization chamber method as the stopping powers have undergone large changes. These changes may also influence the numerical value of W (see Section 4.5), and in some experiments the systematic errors may cancel out. A systematic error, or rather inconsistency in comparison with other methods of absorbed dose determinations, of a few percent would probably have been discovered, as ferrous sulfate dosimetry has been in considerable use for long time. The G value 15.5 per 100 eV would correspond to a value of $\epsilon_m G = 354 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ if the $\epsilon_m = 220.5 \text{ m}^2 \text{ mol}^{-1}$ is used. This ϵ_m value was the mean found from a literature study of 83 investigators³⁰ and used by ICRU report No. 21.⁷⁴ There are, however, some very recent measurements that can be utilized to give a more solid base for evaluation of $\epsilon_m G$.

Experiments that have been carried out by or in close cooperation with some of the national laboratories can be analyzed in order to determine $\epsilon_m G$. These experiments are of particular interest as all the spectrophotometers have been intercompared.⁵⁰ A test sample of potassium dichromate in aqueous perchloric acid solution was used. Measurements of absorbance were made at the wavelengths of 313 and 350 nm, thus close to the 304 nm as used for Fricke dosimeters. The deviations between the absorbance standard and the measured values at the ten participating laboratories were fairly small, of the order of one or two tenths of one percent at an absorbance above 0.2.

The BIPM (Bureau International des Poids et Mesures) undertook irradiation of Fricke dosimeters from some of the laboratories in a water phantom. The absorbed dose rate to water, $\dot{D}_w$, at 5 g cm^{-2} depth at a reference point 1 m from the BIPM ^{60}Co source was calculated from measurements of the dose rate, $\dot{D}_{\text{gr}}$, in a corresponding irradiated

geometry using a graphite phantom. At the BIPM the measurement of absorbed dose in graphite is performed with a graphite wall cavity ionization chamber. From these ionometric measurements and also from calorimetric determinations which were carried out in cooperation with four national laboratories using their instruments at the BIPM, a weighted mean of absorbed dose rate to graphite in the BIPM beam was determined with an uncertainty (one σ) of 0.1%.¹⁷ Calculation of $\dot{D}_w/\dot{D}_{gr}$ was performed by Boutillon. A ratio of 1.050 was derived with an uncertainty of 1% (cf. Fig. 23). The BIPM could thus report the absorbed dose to water at the position of the ferrous sulfate dosimeters.

From the paper by Ellis *et al.*⁵⁰ it is now possible to evaluate the value of $\epsilon_m G$ that should have been used by the various laboratories in order to give the absorbed dose stated by BIPM for their irradiated dosimeters. Table III gives this information. The mean value for $\epsilon_m G$ is $353.3 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$.

One of the centers participating in this intercomparison and reported in Table III (No. 4) also carried out a separate determination of $\epsilon_m G$. Two liquid ionization chambers¹⁴⁷ were used in order to transfer absorbed dose to water determinations from the NBS and to laboratory No. 4 (i.e., University of Gothenburg). The standard deviation in a single measurement was about 0.1% using these chambers. The transfer was carried out with an uncertainty less than about 0.2%. A graphite calorimeter was used by NBS to determine absorbed dose to graphite and a thick wall graphite ionization chamber was used as a transfer instrument by the NBS in order to determine absorbed dose to water at

TABLE III
 $\epsilon_m G$ Determined from Experiment at BIPM^a

Laboratory	$\epsilon_m G \text{ (m}^2 \text{ kg}^{-1} \text{ Gy}^{-1})$ ($\times 10^{-6}$)
2. ^b Laboratoire Metrologie des Rayonnements Ionisants, Saclay, France	354.5
3. National Institute of Radiological Sciences/ ETL., Chiba-shi, Japan	352.7
4. NIRP/University of Gothenburg, Sweden	354.5
5. National Physical Laboratory, Teddington, U.K.	351.6
7. International Atomic Energy Agency, Vienna	353.1
Mean:	353.3

^a A recalculation from Ellis *et al.*⁵⁰

^b The numbers are as given in Table VIII of Ref. 50.

the point of interest in the water phantom. Stopping-power ratios are not used in the calculations (see Ref. 117). From these transfer experiments $\epsilon_m G = 352.3 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ was determined for ^{60}Co gamma radiation⁹⁵ (14 cm $\times$ 14 cm, 5 cm depth in water, source-detector distance 100 cm). This value should be compared with $\epsilon_m G = 354.5 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ (laboratory No. 4 in Table III) based on the BIPM comparison, a difference of 0.6%.

Mail intercomparisons using the ferrous sulfate dosimeter have been in extensive use for many years and it is well known that the storage of Fricke solution in irradiation cells may influence both the G value (generally an increase) and the precision (see, e.g., Ref. 138). Most of the participating laboratories in the experiment reported above used glass irradiation cells which should have an insignificant storage effect⁴⁹ but may instead cause interface problems between glass and liquid.³² In the intercomparison reported above the standard deviation for the single measurements (σ for each laboratory) at the different participating laboratories varied from about 0.1% to 2%. In the transfer experiments by Mattsson *et al.*⁹⁵ based on the transfer with a liquid ionization chamber these problems do not appear, as the storage time is very short, and therefore also plastic irradiation cells can be used. The precision of the measurement was generally better than 0.1% (i.e., σ of a single measurement).

In a further experiment Mattsson *et al.*⁹⁵ compared the absorbed dose determinations with ferrous sulfate dosimeters and with a copy of the Domen water calorimeter.⁴⁰ Taking $\epsilon_m G = 352.3 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ an exothermic chemical (or heat) defect of 3.5% has to be assumed in order to have the two methods agree. Domen⁴² has reported exactly the same value, which further supports the results from Mattsson *et al.* All the analysis above results in $\epsilon_m G$ for Co-60 gamma rays between 352 and $354 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$, and a value of $353 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ is therefore recommended.

In the ICRU report No. 14 somewhat higher G values were recommended for high-energy photons from an accelerator than for Co-60 gamma rays. A very large uncertainty was, however, stated; for 11–30 MV $G = 15.7 \pm 0.6$ per 100 eV. More recent studies instead indicate that the G value may be even lower for these qualities than for Co-60 gamma radiation and high-energy electrons.¹⁰² Conclusive experiments are not available for these qualities.

The value of $\epsilon_m G$ for electron beams in Table II was assumed to be energy independent. However, there might be a weak energy dependence of G , as discussed by Nahum.¹⁰² He carried out a semiempirical evaluation of G values based on Monte Carlo calculated local energy dissipation spectra ("local" meant a cutoff of 100 eV) for various

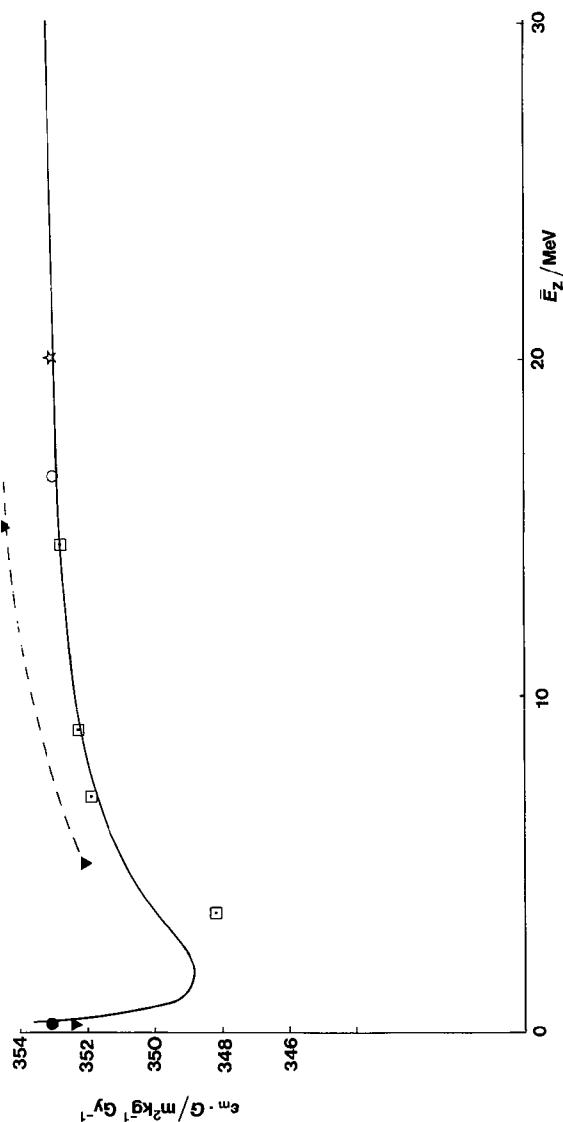


Figure 13. Semitheoretical values of $\epsilon_m G$ ($\times 10^{-6}$) by Nahum¹⁰² (—) normalized after the experimental value at 14.5 MeV by Cottens *et al.*³⁸ (□). Experimental values of $\epsilon_m G$ from Mattsson *et al.*⁹⁵ (▼) and Ellis *et al.*⁵⁰ (○). Petterson¹¹⁴ (★) and Pinkerton¹¹⁵ (○) measured G -values both for ^{60}Co gamma and electron radiation. However, in the figure a normalization to $\epsilon_m G = 353 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ at ^{60}Co gamma quality has been performed, as it is felt that the uncertainty in the absolute values of the product $\epsilon_m G$ may be somewhat larger than for the recent investigations.

radiation qualities. This information, in combination with the application of some experimentally well-established G values of radiations differing widely in quality, made it possible to predict trends also for small quality changes in the case of high-energy photon and electron beams (see Fig. 13).

Also experimentally, it is indicated by the very careful measurements by Cottens *et al.*³⁸ (the systematic and random uncertainty, not including the stopping powers, are totally a few tenths of one percent) that G may increase with the electron energy. An opposite result was obtained by Geisseloder *et al.*⁵⁵ (see Table II). The uncertainty in these earlier measurements seems, however, to be much larger. In Fig. 13 the values by Cottens *et al.*³⁸ and Nahum¹⁰² are compared, normalized to the experimental values at 14.5 MeV (see also Ref. 105). A weak energy dependence is seen.

Of special interest are calorimetric determinations that have been performed by one investigator both for electrons and Co-60 gamma beams. Three experiments of this kind will be analyzed here, namely, those by Pettersson,¹¹⁴ Pinkerton,¹¹⁵ and Mattsson *et al.*⁹⁵ Pettersson used a water calorimeter. His determinations do not involve stopping power ratios, except that for water to ferrous sulfate dosimeter solution, which has an insignificant uncertainty. Practically the same $\epsilon_m G$ was obtained for Co-60 gamma and 20 MeV (mean energy, $\bar{E}_z$) electrons, 15.57 and 15.56 per 100 eV, respectively. Pettersson reported that he overcame part of the chemical defect in water by carrying out preirradiations of the water. A possible chemical effect ought, in any case, to be of the same amount both for Co-60 gamma radiations and electron beams (see Ref. 42). Mattsson *et al.* used the Domen calorimeter to determine $\epsilon_m G$ at $\bar{E}_z = 5$ and 15 MeV. The same chemical defect was assumed as for Co-60 gamma rays, i.e., 3.5%.⁴² This would result in the $\epsilon_m G$ values given in Table II (Ref. 95) and Fig. 13. Finally, the experiment by Pinkerton was carried out using a graphite calorimeter. Pinkerton's results were 15.60 and 15.62 per 100 eV for Co-60 gamma rays and 16.5 MeV ($\bar{E}_z$) electrons, respectively. Pinkerton has not reported the exact stopping power values in use for his experiment but has referred to the table by Berger and Seltzer. In Table II corrections to the new data on stopping powers¹¹ were carried out. In addition a small scattering correction was applied (see Section 3.4.4). If the bulk density of graphite is used, as discussed above, then $\epsilon_m G = 355.7 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ is obtained for $\bar{E}_z$ 16.5 MeV that is exactly the same value as for Co-60 gamma radiation (i.e., 15.60 per 100 eV, which corresponded to this $\epsilon_m G$ value).

In conclusion: comparisons of graphite and nongraphite calorimeter determinations of G values for electron beams indicate a better agreement using the low-density $\rho = 1.7 \text{ g cm}^{-3}$ than the high one. There are

several indications that there is a small energy dependence of the G value for Co-60 gamma rays, photon and electron beams. The following $\varepsilon_m G$ values ($\times 10^{-6}$) are therefore recommended:

Co-60 gamma rays, 2–4 MV: $353 \pm 3 \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$

X rays, 5–50 MV and electrons $E_z = 2\text{--}10 \text{ MeV}$: $352 \pm 4 \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$

electrons $\bar{E}_z = 10\text{--}50 \text{ MeV}$: $353 \pm 3 \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$

4.6. Mean Energy Expended in Gas per Ion Pair Formed, W

It has generally been assumed that the value of W for air (defined by $W = E/\bar{N}$, where $\bar{N}$ is the mean number of ion pairs formed when the initial kinetic energy E of a charged particle is completely dissipated in the gas, ICRU No. 33⁷⁷) does not vary with photon or electron energy in the MeV^{-1} range. There is, however, no experimental proof of this assumption.

Several investigations have been carried out to determine W for Co-60 gamma rays; see ICRU No. 31.⁷⁶ In these measurements the Bragg–Gray relation for an ionization chamber has been made use of [Eq. (21b)]:

$$\frac{W}{e} = \frac{D_{\text{gr}}}{J_{\text{air}} s_{\text{gr,air}}} = \frac{\omega_{\text{gr,air}}}{s_{\text{gr,air}}} \quad (43)$$

The absorbed dose to graphite, D_{gr} , is determined from graphite calorimeter measurements. The mass ionization, J_{air} (i.e., ionization charge of one sign per mass of air in the air cavity of an ionization chamber), is measured with a graphite walled chamber. The $s_{\text{gr,air}}$ is the mean restricted mass stopping power ratio at the point of measurement for the cavity size and radiation quality in use. The ICRU (in report No. 31⁷⁶) has recalculated the various experimental W values for the same value of $s_{\text{gr,air}}$ and also included, if necessary, corrections to the dry air condition. For dry air a value of $W = 33.85 \pm 0.15 \text{ eV}$ was recommended. However, the $s_{\text{gr,air}}$ values applied by ICRU, based on previous calculations by Berger and Seltzer, have now been revised (Ref. 11; see also Section 4.2). The change in $s_{\text{gr,air}}$ will increase W by about 1.3% for Co-60 gamma; see Table IV.

Only a few experimental determinations have been carried out at higher energies.^{84,88,113} They are based on a different type of measurement than for Co-60 gamma rays. A Faraday cage was used for the determination of the number of incident electrons on an ionization chamber. The charge of one sign liberated in this chamber per incident electron and per unit pathlength dJ/dX was then measured. For the

TABLE IV
Experimental W Values Determined by Niatel¹⁰⁸ using the "Domen"-Calorimeter
and a BIPM Graphite Ionization Chamber^a

Quality	$\omega_{gr,air}$ (JC ⁻¹)	$s_{gr,air}^{SA}$	W (eV)	$s_{gr,air}^{SA}$ parameters
Co-60 γ rays depth 10 g cm ⁻²	34.13	1.0105	33.78	Value from Ref. 108
		1.0018	34.07	$\Delta = 14$ keV, $\rho = 1.7$ g cm ⁻³ SC-density correction
		0.9996	34.15	$\Delta = 14$ keV, $\rho = 2.26$ g cm ⁻³ SC-density correction

^a Different assumptions for $s_{gr,air}^{SA}$ are applied.

calculation of W , Kretschko⁸⁴ used the relation

$$\left(\frac{dE}{dX}\right)_r = W \frac{dJ}{dX} \quad (44)$$

where $(dE/dX)_r$ is the energy loss per unit pathlength including those energy transfers that are so small that the secondary electrons produced leave all their energy in the cavity chamber (i.e., a restricted stopping power). In addition, the part of the energy transfer from higher-energy secondary electrons that give ionization in the cavity is included.

The value of $(dE/dX)_r$ is theoretically derived. Its value is critically dependent on the cavity shape and an uncertainty less than a few percent seems hard to obtain. The W value given by Markus was 31.9 ± 1.9 eV at 14.8 MeV, by Ovadia *et al.* 34.3 ± 1 at 17.5 MeV, and by Kretschko varying from 33.3 ± 0.5 to 34.0 ± 0.5 eV from 10 to 25 MeV (these latter values should probably be increased by about 1% applying more recent stopping power data). It is very unsatisfactory to base precision dosimetry on these uncertain W values. Therefore, owing to the lack of reliable experimental data for high energies, it has generally been assumed that $W = 33.85$ eV, obtained at low energies, also applies at high energies. However, a more logical way would be to use the method for determination of W at Co-60 gamma also for electron beams, i.e., Eq. (43). Experimental data are available but have to be reevaluated as the purpose of the authors was to determine $s_{m,i}$ from the assumption of a value of W/e and not the other way around. The flow chart here will follow Fig. 8

The most accurate measurements which may be utilized for W calculations both for Co-60 gamma rays and electrons are based on results from Niatel¹⁰⁸ at the BIPM and Domen and Lamperti⁴³ at the NBS. The NBS graphite calorimeter was used to determine D_{gr} in these

two sets of measurements and plane-parallel graphite chambers were used for the measurements of J_{air} .

Niatel obtained $W_{\text{dry air}} = 33.78 \text{ eV}$ at the depth of 10 g cm^{-2} in graphite for Co-60 irradiation (see Table IV). The $s_{\text{gr,air}}$ value applied in this experiment was higher than those from Berger and Seltzer.¹¹ A recalculation using the latter stopping powers would instead give $W = 34.07 \text{ eV}$ assuming $\rho = 1.7 \text{ g cm}^{-3}$ and 34.15 eV assuming $\rho = 2.265 \text{ g cm}^{-3}$. Domen and Lamperti measured absolute $D_{\text{gr}}/J_{\text{air}}$ for electron beams of energies at the phantom surface of 15, 20, 25, 30, 40, and 50 MeV. W/e is obtained by using $s_{\text{gr,air}}^{\text{SA}}$ from Berger and Seltzer.¹² Here $s_{\text{gr,air}}^{\text{SA}}$ is the restricted stopping power ratio graphite to air using a cutoff energy of 10 keV. The spectral distribution of the electron fluence was computed with the Monte Carlo method. The W values derived in this way (see also Fig. 8) were plotted as a function of the electron energy, $\bar{E}_z$, at the point of measurement, calculated from Eq. (41) (see Fig. 14). Also in this case, computations were carried out using $s_{\text{gr,air}}$ based on both the graphite density $\rho = 1.7$ and 2.265 g cm^{-3} . The experimental points are, however, only indicated for the lower density in Fig. 14. The W value for Co-60 gamma rays by Niatel¹⁰⁸ is also shown in this figure recalculated to new values of $s_{\text{gr,air}}^{\text{SA}}$; see Table IV. The values in Table IV are valid for dry air and have therefore been multiplied by 0.997 to yield the value for humid air at about 50% relative humidity at normal temperature and pressure. It is seen that the W -value for 60 Co γ rays constitutes a point in a realistic curve together with those points obtained for electron radiation, provided the lower density $\rho = 1.7 \text{ g cm}^{-3}$ is accepted for graphite with electron radiation. The mean energy expended in humid air per ion pair determined in this way depends on the value of $s_{w,\text{air}}$. The fairly rapid change, that is 1.5%, in the W value from the Co-60 gamma quality ($\bar{E}_z \approx 0.3 \text{ MeV}$) to 10-MeV electrons may simply depend on an error in $s_{\text{gr,air}}$. It can be argued that the largest differences in the various sets of stopping power data are in this energy region (see Figs. 9 and 10) and therefore the largest uncertainty should be expected here. However, an energy dependence in W cannot be excluded. In order to point out that the experimental W -values plotted in Fig. 14 are based on such assumptions a subscript has been used, i.e., W_{ex} .

A similar analysis of D_w/J_{air} determined from Svensson¹³³ and Mattsson *et al.*⁹⁴ may be performed. In these investigations D_w was determined from ferrous sulfate dosimeter measurements. Recalculations to $\varepsilon_m G = 352 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$ were used. A probable energy dependence of $\varepsilon_m G$ was disregarded. Furthermore, recent data, to correct for the perturbation effect in the ionization chamber dosimetry, were applied (see Ref. 135). W/e may then be determined from $D_w/J_{\text{air}} s_{w,\text{air}}^{\text{SA}} = W/e$, taking $s_{w,\text{air}}^{\text{SA}}$ from Berger and Seltzer.¹²

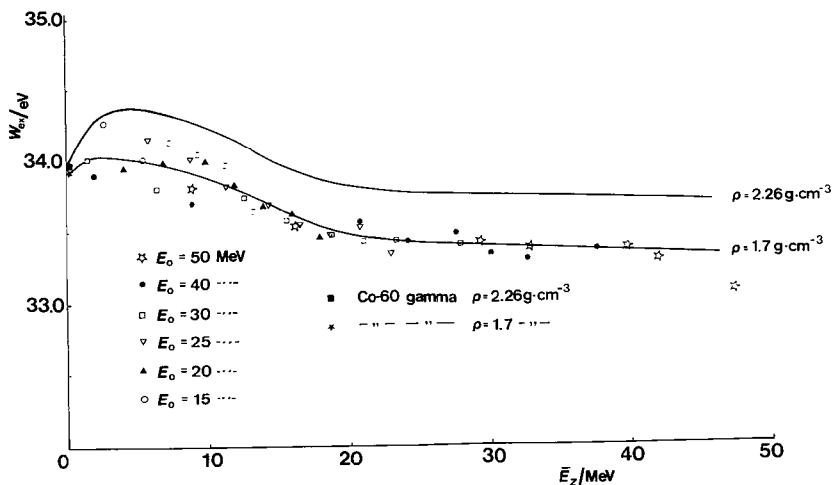


Figure 14. The mean energy expended in humid air per ion pair formed. The experimental points are obtained from the relation $W/e = D_{gr}/J_{air}s_{gr,air}$, where D_{gr}/J_{air} are taken from Ref. 43 for electrons and from Ref. 108 for Co-60 γ rays. The stopping power ratio, $s_{gr,air}$, is taken from Ref. 12. Different sets of data are obtained for the various incident energies, E_0 . The mean energy at a depth z , $\bar{E}_z$ is computed using Eq. (41). The W has been given an index ex in order to stress the semiempirical basis of the values. The lower solid line is a fit to the experimental points assuming $s_{gr,air}$ is calculated for $\rho = 1.7 \text{ g cm}^{-3}$. The upper solid line would have been obtained for $\rho = 2.26 \text{ g cm}^{-3}$.

The data from Svensson and Mattsson *et al.* agree closely with the data in Fig. 14 (see Fig. 15). The W_{ex} "lines" from Fig. 14 are included as a comparison for the two assumptions about the density of graphite (solid lines). The values based on measurements on D_w/J_{air} are scattered between the two solid lines, except at the highest energies. The data from Svensson and Mattsson *et al.* show a somewhat larger spread than those from Domen and Lamperti. One reason is that several different types of medical accelerators have been used as radiation sources in the experiments. The values at the highest energies are from a betatron. It is known that energy-degraded electrons from the collimator appear in this beam, resulting in an underestimate of $s_{w,air}^{SA}$ by possibly 0.5%–1.5%.⁸² Furthermore, an energy dependence of $\epsilon_m G$ as discussed in Section 4.3 would decrease W_{ex} at Co-60 quality and electrons above $E_z \approx 10 \text{ MeV}$; cf. Fig. 13. Considering these facts the best experimental values should be along the broken line, which also falls between the two "graphite experiment" lines.

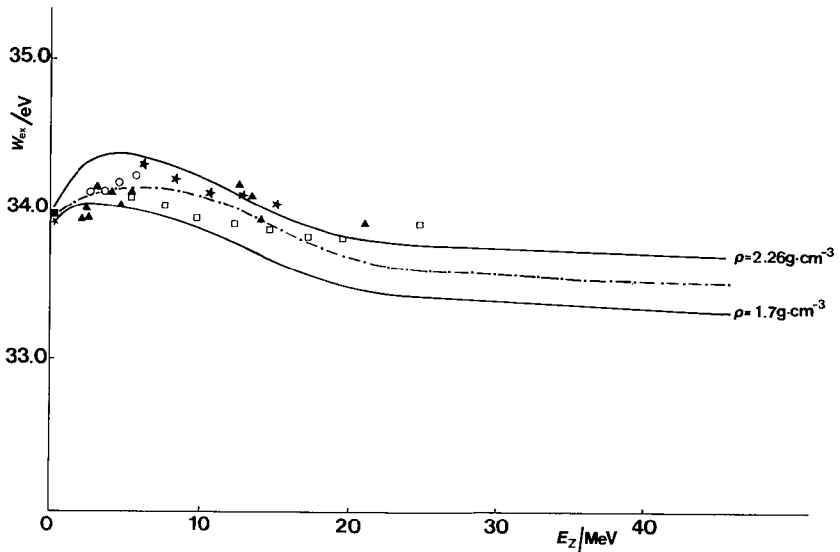


Figure 15. Experimental W values based on $W/e = D_w/J_{\text{air}}s_{w,\text{air}}$, where D_w/J_{air} has been measured by Mattsson *et al.*⁹⁴ (▲) at the reference depth for various beam energies and by Svensson¹³³ at different depths using $\bar{E}_0 = 30, 20$, and 10 MeV. The $s_{w,\text{air}}^{\text{SA}}$ are taken from Ref. 12. The solid lines are explained in Fig. 14 as well as the two experimental data for Co-60 gamma rays (at the lowest energy in the figure).

To avoid problems with different stopping powers used over the years, it would be preferable to investigate the variation of the quantity $\omega_{m,\text{air}} = (W/e)s_{m,\text{air}}$ with energy in a similar way, as $\varepsilon_m G$ is more accurately known than ε_m or G alone. This has the additional advantage that $\omega_{m,\text{air}}$ is exactly the quantity needed for determination of absorbed dose D_m from measurements of J_{air} using the relation $D_m = \omega_{m,\text{air}}J_{\text{air}}$ [cf. Eq. (21c)].

4.7. Absorbed Dose to Mass Ionization Factor $\omega_{m,\text{air}}$

Numerical values of $\omega_{w,\text{air}}$ are given in Figs. 16a and 16b for electron beams between 4 and 50 MeV. These data are based on the W_{ex} from Fig. 15 (broken line), and $s_{w,\text{air}}^{\text{SA}}$ from Berger and Seltzer.¹² A comparison of the present $\omega_{w,\text{air}}$ values with $(W/e)s_{w,\text{air}}$ data from ICRU is shown in Fig. 17. Here a constant value of W/e is assumed for humid air, 33.73 J C^{-1} . Different sets of stopping powers as well as fluence spectra calculations are used. The present $\omega_{w,\text{air}}$ values should be used for a fairly “clean” electron beam, while the data indicated as the dash-dotted line (in Fig. 17) probably apply better for clinical beams, as these beams often

are somewhat contaminated by low-energy collimator electrons at small phantom depths (Ref. 82 see also Section 5.1).

A similar analysis was carried out for photon beams. The result is given in Fig. 18 where the $\omega_{w,\text{air}}$ values are computed using the Monte Carlo method.^{4,5} The Berger and Seltzer¹¹ stopping power data were used as input. The solid lines are $\omega_{w,\text{air}}$ (i.e., $s_{w,\text{air}}^{\text{BG}} W_{\text{ex}}/e$) for different monoenergetic photon beams given as a function of the phantom depth. The broken lines represent ⁶⁰Co gamma rays, and 6- and 21-MV X rays. The spectral distributions of the Bremsstrahlung were taken from Nilsson and Brahme.¹¹⁰ Andreo's results agree fairly well with those by Nahum^{102,103} (recalculated to stopping powers from Berger and Seltzer¹¹) given as dotted lines for the monoenergetic photon beams. The deviations seen at low depths are probably due to the larger energy cutoff used by Nahum¹⁰² for the transport of secondary electrons. In the figure experimental values for ⁶⁰Co gamma rays are included from Niatel.¹⁰⁸ These measurements gave $\omega_{\text{gr},\text{air}}$ (i.e., $D_{\text{gr}}/J_{\text{air}}$) and are corrected to yield $\omega_{w,\text{air}}$ by multiplication by $(S_{\text{col}}/\rho)_{w,\text{gr}}$, taking the collision stopping powers for 0.3 MeV from Berger and Seltzer.¹¹ This experiment agrees very well with the computed values.

To allow calculation for other materials, the mean energies of the primary electrons at a few depths of interest are given in Table V. The corresponding stopping power ratios are also included. The stopping power ratio at the mean electron energy (i.e., $s_{w,\text{air}}^{\text{H}}$) is about 0.5%–1% lower than these values

5. THE CALIBRATION PROCEDURE

5.1. Background

Several steps are needed in the calibration of a medical electron accelerator. The measuring program after installation of a new accelerator and during its servicable life often includes

1. a survey of different safety aspects (radiation, mechanical, and electrical);
2. investigation of the radiation source, e.g., in terms of some of the parameters tabulated in Fig. 19;
3. investigation of the therapeutic beam, with regard to factors like energy, perturbation, uniformity, and penumbra;
4. absorbed dose calibration often based on the ionization chamber method as indicated in the figure.

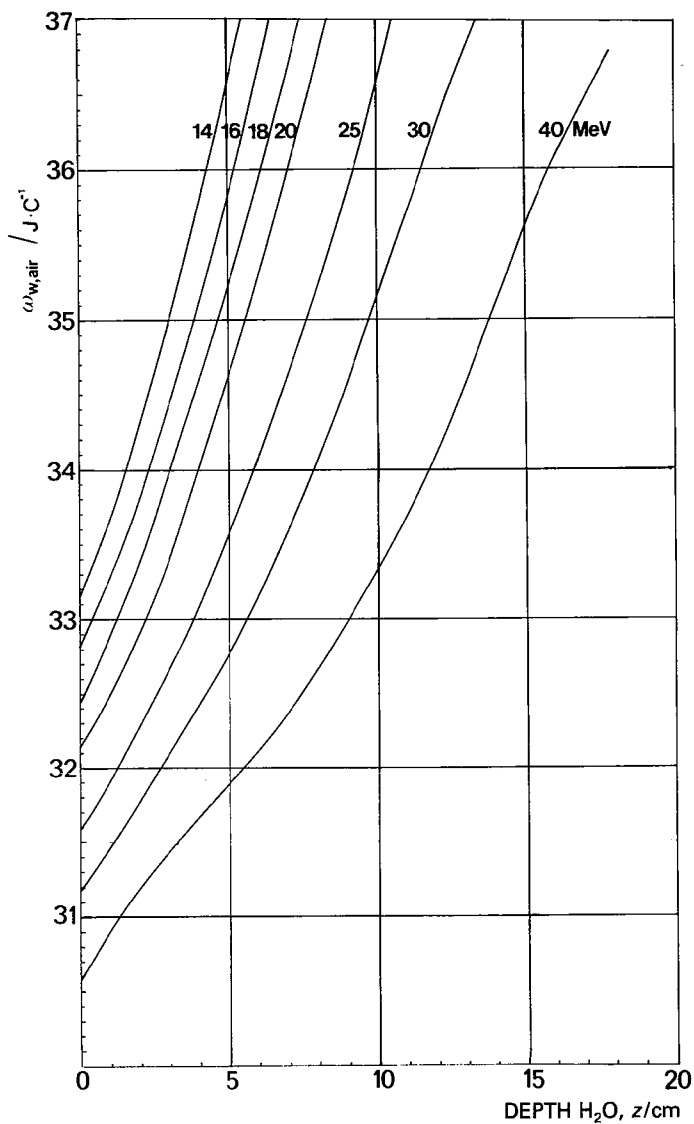


Figure 16. Absorbed dose to mass ionization values for water to air, $\omega_{w,air}$, as a function of depth for different electron energies at the phantom surface. The data are based on W_{ex} from Fig. 15 (broken line) and $s_{w,air}^{SA}$ from Ref. 12.

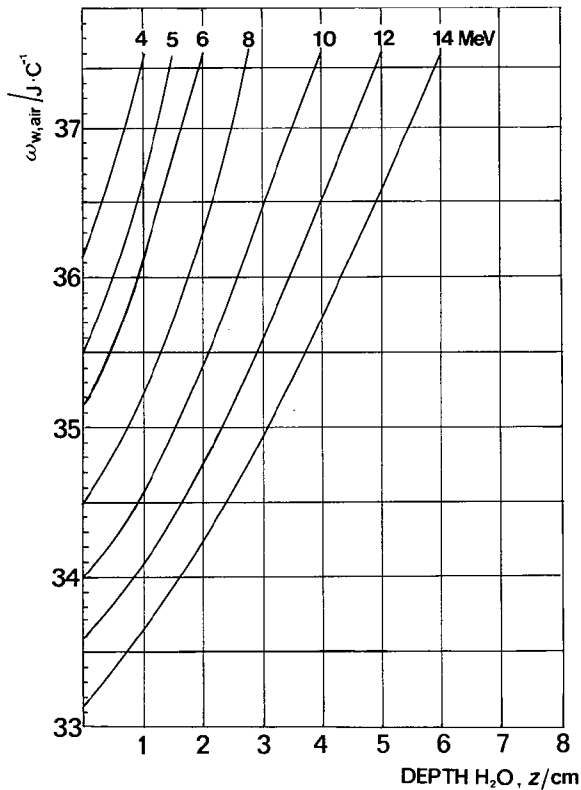


Figure 16. (continued)

Here only those parts of the procedure that are necessary in order to carry out accurate absorbed dose determinations will be discussed.

5.2. Characteristics of Electron Beams for Dosimetric Purposes

The dosimetric properties of a clinical electron beam depend significantly on the spectral distribution of the electrons. Essential parameters are the mean energy $\bar{E}$ and the most probable energy E_p of the incident electrons. Indices a , i , 0 , and z are used to indicate whether the energy is specified for the intrinsic accelerator beam (a), the initial electron beam at the inner side of the vacuum window (i), the beam at the phantom surface (0), and the beam at a depth z inside a phantom (z).

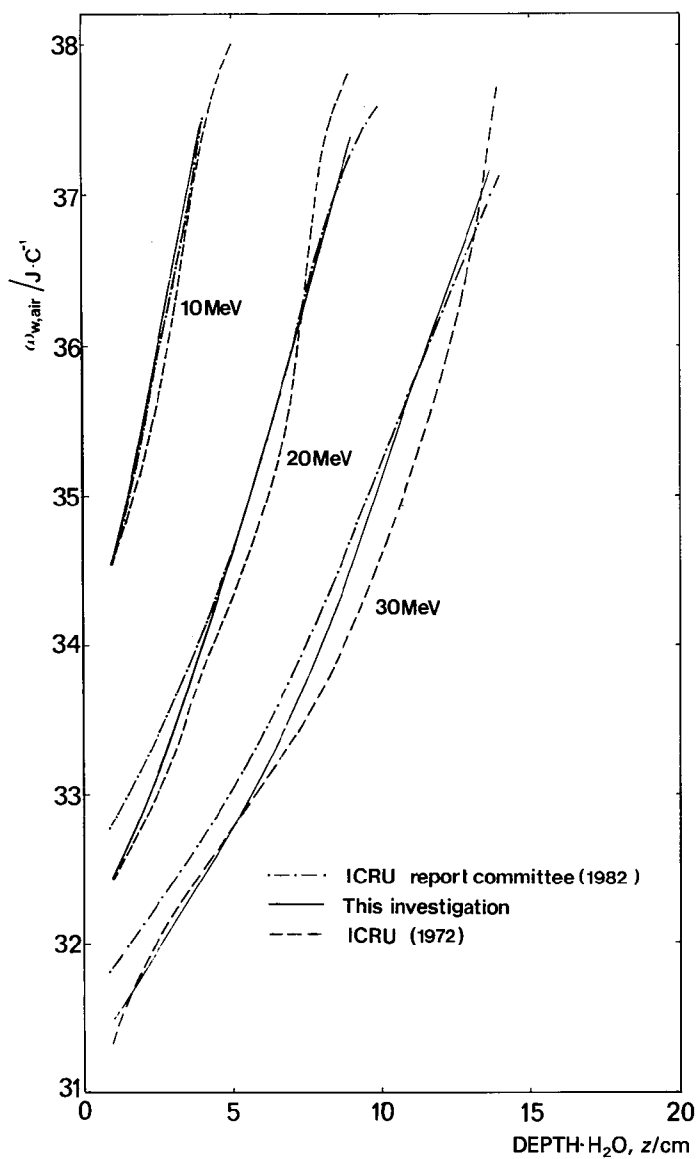


Figure 17. Comparison of $\omega_{w,air}$ as obtained in this report with data based on other assumptions.

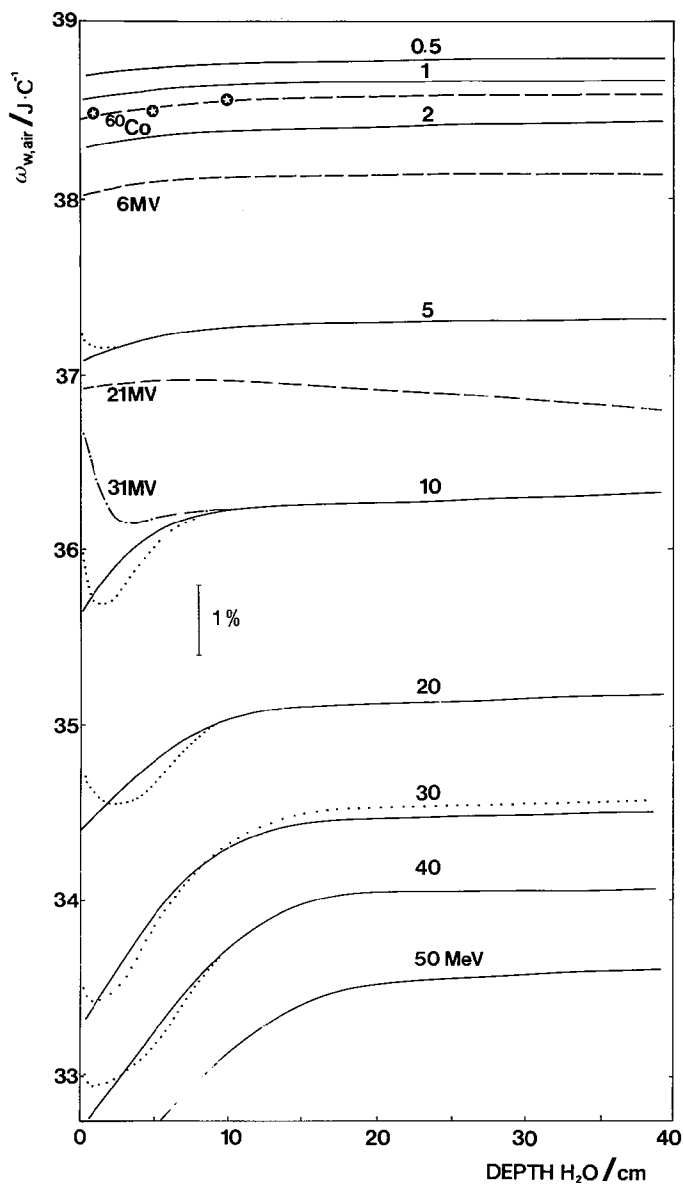


Figure 18. $\omega_{w,air}$ as a function of depth for different photon beam qualities. The notations are as follows: —, monoenergetic photons from Ref. 5; ···, monoenergetic photons from Ref. 104; ---, 6- and 21-MV X rays from Ref. 5; -·-·-, 31-MV X rays from Ref. 102; ○○○, experimental points from Ref. 108.

TABLE V
Mean Energies of Primary Electrons, Stopping Power Ratios, and
Linear Attenuation Coefficients in Photon Beams

Energy or accelerating potential	Depth (cm)	$\bar{E}_{\text{prim}}$ (MeV)	$S_{w,\text{air}}^{\text{BG}}$	$\mu_{\text{MC}} (\text{m}^{-1})$
0.5 MeV	0.5	0.142	1.135	5.24
	10	0.125	1.137	
	40	0.102	1.138	
1.0 MeV	0.5	0.351	1.131	3.94
	10	0.312	1.133	
	40	0.265	1.134	
2 MeV	0.5	0.774	1.119	2.75
	1.5	0.759	1.120	
	10	0.695	1.122	
	40	0.620	1.125	
6 MV	0.5	1.08	1.112	2.87
	1.5	1.08	1.112	
	10	0.95	1.116	
	40	0.95	1.117	
21 MV	0.5	2.10	1.092	2.29
	3.5	1.90	1.086	
	10	1.85	1.090	
	40	1.67	1.085	
5 MeV	0.5	2.65	1.087	2.05
	3.5	2.45	1.091	
	10	2.50	1.093	
	40	2.80	1.098	
10 MeV	0.5	5.02	1.047	1.66
	4.5	3.60	1.062	
	10	3.60	1.064	
	40	3.40	1.067	
20 MeV	1.0	9.26	1.016	1.52
	10	6.85	1.033	
	40	6.30	1.040	
30 MeV	1.0	14.7	0.990	1.18
	10	10.3	1.012	
	40	9.2	1.020	
40 MeV	1.5	18.7	0.979	1.14
	10	13.8	0.996	
	20	12.4	1.005	
	40	12.1	1.007	
50 MeV	1.5	23.0	0.970	1.4
	10	17.6	0.985	
	25	15.0	0.997	
	40	14.5	0.999	

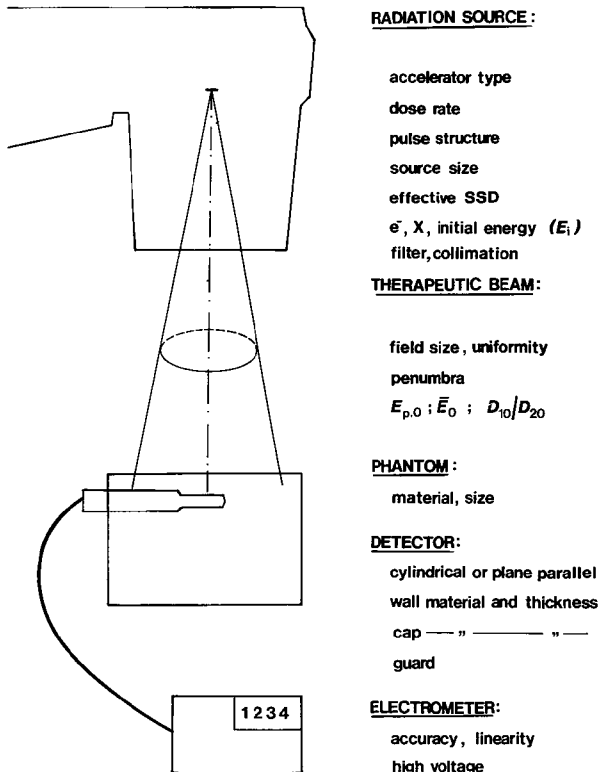


Figure 19. The different factors that affect an absorbed dose calibration.

In theoretical calculations of dose distributions and stopping power ratios it is often assumed that the electron beam is monoenergetic. This is, however, not the case with a clinical beam. For example, the energy spread of the intrinsic accelerator beam is sometimes significant. This energy distribution will often be changed in the beam transport system (i.e., in the bending magnets and associated slits), and further in the scattering material from the inner side of the accelerator window to the patient (phantom).

The most probable energy at the phantom surface, $E_{p,0}$, has in the past often been used as an input parameter in the determination of absorbed dose conversion factors for electron dosimetry, i.e., C_E or $s_{w,air}$ values (see, e.g., Ref. 74). The recommended values of C_E or $s_{w,air}$ have, however, been computed assuming a broad monoenergetic beam perpendicularly incident on a phantom. In Ref. 100 the use of $\bar{E}_0$ as an input

parameter was recommended as this was considered to give more correct values of $s_{w,\text{air}}$, at least at the reference depth.

It has been shown that $\bar{E}_0$ is correlated to the half value depth, R_{50} , of the central axis depth dose curve²⁶ according to

$$\bar{E}_0 = k_1 R_{50} \quad (45)$$

where $k_1 = 2.33 \text{ MeV cm}^2 \text{ g}^{-1}$. The NACP¹⁰⁰ has described a simple procedure for the measurement of R_{50} from a depth ionization curve, which can be recommended.

The most probable energy at the phantom surface $E_{p,0}$ can be obtained from the practical range R_p :

$$E_{p,0} = k_2 + k_3 R_p + k_4 R_p^2 \quad (45)$$

where $k_2 = 0.22 \text{ MeV}$, $k_3 = 1.98 \text{ MeV cm}^{-1}$, and $k_4 = 0.0025 \text{ MeV cm}^{-2}$.^{29,100}

Johansson and Svensson⁸² investigated six different types of accelerators and found that $E_{p,0} - \bar{E}_0$, as determined with the range formulas, was as large as 2 MeV in the energy range 17–22 MeV. Using $E_{p,0}$ instead of $\bar{E}_0$ would therefore have given too low $s_{w,\text{air}}$, in some cases by about 1%. Furthermore, they found that $\bar{E}_0$ values determined from R_{50} , were probably overestimated, the reason being that the relation between R_{50} and $\bar{E}_0$ holds only if the electrons have been degraded in energy when passing through materials in the beam (see Ref. 26), like scattering foils and vacuum windows, but does not fully apply if there is a large energy degradation and scattering of electrons in the collimators. Such electrons will reach the phantom and may be fairly important, at least up to the depth of dose maximum, as shown by Lax and Brahme.⁸⁵ Their presence has therefore only minor influence on the 50% depth.

In order to investigate the consequence for absorbed dose determinations Johansson and Svensson⁸² measured $\omega_{w,\text{air}}$, i.e., D_w/J_{air} . D_w was measured with the ferrous sulfate dosimeter and J_{air} with a plane-parallel chamber. They found that $\omega_{w,\text{air}}$ at the reference depth for a given value of $\bar{E}_0$ (about 20 MeV) determined from R_{50} varied by up to 4% for different accelerators. The highest values were obtained using accelerators with cone types of collimators, since cones give a large contamination of low-energy electrons.

From the results of these experiments it is obvious that the methods applied today often overestimate the mean energy, resulting in an overestimation of $\omega_{w,\text{air}}$ ($s_{w,\text{air}}$) of 0.5%–1.5%, with most accelerators at about $\bar{E}_0 = 20 \text{ MeV}$. This figure can even be larger for some beams. A further increase by about 1% results if $E_{p,0}$ is used instead of $\bar{E}_0$ as an input parameter.

It could thus be concluded that the ionization chamber dosimetry is rather uncertain for beams contaminated with a large contribution of low-energy electrons and that absorbed dose calibrations for such beams should be carried out with a less energy-dependent method, e.g., ferrous sulfate dosimeter or a liquid ionization chamber. It is, therefore, important to be able to analyze the beam quality to find out if there is a large contamination of low-energy electrons. This may be done by studying the shape of the central axis depth absorbed dose curve. The NACP¹⁰⁰ has recommended some minimum values of the dose gradient. These recommendations have been set in order to assure an acceptable beam performance for radiation treatment. It seems, however, that if these recommendations are fulfilled then also an acceptable accuracy is obtained in the dosimetric constants, i.e., an uncertainty below about 3% (see Ref. 134). This is, of course, a fairly large uncertainty and simple methods other than those based on ionization dosimetry ought to be further developed.

5.3. Characteristics of Photon Beams for Dosimetric Purposes

It is not simple to define a unique radiation quality parameter to use in dosimetry of photon beams from therapy accelerators. For instance, Rawlinson and Johns¹¹⁸ found that the central axis depth-dose curve from a linear accelerator at 25-MV X rays was very similar to that from a betatron for 16-MV X rays. The reasons were attributed to differences in the target and flattening filter.

The main features influencing the quality of photon beams are as follows:

1. E_i (the electron energy at the target), and possible energy spread Γ_i . Generally, the intrinsic distribution is very narrow for betatrons, $\Gamma_a < 20$ keV,⁵⁷ and microtrons, $\Gamma_a < 40$ keV,²⁸ and broader for linear accelerators, particularly of the standing wave type, where Γ_a is around 10% (i.e., of $E_{p,a}$) and the traveling wave type, where Γ_a is about half this value.^{39,140,143}

2. For a betatron, the target could be fairly thin as electrons penetrating the target are turned away by the magnetic field. For other medical accelerators, the electrons are instead completely stopped in a thick target. It is clear that the mean energy of the generated photons is greatest in a thin target. A composite target, with a thin layer of high atomic number (Z) material followed by a low- Z electron stop, falls between the thin and thick target situation (see Ref. 27). The high-energy electrons here mainly produce X rays in the first layer. The remaining electron energy is dissipated in collision losses in the low- Z material without stopping the high-energy bremsstrahlung produced in the first layer.

3. Depending on the relative location of the bremsstrahlung spectrum and the absorption minimum of the flattening filter, the spectral distribution of the beam can be influenced considerably. Prodigoršak *et al.*¹¹⁶ showed that a substantial hardening of the photon spectrum was obtained by replacing a lead flattening filter by one of aluminum at 25 MV. However, this will certainly introduce problems if uniform dose distribution is the aim at all phantom depths. The reason is that the mean photon energy decreases with the distance from the central ray, in an unflattened photon beam. This effect will be magnified with a filter that increases the mean photon energy (i.e., filter more low-energy photons than high-energy photons, particularly in the central part of the field, where the filter is thickest). The penetration of the photons is therefore greatest at the beam axis. Even if an optimization of the uniformity is achieved at one phantom depth, poor uniformity may be obtained at other depths. The manufacturers make different types of compromises to achieve both an acceptable uniformity and depth-dose distribution. This fact means that the spectral distributions may differ considerably for accelerators using the same accelerating potential.

4. The beam may be contaminated by secondary photons and electrons from the flattening filter, the collimator system, and the air. This influence is particularly large in the dose buildup region of broad low- and high-energy photon beams and is at low energies mainly due to Compton electrons from the air.¹⁰⁹ The influence from scattered photons is generally of the order of 3% of the primary photon dose.^{6,110}

It appears from the discussion above that the maximum acceleration potential, U , or maximum photon energy $h\nu_{\max}$, in the X-ray spectrum gives an insufficient description of the beam quality. This is also the case from a dosimetric point of view. The increase in C_λ by replacing an aluminum filter by a lead filter was, for instance, calculated by Nahum¹⁰² to 1% at 31 MV.

It was, therefore, suggested by NACP¹⁰⁰ to base the choice of stopping power ratio on the slope of the exponential portion of the depth-dose curve as characterized by the D_{10}/D_{20} dose ratio rather than the acceleration potential U . This dose ratio reflects the attenuation coefficient of the photon spectrum in the beam and should therefore be better related to its dosimetric properties that depends on μ_{en} (cf. Eq. (18)). In fact, if both D_{10} and D_{20} , or more generally D_{z_1} and D_{z_2} , are located on the pure exponential falloff portion of the depth-dose curve, the practical attenuation coefficients (cf. Refs. 27 and 61) can be obtained from

$$\mu_p = \frac{\ln(D_{z_1}/D_{z_2})}{z_2 - z_1} \quad (47)$$

Of course, this practical attenuation coefficient includes the inverse square decrease in photon dose with distance from the effective source, but this can easily be removed by elementary calculations:

$$\bar{\mu} \approx \mu_p - \frac{2}{s_{\text{vir}} + (z_1 + z_2/2)} \quad (48)$$

where $\bar{\mu}$ is the mean attenuation coefficient for the field size and photon spectrum at hand (cf. Refs. 121 and 27). If D_{10}/D_{20} data are used, the negative factor in Eq. (48) becomes 1.74 m^{-1} assuming $s_{\text{vir}} = 100 \text{ cm}$.

To illustrate the interrelation of various types of attenuation coefficients in water, the most important ones are plotted in a single diagram as a function of photon energy in Fig. 20. The total (μ) and the energy (μ_{en}) absorption coefficients are taken from data by Hubbel.⁷² As discussed by Nilsson and Brahme,¹¹¹ the mean attenuation coefficient for various field sizes should fall within these bounds. This is indeed the case with the various experimental data also included in the figure.

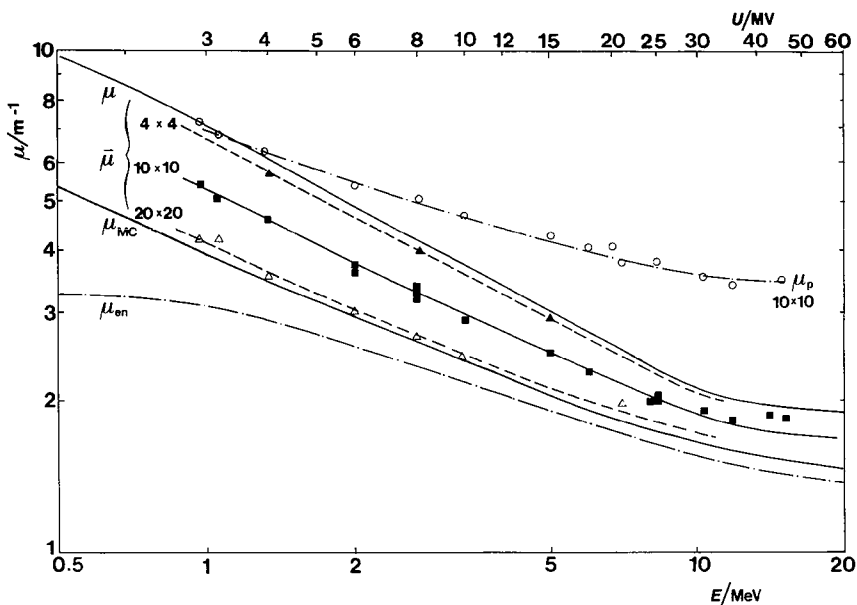


Figure 20. The energy dependence of the most important attenuation coefficients for photon beams in water. The experimental data points for μ_p and $\bar{\mu}$ are mainly taken from Ref. 27, μ_{MC} is from Ref. 5, and μ_{en} and μ from Ref. 72. The lower scale is for monoenergetic photons (μ_{MC} , μ_{en} , and μ). The upper scale for bremsstrahlung spectra is shifted a factor of 3 relative to the monoenergetic scale.

It is seen that, for small fields (<4 cm across), the total attenuation coefficient is approached, whereas for large fields (>20 cm across) the curve is close to the energy absorption coefficient curve. For comparison, the attenuation coefficient in infinite parallel beams calculated by the Monte Carlo method (μ_{MC}) is included. As should be expected, it practically coincides with the experimental broad beam attenuation data. The increasing difference between μ_{MC} and μ_{en} at low photon energies is due to the increased multiple scatter and "local" absorption of the scattered photons that are disregarded in μ_{en} . It is also seen from Fig. 20 that $\bar{\mu}$ for 10×10 cm² fields falls approximately halfway between μ and μ_{MC} .

The experimental bremsstrahlung points are plotted at a monochromatic energy equal to one third of the acceleration potential U in MV (see the upper scales). This potential corresponds precisely to the energy of the electrons incident on the target and thus the energy of most energetic photons $h\nu_{max}$. To illustrate the influence of beam divergence, the μ_p values for a 10×10 cm² field are also included in the diagram. Naturally this curve represents the primary experimental data and the $\bar{\mu}$ values are obtained after application of Eq. (48).

Based on this information, it would be most natural to relate the dosimetric properties of photon beams to $\bar{\mu}$ in broad beams, as this parameter is closest to the dosimetrically important quantity μ_{en} . However, owing to the well-behaved relation between $\bar{\mu}$ for large and small fields and μ_p , it might be more convenient to continue the use of standard depth-dose data at the standard field size of 10×10 cm². It is therefore recommended to use μ_p determined for a field size of 10×10 cm² and an SSD of 100 cm as a beam quality parameter for choice of dosimetric quantities. This has the advantage that previously used D_{10}/D_{20} values can be used simply by using Eq (45).

For this purpose the $\omega_{w,air}$ value for different photon beams is plotted in Fig. 21 as a function of μ_p . The dashed line pertains to monoenergetic photon beams, whereas the solid line represents the typical values for ordinary bremsstrahlung spectra. Both curves are based on Monte Carlo calculations by Andreo.⁵ Because the data are plotted as a function of μ_p , the dependence of $\omega_{w,air}$ on the spectral shape and filtration is considerably reduced. It is natural that the solid line falls below the dashed line for monoenergetic beams, as the low-energy photon component of the bremsstrahlung spectrum decreases the beam penetration more than it increases the $\omega_{w,air}$ value owing to the slow variation of $\omega_{w,air}$ and $s_{w,air}$ with energy when the density effect becomes negligible at low energies. The dashed line thus represents the maximum possible deviation for extremely hard photon spectra. The solid line could thus be recommended for use over a wide range of bremsstrahlung

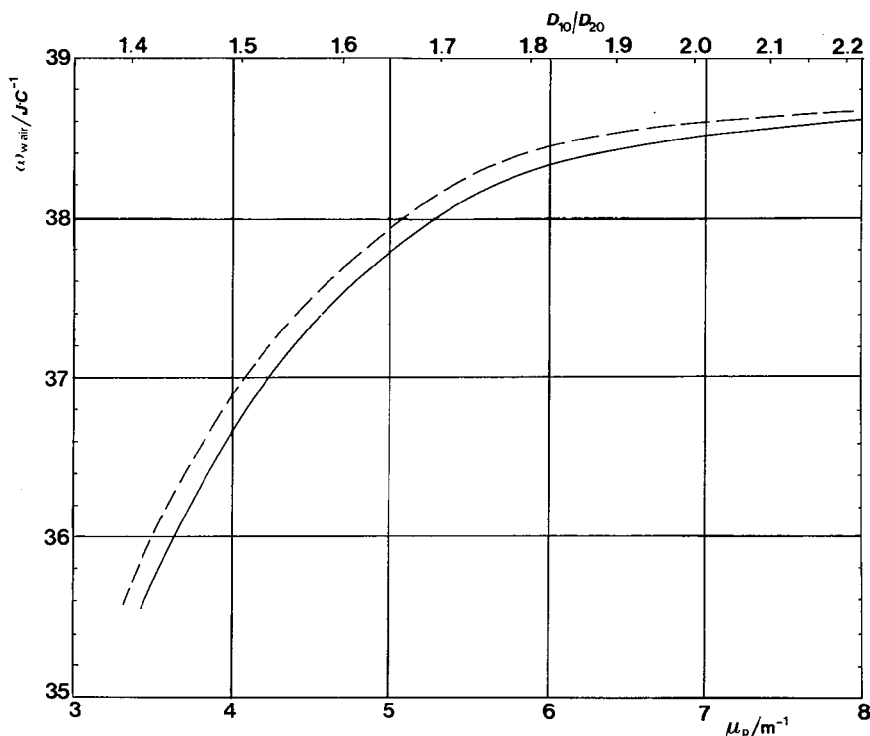


Figure 21. The relation between $\omega_{w,air}$ and the practical attenuation coefficient μ_p . The dashed line holds for monoenergetic photons, whereas the solid line is representative for most common bremsstrahlung spectra.

spectra. Only when the spectrum is unusually soft, e.g., after thick high atomic number filters at high photon energies, or when it is unusually hard after a thick low atomic number filter, will the true values fall slightly below and above the solid line, respectively. The differences should rarely be more than 0.5%.

5.4. Phantom Materials

5.4.1. Background

The general practice in radiation therapy with high-energy photon and electron beams is to state the absorbed dose to the target volume or an organ of interest in units of absorbed dose to water. This is done even though it is well known that the absorbed dose to the tissue in question may differ considerably from the absorbed dose to water.

With the increased use of advanced diagnostic equipment, such as

computerized tomography (CT) and to some extent also ultrasound and magnetic resonance imaging, the accurate delineation of the patient geometry has been greatly improved. In addition the CT scanner and also the Compton scatter tomograph give the possibility of determining the density and even the atomic composition of the different tissues. In some cases it may therefore be desirable to reconsider the old practice of stating absorbed dose to water, especially since failure to do so may add to the total uncertainty in dosimetry.

In a recent analysis,²⁵ it was shown that both muscle and adipose tissues are water equivalent to within 1% from 2 to 40 MeV for electrons and from 0.2 to 20 MV for photons. Deviations of 5% or larger are generally obtained for the absorbed dose to bone as compared to water. Water is thus a very good soft tissue substitute and the natural phantom material for accurate absolute dosimetry in therapy beams. However, in some cases it is more convenient to use solid phantom materials owing to the ease of handling and high geometric reproducibility. When a solid phantom is used to calibrate in units of absorbed dose to water, it is essential that all necessary corrections factors are known, as both the particle fluence and the absorbed dose at dose maximum or the reference point generally are different. When the phantom is made of a solid insulator like polystyrene and polymethylmethacrylate, care should be taken when it is used in electron beams owing to the buildup of long-lasting electrostatic fields inside the insulator that may influence the detector reading. Large errors in dosimetry based on a cylindrical chamber placed in such a preirradiated phantom have been demonstrated.^{54,92}

5.4.2. Electrons

With a given electron fluence incident on a uniform medium, the absorbed dose at the reference point (dose maximum) in the material will generally differ from that in water. This is due to two principal effects. First, the mass stopping power of the medium will generally be different, resulting in different absorbed doses for a given electron fluence. Secondly, the mass scattering power may be different, resulting in a difference in the buildup of the fluence of primary electrons.

The ratio of the absorbed doses in a medium to that in water may under assumption of equilibrium in the transport of secondary electrons be written

$$\frac{D_m}{D_w} = \frac{\int (\Phi_E)_m \left(\frac{S_{\text{col}}}{\rho} \right)_m dE}{\int (\Phi_E)_w \left(\frac{S_{\text{col}}}{\rho} \right)_w dE} = \frac{\Phi_m \left(\frac{\bar{S}_{\text{col}}}{\rho} \right)_m}{\Phi_w \left(\frac{\bar{S}_{\text{col}}}{\rho} \right)_w} \quad (49)$$

where the integrals are taken over the spectrum of primary electrons, Φ_E , and the mean collision stopping powers are defined according to

$$\left(\frac{\bar{S}_{\text{col}}}{\rho}\right)_m = \frac{\int (\Phi_E)_m \left(\frac{S_{\text{col}}}{\rho}\right)_m dE}{\int (\Phi_E)_m dE} \quad (50)$$

There are two main sources for the difference in fluence at dose maximum. First, because different depths of dose maximum in different media, the inverse square law will produce different fluence reductions at dose maximum in divergent beams. Secondly, because of differences in linear scattering powers the fluence at dose maximum will differ owing to differences in the angular distribution of the electrons.

According to Brahme and Lax,²⁵ the ratio of the primary electron fluences at dose maximum may be written

$$\frac{\Phi_m}{\Phi_w} = 1 + \frac{1}{2}(T_m R_{100,m} - T_w R_{100,w}) - 2 \left(\frac{R_{100,m} - R_{100,w}}{s_{\text{vir}}} \right) \quad (51)$$

where T is the linear scattering power, R_{100} the depth of dose maximum, and s_{vir} the distance to the virtual point source.²⁴

The fluence factor given by Eq. (51) is plotted in Fig. 22 as a function of the most probable energy at the surface for characteristic tissues and some phantom materials of interest. Also included in the diagram are experimental data points from measurements by Mattsson *et al.*⁹³ It is seen that for therapeutic electron beams at least in the range 3–30 MeV, PMMA (polymethylmethacrylate) has the interesting property that the response of a small Bragg–Gray detector at dose maximum agrees within half a percent with its response in water. This is essentially because the linear scattering power of PMMA and water are very close and thus the electron fluence at dose maximum is practically the same even though the absolute dose level is about 3% lower in PMMA due to its lower mass stopping power. For graphite (C) and polystyrene (PS) over a wide range of energies, the fluence is about 1% higher and lower, respectively, as compared to water. It should be pointed out that small high atomic number additions may change the properties of the phantom material. This is the case with TiO_2 in white polystyrene, which therefore is more water equivalent than clear polystyrene.

5.4.3. Photons

For a given incident photon energy fluence differential in energy, Ψ_E , the ratio of the absorbed dose at secondary electron equilibrium in a

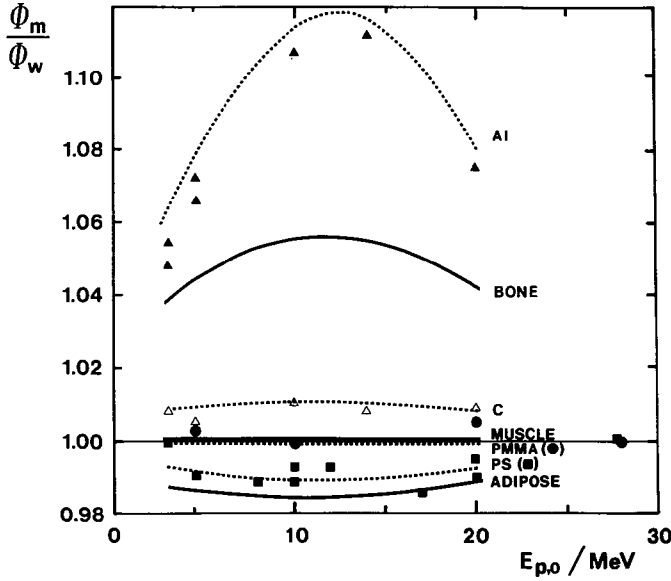


Figure 22. The ratio of the fluence at dose maximum in a medium to that in water as a function of the most probable energy at the surface $E_{p,0}$ (dotted and solid lines from Ref. 25). The experimental points are taken from Ref. 93.

uniform medium to the equilibrium dose in water is given by

$$\frac{D_m}{D_w} = \frac{\int (\Psi_E)_m \left(\frac{\bar{\mu}_{en}}{\rho} \right)_m dE}{\int (\Psi_E)_w \left(\frac{\bar{\mu}_{en}}{\rho} \right)_w dE} = \frac{\Psi_m \left(\frac{\bar{\mu}_{en}}{\rho} \right)_m}{\Psi_w \left(\frac{\bar{\mu}_{en}}{\rho} \right)_w} \quad (52)$$

where the bar indicates a mean mass energy absorption coefficient averaged over the energy fluence differential in energy:

$$\left(\frac{\bar{\mu}_{en}}{\rho} \right)_m = \frac{\int (\Psi_E)_m \left(\frac{\bar{\mu}_{en}}{\rho} \right)_m dE}{\int (\Psi_E)_m dE} \quad (53)$$

The ratio of the signal measured with a small Bragg-Gray detector, that is, the ratio of the electron fluence in the two media, is obtained by

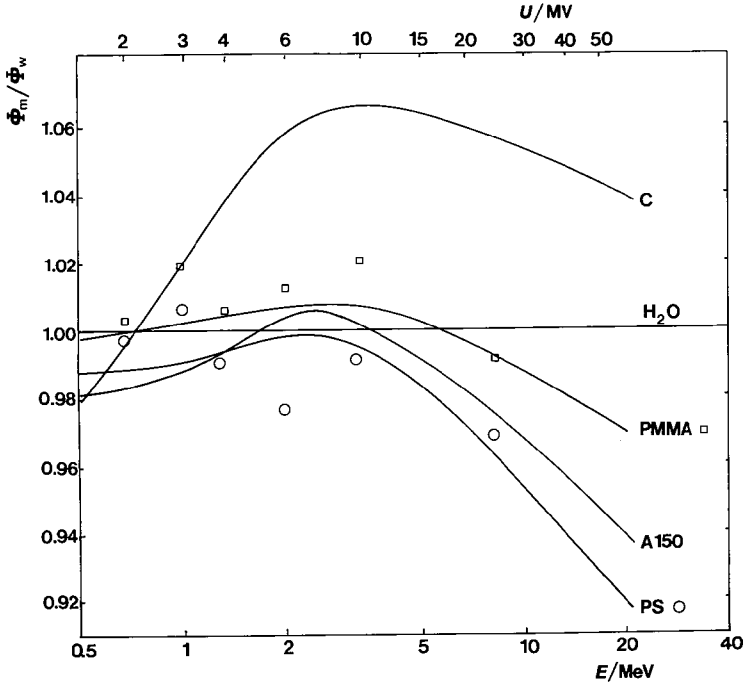


Figure 23. The ratio of the electron fluence at dose maximum in a medium to that in water for different photon beam energies. The experimental data points for PMMA and PS for a $15 \times 15 \text{ cm}^2$ field¹⁰⁷ represent the signal ratio for a Farmer chamber at 5 g/cm^2 depth.

dividing by the respective mean stopping powers according to

$$\frac{\Phi_m}{\Phi_w} = \frac{\Psi_m \left(\frac{\bar{\mu}_{en}}{\rho} \right)_m \left(\frac{\bar{S}_{col}}{\rho} \right)_w}{\Psi_w \left(\frac{\bar{\mu}_{en}}{\rho} \right)_w \left(\frac{\bar{S}_{col}}{\rho} \right)_m} \quad (54)$$

This ratio is plotted in Fig. 23 using stopping powers determined at the mean energy of primary electrons [cf. Eq. (39)] taken from Table III. The fluence ratio was evaluated at dose maximum with correction for the inverse square attenuation according to

$$\frac{\Psi_m}{\Psi_w} \approx 1 + 2 \left(\frac{R_{100,w} - R_{100,m}}{s_{vir}} \right) \approx 1 + \frac{2R_{100,w}}{s_{vir}} \left[1 + \frac{(r_0/\rho)_m}{(r_0/\rho)_w} \right] \quad (55)$$

where r_0 is the continuous slowing down range in g cm^{-2} [cf. Eq. (51) and Ref. 25]. The last step in Eq. (55) is based on the assumption that the buildup depth is a constant fraction of r_0 , independent of the material.

As seen by comparing Figs 23 and 22 PMMA is almost as water equivalent for photons as for electrons, at least from ^{60}Co to about 20 MV. Nath *et al.*¹⁰⁷ showed that the response of a Farmer-type chamber at a depth of 5 g cm^{-2} in PMMA is the same as in water within 1% in the same energy range. Nilsson and Brahme¹¹¹ showed by electron transport calculations based on Eq. (6) that the ratio between the absorbed dose at dose maximum and the collision kerma at the phantom surface for ^{60}Co gamma rays was the same within 0.1% for water, air, and aluminum, at infinite SSD. This indicates that in photon beams, energy fluence differences are mainly due to differences in the inverse square corrections at the depth of dose maximum.

5.5. Absorbed Dose Determination by the Ionization Method

5.5.1. Background

A determination of the absorbed dose in high-energy photon and electron beams by the user is, in principle, carried out as a two-step procedure. First the ionization chamber is calibrated at a reference quality by the user at a standards laboratory, and then the chamber is used to determine the absorbed dose with the user's beam. A number of conversion and correction factors have to be applied. Different sets of factors are needed depending on what physical quantity the calibration refers to (X , K_{air} , etc.), on the calibration geometry (free in air, in phantom, etc.), and on the design of the chamber (size, shape, wall material). Factors may be needed to convert from one physical quantity (X or K_{air}) at the calibration quality to a different quantity of interest for the user, i.e., $s_{w,\text{air}}$; also, these are dependent on the irradiation geometry and chamber design.

The ICRU reports No. 14⁷³ and No. 21⁷⁴ have given some sets of factors to be applied under special conditions (see Fig. 2). Similar factors were recommended by the national protocols, i.e., SCRAD,¹²³ HPA,⁷⁰ and NACP.⁹⁹ The uncertainty in using these procedures seems, however, to be fairly large, as none of the protocols gives a clear description of under what conditions these factors are to be applied. For instance, the wall material in the chamber is seldom water equivalent, as assumed, but not stated, in the ICRU photon report (No. 14),⁷³ or air equivalent, as assumed, but not stated, in the electron report (ICRU No. 21).⁷⁴

A discussion of the influence of the wall material on the calibration and on the measurement of absorbed dose was given by Mutsuzawa,⁹⁸

Greening,⁶⁰ and Almond and Svensson.² It was shown that just an incorrect assumption regarding the wall materials often gave errors of 3% or more.

A very careful analysis of all the factors involved was undertaken by Loevinger.⁸⁶ His analysis covered different calibration assumptions, i.e., exposure, air-kerma, and absorbed dose to water calibration for ⁶⁰Co gamma beams. It appears from this analysis that the dosimetrical procedure, if not restricted to certain well-defined geometries (e.g., chamber design), will be very complicated and that some simplifications are needed.

The NACP¹⁰⁰ described a procedure to be applied for specially designed ionization chambers. Cylindrical chambers, with a cavity diameter between 4 and 6 mm, were recommended for all high-energy photon beams and electron beams above 10 MeV. The chamber wall and buildup caps should be of the same material. Factors were given for walls of graphite, and air equivalent and water ($\approx$ tissue) equivalent materials. For electron beams with $\bar{E}_0$ below 10 MeV, a specific plane-parallel chamber was recommended, in order to reduce the uncertainty in the perturbation correction at these qualities.¹⁰¹

It was shown that very good agreement was obtained in the absorbed dose values as determined from measurements according to the NACP^{99,100} and from ferrous sulfate measurements using $\epsilon_m G = 352 \times 10^{-6} \text{ m}^2 \text{ kg}^{-1} \text{ Gy}^{-1}$, generally within 1% (see Ref. 82). However, it should be pointed out that some of the factors recommended in the NACP (e.g., $s_{w,\text{air}}$) are in error by several percent if newer data from Berger and Seltzer¹¹ are adopted. In most cases the error is canceled by the product of mutually compensating factors in the dose equation. It is therefore important that the stopping power ratios are not changed, if more correct values are made available, without also changing other factors in the protocol such as k_m (see Table VI) and W/e . These factors will be discussed below in order to identify their interrelations.

5.5.2. Calibration Quality

At the standards laboratory much experience has been gained in determining exposure in a ⁶⁰Co gamma beam. The exposure standards from different laboratories generally agree within one or two tenths of one percent. In order to obtain a uniform dosimetry throughout the world this very good agreement should be used.

The exposure is determined from the following relationship (see, e.g., Ref. 87):

$$X_c = J_{\text{air}} \prod_i K_i \quad (56)$$

TABLE VI
The Calibration Procedure by NACP^{100,101} and That Recommended in the Text

	NACP ^{100,101}	This paper
Calibration at calibration quality (index <i>c</i>)	<i>Geometry</i> Co-60 γ rays SSD = 100 cm, to chamber center; field size 10 cm $\times$ 10 cm; free in air <i>Quantity calibrated</i> $K_{\text{air},c}$: air-kerma derived from $K_{\text{air},c}(1 - g) = X_c W/e$ <i>Calibration factor</i> $N_K = \frac{K_{\text{air},c}}{M_c}$ Derived factor: $N_D = \frac{\bar{D}_{\text{air},c}}{M_c} = N_K(1 - g)k_m k_{\text{att}}$	<i>Geometry</i> As NACP^{100,101} <i>Quantity calibrated</i> X_c <i>Calibration factor</i> $N_x = \frac{X_c}{M_c}$ Derived factor: $N_J = \frac{J_{\text{air}}}{M_c} = N_x k_m k_{\text{att}}$
Assumption	$N_D = \frac{\bar{D}_{\text{air},c}}{M_c} = \frac{D_{\text{air},u}}{M_u}$ <i>Geometry</i> Water phantom; chamber center at the reference point; chamber without build-up cap	No assumptions <i>Geometry</i> As NACP (1980)
Absorbed dose measurement at users' quality (index <i>u</i>)	<i>Dose determination</i> $D_{w,u} = \bar{D}_{\text{air},u}(s_{w,\text{air}})_u p_u$ $= N_D M_u (s_{w,\text{air}})_u p_u$	<i>Dose determination</i> $D_{w,u} = J_{\text{air}}(\omega_{w,\text{air}})_u (p_{w,\text{air}})_u$ $= N_J M_u (\omega_{w,\text{air}})_u (p_{w,\text{air}})_u$

where X_c is the exposure at the calibration radiation quality; J_{air} is the mass ionization, i.e., charge of one sign liberated in the air cavity per unit mass of air [see Eqs. (4b) and (4c)]; and $\prod_i K_i$ is the product of the various correction and conversion factors. The factors, K_i , convert J_{air} into exposure. Some of these factors are trivial, e.g., the correction for recombination losses, and are therefore not discussed here. Other factors are of more fundamental character. Corrections thus have to be applied for the non-air-equivalence of the chamber wall, given the symbol K_m , and the attenuation of the photons in the buildup cap and chamber wall, given the symbol K_{att} . The material in the standard chamber is generally graphite, as this material is fairly "air equivalent." K_m is calculated from [see Eq. (55) and Refs. 18 and 87]:

$$K_m = (\mu_{\text{en}}/\rho)_{\text{air,gr}} s_{\text{gr,air}}^{\text{SA}} \quad (57)$$

K_{att} , as applied below, does not include all the attenuation in all the wall

as the electrons giving ionization are produced up-stream in the chamber wall and buildup cap. K_{att} depends on the thickness of the wall and cap and also on its shape. K_{att} can be computed, e.g., using the Monte Carlo method,¹⁰⁶ electron transport calculations,¹¹¹ or measurements using an extrapolation method.

The exposure X_c , for a given irradiation time, is in this way determined for a specified point in the ^{60}Co gamma beam. The very good agreement in exposure, as determined by different standards laboratories, relies on the fact that K_m is computed in the same way.

However, the user is not really interested in a calibration of the ionization chamber in exposure. The main interest is instead to make it possible to determine the relation between the detector signal and the mass ionization, J_{air} [Eqs. (4a-c)]. In practice, dose determinations are based on a knowledge of $\bar{D}_{\text{air}}$. The Bragg-Gray relation Eq. (21) or its generalized form Eqs. (26) and (27) are then applied to determine absorbed dose to water at the user's quality (see Table VI).

It may be possible for the user to measure, Q_{air} , directly, but the mass of air, m_{air} , is more complicated to determine. In effect the "effective mass," $m_{\text{air,eff}}$, is what is determined during calibration. However, if a graphite chamber of design similar to that used by the standards laboratory is available, then it should be very simple to make an exposure calibration of the chamber at the standards laboratory and applying Eq. (56) to determine mass ionization per meter reading J_{air}/M_c ($=N_7$).

As the chamber often is of a very different design from that at the standards laboratory, other correction factors must often be supplied. In the NACP the inverse of the K_m and K_{att} , given the symbols k_m and k_{att} , respectively, are reported for some types of chambers. It is thus obvious that k_m for a graphite chamber is the inverse value used by the standards laboratory. The cavity size dependence of $s_{\text{air,gr}}^{\text{SA}}$ can generally be neglected, as the variation for ordinary cavity sizes is below 0.2%, cf Boutillon and Niatel.¹⁸ An error in K_m is therefore canceled out provided that the same physical data are used both by the standards laboratory and the user.

In the Nordic protocol an absorbed dose to air ionization chamber factor

$$N_D = \frac{\bar{D}_{\text{air,c}}}{M_c} = \frac{J_{\text{air}}(W/e)}{M_c}$$

was defined. Here $\bar{D}_{\text{air,c}}$ is the mean absorbed dose to air inside the air cavity of the chamber [see Eq. (4b)]. J_{air} and therefore also $\bar{D}_{\text{air}}$ may be determined from Eq. (55), i.e., $J_{\text{air}} = k_m k_{\text{att}} X_c$, again leaving out the normal corrections. The Nordic standards laboratories give the ionization chamber calibration in air-kerma rather than exposure. The relation

between these two quantities is

$$K_{\text{air},c}(1 - g) = X_c \frac{W}{e} \quad (58)$$

where g is the fraction of the charged particle energy lost to bremsstrahlung in air ($g = 0.004$ at ^{60}Co ; see Boutillon¹⁶). Thus N_D may be written

$$N_D = \frac{K_{\text{air},c}(1 - g)k_m k_{\text{att}}}{M_c} \quad (59)$$

It is seen [Eq. (58)] that the numerical value of W/e is needed at the calibration quality in order to determine N_D . As the analysis in Section 4 indicates that this value still is fairly uncertain, a more direct way would be to determine a mass ionization factor $N_J = J_{\text{air}}/M_c = X_c k_m k_{\text{att}}/M_c$ at the calibration quality. Furthermore, $(\omega_{w,\text{air}})_u$ for the users quality can be used which takes a possible energy dependence of W/e into consideration (see Section 4.5 and Table VI).

5.5.3. Absorbed Dose to Water Determination at the User's Quality

The calibrated ionization chamber can now be used to determine the absorbed dose to water utilizing the Bragg–Gray equation (see Table VI). The NACP^{100,101} used a total perturbation factor, p_w , that corrects both for the fluence perturbation of the wall and cavity as shown in Fig. 5 (Section 3.4.4) and for the shift of effective point of measurement as the center of the chamber is placed at the reference point. Experimental values of p_u have been recommended by NACP¹⁰⁰ and are given for a cylindrical chamber that has an inner diameter of 5 mm with a 0.5-mm-thick graphite or water equivalent wall. The modified procedure, described in Table VI, uses the equation

$$D_{w,u} = N_J M_u (\omega_{w,\text{air}})_u (p_{m,\text{air}})_u \quad (60)$$

Values of $(\omega_{w,\text{air}})_u$ are given in Fig. 16 for electron radiation and in Fig. 18 for photon radiation.

$(p_{m,\text{air}})_u$ is the fluence perturbation correction correcting for these effects indicated in Fig. 5. However, it is here assumed that the ionization chamber is placed with its effective point of measurement at the reference depth. In most practical cases with electron and photon beams the effective point of measurement is located at $0.55r$ in front of the chamber center for cylindrical chambers (see Section 3.4.3.1).

The experimental $(p_{m,\text{air}})_u$ values from Fig. 7 are recommended for electron beams (the broken line). The values are plotted as a function of the mean energy of the primary electrons at the point of measurements. This energy could be derived from Eq. (41). A more simple estimate could, however, be obtained by assuming that the electron energy decrease with 2 MeV per cm penetration in the water phantom. These values can also be used for chambers having thin walls of low atomic number materials other than water as the primary electrons mainly are responsible for the absorbed dose to the air in the cavity and the difference between in- and outflow from the cavity of secondary electrons should be fairly small.

The $(p_{m,\text{air}})_u$ for photon beams are generally assumed to be equal to 1.000 if the chamber wall is made of water equivalent material ($m = w$) and if the chamber is placed with the effective point of measurement at the depth of interest. The wall material (m) is, however, more critical than for electron radiation as the photon generated electrons in the wall may give a large contribution to J_{air} . Almond and Svensson² used a simple two-component theory, assuming that a fraction β of the ionization comes from electrons generated in the water and a fraction $(1 - \beta)$ from electrons generated in the wall. Using their approach $p_{m,\text{air}}$ would be approximated by

$$(p_{m,\text{air}})_u = \frac{\beta(s_{w,\text{air}})_u + (1 - \beta)(s_{m,\text{air}})_u[(\mu_{\text{en}}/\rho)_{w,m}]_u}{(s_{w,\text{air}})_u} \quad (61)$$

As an example, in Table VII, $p_{\text{gr,air}}$ values are shown for $\beta = 0.2$ and $\beta = 0.8$. The lower value, $\beta = 0.2$, is relevant for a ^{60}Co gamma beam using a common thickness of the graphite wall, i.e., 0.5–1 mm, while for acceleration potentials >20 MV values of $\beta > 0.8$ should be applied instead. The agreement between theory and experiment is reasonable. A minor systematic error may, however, be introduced in the experiment as it was assumed that the reference chamber made of A-150 was completely water equivalent, which may not be the case (see Fig. 21), particularly if the recent stopping power data from Berger and Seltzer¹¹ are trusted.

5.6. Calibration Procedures Based on Solid Dosimeters

The ionization chamber is today the base for most dose measurements at a hospital. It is therefore very unfortunate that some fundamental features of this dosimeter always will cause problems resulting in fairly large uncertainties, e.g., the quality dependence and perturbation correction. The ferrous sulfate dosimeter does not have these drawbacks and

TABLE VII
Values of $(p_{gr,air})_u$ Given as a Function of the
Beam Quality^a

Radiation quality	$(p_{gr,air})$		
	Experimental 0.5 mm wall	Theoretical	
		$\beta = 0.2$	$\beta = 0.8$
⁶⁰ Co γ rays	0.975	0.985	0.996
4 MeV	0.978	0.983	0.996
6 MeV	0.980	0.978	0.995
8 MeV	0.982	0.980	0.995
10 MeV	0.984	0.981	0.996
15 MeV	0.989	0.986	0.996
20 MeV	0.992	0.990	0.997
25 MeV	0.993	0.996	0.998
30 MeV	0.995	0.999	1.000
40 MeV	0.997	1.005	1.001

^a The experimental values have been measured as the ratio of the response from a cylindrical chamber made of A-150 and from a graphite chamber at 10 cm depth in a water phantom. The responses of the chambers were normalized in a high-energy electron beam. The theoretical values were obtained from Eq. (60). The stopping power ratios $s_{m,i}^{BG}$ were calculated for 10 cm depth using data from Berger and Seltzer.¹¹ The values of μ_{en}/ρ were taken from Hubbell.⁷¹

seems at present to be the best choice when accuracy is most important. However, this dosimeter system has been in use for many years and has still not become very popular owing to the laborious technique. Therefore there is still a need for an accurate method that is attractive to the hospital physicist.

Recently Domen⁴⁰ developed a new type of calorimeter which made it possible to measure the temperature increase at a point of interest in water after it had been irradiated at common dose rates. The method is based on the use of a microthermistor sandwiched between two very thin insulating polyethylene films in a water phantom. With a dose rate of 1 Gy min^{-1} and an exposure time of 3 min it was possible to achieve a standard deviation of about 0.6%. This technique is surely possible to improve. Also it needs to be explained why this calorimeter measure an absorbed dose value 3.5% higher for ⁶⁰Co gamma rays than if the dosimetry is based on a graphite calorimeter measurement transferred to water with the ionization chamber method; see Domen.⁴¹ This difference is overcome by using the microthermistor in a polystyrene phantom,⁴² but then absorbed dose to water is not directly measured.

Still, the transfer from an absorbed dose determination at the standards laboratory to the user's quality must be carried out. If an ionization chamber method is applied, then again most of the uncertainties discussed in Sections 4 and 5 will be introduced. Thus, there is a great need for a precision dosimeter with small energy dependence. Also, the dosimeter should not disturb the electron and photon fluence at the point of interest in the water phantom and should have a high spatial resolution. It seems that the liquid ionization chamber would satisfy these criteria; cf. the transfer of dosimetry carried out in the experiment reported by Mattsson *et al.*⁹⁵ and Wickman and Svensson¹⁴⁷ to determine $\epsilon_m G$ (see Section 4.3).

The liquid ionization chamber was described already by Strahel.¹³¹ However, a more detailed analysis of the physical properties was first carried out by a group in Toulouse (Refs. 34, 90, 91, etc.). A practical chamber design was developed by Wickman,^{144,145} who used 2,2,4-trimethylpentane as the radiation sensitive liquid. The chambers by Wickman made high-resolution and precision measurements possible. The corrections for general recombination and temperature were of about the same magnitude as for an air ionization chamber, that is, generally below a few percent. Furthermore, the chambers seemed to have an insignificant quality dependence for electron beams, that is, a constant response per unit absorbed dose independent of the energy at the phantom surface and the depth in the phantom. Measurements with this type of chamber by Svensson *et al.*¹³⁹ and Johansson and Svensson⁸² confirmed these results and showed, in addition, that the quality dependence for photon beams was fairly small (2% between 6 and 40 MV).

The liquid ionization chamber thus has many advantages for use as a transfer instrument. The chamber has until now been in use in only a few departments. The reason seems to be that the chamber needs some special care in handling. However, recently, Wickman has constructed a chamber that also seems to be sufficiently rugged for use in most departments.¹⁴⁶

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