CHAPTER 9.

CALIBRATION OF PHOTON AND ELECTRON BEAMS

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9.1.   INTRODUCTION

Modern radiotherapy relies on an accurate dose delivery to the prescribed target volume. The International Commission on Radiation Units and Measurements (ICRU) has recommended an overall accuracy in tumour dose delivery of $\pm 5\%$ based on an analysis of dose response data and on an evaluation of errors in dose delivery in a clinical setting. Considering all uncertainties involved in the dose delivery to the patient, the $\pm 5\%$ accuracy recommendation is by no means easy to attain.

Before clinical use, the output of photon and electron beams produced by external beam radiotherapy machines must be calibrated. This basic output calibration is but one, albeit very important, of the links constituting the chain representing an accurate dose delivery to the patient. The other links refer to: (i) the procedures for measurement of relative dose data, equipment commissioning and quality assurance; (ii) treatment planning, and finally (iii) the actual patient setup on the treatment machine.

- The basic output for a radiotherapy machine is usually stated as the dose rate for a point *P* at a reference depth z_{ref} (often the depth of dose maximum z_{max}) in a water phantom for a nominal source-surface or source-axis distance (*SSD or SAD*) and a reference field size (often 10×10 cm^2) on the phantom surface or the isocenter. The output for kV x-ray generators and teletherapy units is usually given in Gy/min, while for clinical accelerators it is given in Gy per monitor unit (Gy/MU).

- For superficial and orthovoltage beams and occasionally for beams produced by teletherapy radioisotope machines, the basic beam output may also be stated as the *air-kerma rate* in air (in Gy/min) at a given distance from the source and for a given nominal collimator or applicator setting.

The basic output calibration of photon and electron beams is carried out with radiation dosimeters and special radiation dosimetry techniques.

- *Radiation dosimetry* refers to a determination by measurement and/or calculation of the absorbed dose or some other physically relevant quantity, such as air-kerma, fluence, equivalent dose, etc. at a given point-of-interest in a given medium.

- A *radiation dosimeter* is defined as any device that is capable of providing a reading M that is a measure of the dose D deposited in dosimeter's sensitive volume V by ionizing radiation.

  - A dosimeter that produces a signal from which the dose in its sensitive volume can be determined without requiring calibration in a known field of radiation is referred to as an *absolute dosimeter*.
  - Dosimeters requiring calibration in a known radiation field are called *relative dosimeters*.

The basic output calibration of a clinical radiation beam, by virtue of a direct measurement of dose or dose rate in water under specific reference conditions, is referred to as *reference dosimetry*. Three types of reference dosimetry techniques are currently known:

(1)   *Calorimetry*;
(2)   *Fricke dosimetry*;
(3)   *Ionisation chamber dosimetry*.

These dosimeters can be used as absolute dosimeters but are seldom used as such in clinics because their use in absolute dosimetry is cumbersome and, moreover, calibration in a known radiation field offers certain advantages such as traceabilty to a standards laboratory. When an absolute dosimeter is used independently, it relies on its own accuracy instead of referring to a standard in common with other radiation users.

9.1.1.   Calorimetry

Calorimetry is the most absolute of the three reference dosimetry techniques, since it relies on basic definitions of either electrical energy or temperature. In principle, calorimetric dosimetry is simple; in practice, however, the need for measuring extremely small temperature differences makes the technique very complex and relegates it to sophisticated standards laboratories.

Two main types of absorbed dose calorimeters are currently used in standards laboratories:

(1)   *Graphite calorimeters*;
(2)   *Sealed water calorimeters*.

In *graphite calorimeters* the average temperature rise is measured in a body that is thermally insulated from surrounding bodies ("jackets") by evacuated vacuum gaps. Gap corrections and dose transfer procedures are used in conjunction with graphite calorimeters, to allow for the transfer of absorbed dose from graphite to water.

In stagnant *sealed water calorimeters,* use is made of the low thermal diffusivity of water that enables the temperature rise to be measured directly at a point in (continuous) water. Dose transfer procedures are not needed but the measurement and analysis is complicated by the presence of conductive heat loss (or gain) and by the heat defect induced by radiolysis.

9.1.2. Fricke dosimetry

The energy of ionizing radiation absorbed in certain media produces a chemical change in the absorbing medium and the amount of this chemical change may be used as a measure of absorbed dose. The best-known chemical radiation dosimeter is the Fricke dosimeter which relies on oxidation of ferrous ions into ferric ions in an irradiated ferrous sulfate solution. The amount of ferric ion produced in the solution is measured by absorption spectrometry with ultraviolet light at 304 nm that is strongly absorbed by the ferric ion.

Fricke dosimetry (sometimes referred to as chemical dosimetry or ferrous sulfate dosimetry) depends on an accurate knowledge of the radiation chemical yield of ferric ions, measured in moles produced per 1 J of energy absorbed in the solution. The chemical yield is related to the older parameter, the G-value, defined as the number of ferric molecules produced in the ferrous sulfate solution by 100 eV of absorbed energy. The accurate value of the chemical yield is difficult to ascertain because the chemical yield is affected to a certain degree by the energy of radiation, dose rate, and temperature of the solution during irradiation and readout. The best G-value for cobalt-60 gamma rays is 15.6 molecules per 100 eV corresponding to a chemical yield of 1.607×10^{-6} mol/J. The typical dynamic range for ferrous sulfate Fricke dosimeters is from a few Gy to about 400 Gy making Fricke dosimetry impractical for routine use in a clinic.

9.1.3. Ionisation chamber dosimetry

The ionisation chamber is the most practical and most widely used type of dosimeter for accurate measurement of machine output in radiotherapy. It may be used as an absolute or as a relative dosimeter. Its sensitive volume is usually filled with ambient air and the dose-related or dose rate-related measured quantities are the ionisation charge Q or ionisation current I, respectively, produced by radiation in the chamber sensitive air mass m_{air}. Charge Q and air mass m_{air} are related to absorbed dose in air D_{air} by:

$$D_{air} = \frac{Q}{m_{air}} \left(\frac{W_{air}}{e} \right),$$

(9.1)

where (W_{air}/e) is the mean energy required to produce an ion pair in air per unit charge (the current value for dry air is 33.97 eV/ion pair or 33.97 J/C).

The subsequent conversion of the air cavity dose D_{air} to *dose-to-medium* (usually water) D_w is based on the Bragg-Gray or Spencer-Attix cavity theories (see Chapter 2 and Section 9.4. in this chapter).

The sensitive air volume or mass in an ionisation chamber is determined:

(1)   *Directly by measurement* (chamber becomes an absolute dosimeter under special circumstances);

(2)   *Indirectly through calibration* of the chamber response in a known radiation field (chamber is used as a relative dosimeter).

Mean energy expended in air per ion pair formed. It is generally assumed that a constant value of (W_{air}/e) can be used for the complete photon and electron energy range used in radiotherapy dosimetry. However, there is no direct experimental support for such an assumption as the data available has been obtained only from measurements with cobalt-60 and cesium-137 gamma ray beams and 2 MV x rays. The value (W_{air}/e) = (33.85 ± 0.15) J/C early recommended by the ICRU came from a weighted mean value of the available experimental data, obtained mainly from absorbed dose measurements using a graphite calorimeter and a graphite ionisation chamber in a graphite phantom. The two methods for deriving the absorbed dose to graphite must yield the same dose value and one gets:

$$(W_{air}/e) = \frac{D_{calorimetry}}{(Q/m_{air})\, s_{graphite.air}},$$
(9.2)

where

Q/m_{air} is the charge Q collected in air mass m_{air} and corrected for influence quantities and

$s_{graphite,air}$ is the ratio of stopping powers for graphite and air calculated for the photon or electron beam energy used.

This method of evaluation requires a change in (W_{air}/e) when the stopping-power ratio $s_{graphite,\,air}$ is changed. Following the introduction of new electron stopping power data by the ICRU in 1984, the value of (W_{air}/e) has been modified to (33.97 ± 0.06) J/C for dry air.

Analysis of the available experimental data at higher energies, mainly for electron beams, has suggested that energy dependence in (W_{air}/e) cannot be ruled out but experimental uncertainties and the use of different stopping-power ratios over the years do not allow a definitive conclusion to be reached on this question.

It is known that the (W_{air}/e) value for air at a temperature of 20°C, pressure of 101.325 kPa and 50% relative humidity is 0.6% lower than that for dry air at the same temperature and pressure, resulting in a value of 33.77 J/C instead of 33.97 J/C. Thus, for the same amount of energy available for creating charge, 0.6% more charge will be created in air at 50% relative humidity than in dry air (at 20°C and 101.325 kPa).

9.1.4.   Reference dosimetry with ionisation chambers

Three types of *ionisation chambers* may be used in reference dosimetry as absolute dosimeters:

(1)   *Standard free air ionisation chamber*;
(2)   *Cavity ionisation chamber*; and
(3)   *Phantom-embedded extrapolation chamber*.

Standard free air ionisation chamber measures the *air-kerma in air* according to its definition by collecting all ions produced by the radiation beam that result from the direct transfer of energy from photon to primary electrons in a defined volume of air. Determination of the *air-kerma in air* or *air-kerma rate in air* requires the accurate knowledge of (W_{air}/e). For practical reasons related to the range of charge carriers in air the use of the standard free air ionisation chamber is limited to photon energies below 0.3 MeV.

Cavity ionisation chamber measures the *air-kerma in air* for energies in the range from 0.6 MeV to 1.5 MeV by making use of the Bragg-Gray cavity relation. Analogously to the standard free-air chamber, ions are collected in air, but this time inside a cavity with a *known cavity volume* surrounded by a graphite wall thick enough to provide full build-up of secondary electrons. The Bragg-Gray equation relates the *dose-to-air* in the cavity of known volume to the *dose-to-medium* in which the secondary electron spectrum is being built up, *i.e.*, the graphite wall (for the thick-walled chambers used in PSDLs). The absorbed dose to the wall is related to the collision *air-kerma in air* through the mass energy absorption coefficient ratio, wall to air. The collision *air-kerma in air* is related to the total *air-kerma in air* by correcting for the fractional energy expended in radiative interactions. In addition to the need for an accurate knowledge of the sensitive air volume, wall correction factors are required to account for the effect of photon attenuation and scattering in the chamber wall. An accurate knowledge of (W_{air}/e) as well as cavity volume and bremsstrahlung fraction is required to determine the *air-kerma (rate) in air*. Finally, standards laboratories implement additional correction factors, such as point source non-uniformity correction factor and factors that account for deviations from the Spencer-Attix cavity theory.

Phantom-embedded extrapolation chamber. Un-calibrated, variable air-volume extrapolation chambers, built as an integral part of a water-equivalent phantom in which the dose is measured, can serve as radiation dosimeters in the measurement of absorbed dose for megavoltage photon and electron beams. Standard dosimetry protocols are based on Bragg-Gray or Spencer-Attix cavity theories that provide a simple linear relationship between the dose at a given point in the medium and the ratio *Q/m* where *Q* is the ionisation charge collected in mass *m* of air in the measuring cavity inside the medium. In extrapolation chambers, the ratio *Q/m* is constant and may be replaced in the cavity relationship by the derivative *dQ/dm* that can be measured accurately through a controlled variation in electrode separation. The conversion of cavity dose to dose in medium is based on the Spencer-Attix cavity theory. Just like in the standard free air ionisation chamber and in the cavity ionisation chamber, the extrapolation chamber dosimetry relies on an accurate knowledge of the value of (W_{air}/e).

9.1.5.   Clinical beam calibration and measurement chain

The theoretical aspects of the three reference dosimetry techniques discussed above are all well understood; however, none of the three techniques, for one reason or another, is practical for routine clinical use. Therefore, clinical photon and electron beams are most commonly calibrated with ionisation chambers that are used as relative dosimeters and have calibration coefficients determined either in air or in water and traceable to a national primary standards dosimetry laboratory. The chamber calibration coefficient essentially obviates the need for an accurate knowledge of the chamber sensitive air volume.

The standard ISO 31-0 on Quantities and Units has provided guidelines with regard to the use of terms *coefficient* and *factor*. The former should be used for a multiplier possessing dimensions; the latter should be reserved for a dimensionless multiplier. For consistency, the widely disseminated practice of using the term calibration factor is updated here to *calibration coefficient*.

The traceability of a calibration factor to a national primary standards dosimetry laboratory (PSDL) implies that:

    (1)   The chamber was calibrated directly at the PSDL in terms of the *air-kerma in air* or absorbed *dose-to-water*; or

    (2)   The chamber was calibrated directly at an accredited dosimetry calibration laboratory (ADCL) or secondary standards dosimetry laboratory (SSDL) that trace their calibration to a PSDL; or

    (3)   The chamber calibration coefficient was obtained through a cross-calibration with another ionisation chamber (user's secondary standard), the calibration coefficient of which was measured directly at a PSDL, ADCL, or SSDL.

9.1.6. Dosimetry protocols

The procedures to be followed when calibrating a clinical photon or electron beam are described in international or national radiation *dosimetry protocols* or *dosimetry codes of practice* and the choice of which protocol to use is largely left to individual radiotherapy departments. Dosimetry protocols are generally issued by national or regional organizations, such as the AAPM (North America), IPEMB (UK), DIN (Germany), NCS (The Netherlands and Belgium) and NACP (Scandinavia), or by international bodies such as the IAEA. This procedure ensures a high level of consistency in dose determination among different radiotherapy clinics in a given country and also between one country and another.

9.2. IONISATION CHAMBER-BASED DOSIMETRY SYSTEMS

As shown schematically in Fig. 9.1, ionisation chamber-based dosimetry systems are in principle quite simple and consist of three main components:

    (1)   Suitable ionisation chamber;
    (2)   Electrometer; and
    (3)   Power supply.

The circuitry of a simple ionisation chamber-based dosimetry system resembles a capacitor (ionisation chamber) connected to a battery (power supply) with the electrometer measuring the "capacitor" charging or discharging current.

9.2.1. Ionisation chambers

Ionisation chambers incorporate three electrodes that define the chamber sensitive air volume. The sensitive volume is typically on the order of 0.1 to 1 cm^3 in ionisation chambers used for calibration of clinical photon and electron beams. The three electrodes are:

    (1)   *Polarizing electrode* which is connected directly to the power supply;

    (2)   *Measuring electrode* which is connected to ground through the low impedance electrometer to measure charge or current produced in the chamber sensitive volume; and

    (3)   *Guard electrode* which is directly grounded and serves two purposes: (i) it defines the chamber sensitive volume and (ii) it prevents the measurement of chamber leakage currents.
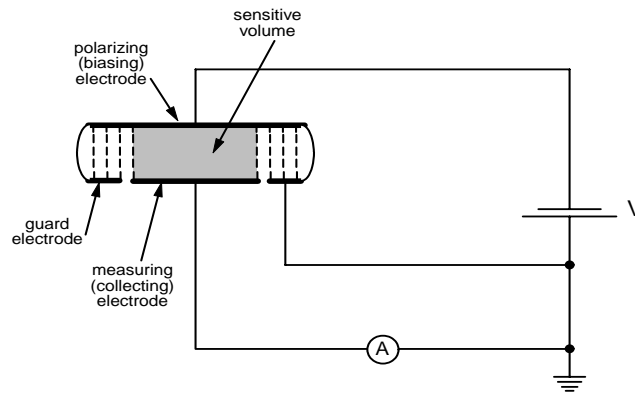
FIG. 9.1.   Schematic diagram of the circuitry for an ionisation chamber-based dosimetry system. A represents the electrometer, V the power supply. The ionisation chamber is usually connected to the electrometer through a shielded low noise triaxial cable.

Two types of ionisation chambers are used in routine beam calibration:

(1)   *Cylindrical* (often referred to as thimble) chambers and
(2)   *Parallel-plate* (sometimes called end-window or plane-parallel) chambers.

The more common *cylindrical chambers* are used in calibration of orthovoltage and megavoltage x-ray beams and electron beams of 10 MeV and above, while parallel-plate chambers are used in calibrations of superficial x-ray beams, in calibrations of low energy electron beams, and in surface dose measurements for megavoltage photon beams.
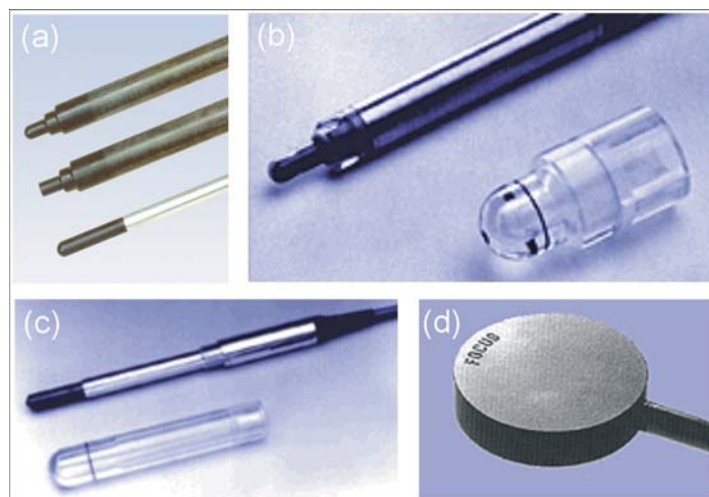

FIG 9.2.   *Examples of typical ionisation chambers used in radiation therapy: (a) cylindrical ionisation chambers used for relative dosimetry, (b) pinpoint mini-chamber, (c) Farmer-type cylindrical chamber (top) with buildup cap (bottom), (d) parallel plate Roos-type electron beam ionisation chamber*

Air is usually used as the sensitive gas in an ionisation chamber. The initial event of inter-action of indirectly ionizing radiation with the chamber is characterized by a release of high-energy electrons in the chamber wall or phantom through photoelectric effect, Compton effect, or pair production. Some of these electrons enter the chamber sensitive volume and ionize air molecules producing positive ions and low-energy electrons in the chamber sensitive gas. The low-energy electrons attach themselves to electronegative oxygen mole-cules in air forming negative ions. Thus, in an air-based ionisation chamber the charged particles that are collected are the positive and negative ions (*ion pairs*) rather than positive ions and electrons.

9.2.2. Electrometer and power supply

An ionisation chamber is essentially a capacitor in which leakage current or leakage charge is induced through the action of the radiation beam. The charge or current that is induced in the chamber is very small and must be measured by a very sensitive charge or current measuring device (electrometer). The power supply in ionisation chamber/electrometer circuits is either a stand-alone unit or it forms part of the electrometer. In either case, it is important that one can change the magnitude and polarity of the voltage produced by the power supply, so that the ion collection efficiency of the chamber may be determined for a particular radiation beam (see Section 9.3.).

9.2.3. Phantoms

Water is the standard phantom material for dosimetry measurements of photon and electron beams; however, dosimetric measurements are often carried out in more practical solid materials, such as polystyrene, Lucite, A-150 tissue equivalent plastic, Solid Water (WT1), Solid Water (RMI-457), Plastic water, Virtual water, etc., that mimic water in terms of mass density, number of electrons per gram, and effective atomic number.

- The effective atomic number Z_{eff} depends on the atomic composition of the mixture as well as on the type and quality of the radiation beam.

- For low energy photons where photoelectric effect is dominant over the Compton process, Z_{eff} of a mixture is defined by:

$$Z_{eff} = \sqrt[3.5]{\sum_i a_i Z_i^{3.5}} \quad ,$$ (9.3)

where

a_i is the mass fraction,
Z_i is the atomic number

Using Eq. (9.3) we obtain a Z_{eff} of 7.8 for air and 7.5 for water.

- For megavoltage photon and electron beams Z_{eff} of a mixture is defined by:

$$Z_{eff} = \frac{\sum_i a_i \dfrac{Z_i^2}{A_i}}{\sum_i a_i \dfrac{Z_i}{A_i}} \quad ,$$ (9.4)

where

a_i is the mass fraction,

Z_i is the atomic number,

A_i is the atomic mass of element *i*.

Water is the most universal soft tissue substitute material, useful in both photon and electron beam measurements. Plastic solid materials are often used in phantom measurements; however, they are not universal tissue substitutes, since not all three equivalency parameters for plastics can be matched adequately with those of water.

- For photon beams, tissue-equivalency or water-equivalency implies a match in mass energy absorption coefficient, mass stopping power and mass scattering power.

- For a phantom to be water-equivalent for electron dosimetry, it must match the linear stopping power and the linear scattering power of water. This is approximately achieved, if the phantom material has the same electron density and the same atomic number as water.

- Generally, water is recommended as phantom material for calibration of megavoltage photon and electron beams. The depth of calibration for megavoltage x-ray beams is 10 cm, while for electron beams it is at a reference depth z_{ref}. The margin on the phantom around the nominal field size must be at least 5 cm of water in all directions and there should be at least 10 cm of water beyond the chamber.

- For kilovoltage x-ray beams, the current plastics used in dosimetry cannot be considered truly water-equivalent and their use for calibration of x-ray beam output should be approached with care.

9.3.   CHAMBER SIGNAL CORRECTION FOR INFLUENCE QUANTITIES

For each ionisation chamber, reference conditions are described by a set of influence quantities for which a chamber calibration coefficient is valid without any further corrections. Influence quantities are defined as quantities that are not the subject of a measurement, but yet influence the quantity being measured. Examples of influence quantities in ionisation chamber dosimetry are: ambient air *temperature, pressure* and *humidity*; *applied chamber voltage* and *polarity*; chamber *leakage currents*; and *chamber stem effects*. If the chamber is used under conditions that differ from the reference conditions, then the measured signal must be corrected for the effects of influence quantities to obtain the correct signal value.

9.3.1.   Air temperature, pressure and humidity effects: $k_{T,P}$

The mass of air contained in the sensitive volume of the chamber is equal to $\rho_{air} V_{eff}$, with ρ_{air} the air density and V_{eff} the effective sensitive volume of the chamber. Since most ionisation chambers are open to the ambient atmosphere, ρ_{air} is a function of the atmospheric pressure, temperature and humidity and so is the charge, collected by the chamber, as both are correlated.

It is common practice to fix the value of ρ_{air} to certain conditions and convert the chamber reading to these conditions. Most standards laboratories use the value of 1.2930 kg/m³ for the dry air density value at *standard conditions* of 0°C and 101.325 kPa. Considering air as an ideal gas, the density $\rho_{air}(T,P)$ at an arbitrary temperature T (in degree C) and pressure P (in kPa) is then given by:

$$\rho_{air}(T,P) = \rho_{air}(0°C, 101.325 \text{ kPa}) \frac{273.2}{(273.2+T)} \frac{P}{101.325} . \tag{9.5}$$

When calibrating an ionisation chamber, the charge measured by the chamber depends on air temperature, pressure and humidity, and therefore the calibration coefficient must be given for stated reference values of these parameters. At most standards laboratories the chamber signal is corrected to *normal conditions* of 20°C (22°C in North America) and 101.325 kPa, but no correction is applied for humidity. Instead, the relative humidity during calibration is controlled within the range from 45% to 55%, so that the calibration factor applies for relative humidities around 50%.

In the users' beam, the correction factor for air temperature and air pressure $k_{T,P}$ is given as :

$$k_{T,P} = \frac{(273.2+T)}{(273.2+T_o)} \frac{P_o}{P} \tag{9.6}$$

and is applied to convert the measured signal to the reference conditions used for the chamber calibration at the standards laboratory. Note that P and T (in °C) are chamber air pressure and temperature, respectively, at the time of measurement, while P_o and T_o (in °C) are the normal conditions used in the standards laboratory.

The temperature of the air in a chamber cavity should be taken as that of the phantom and this is not necessarily the same as the temperature of the surrounding air. For measurements in a water phantom the chamber waterproof sleeve should be vented to the atmosphere in order to obtain a rapid equilibrium between the ambient air and the air in the chamber cavity.

(W_{air}/e) and stopping powers that are used in dosimetry protocols are stated for dry air but are affected by chamber air humidity. This results in an overall humidity correction factor of 0.997 for a cobalt-60 beam, correcting measurements at the 50% humidity level to those that would be obtained under dry air conditions and consisting of a 0.994 correction to the (W_{air}/e) dry air value of 33.97 J/C and a 1.003 correction to stopping powers.

9.3.2.   Chamber polarity effects: polarity correction factor k_{pol}

Under identical irradiation conditions, the use of polarizing potentials of opposite polarity in an ionisation chamber may yield different readings, a phenomenon that is referred to as the polarity effect. For most ionisation chamber types the effect is practically negligible in megavoltage photon beams, but in electron beams, notably at low energies, as well as in very low energy x-ray beams the effect may be significant.

In electron beams the polarity effect is considered a charge balance effect that depends on the energy and angular distribution of the incident radiation, the measurement depth in phantom, and field size. The polarity effect may actually change its sign with depth in phantom.

- When a chamber is used in a beam that produces a measurable polarity effect, the true reading is taken to be the mean of the absolute values of readings taken at the two polarities.

- The polarity correction factor k_{pol} is thus given by the following relationship:

$$k_{pol} = \frac{|M_+| + |M_-|}{2M},$$
(9.7)

where

M_+ and M_- are the chamber signals obtained under identical irradiation conditions at positive and negative chamber polarities, respectively, and M is the signal obtained at the polarity used routinely (either positive or negative).

- If the polarity effect for a particular chamber is larger than 3%, the chamber should not be used for absolute dose measurement.

- Whenever the polarity has been changed, charge equilibrium and stable operating conditions should be re-established by pre-irradiating the chamber and waiting several minutes before the next measurement.

- Detailed procedures on how to perform the polarity effect corrections can be found in the IAEA TRS-398 dosimetry protocol.

9.3.3.   Chamber voltage effects: recombination correction factor k_{sat}

The response of a given ionisation chamber depends not only on the radiation dose, dose rate and chamber polarity but also on the voltage applied between the measuring and collecting electrodes of the chamber. The charges produced in the chamber by radiation may differ from the charges that are actually collected and these discrepancies (charge losses or excess charges) occur as a result of constraints imposed by the physics of ion transport in the chamber sensitive volume and the chamber electrical design.

Charge losses in the chamber are caused by ion recombination; excess charges by charge multiplication and electrical breakdown. Both the charge recombination and charge multiplication are influenced by the potential applied to the ionisation chamber.

- A plot of chamber response, *i.e.*, current *I* or charge *Q* against applied voltage *V* for a constant dose rate or dose, respectively, is called a *saturation curve*, first rising linearly with voltage at low voltages, then reaching a saturation at high voltages and eventually breaking down at even higher voltages. A sketch of a typical saturation curve is shown in Fig. 9.3.

- The ratio $Q(V)/Q_{sat}$ or $I(V)/I_{sat}$, where Q_{sat} and I_{sat} are the saturation values of *Q* and *I*, respectively, is referred to as the collection efficiency *f*. In radiation dosimetry, ionisation chambers are commonly used in the near-saturation region where $f > 0.98$ or even in the saturation region where $f \approx 1$.
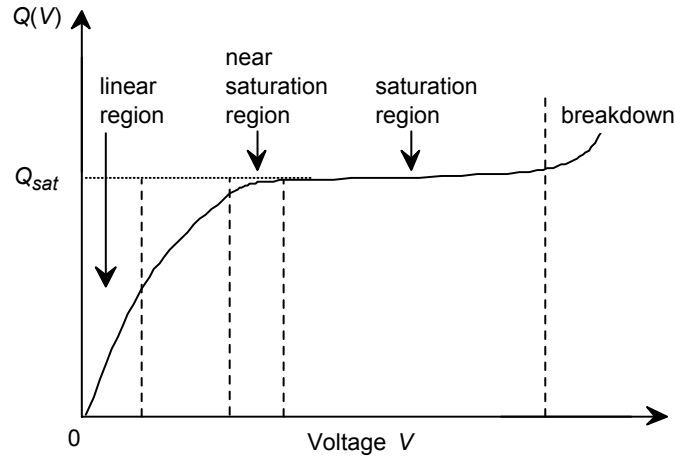
FIG. 9.3. Typical saturation curve for an ionisation chamber. The saturation charge is represented by Q_{sat} and is used in dosimetry protocols as the appropriate parameter describing the radiation signal. Ionisation chambers are usually operated in the near-saturation region and Q_{sat} is calculated by dividing the measured signal by the collection efficiency f.

- In saturation, all charges produced by radiation are collected and produce directly the Q_{sat} and I_{sat} for use in dosimetry protocols. When the chamber is used below saturation, some of the charges produced by radiation actually recombine and are lost to the dosimetric signal. This charge loss occurs through three different mechanisms:

  (1) *general recombination*: opposite charges from different tracks collide and recombine.
  (2) *initial recombination*: opposite charges from same tracks collide and recombine.
  (3) *ionic diffusion loss*: charges diffuse against the electric field.

For studies of ionic recombination losses, ionizing radiations are placed into three categories:

  (1)   *continuous radiation* (*e.g.*, cobalt beams and orthovoltage x rays);
  (2)   *pulsed beams* (*e.g.*, non-scanned linac x ray beams and electrons); and
  (3)   *scanned pulsed beams* (*e.g.*, scanned linac beams).

The ionic recombination correction factor k_{sat} (labeled P_{ion} in the AAPM TG-21 and TG-51 notation and equal to $1/f$ in recombination theory) accounts for the loss of ions in the chamber sensitive volume due to initial recombination, general recombination, and diffusion against the electric field. General recombination is by far the predominant of the three effects.

According to Boag, in the near saturation region f_g^c, the collection efficiency for the general recombination in a continuous radiation beam, may be written as:

$$f_g^c = \frac{Q}{Q_{sat}} = \frac{1}{1 + \dfrac{\Lambda_g}{V^2}} \tag{9.8}$$

or

$$\frac{1}{Q} = \frac{1}{Q_{sat}} + \frac{\Lambda_g / Q_{sat}}{V^2} = \frac{1}{Q_{sat}} + \frac{\lambda_g}{V^2} \tag{9.9}$$

and in a pulsed beam:

$$f_g^p = \frac{Q}{Q_{sat}} = \frac{V}{C} \ln\left(1 + \frac{C}{V}\right) \tag{9.10}$$

or

$$\frac{1}{Q} = \frac{1}{Q_{sat}} + \frac{C / Q_{sat}}{2V} = \frac{1}{Q_{sat}} + \frac{C'}{V} \quad , \tag{9.11}$$

with Λ_g, C and C' constants, Q the measured signal and Q_{sat} the saturation value of the signal.

The relationship for $1/Q$ suggests a linear behavior when plotted against $1/V^2$ for continuous beams (Eq. (9.9)) and against $1/V$ for pulsed beams (Eq. (9.11)) with $1/Q_{sat}$ the intercept of the linear plot with the ordinate, *i.e.,* for $1/V \rightarrow 0$ or $V \rightarrow \infty$.

- Assuming the predominance of general recombination and based on the linear relationship of $1/Q$ with either $1/V^2$ in continuous radiation or $1/V$ in pulsed radiation, one can determine the collection efficiencies f_g^c and f_g^p for continuous and pulsed beams, respectively, with the so-called *two-voltage technique*. Chamber signals M are determined under the same irradiation conditions at two voltages, the normal operating voltage V_N and a lower voltage V_L. The collection efficiencies at the normal chamber operating voltage V_N are then expressed as:

$$f_g^c(V_N) = \frac{M_N}{M_{sat}} = \frac{\dfrac{M_N}{M_L} - \left(\dfrac{V_N}{V_L}\right)^2}{1 - \left(\dfrac{V_N}{V_L}\right)^2} \tag{9.12}$$

for continuous beams and

$$f_g^p(V_N) = \frac{M_N}{M_{sat}} = \frac{\dfrac{M_N}{M_L} - \dfrac{V_N}{V_L}}{1 - \dfrac{V_N}{V_L}} \quad , \tag{9.13}$$

for pulsed beams, where

M_N	is the chamber signal determined at the normal operating voltage V_N,
M_L	is the chamber signal determined at a lower voltage V_L, and
M_{sat}	is the saturation signal at $V = \infty$.

- The polarity effect will change with the voltage and both M_N and M_L should be corrected for this effect using Eq. (9.7).

- For pulsed and pulsed-scanned megavoltage radiation beams, dosimetry protocols recommend that the recombination correction factor $k_{sat}(V_N)$ be determined :

  (1) assuming a linear relationship between $1/M$ and $1/V$ and
  (2) using the two-voltage technique and the following quadratic polynomial:

$$k_{sat}(V_N) = a_o + a_1 \frac{M_N}{M_L} + a_2 \left(\frac{M_N}{M_L} \right)^2 , \qquad (9.14)$$

  where a_i are constants tabulated for pulsed and pulsed-scanned beams (see, for example, the IAEA TRS-398 protocol; p. 52).

- For $k_{sat}(V_N) \leq 1.03$ (*i.e.*, $f \geq 0.97$) the recombination correction factor may be approximated to within 0.1% using the following relationship obtained from the general recombination theory:

$$k_{sat}(V_N) = 1 + \frac{\dfrac{M_N}{M_L} - 1}{\dfrac{V_N}{V_L} - 1} , \qquad (9.15)$$

  where, as defined above,

  M_N and M_L represent the chamber signals obtained with the normal applied potential V_N and low applied potential V_L, respectively.

- The ratio V_N / V_L should be equal to or larger than 3, and V_N must not be too large in order to ensure that charge multiplication effects do not contribute to the measured chamber signal.

- It is important to reestablish charge equilibrium after the bias voltage has been changed. This can be achieved by pre-irradiating the chamber with a dose of 2 Gy to 5 Gy before the next measurement.

9.3.4. Chamber leakage currents

Leakage currents present a difficult challenge in design of ionisation chamber-based dosimetric systems. Their effects on the true radiation-induced currents are minimized with guard electrodes, low noise triaxial cables, and sophisticated electrometers. The leakage currents fall into three categories:

(1) Intrinsic (dark) leakage currents
(2) Radiation-induced leakage currents
(3) Mechanical stress-induced and friction-induced spurious cable currents

- No matter how well an ionisation chamber dosimetric system is designed there will always be a small, non-radiation-related, signal present when the system is in a ready mode to respond to radiation. This intrinsic (dark) current results from surface and volume leakage currents flowing between the polarizing and measuring electrodes of the ionisation chamber.

- In a well designed ionisation chamber system the intrinsic leakage currents are at least two orders of magnitude lower than the measured radiation induced signals and are thus either negligible or can be suppressed from the actual radiation signal.

- Electric leakage in the ionisation chamber and electrometer may also occur as a consequence of the irradiation of insulators and chamber parts, cables, and electronics of the measuring equipment. This is termed *post-irradiation leakage*, an effect which continues after the irradiation has ceased and commonly decreases exponentially with time.

- The IEC 60731 document recommends that within 5 s after the end of a 10 min irradiation the leakage current shall have decreased to ±1.0 % or less of the ionisation current produced in the measuring volume during the irradiation, *i.e.*, it will fall to the intrinsic leakage current level of the dosimetric system.

- Another effect in insulators, which received considerable attention in the mid-1980s was the charge accumulation in non-conductive plastic phantoms, which causes a very large electric field around the chamber directing the flow of electrons towards the chamber cavity, yielding an increased signal and an erroneous result for the collection efficiency.

- Mechanical stress on cable insulators can also cause a leakage current and for this reason bending and twisting the cables should be avoided.

9.3.5. Chamber stem effects

Irradiating the chamber stem often cannot be avoided but it results in a different type of leakage current that is generally referred to as the *stem effect*. Two mechanisms have been described by the IEC, namely *stem scatter* and *stem leakage*.

- Stem scatter arises from the effect of scattered radiation in the stem, which reaches the chamber volume. This effect can be determined using a dummy stem, and the chamber is irradiated successively with and without the presence of the dummy stem; the ratio of the readings allows a correction factor for the effect to be determined.

- Stem leakage arises as a consequence of a direct irradiation of this chamber volume as well as the insulators and cables in the chamber. The effect can be determined by irradiating a chamber twice with a narrow rectangular field, once parallel and perpendicularly to the chamber central axis, and a correction factor is derived as above.

9.4. DETERMINATION OF ABSORBED DOSE USING CALIBRATED IONISATION CHAMBERS

For practical reasons, outputs of clinical photon and electron beams are usually measured with ionisation chambers that have calibration coefficients traceable to a standards laboratory and are thus used as relative dosimeters. Before such a chamber is used in radiotherapy machine output calibration, the user must identify a dosimetry protocol (code of practice) appropriate for the given radiation beam. A dosimetry protocol provides the formalism and the data to relate a calibration of a chamber at a standards laboratory to the measurement of absorbed dose to water under reference conditions in the clinical beam. Two types of dosimetry protocols are available:

(1)   Protocols based on *air-kerma in air* calibration coefficients and
(2)   Protocols based on absorbed *dose-to-water* calibration coefficients.

Most current megavoltage dosimetry protocols rely on chamber calibration coefficients determined in cobalt-60 beams at standards laboratories. It is expected that the use of megavoltage beam calibration qualities (x rays and electrons), today available only in a few PSDLs, will become more wide spread in the future.

- Conceptually, both types of protocols are similar and are based on several steps in the process of determining absorbed dose or dose rate from a charge or current measurement, respectively, with an ionisation chamber.

- The first step in the use of dosimetry protocols involves the determination of the chamber signal M_Q through correcting the measured chamber charge or current for influence quantities known to affect the measured chamber signal, as discussed in Section 9.3. The subscript Q denotes the quality index of the beam being calibrated, as discussed in Section 9.8.

- It should be noted that the formalisms presented here, based on a cobalt-60 calibration coefficient, work well for megavoltage photon and electron beams. The calibration of superficial and orthovoltage x-ray beams, on the other hand, relies on different principles and the chamber calibration coefficient should be obtained for the particular x-ray beam quality that is being calibrated. The physics of kilovoltage dosimetry is discussed in more detail in Section 9.10.

9.4.1. *Air-kerma*-based protocols

The air-kerma-based protocols use the *air-kerma in air* calibration coefficient $N_{K,Co}$ obtained for a local reference ionisation chamber in a cobalt-60 beam at a standards laboratory. Routine ionisation chambers are then cross-calibrated with the reference ionisation chamber in a local cobalt-60 beam. Two steps are involved in an *air-kerma*-based protocol for calibration of megavoltage photon and electron beams:

(1)   The cavity air calibration coefficient $N_{D,air}$ is calculated from the $N_{K,Co}$ calibration coefficient.
(2)   Absorbed *dose-to-water* is determined using the Bragg-Gray relationship in conjunction with the chamber signal M_Q and the cavity air calibration coefficient $N_{D,air}$.

- In a cobalt-60 beam at a standards laboratory the mean absorbed *dose-to-air* in the cavity is determined from total *air-kerma in air* $(K_{air})_{air}$ using the relationship:

$$D_{air} = (K_{air})_{air} (1-g)k_m k_{att} k_{cel} \quad , \tag{9.16}$$

where

g is the fraction of the total transferred energy expended in radiative interactions upon the slowing down of secondary electrons in air,

k_m is a correction factor for the non-air equivalence of chamber wall and build-up cap needed for an *air-kerma in air* measurement,

k_{att} is a correction factor for photon attenuation and scatter in the chamber wall (equivalent to the A_{wall} correction in the AAPM TG-21 nomenclature), and

k_{cel} is a correction factor for the non-air equivalence of the central electrode of the cylindrical ionisation chamber.

- The cavity air calibration coefficient, $N_{D,air}$, is defined as:

$$N_{D,air} = D_{air}/M_Q , \tag{9.17}$$

where M_Q is the chamber signal corrected for influence quantities.

- The *air-kerma in air* calibration coefficient $N_{K,Co}$ is defined as:

$$N_{K,Co} = (K_{air})_{air}/M_Q . \tag{9.18}$$

If the electrometer device has its read-out in nC, both the cavity calibration coefficient and the *air-kerma in air* calibration coefficient have the units cGy/nC.

- By dividing the left and right hand sides of Eq. (9.16) by the corrected chamber signal in the calibration beam M_Q, the cavity air calibration coefficient can be determined from the *air-kerma in air* calibration coefficient, determined at the cobalt-60 beam quality, using the relation

$$N_{D,air} = N_{K,Co}(1-g)k_m k_{att} k_{cel} . \tag{9.19}$$

- The cavity air calibration coefficient is also directly related to the effective volume V_{eff} of the chamber by:

$$N_{D,air} = \frac{D_{air}}{M_Q} = \frac{1}{m_{air}} \frac{W_{air}}{e} = \frac{1}{\rho_{air} V_{eff}} \frac{W_{air}}{e} \tag{9.20}$$

where

(W_{air}/e) is the average energy required to produce an ion pair in air,

m_{air} is the mass of air in the chamber cavity,

ρ_{air} is the ▒ density at standard conditions of temperature and pressure,

V_{eff} is the effective air volume in the chamber collecting ions.

Equation (9.20) shows clearly that $N_{\mathrm{D,air}}$ is a characteristic of the dosimetric device and depends only on the effective mass of air in the chamber cavity and does not depend on radiation quality as long as (W_{air}/e) is independent of the radiation quality. Hence, the $N_{\mathrm{D,air}}$ calibration coefficient determined at the cobalt-60 beam quality at the standards laboratory is also valid at the user's megavoltage beam quality Q.

If the effective chamber cavity volume V_{eff} were accurately known, the $N_{\mathrm{D,air}}$ calibration coefficient could in principle be determined using Eq. (9.20). This is the case for cavity ionisation chambers used to establish the *air-kerma in air* for cobalt units at standards laboratories (see Section 9.1.4.). For typical ionisation chambers used in clinic, however, V_{eff} is not known with sufficient accuracy and the $N_{\mathrm{D,air}}$ must then be determined from the *air-kerma in air* calibration coefficient $N_{\mathrm{K,Co}}$ using Eq. (9.19)

The absorbed *dose-to-air* $D_{\mathrm{air},Q}$ in the air cavity can be converted into absorbed *dose-to-medium* (*e.g.*, water) $D_{\mathrm{w},Q}$ by making use of the Bragg-Gray cavity relation. With a known value of $N_{\mathrm{D,air}}$ for a specific chamber, the fully corrected chamber signal M_Q at a point in a phantom allows determining the absorbed dose-to-water as follows:

$$D_{\mathrm{w},Q} = D_{\mathrm{air},Q}(s_{\mathrm{w,air}})_Q \, p_Q = M_Q N_{\mathrm{D,air}}(s_{\mathrm{w,air}})_Q \, p_Q \qquad (9.21)$$

where

$(s_{\mathrm{w,air}})_Q$ is the ratio of restricted collision stopping powers of medium to air and

p_Q is a perturbation correction factor accounting for perturbations caused by the chamber inserted into the medium, as discussed in detail in Section 9.7.

9.4.2. Absorbed *dose-to-water*-based protocols

All dosimetry protocols aim at the determination of the quantity absorbed *dose-to-water*. It is therefore logical to provide ionisation chambers directly with a calibration coefficient in terms of this quantity, rather than in terms of the *air-kerma in air*, if at all possible. Recent developments have provided support for a change in the quantity used at present to calibrate ionisation chambers and provide calibration coefficients in terms of absorbed *dose-to-water*, $N_{D,w}$, for use in radiotherapy beams. Many PSDLs now provide $N_{D,w}$ calibrations at cobalt-60 gamma ray beams and some laboratories have already extended these calibration procedures to high-energy photon and electron beams.

- The absorbed *dose-to-water* D_{w,Q_o} at the reference depth z_{ref} in water for a reference beam of quality Q_o and in the absence of the chamber is directly given by:

$$D_{\mathrm{w},Q_o} = M_{Q_o} N_{\mathrm{D,w},Q_o}, \qquad (9.22)$$

where M_{Q_o} is the fully corrected chamber reading under the reference conditions used in the standards laboratory and $N_{\mathrm{D,w},Q_o}$ is the calibration coefficient in terms of the absorbed *dose-to-water* of the chamber obtained from the standards laboratory.

- When a chamber is used in a beam of quality Q that differs from the quality Q_o that was used in its calibration, the absorbed *dose-to-water* is given by:

$$D_{w,Q} = M_Q \, N_{D,w,Q_o} k_{Q,Q_o},$$ (9.23)

where the factor k_{Q,Q_o} corrects for the differences between the reference beam quality Q_o and the actual user quality Q.

- The beam quality correction factor k_{Q,Q_o} is defined as the ratio, at beam qualities Q and Q_o, of the calibration coefficients in terms of absorbed *dose-to-water* of the ionisation chamber:

$$k_{Q,Q_o} = \frac{N_{D,w,Q}}{N_{D,w,Q_o}}.$$ (9.24)

- Currently, the common reference quality Q_o used for the calibration of ionisation chambers is the cobalt-60 gamma radiation, and the symbol $k_{Q,Co}$, abbreviated to k_Q, is often used for the beam quality correction factor.

- At some PSDLs high-energy photon and electron beams are directly used for calibration purposes and the symbol k_{Q,Q_o} is used in these cases with Q_o specifying the calibration beam. Ideally, the beam quality correction factor should be measured directly for each chamber at the same quality as the user's beam. However, this is not achievable in most standards laboratories. Such measurements can be performed only in laboratories having access to the appropriate beam qualities; for this reason the technique is at present restricted to a few PSDLs around the world, as the procedure requires the availability of an energy-independent dosimetry system, such as a calorimeter, operating at these beam qualities.

- When no experimental data are available, or when it is difficult to measure k_{Q,Q_o} directly for realistic clinical beams, the correction factors can, in many cases, be calculated theoretically. By comparing Eq. (9.24) with the $N_{D,air}$ formalism given above, k_{Q,Q_o} can be written as:

$$k_{Q,Q_o} = \frac{(s_{w,air})_Q}{(s_{w,air})_{Q_o}} \frac{p_Q}{p_{Q_o}},$$ (9.25)

including ratios, at beam qualities Q and Q_o, of: (i) Spencer-Attix water/air stopping-power ratios, $s_{w,air}$, and (ii) the perturbation factors p_Q and p_{Q_o} for departures from the ideal Bragg-Gray detector conditions.

- The calculations of k_{Q,Q_o} are based on exactly the same data that are used in the calculations in the *air-kerma*-based approach, but the parameters are used as ratios, which have reduced uncertainties, in comparison with the individual values.

- Most protocols provide a modified formalism for electron beams for use when a chamber is cross-calibrated (does not have a direct $N_{D,w,Co}$ calibration coefficient). The details can be found in the IAEA TRS-398 and AAPM TG-51 protocols.

- A still frequently used quantity is the exposure calibration coefficient N_X which is related to the *air-kerma in air* calibration coefficient N_K through the following relationship:

$$N_K = N_X \frac{W_{air}}{e} \frac{1}{1-g}, \tag{9.26}$$

  where g is the fraction of the energy loss in air expended in radiative interactions (the bremsstrahlung fraction). For cobalt-60 gamma rays in air $g = 0.003$, for super-ficial x rays in air $g < 0.0002$).

- Typical units of N_X and N_K are (R/nC) and (Gy/nC), respectively. Typical unit for both $N_{D,air}$ and $N_{D,w}$ is (Gy/nC).

- A schematic summary of the steps involved in *air-kerma in air*-based and absorbed *dose-to-water*-based calibration routes is given in Fig. 9.4. The physics and characteristics of stopping power ratios and perturbation correction factors are discussed in more detail in Sections 9.5.and 9.7.
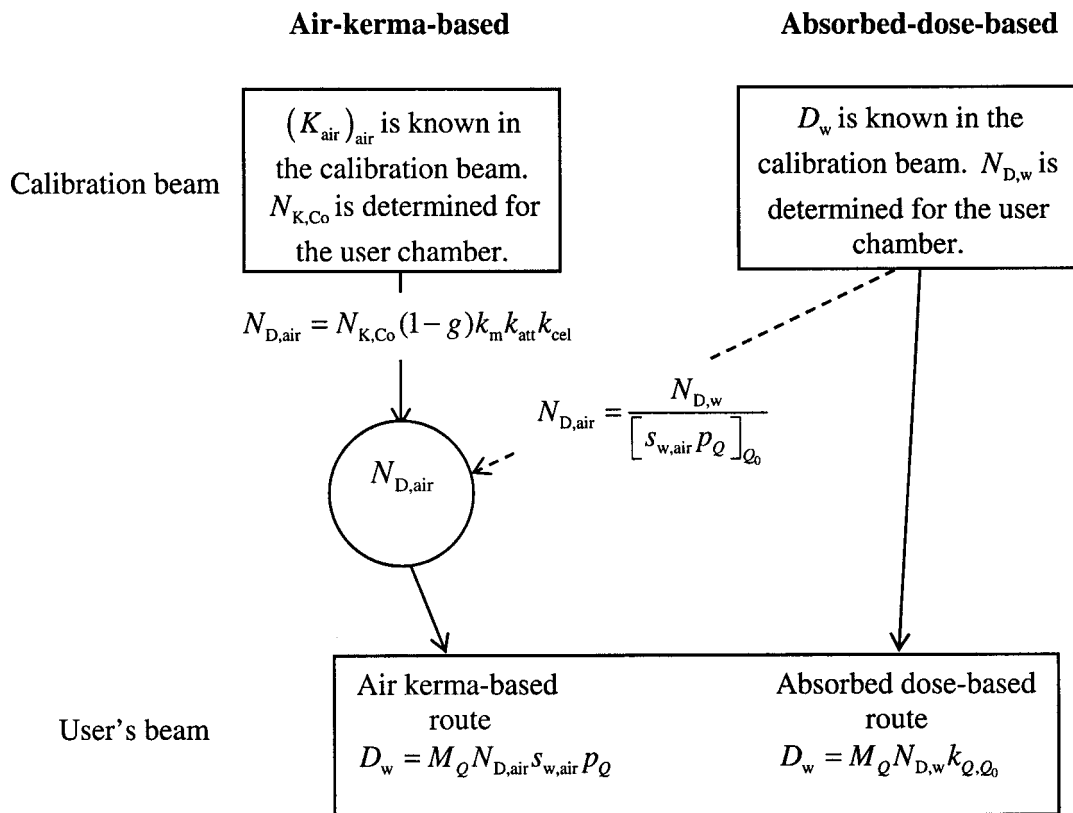

FIG. 9.4. Schematic diagram of steps involved in ionisation chamber-based reference dosimetry: air-kerma in air-based on the left, absorbed dose-to-water-based on the right.

The *air-kerma in air*-based formalism as well as the absorbed *dose-to-water*-based formalism for the determination of absorbed *dose-to-water* in reference conditions includes *stopping-power ratios* and *correction factors for perturbation effects*, the latter being detector-dependent. Some of the analytical models available for the calculation of perturbation correction factors also include *mass energy-absorption coefficient ratios*.

Although ideally the formalism in terms of absorbed *dose-to-water* is based on experimentally determined quantities, the approach most common today relies on theoretically determined beam quality factors k_{Q,Q_o}, which are also based on stopping-power ratios (see Section 9.5) and perturbation correction factors (see Section 9.7).

9.5.   STOPPING-POWER RATIOS

As already mentioned, the determination of absorbed dose in a medium using an ionisation chamber is based on the Bragg-Gray principle relating the absorbed dose at a point in the medium (water) D_w to the mean absorbed dose in the detector (air) $\overline{D}_{air}$ through a proportionality factor that classically has been identified as the ratio of the mass (collision) stopping powers, water to air:

$$D_w = \overline{D}_{air} s_{w,air} \quad . \tag{9.27}$$

The key Bragg-Gray assumption is that the electron fluence present in the detector is identical to that in the (undisturbed) medium at the point-of-interest in the water phantom. The gas-filled ionisation chamber in a high-energy photon or electron beam behaves to a good approximation as a Bragg-Gray detector. Any deviations from perfect Bragg-Gray behaviour are accounted for by perturbation factors that are discussed in detail in Section 9.7. below. The stopping-power ratio applies to the electron spectrum at the point-of-interest in the undisturbed medium and is independent of the detector (except for the minor influence of the Spencer-Attix cutoff).

9.5.1.   Stopping-power ratios for electron beams

The most important characteristic of the water/air stopping-power ratios for monoenergetic electrons is their strong dependence on energy and depth, as shown in Fig. 9.4, resulting mainly from the considerable variation in energy spectra at the various depths in water.

Until lately, the selection of stopping-power ratios for the user's beam in electron dosimetry protocols has relied on the use of mono-energetic data using Harder's procedure based on the characterization of the electron beam through the mean electron energy at the phantom surface $\overline{E}_o$ together with depth of measurement z. Clinical beams are, however, far from mono-energetic and mono-directional at the phantom surface and even less so at depths in phantom.

The validity of the $s_{w,air}(\overline{E}_o, z)$ selection procedure has been reviewed in detail in the IAEA protocol for parallel-plate ionisation chambers (IAEA TRS-381) and a conclusion was reached that, even for beams with large energy and angular spread, the maximum error produced by such a procedure is always within 1%. For most beams used in clinical practice, even for those with a certain degree of photon contamination, the agreement was within the estimated uncertainty of the calculated stopping-power ratios, being of the order of 0.6%.
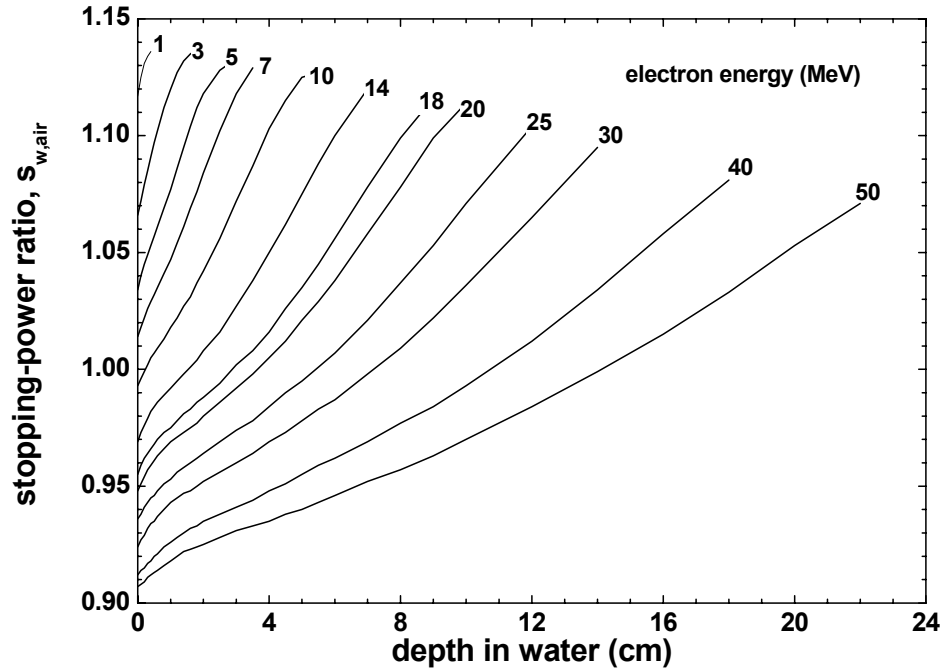
FIG. 9.5. Depth variation of the Spencer-Attix water/air stopping-power ratio $s_{w,air}$ for $\Delta=10\,keV$, derived from Monte Carlo generated electron spectra for monoenergetic, plane-parallel, broad electron beams.

At present stopping-power ratios for realistic electron beams, obtained by simulating in detail the treatment head of some clinical accelerators, have become available and are used in the most recent dosimetry protocols based on standards of absorbed *dose-to-water*. However, it has been verified that no dramatic changes occur in electron beam dosimetry solely due to this improvement in the calculation of stopping-power ratios.

9.5.2. Stopping-power ratios for photon beams

The most important characteristic of the depth variation of the stopping-power ratios of monoenergetic photons is that the ratios are almost constant beyond the depth of transient electronic equilibrium, as Fig. 9.6 clearly shows. The range of variation of the stopping-power ratio data with energy is also much smaller than in the case of electrons with similar energies. In the case of photon bremsstrahlung spectra produced by clinical accelerators the constancy of the stopping-power ratio is reached at shallower depths due to the presence of low-energy photons in the spectrum.

9.6. MASS ENERGY-ABSORPTION COEFFICIENT RATIOS

The role of spectrum averaged mass energy-absorption coefficient ratios in modern dosimetry protocols is mainly restricted to their use in calculating perturbation and other correction factors for ionisation chambers in cobalt-60 and high-energy photon beams. In general, they are associated with the fraction of energy deposited within a detector due to electrons generated by photon interactions in the detector material itself.
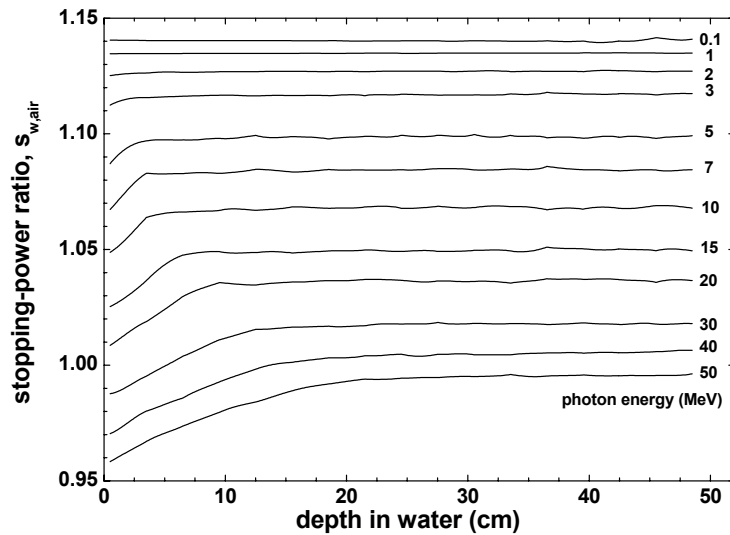
FIG. 9.6. Depth variation of the Spencer-Attix water/air stopping-power ratio $s_{w,air}$ *for* $\Delta=10\,keV$, *derived from Monte Carlo-generated electron spectra for monoenergetic, plane-parallel photon beams.*

Depending on the medium, the photon fluence spectra may change appreciably with depth or material thickness and also depend on the field size of the incident beam. It has been shown that the effects of spectral changes within a phantom on the mean mass energy absorption coefficients are of importance only for large field sizes or for low-energy photon beams, where there is more than a 0.5% variation in $\left(\mu_{en}/\rho\right)_{w,m}$, the ratio of mass-energy absorption coefficients, for tissue-like materials (m) with respect to water (w) due to this effect.

A consistent set of mass energy absorption coefficient ratios for photon dosimetry that is used in most dosimetry protocols was given in the IAEA TRS-277 protocol. These data have not yet been superseded by any other new set of data.

9.7.   PERTURBATION CORRECTION FACTORS

For a detector to behave as a Bragg-Gray cavity, the electron fluence in the sensitive medium of the detector must be identical to that at a specified point in the uniform medium. The only possible true Bragg-Gray detector would be a small air bubble; all protocols for absolute dose determination are, in fact, based on air-filled ionisation chambers.

For megavoltage photon radiation the Bragg-Gray conditions are adequately fulfilled for air cavities of the size encountered in practical ionisation chambers, *i.e.*, the ranges in air of the secondary electrons generated in megavoltage photon beams are much greater than the cavity dimensions. However, an ionisation chamber does not only consist of an air cavity. There will always be a wall that, in general, is not perfectly medium-equivalent. Often this wall is made of graphite, whereas the medium is water. Moreover, for cylindrical chambers there must be a central electrode, which is frequently made of aluminum. And there may be other materials around the chamber, such as a stem for cylindrical chambers and a back wall in the case of parallel-plate designs. All of these features can introduce deviations from perfect Bragg-Gray behaviour.

These deviations are generally dealt with by introducing one or more correction factors, often known as *perturbation* factors, into the expression for the absorbed dose, *i.e.*, the factor p_Q in Eq. (9.18). This overall factor is often written as a product of four perturbation factors, each one accounting for a different effect, assumed to be independent of the others, as follows:

$$p_Q = (p_{dis} p_{wall} p_{cel} p_{cav})_Q \ , \tag{9.28}$$

where

p_{dis}	is a factor that accounts for the effect of replacing a volume of water with the chamber cavity (cylindrical chambers);
p_{wall}	is a factor that corrects the response of the ionisation chamber for the non-water equivalence of the chamber wall and any waterproofing material;
p_{cel}	is a factor that corrects the response of the chamber for the effect of the central electrode during in-phantom measurements; and
p_{cav}	is a factor that corrects the response of the ionisation chamber for effects related to the air cavity, predominantly the in-scattering of electrons that makes the electron fluence inside a cavity different from that in water in the absence of the cavity.

The word *perturbation* is used in the sense of a *perturbation* by the detector of the electron fluence $\phi_{med}(P)$ present at the point-of-interest P in a uniform medium where the relevant fluence in the detector, inevitably a mean value over a finite volume, $\bar{\phi}_{det}$, is that which gives rise to the signal, *i.e.*, the fluence in the air in the case of an ionisation chamber.

The following subsections deal with the four different sources of the Bragg-Gray cavity perturbation. The emphasis here is on the *physics* of these correction factors. A complete account of numerical values for the particular chamber and radiation quality of interest can be taken from the particular protocol being followed and a concise summary of the protocol recommendations is given in Section 9.9.

9.7.1. Displacement perturbation factor p_{dis} and effective point of measurement

The ionisation chamber placed into a phantom will displace a certain volume of the phantom medium. Even if the chamber wall is medium-equivalent, one still must consider the effect of the volume occupied by the air cavity. In general the dimensions of this volume are not negligible compared to any changes in the radiation field and hence in the dose distribution.

For example, the dose may change by a few per cent in a distance equal to the diameter of the chamber. Clearly then the chamber reading will be affected by this "missing" medium. In simple terms one can expect that the reduced attenuation, in the case of photon beams, will result in a *higher* chamber reading, compared to that in a very small "air bubble" situated at the center of the detector.

However, there is another effect; the missing material means that there is less *scatter*. This will counterbalance the first effect. The net result is still generally an increase in the signal that results in a correction factor known as the *displacement perturbation factor*, usually denoted by p_{dis}, which will thus be less than unity.

- The value of p_{dis} will in general depend on both the radiation quality and on the physical dimensions of the air cavity in the direction of the beam, as well as on the depth of measurement. In photon beams p_{dis} will be practically constant beyond the depth of dose maximum, due to the exponential falloff in dose. However, in the buildup region it will vary in a complicated fashion with depth. For a Farmer chamber, which has an internal radius of 3 mm, the value is close to 0.988 in a cobalt-60 beam.

- The correction for displacement can be viewed in an alternative way. Instead of applying a factor to correct the chamber reading by assuming that the chamber is positioned so that its center is at the depth of interest, a *shift* in the position of the chamber can be made. For a cylindrical chamber the electrons enter the wall at various depths, generally forward of its centre, and hence the electron fluence in the air cavity is representative of that existing at some point in the uniform medium shifted forward of the chamber centre. In fact, it was found that the readings of different chambers could be brought into coincidence with one another by performing shifts depending on the chamber dimensions. Thus the concept of the *effective point of measurement P_{eff}* was developed.

- The newer absorbed *dose-to-water*-based dosimetry protocols favour the p_{dis} approach. However, *air-kerma in air*-based protocols use the P_{eff} concept in preference to p_{dis}. The IAEA TRS-277 protocol recommended a shift of $0.5r$ for cobalt-60 gamma rays, increasing to $0.75r$ for all higher energy photon beams. More recent reviews of the experimental evidence on the magnitude of the shift led the IAEA to recommend a single value of $0.6r$ for all high-energy photon beams (IAEA TRS-398).

- In electron beams the use of p_{dis} is impractical as the depth dose curve is very irregular in shape in contrast to the quasi-exponential decrease in photon beams at depths beyond the buildup region. Since p_{dis} would vary rapidly and in an irregular way with depth in an electron beam, the P_{eff} concept is universally employed in electron beams.

  - For cylindrical chambers the recommended shift is $0.5r$ (IAEA TRS-277 and TRS-398).

  - For parallel-plate chambers P_{eff} is assumed to be situated in the centre of the inside face of the front wall, as illustrated in Fig. 9.7; this is logical since, in a well-guarded chamber, it can be assumed that all the electrons entering the sensitive air volume do so through the front window.

9.7.2. The chamber wall perturbation factor p_{wall}

Compliance with Bragg-Gray conditions implies that the electron fluence in the sensitive volume of the detector is identical (strictly in magnitude, energy and angular distribution) to that present in the undisturbed medium at the position of interest. However, an ionisation chamber has a wall that in general is not made of medium-equivalent material. In the case of photon beams the electron fluence in the air cavity in an ionisation chamber, assumed cylindrical, with a wall of a typical thickness, will consist partly of electrons generated in the (uniform) medium surrounding the wall and which have traveled through the wall and partly of electrons generated by photon interactions with the wall material.
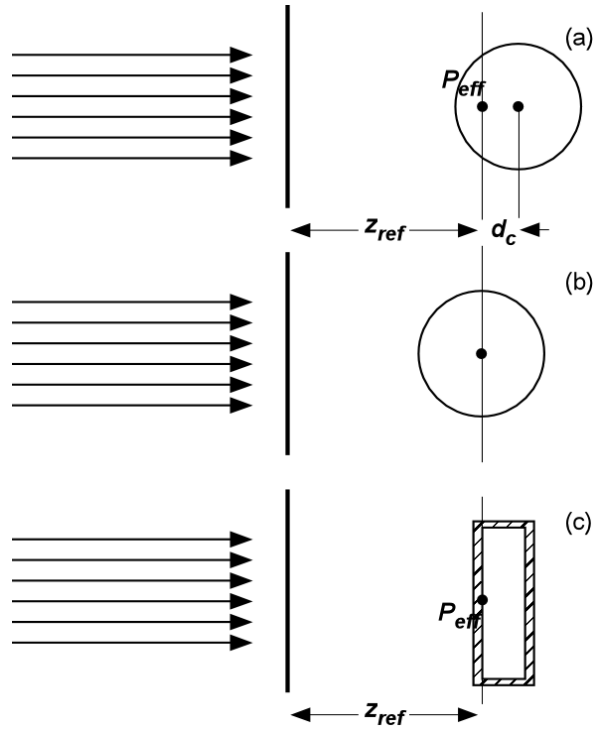
FIG. 9.7. (a) In most N_K-based dosimetry protocols the effective point of measurement of a cylindrical ionisation chamber is positioned at the reference depth z_{ref} where the absorbed dose is required; the chamber centre is deeper than z_{ref} a distance d_c equal to the shift of P_{eff}. (b) Except in electron and heavy-ion beams, in $N_{D,w}$-based protocols the centre of a cylindrical chamber is positioned at the reference depth z_{ref} and the absorbed dose is determined at this position. (c) For plane-parallel chambers all protocols position the effective point of measurement (front of the air cavity) at the reference depth z_{ref}.

Quite clearly the number and energy distribution of these wall-generated secondary electrons will be characteristic of photon interactions with the material of the wall and not of the medium as demanded by Bragg-Gray conditions.

- For ionisation chambers with walls of intermediate thickness, in practical use in radiotherapy, an approximate empirical two-component expression is in common use:

$$p_{wall} = \frac{\alpha s_{wall,air} (\mu_{en}/\rho)_{w,wall} + (1-\alpha) s_{w,air}}{s_{w,air}}, \qquad (9.29)$$

where α is the fraction of the dose to the air in the cavity due to electrons generated in the chamber wall; thus if this is zero, then p_{wall} reduces to unity as expected.

- An additional small correction has been implemented for the case when a water-proofing sleeve is used, where Eq (9.29) is extended to a three-component model, with a third term $\tau s_{sleeve,air} (\mu_{en}/\rho)_{w,sleeve}$ where τ is the fraction of the ionisation due to electrons generated in the sheath as follows:

$$p_{\text{wall}} = \frac{\alpha s_{\text{wall,air}} (\mu_{\text{en}} / \rho)_{\text{w,wall}} + \tau s_{\text{sleeve,air}} (\mu_{\text{en}} / \rho)_{\text{w,sleeve}} + (1 - \alpha - \tau) s_{\text{w,air}}}{s_{\text{w,air}}} \qquad (9.30)$$

where α and τ are the fractional contributions to ionisation resulting from photon interactions in wall and sleeve, respectively.

- The two parameters α and τ can be estimated for cobalt-60 beams from the thickness of the wall t_{wall} and the waterproofing sleeve t_{sleeve} (in g/cm^2), if present, using

$$\alpha = 1 - \exp(-11.88\, t_{\text{wall}}) \qquad (9.31)$$

and

$$\tau = \exp(-11.88\, t_{\text{wall}}) - \exp(-11.88\, (t_{\text{wall}} + t_{\text{sleeve}})). \qquad (9.32)$$

- For high energy beams, the fractional ionisations α and τ are derived from the data given by the IAEA TRS-398 protocol. In the case of electron beams, it is generally assumed that the effect of the chamber wall is negligible.

9.7.3. Central electrode perturbation p_{cel}

Cylindrical chambers have a central electrode which is usually made of aluminum but can be made of graphite. The central electrode will produce an increase in the chamber signal compared with what would be obtained in an air bubble, and a correction for the non-air equivalence of the electrode is in principle necessary; this is denoted by p_{cel}.

- The effect of a central electrode made of graphite has been shown to be practically negligible in photon beams but decreases with energy from 1.008 to 1.004 for a 1-mm diameter aluminum electrode.

- In electron beams the effect is negligible for graphite, and never greater than 0.2% at any energy (5-20 MeV) or depth for a 1-mm diameter aluminum electrode.

9.7.4. Cavity or fluence perturbation correction p_{cav}

An ionisation chamber introduces a low-density heterogeneity into a medium. In an electron beam, density changes can cause *hot* or *cold* spots as a result of electron scattering. The reason for this is clear from Fig. 9.8 due to Harder.

- As a result of (elastic nuclear) scattering, the angular distribution of electrons broadens with depth; a low density cavity will consequently scatter *out* fewer electrons than are scattered *in*, resulting in an increase in the electron fluence toward the *downstream* end of the cavity, in comparison with the fluence in a uniform medium at that depth.

- All modern *air-kerma in air*-based dosimetry protocols include values of perturbation factors determined experimentally.
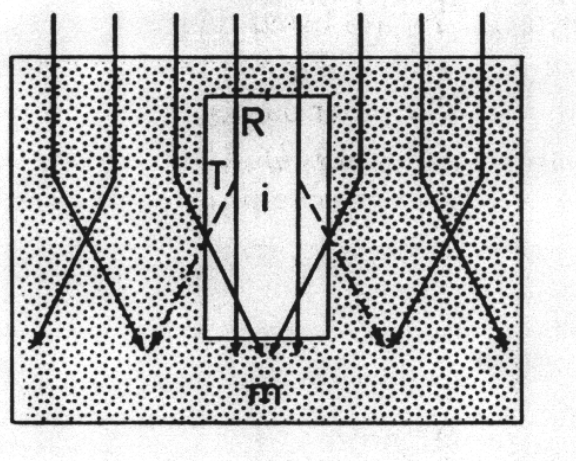
FIG. 9.8. Perturbation of the electron fluence caused by a gas-filled cavity in a solid or liquid phantom. Electron tracks are idealized to emphasize the effects being shown. The dominance of in-scattering over out-scattering gives rise to an increase in the fluence towards the back of the cavity and hence this produces an increase in the chamber signal.

- The magnitude of the *in-scattering* perturbation exceeds 3% for Farmer-type chambers for E_z below 8 MeV. This is one of the principal reasons why parallel-plate chambers are recommended in low energy electron beams. In a parallel-plate chamber the diameter of the air cavity (typically between 13 mm and 20 mm) is deliberately made very much greater than its thickness (the electrode spacing) that is 2 mm in almost all commercial designs. Thus, most of the electrons enter the air cavity through the front face of the chamber and only a small fraction through the side walls.

- Furthermore, well-designed parallel-plate chambers have a relatively wide guard ring, 3 mm or more, which ensures that almost no electrons entering through the short side walls can contribute to the chamber signal. Consequently the *in-scattering* is virtually eliminated. The electron fluence in the sensitive volume of such a chamber is therefore that existing in the uniform medium at the depth of the inside face of the front window, which is the position of the effective point of measurement P_{eff}. For cylindrical chambers this guarding capability is virtually absent and the electron fluence is significantly perturbed. For these chambers the cavity perturbation correction factor p_{cav} is given by:

$$p_{\text{cav}}(\bar{E}_{\text{o}}, r) = 1 - 0.02155 \, r \, \exp(-0.1224 \, \bar{E}_z), \tag{9.33}$$

with r the cavity inner radius in mm and $\bar{E}_z$ the mean electron energy at depth z as obtained from the Harder relationship (see Eq. 9.36).

In photon beams there is generally charged particle equilibrium (or a very good approximation to it), and therefore no change in either the energy or angular distribution of the secondary electrons with position in the irradiated medium. The electron fluence perturbation effect is therefore negligible in photon beams. However, in the buildup region in photon beams, where there is no charged particle equilibrium, significant perturbation effects have been demonstrated.

9.8.   BEAM QUALITY SPECIFICATION

The signal (current or charge) that is produced by an ionisation chamber and measured by an electrometer must be multiplied by factors correcting for influence quantities (see Section 9.3.) and various dosimetric physical quantities described in the previous sections to yield the absorbed *dose-to-water* at a reference point in water, the quantity in terms of which radiotherapy machine output is specified. Some of these quantities depend upon photon or electron beam energy, thus the beam quality needs to be specified for dosimetric calculations.

The most logical means to characterize the quality of a clinical radiation beam is to state its spectral distribution. However, since beam spectra are difficult to measure directly and cumbersome to determine in an absolute sense with Monte Carlo techniques, other more practical approaches to beam quality specification have been developed. These approaches are specific to three distinct ionizing radiation beam categories:

(1)   *Kilovoltage* (superficial and orthovoltage) *x-ray beams.*
(2)   *Megavoltage x-ray beams.*
(3)   *Megavoltage electron beams.*

9.8.1.   Beam quality specification for kilovoltage photon beams

For low energy photon beams, the quality of the beam is most conveniently expressed in terms of the half-value layer (*HVL*) of the beam, with *HVL* representing the thickness of an attenuator that decreases the measured *air-kerma rate in air* to half of its original value.

- To minimize the effects of radiation scattered in the attenuator the *HVL* must be measured under "good geometry" conditions that imply the use of:

  - Narrow beam geometry to minimize scattering from the attenuator,
  - Reasonable distance between the attenuator and the measuring device (ionisation chamber) to minimize the number of scattered photons reaching the detector,
  - Ionisation chamber with air-equivalent walls and with a flat photon energy response for the spectrum of radiations comprising the beam.

- For superficial x-ray beams (10 kVp to 100 kVp) *HVL*s are usually given in millimeters of pure aluminum (typical *HVL*s from 0.01 mm to 10 mm of aluminum), while for orthovoltage x-ray beams (above 100 kVp) they are usually given in millimeters of pure copper (typical *HVL*s from 0.5 mm to 4 mm of copper).

- The specification of beam quality in terms of *HVL* is really a very crude beam specification, since it tells little about the energy distribution of the photons present in the beam. However, the beam specification through the *HVL* provides a general idea of the effective energy of the photon beam that may be used: (i) to assess the beam penetration into tissue and (ii) to determine the appropriate values of quantities used in dosimetry protocols.

- Since two beams with widely differing potentials can have similar *HVL*s due to the marked effect of different filtrations, it is customary to state, in addition to *HVL*, the x-ray potential and total filtration used in generating a given x-ray beam.

- Often the low energy x-ray beams are also characterized by stating their homogeneity coefficient κ that is defined as the ratio between the first and second *HVL*, *i.e.*, HVL_1 / HVL_2. For heterogeneous low energy x-ray beams $HVL_2 > HVL_1$, resulting in $\kappa < 1$; for monochromatic beams, on the other hand, $HVL_2 = HVL_1$ and $\kappa = 1$.

- Another quantity that is often used in beam quality specification is the equivalent or effective photon energy, defined as the quantum energy of a monoenergetic beam having an *HVL* equal to the HVL_1 of the beam being specified.

9.8.2.   Beam quality specification for megavoltage photon beams

In the megavoltage photon energy range, *HVL*s vary little with photon energy making *HVL*s unsuitable for beam quality specification. Therefore, other indices were developed relating to: (i) the energy of the electron beam as it strikes the target (nominal accelerating potential, *NAP*) and (ii) radiation beam attenuation as the beam penetrates into water or tissue. Older radiation protocols were based on the nominal accelerating potential, while the recent ones are based on quantities that are related to beam penetration into water, such as the tissue-phantom ratio (*TPR*) or the percentage depth dose (*PDD*).

A considerable improvement was made to *air-kerma in air*-based dosimetry protocols when stopping power ratios and mass energy absorption coefficient ratios were correlated with clinically measured ionisation ratios, such as the $TPR_{20,10}$, rather than with nominal accelerating potentials.

- The parameter $TPR_{20,10}$ is defined as the ratio of doses on the beam central axis at depths of 20 cm and 10 cm in water obtained with a constant source-detector distance of 100 cm and a field size of 10×10 cm^2 at the position of the detector. $TPR_{20,10}$ is a measure of the effective attenuation coefficient describing the approximately exponential decrease of a photon depth dose curve beyond the depth of maximum dose z_{max}, and, more importantly, it is independent of electron contamination of the incident photon beam.

- The $TPR_{20,10}$ can be related to measured $PDD_{20,10}$ using the following relationship:

$$TPR_{20,10} = 1.2661 \, PDD_{20,10} - 0.0595 \,, \tag{9.34}$$

   where $PDD_{20,10}$ is the ratio of percentage depth doses at depths of 20 cm and 10 cm for a field of 10×10 cm^2 defined at the water phantom surface with an *SSD* of 100 cm. This empirical relation was obtained from a sample of almost 700 linacs.

Other beam quality indices have been proposed for megavoltage photon dosimetry and they are, in most cases, related to the depth of maximum dose z_{max}, making them susceptible to the electron contamination of the beam at this depth in water phantom.

- Based on *PDD* distributions a widely disseminated recommendation for specifying the quality of high-energy photon beams was made in the BJR Supplement 17 report that defined a parameter d_{80} as the depth of the 80% depth dose for a 10×10 cm^2 field at an *SSD* of 100 cm.

- The BJR Supplement 17 report clearly points out that electron contamination of the photon beam should be considered a practical shortcoming of the d_{80} method. The use of $TPR_{20,10}$ as a photon beam quality index has also been endorsed by the recent BJR Supplement 25 report, although other beam quality specifiers, such as the $PDD(10)$, are also considered.

- The parameter $PDD(10)$, the percentage depth dose at 10 cm depth in water, determined under the same conditions of field size and SSD as the parameter d_{80}, has in principle the same limitation with regard to the effect of electron contamination as d_{80}. A recommendation has been made that a 1 mm thin lead foil be used in the measurements to remove the unknown electron contamination from the reading at z_{max} and replace it by a known amount of electron contamination to arrive at the percentage depth dose with presence of the lead foil, $PDD(10)_{Pb}$. A correction formula is provided to convert $PDD(10)_{Pb}$ into $PDD(10)_x$, the percent depth dose at a depth of 10 cm in water in a pure photon beam excluding the electron contamination at z_{max}.

Advantages and limitations of the different photon beam quality indices have been discussed at great lengths in the scientific literature. The general conclusion is that there is no unique beam quality index that works satisfactorily in all possible conditions for the entire energy range of megavoltage photon energies used in radiotherapy and all possible linacs used in hospitals and in standards laboratories.

Most recent dosimetry protocols based on in-water calibrations of ionisation chambers use the $TPR_{20,10}$ as the beam quality index (IPEM, IAEA-TRS-398, etc); the AAPM TG-51 protocol, on the other hand, uses the $PDD(10)_x$. For a user in a hospital or clinic there is strictly no advantage of one specifier (index) over the other, as both lead to the same dose conversion factors and hence yield the same *dose-to-water* in the user's beam.

9.8.3. Beam quality specification for megavoltage electron beams

Electron beams are essentially monoenergetic when exiting the accelerator waveguide; however, the electron beam striking the phantom or patient surface at nominal source-surface distance (SSD) exhibits a spectrum that results from the energy spread caused by interactions between electrons and air as well as interactions between electrons and the linac components, such as the collimator, scattering foil, ionisation chamber, and treatment cone. A typical electron beam percentage depth dose distribution is shown in Fig. 9.9.

Until lately, the quality of clinical electron beams has been specified in practically all dosimetry protocols by $\bar{E}_o$, the mean electron energy of the incident spectrum striking the phantom surface. This beam quality index was derived from the measurement of the half-value depth R_{50} defined as the depth at which the electron beam depth dose decreases to 50% of its maximum value. The well-known empirical relationship between $\bar{E}_o$ and R_{50} is:

$$\bar{E}_o = CR_{50} \quad ,$$

$$(9.35)$$

where

C   is a constant (2.33 MeV/cm) and
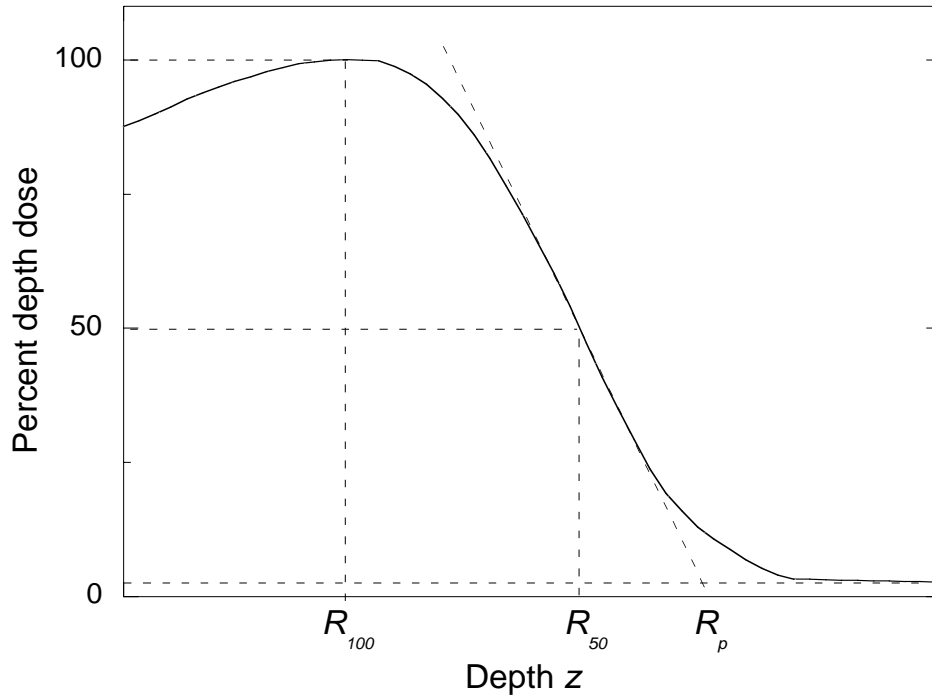R_{50} is the half-value depth in centimeters.

FIG. 9.9. Typical electron beam depth dose distribution with R_{100} the depth of dose maximum, R_{50} the depth of the 50% dose, and R_{p} the practical range of electrons. The dose beyond R_{p} is contributed by the bremsstrahlung contamination of the electron beam produced in linac components, air and water phantom.

Strictly speaking, Eq. (9.35) is valid only for large field sizes (broad beams), for electron energies between 5 MeV and 30 MeV, and for the R_{50} determined from depth dose distributions measured in water with a constant source-chamber distance. The criterion for broad beam is satisfied when the depth dose distribution is independent of field size and this is achieved for field sizes exceeding 12×12 cm^2 for electron beam energies below 15 MeV and at least 20×20 cm^2 for energies above 15 MeV.

The mean energy at depth z in phantom, $\bar{E}_z$, is also a quantity of general use in electron beam dosimetry. An empirical relationship, originally proposed for the most probable energy of an electron spectrum at a depth z in a water phantom, has been recommended by many electron dosimetry protocols to determine $\bar{E}_z$ according to:

$$\bar{E}_z = \bar{E}_\mathrm{o}\left(1 - z/R_\mathrm{P}\right),$$
(9.36)

where R_p is the practical range of the electron beam defined as the depth where the tangent at the steepest point (the inflection point) on the almost straight descending portion of the depth dose curve meets the extrapolated bremsstrahlung background.

Equation (9.36) is only an acceptable approximation of the mean energy at depth in water for electron beams with incident energies less than 10 MeV or for only small depths at higher energies. In other situations the mean electron energy $\bar{E}_z$ is obtained from tabulated data of $\bar{E}_z / \bar{E}_\mathrm{o}$ versus the scaled depth z/R_P (IAEA TRS-277).

- Much of the available data for electron dosimetry that were originally given in terms of $\overline{E}_z$ determined from Eq. (9.36) has been recast to Monte Carlo-determined $\overline{E}_z$ in recent dosimetry protocols.

- When R_{50} and R_p have been determined in a plastic phantom rather than in water, the electron ranges in plastic R_{pl} should be converted into ranges in water as follows:

$$R_{water} = R_{pl}C_{pl},\tag{9.37}$$

  where C_{pl} is a material-dependent scaling factor that was defined by the IAEA TRS-381 protocol as the ratio of average electron penetrations in water and plastic and is equivalent to the effective density of the AAPM TG-25 dosimetry protocol.

- Percentage depth dose distributions for clinical electron beams are most commonly determined from ionisation measurements carried out in water or water-equivalent phantoms with diodes or ionisation chambers.

  - The percent ionisation curve measured with a diode represents the percent depth dose curve, since the stopping power ratios silicon-to-water are essentially constant with depth in phantom, *i.e.*, with electron beam energy.

  - Percentage ionisation curves measured with an ionisation chamber, on the other hand, must be corrected for gradient effects as well as for variations in stopping power ratios water to air with electron energy when determining the percentage depth doses from ionisation measurements.

- R_{50} may be determined from the I_{50}, the 50% value on the percentage ionisation curve measured with an ionisation chamber in water, as follows:

$$R_{50} = 1.029\, I_{50}(in\ cm)\ \text{-}\ 0.06\ \text{cm} \qquad (\text{for } 2\text{ cm} \le I_{50} \le 10\text{ cm})\tag{9.38}$$

  and

$$R_{50} = 1.059\, I_{50}(in\ cm)\ \text{-}\ 0.37\ \text{cm} \qquad (\text{for } I_{50} > 10\text{ cm}),\tag{9.39}$$

- $\overline{E}_o$ and $\overline{E}_z$ are determined from the measured R_{50} and R_p, respectively, and their use in electron beam dosimetry should be considered an approximation that facilitates the selection of dosimetric quantities and correction coefficients, rather than being an accurate statement on the energy parameters of clinical electron beams.

- To avoid potential misunderstandings of the meaning of the energy-based relationships, the recent dosimetry protocols use R_{50} directly as a beam quality index for selecting stopping power ratios and reference depths. This parallels the longstanding practice in photon dosimetry where beam qualities are expressed in terms of the penetration of the beam into a water phantom.

- The choice of R_{50} as the beam quality index is a change from the previous practice of specifying beam quality in terms of the mean energy at the phantom surface. Since $\bar{E}_o$ is normally derived from R_{50}, this change in beam quality index is merely a simplification that avoids the need for a conversion to energy. By choosing a specific reference depth z_{ref} for the calibration in water the stopping power ratio water-to-air has been shown to depend only on the R_{50}.

- The recent dosimetry protocols based on in-water calibration by the IAEA (TRS-398) and the AAPM (TG-51) have endorsed this approach, and all data are expressed in terms of R_{50}. The reference depth z_{ref} for electron beam calibration in water is expressed in terms of the R_{50} as follows:

$$z_{ref} = 0.6\ R_{50}(in\ cm) - 0.1\ \text{cm}. \tag{9.40}$$

- The reference depth in water is close to the depth of dose maximum z_{max} for beams with $R_{50} < 4$ cm ($\bar{E}_o < 10$ MeV); however, for beams with $R_{50} \geq 4$ cm, z_{ref} is deeper than z_{max}. This choice of reference depth may be less convenient than that recommended in previous protocols, since for a given linac no two reference beams will have the same reference depth. However, the new reference depth defined by Eq. (9.40) has been shown to reduce significantly machine-to-machine variations in chamber calibration factors and the gained accuracy justifies its use.

9.9. CALIBRATION OF MEGAVOLTAGE PHOTON AND ELECTRON BEAMS: *PRACTICAL ASPECTS*

This section summarizes the practical aspects of the recommendations of *air-kerma in air*-based and absorbed *dose-to-water*-based codes of practice or protocols for calibration of megavoltage photon and electron beams. For numerical values on the various correction and conversion factors we refer to the IAEA TRS-277 or the IAEA TRS-398 protocols. For background on the physical meaning of the factors we refer to the previous sections.

9.9.1. Calibration of megavoltage photon beams based upon *air-kerma in air* calibration coefficient $N_{K,Co}$

- A cylindrical ionisation chamber is used at a given depth z in a water phantom (typically z is 5 cm or 10 cm).

- The calibration is based on an *air-kerma in air* calibration factor $N_{K,Co}$ obtained in a cobalt-60 beam at a standards laboratory.

- Beam quality is specified with a ratio of tissue-phantom ratios, $TPR_{20,10,}$ at depths of 20 and 10 cm in water or with the percentage depth dose at a depth of 10 cm in water with electron contamination removed, $PDD(10)_x$, as discussed in Section 9.8.2.

- The Bragg-Gray or Spencer-Attix cavity theory is used to determine the absorbed dose or dose rate rate at the point-of-interest at depth z in the water phantom from the measured signal M_Q (charge or current) as follows:

$$D_{\mathrm{w}}(z) = M_Q\, N_{\mathrm{D,air}}\, s_{\mathrm{w,air}}\, p_{\mathrm{wall}} p_{\mathrm{cel}} \quad , \tag{9.41}$$

where

M_Q is the chamber current or charge corrected for influence quantities and measured at beam quality Q;

$N_{\mathrm{D,air}}$ is related to N_{K} through Eq. (9.17);

$s_{\mathrm{w,air}}$ is the restricted stopping power ratio between water and air averaged over the electron slowing down spectrum resulting from the photon spectrum;

p_{wall} is the wall correction factor that accounts for the non-equivalence of medium and wall; and

p_{cel} is the central electrode correction factor that accounts for scatter and absorption of radiation on the central electrode (the factor was ignored in the AAPM TG-21 protocol but introduced in the IAEA TRS-277 and subsequent IAEA dosimetry protocols).

- In the IAEA *air-kerma in air*-based protocols, the displacement effect resulting from the insertion of an air cavity into a phantom is accounted for by defining an effective point of measurement while the cavity perturbation effect is negligible. For cylindrical chambers in high-energy photon beams, the effective point of measurement is located $0.6r$ upstream of the chamber center, r being the cavity inner radius.

- For the purpose of absorbed dose measurements, in the absorbed *dose-to-water*-based protocols, the point of measurement is defined as the center of the chamber and the displacement effects are accounted for by the introduction of the "gradient correction factor" equivalent to p_{dis}.

- The cavity fluence perturbation correction factor p_{cav} is unity in high energy photon beams.

9.9.2. Calibration of megavoltage photon beams based on *dose-to-water* calibration coefficient $N_{\mathrm{D,w,Co}}$

- A cylindrical ionisation chamber is used at a given depth z in a water phantom (typically z is 10 cm).

- The calibration is based on a *dose-to-water* chamber calibration factor $N_{\mathrm{D,w,Co}}$ obtained from a standards laboratory with the chamber irradiated with a cobalt-60 beam at a reference depth z_{ref} in a water phantom.

- The absorbed dose to water $D_{\mathrm{w,Co}}$ at a given depth in phantom in a cobalt beam in the absence of the ionisation chamber is given by:

$$D_{\mathrm{w,Co}} = M_{\mathrm{Co}}\, N_{\mathrm{D,w,Co}} \quad , \tag{9.42}$$

where M_{Co} is the chamber signal (charge or current) corrected for influence quantities, as discussed in Section 9.5.

- When the ionisation chamber is used in a beam quality Q different from cobalt-60 used in its calibration, the absorbed dose to water is given by:

$$D_{w,Q} = M_Q \, N_{D,w,Co} \, k_{Q,Co},$$ (9.43)

where the correction factor $k_{Q,Co}$ corrects for the effects of the difference between the reference cobalt-60 beam quality (Co) and the actual user beam quality Q, and the chamber signal M_Q has been corrected to the reference values of influence quantities, other than beam quality, for which the calibration factor is valid.

- Beam quality Q of megavoltage photon beams is specified, as discussed in Section 9.8.2, either with a ratio of tissue-phantom ratios ($TPR_{20,10}(Q)$) or with the percentage depth dose $PDD(10, 10 \times 10, SSD, Q)_x$.

- The IAEA TRS-398 dosimetry protocol recommends the use of the ratio of *TPR*s, while the AAPM TG-51 protocol recommends the use of PDD_X for megavoltage photon beam quality specification. Despite considerable polemics on the merits of each of the two approaches, in practice they both give essentially the same results for megavoltage photon beams currently used in clinical practice.

- The beam quality correction factor $k_{Q,Co}$ is defined as the ratio, at the beam qualities Q and Co, of the calibration coefficients in terms of the absorbed dose to water of the ionisation chamber:

$$k_{Q,Co} = \frac{N_{D,w,Q}}{N_{D,w,Co}}.$$ (9.44)

- Ideally, the beam quality correction factors should be measured directly for each ionisation chamber at the same quality as the user's beam. In practice, this is generally not possible, so the factors are calculated theoretically assuming the validity of Bragg-Gray cavity theory and using the $N_{D,air}$ concept (see Section 9.4.

9.9.3. Calibration of megavoltage electron beams based upon *air-kerma in air* calibration coefficient $N_{K,Co}$

- For electron beams with energies equal to or above 10 MeV a cylindrical or a parallel-plate ionisation chamber is used at a reference depth in a water phantom (usually close to z_{max}). For electron energies below 10 MeV a parallel-plate ionisation chamber must be used.

- The calibration is based on *air-kerma in air* calibration coefficient N_K obtained in a cobalt-60 beam at the standards laboratory, but parallel-plate chambers a cross-calibration against a cylindrical chamber allows a direct determination of the $N_{D,air}$ coefficient.

- $\bar{E}_o$, the mean electron energy on phantom surface, is used for specifying the electron beam quality (Eq. 9.35).

- The Bragg-Gray cavity relationship is used to determine the absorbed dose at the reference point in water phantom as follows:

$$D_{\text{w}}(z) = M_Q \, N_{\text{D,air}} \, s_{\text{w, air}} \, p_{\text{cav}} p_{\text{cel}} \,, \tag{9.45}$$

where

M_Q is the corrected measured chamber current or charge;

$N_{\text{D,air}}$ is related to N_{K} through Eq. (9.20);

$s_{\text{w,air}}$ is the restricted stopping power ratio between water and air; and

p_{cav} is the cavity fluence perturbation correction factor accounting for the in-scattering effect as discussed in Section 9.3. It is unity for well-guarded parallel-plate chambers and is given by (Eq. 9.33) for Farmer-type cylindrical chambers.

p_{cel} is the central electrode correction factor that accounts for scatter and absorption of radiation on the central electrode of a cylindrical chamber. The factor was ignored in the AAPM TG-21 protocol, but was taken into account in the IAEA TRS-277 and subsequent IAEA protocols, as well as in the AAPM TG-51 protocol.

- For parallel-plate chambers the effective point of measurement is located on the inner surface of the window at its center and no gradient correction is required.

- For cylindrical ionisation chambers, on the other hand, the effective point of measurement is located $0.5r$ upstream from the chamber center. In the AAPM protocols, in the latter case, the point of measurement is defined as the center of the chamber and the gradient effects are to be accounted for by the introduction of the gradient or displacement correction factor p_{dis}.

- The wall correction factor p_{wall} is considered unity in electron beam dosimetry.

9.9.4. Calibration of high-energy electron beams based upon *dose-to-water* calibration coefficient $N_{\text{D,w,Co}}$

- The output calibration is based on a dose-to-water chamber calibration factor $N_{\text{D,w,Co}}$ obtained from a standards laboratory with the chamber irradiated in a reference beam of quality Q_{o}. This reference quality is most commonly a cobalt-60 gamma ray beam ($N_{\text{D,w,Co}}$) but may also be an electron beam.

- Parallel-plate ionisation chambers are recommended for all electron beam qualities and must be used for electron beams of energies below 10 MeV. At electron energies equal or above 10 MeV the use of cylindrical chambers is allowed.

- The reference point for parallel-plate chambers is taken to be on the inner surface of the entrance window, at the center of the window.

- Water is recommended as the reference medium. The water phantom should extend to at least 5 cm beyond all four sides of the largest field size employed. For electron energies below 10 MeV a plastic phantom may be used but all depths must be scaled appropriately.

- The beam quality index for electron beams is R_{50}, the half-value depth in water, measured with a field size of at least 10×10 cm^2 for $R_{50} \leq 7$ g/cm^2 and at least 20×20 cm^2 for $R_{50} > 7$ g/cm^2. The preferred choice of detector for the measurement of the R_{50} is a well-guarded parallel-plate ionisation chamber, the preferred choice of phantom medium is water.

- Output calibration is carried out in a water phantom at a reference depth z_{ref} with a field of 10×10 cm^2. The reference depth is given by:

$$z_{ref} = 0.6\, R_{50} - 0.1\,\text{g/cm}^2, \tag{9.46}$$

  with R_{50} given in g/cm^2. This depth is close to the depth of dose maximum z_{max} at beam qualities $R_{50} < 4$ g/cm^2 ($E_0 \leq 10$ MeV); at higher beam energies it is deeper than z_{max}.

- The absorbed dose to water at the reference depth z_{ref} in an electron beam of quality Q, in the absence of the chamber, is given by:

$$D_{w,Q} = M_Q N_{D,w,Co} k_{Q,Co}, \tag{9.47}$$

  where

  M_Q is the chamber signal corrected for influence quantities;

  $N_{D,w,Co}$ is the calibration factor in terms of absorbed dose to water for the chamber irradiated in a cobalt-60 beam at standards laboratory; and

  $k_{Q,Co}$ is a chamber correction factor which corrects for differences between the reference beam quality (Co) and the actual electron beam quality Q.

- Calculated values of $k_{Q,Co}$ against R_{50} are available in dosimetry protocol documents for a wide variety of parallel-plate and cylindrical ionisation chambers. They are tabulated directly in the IAEA TRS-398 protocol, or determined as a product of conversion and correction factors termed k_{ecal}, $k'_{R_{50}}$ and P_{gr} in the AAPM TG-51 protocol.

9.10.   KILOVOLTAGE DOSIMETRY

During the past thirty years there has been a great deal of development in the dosimetry of high-energy, *i.e.*, megavoltage photon and electron beams. The dosimetry of *kilovoltage* x-ray beams (low and medium energy or orthovoltage x-ray beams), on the other hand, remained more or less static in that period until the late 1980s. Despite this, kilovoltage beams are still in widespread use for the treatment of superficial lesions.

The IAEA TRS-277 protocol devoted a separate, detailed section to kV x rays, setting out a new *air-kerma in air*-based formalism, and this has recently been followed by several national dosimetry protocols on kV x-ray dosimetry. Note that the second edition of the IAEA TRS-277 protocol provided substantial changes to the numerical data given in the original publication of 1987.

9.10.1. Specificities of kilovoltage beams

When kilovoltage x rays interact with medium, the secondary electrons have extremely short ranges due to their much lower initial energy coupled with the rapid increase of the collision stopping power at sub-relativistic energies. This results in several important differences between kilovoltage and megavoltage beams as far as radiation dosimetry is concerned:

- The Bragg-Gray principle can no longer be applied to such beam qualities, *i.e.*, the electron fluence in the air-cavity of the ionisation chamber is not exclusively determined by the electron interactions in the surrounding medium.

- Because of the short electron ranges, absorbed dose can be equated to collision kerma to a very good approximation.

- Radiative losses can be ignored in low atomic number materials, so that absorbed dose and kerma are essentially equivalent.

An ionisation chamber calibration coefficient in kilovoltage x rays is determined with reference to a *free-air ionisation chamber* at a set of kilovoltage radiation qualities in contrast to a single *air-kerma in air* calibration coefficient at cobalt-60 for megavoltage beams. Since the wall thickness of a typical cylindrical ionisation chamber is larger than the range of secondary electrons created in it, charged particle equilibrium is established in the wall without a build-up cap. For this reason and since the chamber calibration coefficient is in terms of *air-kerma in air*, the calibrated chamber acts as a kerma detector even when used in a phantom.

The amount of photon scatter in a tissue-equivalent phantom at kilovoltage energies is much larger than in high-energy photon beams. This fact makes ratios of mass energy absorption coefficients and, to a lesser extent, other, detector-related, dosimetric quantities depth and field size dependent.

Kilovoltage beam quality is specified differently than the megavoltage beam quality. As discussed in Section 9.8, for kilovoltage beams, the beam quality is specified in terms of the *Half-Value Layer* (*HVL*) generally expressed in millimeters of aluminum or, at the top end of the energy range, in millimeters of copper.

It should be noted that beams with widely differing tube potential can have similar *HVL*s, due to the marked effect of different filtrations. Thus, the user determines the *HVL* of the beams of interest and then chooses N_K values for the calibrated chamber for the beam using the calibration curve supplied by the standards lab.

9.10.2. The *air-kerma*-based in-phantom calibration method (medium energies)

For so-called *medium*-energy x-ray beams, typically above 100 kV, various dosimetry protocols recommend that the dose be determined at a depth in a water phantom, similarly to recommendations for megavoltage beam qualities. The various dosimetry protocols differ, however, in their specification of the reference depth.

- The IAEA TRS-277 protocol followed early recommendations of the ICRU and specifies a depth of 5 cm in water.

- The UK protocol (IPEMB, 1996) chose instead 2 cm for the reference depth, considering this to be much more representative of clinical practice; this has also been adopted by several other recent protocols.

The formalism for the determination of the absorbed dose to water is:

$$D_{w,Q} = M_Q N_{K,Q} \left[(\mu_{en}/\rho)_{w,air} \right]_Q \; p_Q, \tag{9.48}$$

where, for the *HVL* of the user's beam (*Q*),

M_Q	is the instrument reading corrected for influence quantities (see Section 9.3.),
N_K	is the *air-kerma in air* chamber calibration coefficient for beam quality *Q*,
$(\mu_{en}/\rho)_{w,air}$	is the mass energy-absorption coefficient ratio, water-to-air, for the photon spectrum at the depth of measurement in water and for field size of the user's beam and
p_Q	is an overall perturbation correction factor which is not to be confused with the p_Q factor of Eq. (9.28) and which consists of multiplicative components of a different nature than those involved in Eq. (9.28).

For a detailed description of these components please refer to, *e.g.,* the AAPM TG-61 protocol.

There is only a weak dependence on field size or depth in the values of $(\mu_{en}/\rho)_{w,air}$ in general and in the values p_{Q_0} for Farmer-type chambers. Cylindrical chambers of the Farmer type are commonly used at this energy range. All recent kV dosimetry protocols agree within about 1.5% in the determination of the absorbed *dose-to-water* at 2 cm depth.

9.10.3. The *air-kerma*-based backscatter method (low and medium photon energies)

Clinically, for these beams, the dose is most often prescribed on the skin surface (strictly just below the surface, where CPE is established). This has led to the most important and widely used method of determining the absorbed dose. The principle is straightforward. A chamber is positioned free-in-air, *i.e.,* with no phantom involved, at a position corresponding to the center of the field on the patient's skin surface. The reading of the (calibrated) chamber yields the *air-kerma in air* $(K_{air})_{air}$. This is then converted into *dose-to-water* at the surface of a phantom at the field size of interest. The energy or quality range for this method differs slightly from protocol to protocol but all of the protocols denote it by the term *low energy*; in the IAEA TRS-277 protocol this range is 10 – 160 kV.

The theoretical route is as follows:

(1)   The *air-kerma in air* $(K_{air})_{air}$ is converted into *water-kerma in air* $(K_w)_{air}$ through the mass-energy absorption coefficient ratio, water to air, but still under *free-in-air* conditions, *i.e.*, for the primary spectrum; this has the advantage that $(\mu_{ab}/\rho)_{w,air}$ is independent of field size.

(2)   Next the *backscatter factor BSF* converts the *water-kerma in air* $(K_w)_{air}$ into the *water-kerma in water* $(K_w)_w$ at the surface of a water phantom.

The formalism for this procedure is:

$$D_{w,Q}^{surface} = M_{free\,air,Q} N_{K,Q} BSF \left[(\mu_{en}/\rho)_{w,air} \right]_{free\,air,Q}, \tag{9.49}$$

where, for the *HVL* of the user's beam (Q),

$M_{free\,air,Q}$ is the instrument reading corrected for influence quantities,
N_K is the *air-kerma in air* chamber calibration coefficient, and
BSF is the backscatter factor for the field size, *HVL* and *SSD* and
$\left[(\mu_{en}/\rho)_{w,air} \right]_{free\,air,Q}$ is the mass energy-absorption coefficient ratio, water to air, for the *free-in-air* (primary) spectrum.

Note that, in principle, the type of ionisation chamber has no significance when one uses the *backscatter* method to determine the surface dose; one is merely using the chamber as a means of transferring the *air-kerma in air* from the standards laboratory to the user's beam. In practice, however, one requires the chamber to exhibit a small variation in N_K with *HVL* over the full quality range.

- The IAEA TRS-277 protocol recommended a thin-window parallel-plate ionisation chamber for this *low-energy* range.

- In the IPEMB protocol a cylindrical chamber of the type NE2561 or NE2611 is to be used over the complete *medium* and *low* energy range.

- The AAPM TG-61 protocol explicitly incorporates in an equation of the Eq. (9.49) type a chamber stem correction termed $P_{stem,air}$ that accounts for the change in chamber calibration factor for the difference in field size between the standards laboratory beam and the clinical beam. For an ideal stem-free ionisation chamber this correction is unity. For a cylindrical Farmer-type chamber the correction is less than 1% but it can be very significant for certain types of thin window chambers due to their significant chamber bodies.

9.10.4 *Air-kerma in air*-based calibration method for very low energies

In German and UK protocols there is even a third method at the lowest energies, the so-called *Bucky* therapy range, which corresponds approximately to $8 - 50$ kV. In this energy range a thin-window parallel-plate chamber is recommended for calibration purposes.

- The backscatter method may be invalid for the very small field sizes sometimes employed clinically in such low-energy beams, *i.e.*, the field size can be insufficient to completely cover the chamber and hence the value of the product $M N_K$ will no longer yield the correct value for *air-kerma in air* in the user's beam.

- For these beams the parallel-plate chamber is placed at the surface of a phantom and the dose at the surface is determined. The relevant expression is identical to that for the in-phantom medium-energy method except that now the factor p_Q (denoted as k_{ch}) refers to the specific parallel-plate chamber employed and pertains to the surface dose rather than that at 2 cm depth.

- The lack of data available for the k_{ch} factor led to the assumption that it is equal to unity. However, this assumption cannot be correct as it assumes that the scatter from the body of the chamber is negligible. Recent experimental values have found that it varies from about 1.01 to 1.08 for a field diameter of 5.4 cm (at a focal distance of 50 cm) depending on the chamber, beam quality, and phantom. An update to the IPEMB protocol will recommend these significant correction factors.

9.10.5. Absorbed *dose-to-water*-based calibration method

Standards of absorbed *dose-to-water* in the kilovoltage x-ray range are not generally available. However, it is possible to derive calibration coefficients in terms of absorbed dose-to-water from *air-kerma in air* calibration coefficients using one of the accepted dosimetry protocols (*e.g.*, the IAEA TRS-398). Thus, any calibration laboratory with standards of *air-kerma in air* can in this way provide (derived) calibration coefficients in terms of absorbed dose to water. Even though this is formally equivalent to the user obtaining an *air-kerma in air* calibration and individually applying the same *air-kerma* protocol, it has the advantage of permitting the widespread use of the unified methodology of absorbed *dose-to-water* standards in an area of dosimetry, where standard methods are notably lacking.

9.11 ERROR AND UNCERTAINTY ANALYSIS FOR IONISATION CHAMBER MEASUREMENTS

9.11.1. Errors and uncertainties

An *error* is defined as the difference between the measured value of a measurand and the true value. An error has a sign and a correction factor can be associated with it. When the error is known, the true value of the measurand can be calculated from the measured value. An *uncertainty* associated with a measurement is a parameter that characterizes the dispersion of the values that can be attributed to the measurand. The value of the uncertainty is usually an estimated standard deviation, has no sign and is assumed to be symmetrical with respect to the estimated value of the quantity. It is a measure of our lack of exact knowledge after all recognized systematic effects have been eliminated by applying appropriate corrections.

9.11.2.   Classification of uncertainties

Uncertainties of measurements are expressed as relative standard uncertainties and the evaluation of standard uncertainties is classified into type A and type B.

- *Type A* uncertainties are inherently random and obtained by a statistical analysis of a series of observations. A 1 sigma *type A* uncertainty corresponds to the standard error on the mean of a set of observations at the 68% confidence level.

- *Type B* uncertainties are determined through other than statistical, but often subjective, methods and account for systematic effects in the determination of a quantity.

- Although of totally different nature, *type A* and *type B* uncertainties are often combined assuming they are independent, using the propagation law of uncertainties without cross-correlation (*i.e.*, relative standard uncertainties are quadratically summed).

9.11.3.   Uncertainties in the calibration chain

The IAEA TRS-398 dosimetry code of practice carries out an extensive uncertainty analysis on the *calculated* values of the beam quality conversion factors k_Q for photon and electron beams. For photon beams the estimated relative standard uncertainty for the calculated beam quality conversion factors is 1.0%. For electron beams, the value amounts to 1.2% for cylindrical chambers and 1.7% for parallel-plate chambers when based on a cobalt-60 calibration technique and 0.9% for cylindrical chambers and 0.6% for plane parallel chambers when based on a cross-calibration technique in an electron beam.

In order to obtain the uncertainty on a beam calibration, the above mentioned uncertainties need to be combined with the uncertainties on:

- The absorbed dose calibration coefficient at cobalt-60 or in a high energy electron beam, if a cross calibration technique is used;

- Issues related to the *in-phantom* measurement of absorbed dose in the clinic. These comprise type A and type B uncertainties on positioning of the chamber in the water phantom, the temperature and pressure measurement, the ion recombination, polarity and electrometer correction factor (if present) and on the linac stability during the measurements of absorbed dose. For a detailed analysis we refer to the IAEA TRS 398 protocol.

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