from "Clinical Dosimetry Measurements in Radiotherapy", eds D.W.O. Rogers and J.E. Cygler
Medical Physics Publishing (Madison WI) 2009

Chapter 4

General Characteristics of Radiation Dosimeters and a Terminology To Describe Them

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1. Introduction

When discussing any type of radiation detector, there are a variety of characteristics that need to be described and quantified if possible. The purpose of this chapter is both to describe these characteristics and to define a terminology and notation that will be used throughout this book in an effort to clarify the concepts.

2. Basic Quantities and Notation

We start by defining some basic quantities and their dependencies on influence factors. Many of these quantities are, or may be, dependent on many other quantities, and typically only those dependencies that are relevant in a given situation are made explicit. So for example, the reading from an ion chamber will usually be thought of as dependent on the absorbed dose to water when it is placed in a water phantom but will be thought of as dependent on the air kerma when used free in air.

M_{det} $(D,\ \dot{D}\ ,\ Q,\ \theta,\ \phi)$: the reading of the detector (detector reading). This is the quantified output of the detector and can be some sort of meter reading or other output related to charge, or current, or the light output from a detector or the signal from an Electron Spin Resonance (ESR) spectrometer, etc.

This reading may be dependent on many different quantities such as: the dose to the medium the detector is located in, D; the dose rate at the same point, $\dot{D}$; the beam energy or beam quality, Q; the angular orientation of the detector with respect to the radiation source and geometry, θ, ϕ; etc. The reading may depend on environmental quantities such as temperature at the time of irradiation or reading (which may be at the same time), pressure or humidity, etc. The reading may also depend on certain measurement corrections, e.g., polarity effect corrections or possibly dose-rate dependent effects such as ion recombination. The quantity M_{det} will be considered to be corrected to some reference environmental conditions from the original raw reading $M_{\text{det}}^{raw}(D, \dot{D}, Q, \theta, \phi, T, P, H)$ (see section 3.1).

$D_{\text{det}}(D, \dot{D}, Q, \theta, \phi)$: the average absorbed dose to the material of the detector in gray (Gy) (detector dose or dose to the detector). This quantity is related to the active material in the detector system. For an ion chamber it is the dose to the gas in the cavity and in a diode detector it is the dose to the active region of the silicon wafer. D_{det} can have all of the same dependencies as M_{det}.

$D_{med}(Q)$: the dose to the medium at the point of measurement in the absence of the detector. This is frequently the quantity we are interested in determining with a detector. It may be dependent on many different quantities such as the beam quality, Q, the depth in a phantom, z, etc. The point of measurement is frequently taken as the midpoint of the detector [e.g., for a thimble ion chamber or a thermoluminescent dosimeter (TLD)], but it is not necessarily so (e.g., it is taken as the front face of the air cavity for a parallel-plate ion chamber in phantom). Significant corrections to account for the difference between the dose to the medium in the absence of the detector and the dose to the detector's material may be needed for detectors with very different materials from the medium of interest or in rapidly changing regions of dose (e.g., the gradient effect in ion chambers) or for detectors which attenuate the radiation field significantly (e.g., a TLD chip in a low-energy photon beam).

$K_{med}(Q)$: the kerma to the medium at the point of measurement in the absence of the detector. The quantity that we are often interested in for measurements in air.

$S_{AD,med}(D, \dot{D}, Q, \theta, \phi)$: the detector's absorbed dose sensitivity defined as:

$$S_{AD,med}\left(D,\dot{D},Q,\theta,\phi\right) = M_{\text{det}}\left(D,\dot{D},Q,\theta,\phi\right)/D_{med}\left(Q\right) \qquad \left[\text{rdg/Gy}\right], \quad (4.1)$$

i.e., the detector's reading divided by the absorbed dose to the medium at the point of measurement of the detector in the absence of the detector. In the literature this quantity is sometimes just referred to as the detector's sensitivity, but for clarity it is important to specify exactly what the sensitivity is relative to, i.e., absorbed dose. This quantity is sometimes loosely referred to as a "detector's response," but that term is used in many circum-

stances with varying meanings so it is best to use the precise terminology of "absorbed dose sensitivity." This is a measured quantity that cannot be calculated by Monte Carlo techniques without an explicit model for the intrinsic energy dependence discussed in section 3.4.1 below.

$S_{K,med}(K, \dot{K}, Q, \theta, \phi)$: the detector's kerma sensitivity, which by definition is:

$$S_{K,med}\left(K,\dot{K},Q,\theta,\phi\right) = M_{\text{det}}\left(K,\dot{K},Q,\theta,\phi\right)/K_{med}\left(Q\right) \qquad \left[\text{rdg/Gy}\right], \quad (4.2)$$

which is analogous to $S_{AD,med}$. It is especially important in brachytherapy applications to retain the ",med" subscript (usually air or water) to distinguish this quantity from the air-kerma strength which is denoted by S_K.

$S_{AD}^{rel}\left(D,\dot{D},Q,\theta,\phi\right)$: the detector's relative absorbed dose sensitivity or the detector's absorbed dose sensitivity relative to a reference beam quality, Q_0, is a dimensionless quantity given by:

$$S_{AD,med}^{rel}\left(D,\dot{D},Q,\theta,\phi\right) = S_{AD,med}\left(D,\dot{D},Q,\theta,\phi\right)/S_{AD,med}\left(D,\dot{D},Q_0,\theta,\phi\right), \quad (4.3)$$

where here the emphasis is on the variation with beam quality relative to the reference beam quality (e.g., ^{60}Co). This quantity is referred to in the literature in various ways, sometimes as just the "response" or the "sensitivity," which are not adequate terms.

$S_{K,med}^{rel}\left(K,\dot{K},Q,\theta,\phi\right)$: the detector's relative kerma sensitivity is, by analogy,

$$S_{K,med}^{rel}\left(K,\dot{K},Q,\theta,\phi\right) = S_{K,med}\left(K,\dot{K},Q,\theta,\phi\right)/S_{K,med}\left(K,\dot{K},Q_0,\theta,\phi\right). \quad (4.4)$$

$N_{D,w}$, N_K: the detector's absorbed dose to water and air-kerma calibration coefficients are defined by:

$$N_{D,w}\left(Q\right) = \frac{D_{water}\left(Q\right)}{M_{\text{det}}\left(Q\right)} \qquad \left[\text{Gy/rdg}\right], \quad (4.5)$$

and

$$N_K\left(Q\right) = \frac{K_{air}\left(Q\right)}{M_{\text{det}}\left(Q\right)} \qquad \left[\text{Gy/rdg}\right], \quad (4.6)$$

where in these cases the absorbed dose or the kerma refers to the values of these quantities at the point of measurement of the detector when the detector is absent. These are calibration coefficients rather than factors because a coefficient is needed when the units are being changed (i.e., the coefficient is not dimensionless) whereas a factor is a dimensionless quantity (e.g., a scaling factor or a correction factor).

Note that the calibration coefficients are just the inverse of the detector's absorbed dose to water sensitivity.

$$N_{D,w}\left(Q\right)=\frac{D_{water}\left(Q\right)}{M_{\text{det}}\left(Q\right)}=\frac{1}{S_{AD,water}\left(Q\right)}. \tag{4.7}$$

In the above, θ is the angle between the axis of the beam and either, as appropriate, the cylindrical axis of the detector or the normal to the front face of the detector and ϕ is the angle about the same axis starting from some arbitrary angle related to the detector. As concrete examples, for a cylindrical Farmer-like chamber, θ is often 90° (i.e., the cylindrical axis is perpendicular to the beam) and θ is defined with respect to some mark on the ion chamber so that the chamber is always used with the "front" facing the beam. For a TLD chip used in a brachytherapy measurement, the angle θ is that between the normal to the flat face of the chip and the line to the center of the seed/source being measured and ϕ needs to be arbitrarily defined.

Only the functional dependencies relevant to a particular discussion are usually shown but all of the dependencies are there. So, for example, if one writes $D_{\text{det}}(D)$, this means we are talking about the change in the dose to the detector as the dose to the medium changes and all other variables are nominally held constant.

3. Dosimeter Characteristics

With the above quantities and notation in place, we are in a position to discuss the various characteristics of dosimeters. While we often make many assumptions about these characteristics for a given class of detector, it is worthwhile to remember that each of these characteristics must be understood in order to use an instrument correctly.

The details of each characteristic for specific dosimeters will be discussed in the chapters about the various dosimeters, and in this section examples will be given for illustrative purposes. Each of the following subsections focuses on a specific quality or effect, but it must be emphasized that they are all in play at the same time. For example, when we quantify the energy dependence, we assume that the dose-rate dependence has been accounted for, or when we talk of the dose-rate dependence the assumption is that we are dealing with a specific beam quality and dose or dose rate. The hope is that various characteristics are independent, but this is not necessarily the case.

3.1 Environmental and Measurement Corrections

If a detector's reading is sensitive to environmental and measurement conditions (most commonly *T, P, H,* and dose-rate dependencies such as P_{ion}), then it is usual to correct the reading to some reference or standard set of conditions using an environmental correction k_{env} and a dose-rate correction k_{dr}:

$$M_{\text{det}}\left(T_0,P_0,H_0\right)=k_{env}\left(T,P,H\right)k_{dr}\left(M_{\text{det}}^{raw}\left(\dot{D}\right)\right)M_{\text{det}}^{raw}\left(T,P,H,\dot{D}\right). \tag{4.8}$$

So for a vented ion chamber it is common to correct the reading by the temperature pressure correction discussed in chapter 6. For very precise Fricke dosimetry it is necessary to correct the results for the temperature when the Fricke solution is irradiated and the temperature when the solution is read out in the spectrophotometer. Another component of the environmental corrections is the subtraction of a background reading determined from unirradiated control detectors (e.g., when using TLDs) or other forms of measurement of leakage or background readings. The background component of the environmental correction is given by:

$$k_{env}^{bkgd} = \left(1.0 - \frac{M_{\text{det}}^{bkgd}}{M_{\text{det}}}\right), \tag{4.9}$$

where M_{det}^{bkgd} is the detector's background reading during the measurement of M_{det}.

There can be other instrumental corrections, such as the polarity correction for ion chambers, which are needed to correct a detector reading. These can be considered part of k_{env} unless they are dose-rate dependent.

3.1.1 Dose-Rate Dependence

A detector's reading may also depend on the dose rate being measured, i.e.,

$$M'_{\text{det}}\left(\dot{D}\right) = k_{dr}\left(M_{\text{det}}^{raw}\left(\dot{D}\right)\right) M_{\text{det}}^{raw}\left(\dot{D}\right), \tag{4.10}$$

where $k_{dr}\left(M_{\text{det}}\left(\dot{D}\right)\right)$, the **dose-rate dependence** is defined to be 1.00 for some reference conditions. Ideally, k_{dr} is unity, but as a minimum its dependence on $\dot{D}$ is known. For example, for an ion chamber the reference conditions are 100% collection efficiency. Although general ion recombination increases as the dose rate increases, such detectors can still be used by applying the ion recombination correction, $P_{ion} = k_{dr}(M_{\text{det}})$, although an ion chamber with $P_{ion} = 1.00$ is easier to use.

3.2 Intrinsic Linearity

For any detector it is essential to know whether the reading of the detector is directly proportional to the dose being measured, i.e., the dose to the detector material. In general one has:

$$D_{\text{det}}(D) = \alpha k_l[M_{\text{det}}(D)]M_{\text{det}}(D), \tag{4.11}$$

where $k_l(M_{\text{det}}(D))$ is the **intrinsic linearity** defined to be 1.00 for some reference dose, D_0, and α is a quantity which relates $M_{\text{det}}(D_0)$ to $D_{\text{det}}(D_0)$. If k_l is independent of D and $M_{\text{det}}(D)$, then the detector's response is said to be linear. This simply means that if the dose to the detector is doubled, the reading of the detector doubles. However, there are detectors for which this is not the case and it must be taken into account. Figure 4.1

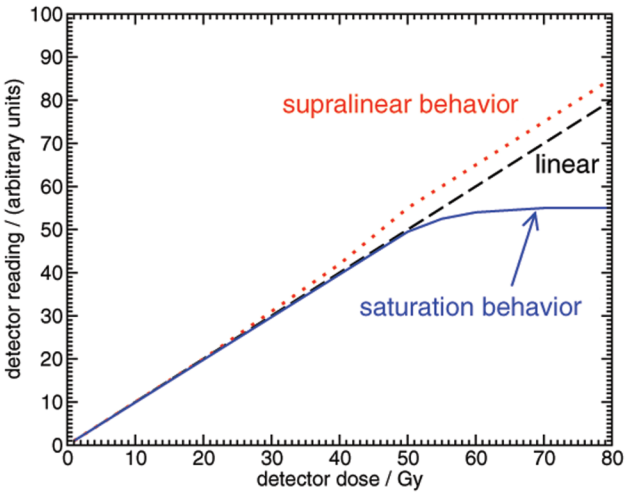

Figure 4-1. Examples of supralinear behavior and saturation behavior of the detector reading, $M_{det}(D)$ and the dose to the detector materials, $D_{det}(D)$.

demonstrates schematically two such situations. One example is the supralinearity of TLDs wherein the reading increases faster than the dose to the detector for certain reasonably large doses. Another, more common example is signal saturation which is observed for detectors when they are exposed to very large doses. The range of doses over which a detector is said to be linear should always be specified.

3.3 Energy Dependence of the Detector

3.3.1 Intrinsic Energy Dependence

The overall energy dependence or beam quality dependence of a detector's reading is broken into two components. The first relates the detector's reading to the average dose to the material of the sensitive detecting element.

$$D_{det}(Q) = k_{bq}(Q)M_{det}(Q). \tag{4.12}$$

For example, for ion chambers used in electron and photon beams, $k_{bq}(Q)$ is generally assumed to be constant (i.e., independent of beam quality Q) because the charge released is directly proportional to the dose to the gas via the quantity W/e, which is taken to be a constant. Although only the beam quality dependence is explicitly recognized in equation (4.12), k_{bq} could, in principle, be dependent on all of the other variables mentioned above except the dose D since the term "intrinsic

linearity" is reserved for that dependence. It is often implicitly assumed that k_{bq} is independent of Q, but this assumption must be verified for all detectors. For example, there is a considerable amount of data showing that TLD response per unit dose to the TLD material varies by 5% to 15% for low-energy photons, but many people have ignored this variation in the past (see chapters 14 and 24). Similarly the absorbed dose response of the Fricke dosimeter has been shown to vary by about 1% between ^{60}Co beams and 20 MV beams (see chapter 31).

The above beam quality dependence, $k_{bq}(Q)$, needs a unique name, but there is no widely adopted term. It will be called the **intrinsic energy dependence of the detector.** Note that $k_{bq}(Q)$ relates the same two quantities, viz., $M_{\text{det}}(Q)$ and $D_{\text{det}}(Q)$ that are related by $k_l(D)$, the intrinsic linearity and $k_{dr}\left(\dot{D}\right)$, the dose-rate dependence, but the three are concerned with different functional dependencies, which one hopes are independent.

3.3.2 Absorbed-Dose Energy Dependence

The other energy dependence of a detector is widely recognized. It relates the dose to the detector material to the dose to the medium at the point of measurement of the detector in the absence of the detector.

$$D_{med}(Q) = f(Q)D_{\text{det}}(Q). \tag{4.13}$$

To help distinguish the two aspects of the energy dependence, this will be called the **absorbed-dose energy dependence of the detector.** In ion chamber dosimetry we have:

$$f\left(Q\right) = \left(\frac{\bar{L}}{\rho}\right)_{gas}^{med} P_{wall}P_{cel}P_{stem}P_{repl}\,, \tag{4.14}$$

where all the quantities on the right, which are defined in chapters 3 and 9, are dependent on the beam quality Q. In TLD dosimetry, a simple approximation is that:

$$f\left(Q\right) = \left(\frac{\bar{\mu}_{en}}{\rho}\right)_{TLD}^{med}, \tag{4.15}$$

although this is known to be too simplistic. For low-energy photons self-shielding becomes an issue (i.e., the TLD attenuates the photons more than the phantom material) and for high-energy photons, the TLD is no longer a photon detector and is responding to electrons originating in the medium as well as in the TLD. In general, one needs to use Monte Carlo techniques to calculate $f(Q)$ for TLDs and many other detectors. But it must be emphasized that Monte Carlo calculations only provide $f(Q)$ and one needs an explicit model of the intrinsic energy dependence to link a

Monte Carlo calculation to the clinically relevant absorbed dose calibration coefficient given by equation (4.1). In other words, the overall energy dependence of a detector is given by the product of $k_{bq}(Q)f(Q)$, which is the detector's calibration coefficient:

$$f(Q)k_{bq}(Q) = N_{D,med}(Q). \tag{4.16}$$

Note that f and k_{bq} may also depend on D, $\dot{D}$, θ, and ϕ and in general depend on the physical characteristics of the detector (e.g., P_{wall} clearly depends on the wall material and thickness, and similarly $f(Q)$ for a TLD will depend on the size and orientation of the TLD material).

3.4 Overall Equation

Putting all of the above together, one has:

$$D_{med}\left(Q\right) = fk_{bq}k_l M_{\text{det}}\left(Q,D,\dot{D},\theta,\phi\right) = fk_{bq}k_l k_{env}k_{dr}M^{raw}_{\text{det}}\left(Q,D,\dot{D},\theta,\phi,T,P,H\right), \tag{4.17}$$

where, in principle, each factor is a function of all the possible variables, but in practice the values of θ, ϕ are usually fixed and thus ignored. If k_l and k_{dr} are not constants, then the hope is that they are independent of the beam quality Q. Likewise, one would hope that f and k_{bq} are independent of D and $\dot{D}$. So in favorable situations the equation becomes:

$$D_{med}\left(Q\right) = f\left(Q\right)k_{bq}\left(Q\right)k_l k_{env}\left(T,P,H\right)k_{dr}\left(\dot{D}\right)M^{raw}_{\text{det}}\left(Q,D,\dot{D},\theta,\phi,T,P,H\right). \tag{4.18}$$

However, even for an ion chamber we should not assume k_{dr} is only a function of $\dot{D}$ since it may also depend on Q as well. Thus one measures P_{ion} in the beam of interest—although the major practical concern is that in different beams, $k_{dr}\left(\dot{D}\right)$ (i.e., P_{ion}) may vary even with the same dose rate since different doses per pulse may give the same dose rate if the pulse rate changes.

3.5 Directional Dependence

Many detectors have a directional dependence and this dependence is properly accounted for in the f and k_{bq} factors defined above. Furthermore, for detectors with a strong directional dependence, any standard protocol for its use will need to define all factors for a single, well-specified orientation of the detector relative to the beam.

However, it is sometimes useful to discuss this dependence separately, in which case the directional dependence is defined by:

$$D_{\text{det}}(\theta_0,\phi_0) = h(\theta,\phi)D_{\text{det}}(\theta,\phi), \tag{4.19}$$

where θ_0,ϕ_0 specify some reference situation. Ideally, there is no directional dependence. A cylindrical ion chamber should be axially symmetric (i.e., independent of ϕ) except for low-energy x-ray beams where there may be some small dependence. However there may be slightly more dependence for the θ direction. These dependencies can be critical for solid-state detectors.

3.6 Spatial Resolution/Size Effects

Any detector averages the dose over some sort of volume and this must be taken into account in some cases, especially in small beams, or near the edges of beams, or in intensity-modulated radiation therapy (IMRT) beams where the detector size may be significant relative to spatial variations in the dose. The gradient effect for ion chambers is another example of a spatial effect. These effects are in principle included in the f and k_{bq} factors, but it may prove useful to extract a separate correction which accounts for spatial effects on the detector's reading or absorbed-dose response in a specific beam relative to that in reference conditions (e.g., central axis of a 10×10 cm^2 field for the same beam quality).

Acknowledgments

I would like to thank Alan Nahum, Jan Seuntjens, and Jeff Williamson for helpful suggestions and comments on early drafts of this chapter. Work supported by Natural Sciences and Engineering Research Council of Canada (NSERC) and the Canada Research Chairs program.