Ionizing Radiation Dosimetry and Medical Physics

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Published in Phyics in Canada, Vol 51(#4) 1995, 178 – 181

July 12, 1995

Introduction

In applications of ionizing radiation to problems related to medicine, it is important to measure the amount of radiation delivered. In diagnostic procedures such as x-ray examinations, nuclear medicine, CT scans, PET etc, this measurement is both for the optimization of image quality, and for radiation protection purposes. However, the need for accurate dosimetry is greatest in radiation therapy for cancer. In radiotherapy, a large dose of radiation is delivered to a tumour (typically 10 times the dose which would kill a person receiving this dose to the entire body) and the effectiveness of the treatment depends on delivering the dose with an accuracy of 5% or better in some situations.

Thus one of the major roles of physicists in radiotherapy is to ensure the accurate delivery of dose to a tumour and at the same time plan the treatment so that the dose to healthy tissue is minimized. An active area of research is treatment optimization taking into account the almost infinite variety of treatments possible with modern therapy accelerators controlled by computers. Part of this work concerns what exactly should be optimized, with a significant effort being made to take into account the biological response of different tissues. While the use of dosimetry based on biological response is the long term goal, practical clinical dosimetry today is based on the quantity absorbed dose, and accurate measurement of absorbed dose represents one of the major responsibilities of clinical medical physicists.

With the accepted need for accurate delivery of dose, and because information about the optimal tumour dose for treating various cancers is based on medical experience gained from around the world, a sophisticated international radiation dosimetry system is in place with the goal of ensuring accurate and uniform dose measurement. At the center of this system are the national measurement standards for ionizing radiation. In Canada these are the responsibility of the NRC (National Research Council) through the Ionizing Radiation Standards Group

of the Institute for National Measurement Standards. Our role is to develop, maintain and disseminate these standards. This includes comparing them to other national standards, often under the auspices of the BIPM (Bureau International des Poids et Mesures) in Paris.

Clinical dosimetry plays a critical role in the dosimetry chain, and hence it is based on dosimetry protocols developed and approved by scientific societies such as the American Association of Physicists in Medicine (AAPM) which includes many Canadian members. As an aside, the field of Medical Physics has a very strong tradition in Canada and thus Canadians play a significant role in medical physics research, *e.g.* Canadian physicists publish more than 3 times as many papers per capita as Americans in the AAPM's journal, Medical Physics.

Since ion chambers are the most precise and easily used instruments for measuring absorbed dose in beams of ionizing radiation, clinical dosimetry protocols are based on using ion chambers[1]. The first step in the protocol is to have the chamber regularly calibrated against NRC's national measurement standards. The protocols are very complex, partially because they start with a calibration for a quantity, exposure, which is different from the quantity of interest, absorbed dose, and partially because the calibration field is ^{60}Co whereas much of radiotherapy is now done with high-energy photon or electron beams from accelerators (5 to 25 MeV).

At this point in history there is a major change occurring in how clinical dosimetry will be done. NRC and primary standards labs in other countries are developing new primary standards for absorbed dose in accelerator beams and new clinical protocols are being developed to exploit the new standards. The goal is to make clinical dosimetry much simpler and to improve its accuracy. The rest of this article defines some of the major quantities of interest and then outlines how clinical ion chambers work, how they are used at present for clinical dosimetry, what the new primary standards are, and how they will likely be used in the future.

Exposure, air kerma and absorbed dose

These 3 quantities play a central role in clinical radiation dosimetry. A detailed discussion and history is given in the article in this issue by Cormack and Carrier[1] and various international reports discuss the definitions in great detail[2]. I wish to give a brief outline here for completeness.

The quantity exposure is defined as the total charge of one sign produced in dry air when all electrons liberated by photons in a unit mass of air are completely stopped in air. The energetic knock-on electrons produce about 30 electrons for every keV of energy they lose, thus a single 100 keV knock-on electron produces 3000 free electrons. The SI unit for exposure is C/kg as suggested by the definition, but the old unit, the roentgen (1R=2.58 10^{-4} C/kg) is still often used. The definition refers to all electrons set in motion by photons in the volume of air with mass m, and the charge is collected from throughout the electron's path as it

[1]Many other types of dosimetry system are used clinically for relative as opposed to the absolute dosimetry discussed here, *e.g.* solid state diodes, TLDs, film, Fricke dosimeters.

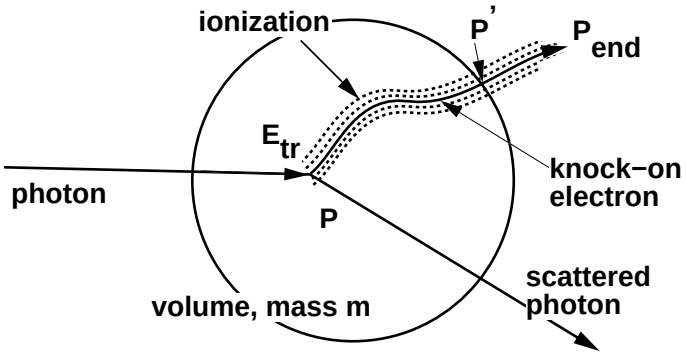

Figure 1: The exposure, air kerma and absorbed dose for a single photon which Compton scatters and transfers an energy E_{tr} to an electron at point P. The volume of interest is shown as a circle and the mass of this volume is m. The energetic electron set in motion at P slows down and stops at P_{end}. As it slows down it loses energy which results in 30 ion pairs being created near the track, per keV of energy lost.

slows down (fig 1). This quantity has the advantage of being directly measurable, at least for low-energy photons (<300 keV or so) where the electrons don't move too far as they slow down. The primary measurement standard is a free air chamber which has two parallel plates with a potential across them for collecting the charge liberated by a well defined photon beam passing between them. This works very well for low-energy photons, and allows a direct measurement with an accuracy of about 0.5%. The measurement technique breaks down for photons with energies typical of ^{60}Co beams (up to 1.33 MeV). In this case very different techniques using ion chambers and Bragg-Gray cavity theory are needed to measure exposure (discussed below). However, exposure suffers from two fundamental flaws. The first is that it is only defined for photons interacting in air; the second is that the quantity becomes ill-defined as photon energies become higher as in accelerator beams because the range of the electrons slowing down becomes so large.

These problems are both overcome by introducing the quantity kerma, which is the Kinetic Energy Released per unit MAss (unit J/kg or gray). For photon beams the kinetic energy released is the kinetic energy transferred to electrons in the material. The quantity must always be defined with respect to the specific material in which the interactions are taking place (*e.g.* air kerma, water kerma *etc*). This quantity is well defined at all energies and for all materials and in fig 1 it is just the sum of all the energy transfers to charged particles, E_{tr}, in the volume divided by the mass m. For kerma, it does not matter whether the charged particles slow down inside the volume or not. As we shall see below, air kerma and exposure are closely related, and although the measurement standards are embodied in the same equipment, air kerma is not directly measurable in the same sense as exposure. Nonetheless kerma plays an important role in radiation dosimetry because it is the energy *released* per unit mass of material, and not surprisingly this is closely related to the energy *absorbed* per unit mass of material.

The absorbed dose to a material, D_{med}, is:

$$D_{med} = E_{med}/m \qquad \text{[J/kg or gray]} \tag{1}$$

where E_{med} is the energy absorbed (J) in a mass m (kg) of the material[2]. In fig 1 the absorbed dose sums the energy deposition on all the tracks from P to P′. For volumes which are large compared to the tracklength, the kerma and absorbed dose are virtually identical, especially since absorbed dose also includes energy deposition within the volume by electrons set in motion outside the volume. This tends to balance the energy deposited outside the volume by those electrons starting inside the volume.

Absorbed dose is currently taken as the best physical indicator of biological response. Because of this, absorbed dose to water (which closely resembles human tissue but is well defined) is the quantity which is used to specify the amount of radiation to be used in clinical practice. Absorbed dose has the further advantage that it is directly measurable in a variety of ways. The most straightforward is by calorimetry which determines the energy deposited per unit mass of material by measuring the temperature rise. The temperature rise in water for a typical radiotherapy dose of 200 cGy (2 J/kg), delivered in 1 or 2 minutes, is 500 μK (specific heat capacity of water is 4181 J/(kg K)).

Bragg-Gray cavity theory

Another fundamental of radiation dosimetry is Bragg-Gray cavity theory. This "theory" relates D_{cav}, the absorbed dose to the material in a small cavity in a uniformly irradiated piece of matter, to D_{med}, the absorbed dose to the surrounding material. Specifically:

$$D_{med} = D_{cav} \left(\frac{\overline{L}}{\rho}\right)^{med}_{cav} \qquad [Gy], \tag{2}$$

where $\left(\overline{L}/\rho\right)^{med}_{cav}$ is the ratio of spectrum-averaged electron mass collision stopping powers in the surrounding medium and the cavity. This quantity is complex in detail, but basically stems from the fact that as an electron slows down, the dose it deposits is proportional to the mass collision stopping power of the medium (energy lost per g/cm^2) and thus the ratio of the doses in the cavity and medium is obtained by averaging these stopping powers over the entire energy spectrum of electrons present. Sophisticated Monte Carlo techniques have been developed to simulate the passage of radiation through materials and these can be used to calculate the required values of $\left(\overline{L}/\rho\right)^{med}_{cav}$ (see *e.g.*, ICRU Report 35[3]), the major uncertainty being the uncertainties in the stopping powers themselves. The essential point is that $\left(\overline{L}/\rho\right)^{med}_{cav}$ is a calculable quantity and conceptually it is best thought of as D_{med}/D_{cav}.

[2]Strictly the definition is based on average values as the mass shrinks to zero so that dose is defined at a point.

Ion chambers

Because of their long term stability, high precision (≈0.1% for a good clinical chamber), direct readout, and relative ease of use, ion chambers have become the standard instrument for clinical dosimetry measurements. An ion chamber for clinical radiation dosimetry typically consists of a thin(0.5 mm) wall of material such as graphite surrounding a small (0.6 cm^3), well-known volume of air with a voltage applied between the wall and an electrode to collect the charge produced in the air by the ionizing radiation (see figure 2). The voltage (typically 300 V) is high enough that the charge produced is collected but not so high that the accelerated

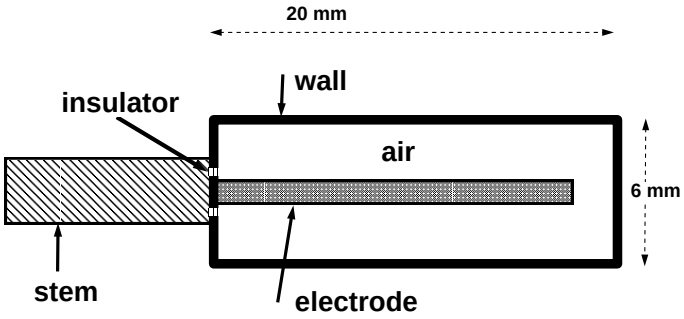

Figure 2: Typical Farmer ion chamber used for high-quality clinical dosimetry.

electrons themselves create more ionization (as in a proportional counter) nor so high that a single charged particle entering the detector causes an avalanche and a pulse (as in a Geiger counter).

For electron and high-energy photon beams used in radiotherapy, ion chambers are always inserted in some material in order to measure the absorbed dose to that material by using Bragg-Gray cavity theory. This material can be either a small "build up" cap if the measurement is done in air, or a water tank which is taken as an approximation to the human body. In either case, essentially all of the ionization in the air cavity is generated by electrons which enter from the surrounding material.

The use of ion chambers in clinical dosimetry is predicated on a fortunate fact, *viz.* as an electron slows down and loses energy in air, the amount of charge it produces per unit energy lost in the air is a constant, independent of the electron's energy. This means we can write a direct relationship between the charge released in the air, Q_{air}, and the energy lost by electrons in the air, E_{air}.

$$E_{air} = (W/e)_{air} Q_{air} \qquad [J] \tag{3}$$

where $(W/e)_{air}$ is the constant energy loss per unit charge produced (33.9 J/C or 33.9 eV per ion pair created).

As an aside, this relationship leads to the close link mentioned above between exposure and air kerma. Air kerma is the kinetic energy transferred to electrons per unit mass and, ignoring bremsstrahlung losses ($< 0.3\%$), if these electrons lose all their energy slowing in the air (*i.e.* $E_{air} = K_{air}$), we can use eqn(3) to deduce the charge released by photons interacting per unit mass of air as $K_{air}/(W/e)_{air}$; but this is just the exposure, X, *i.e.*:

$$K_{air} = X(W/e)_{air} \qquad [Gy]. \tag{4}$$

Considering the ion chamber again, if m, the mass of the air in the chamber is known, then from the definition of absorbed dose (eqn(1)):

$$D_{air} = \frac{E_{air}}{m} = \frac{Q_{air}}{m}\left(\frac{W}{e}\right)_{air} \quad [Gy]. \tag{5}$$

This equation tells us that by measuring the charge collected from the ion chamber, and knowing the mass of the air in it, we know the absorbed dose to the air in the chamber. Applying Bragg-Gray cavity theory (eqn(1)) allows us to calculate the absorbed dose in the medium in which the ion chamber has been placed. In a water tank, one can deduce:

$$D_{water} = \frac{Q_{air}}{m}\left(\frac{W}{e}\right)_{air}\left(\frac{\overline{L}}{\rho}\right)_{air}^{water} \quad [Gy]. \tag{6}$$

As long as one uses an ion chamber for which the mass of the air is known, then by measuring the charge released in the cavity one can deduce the absorbed dose to the water at the position of the ion chamber by using the known value of $(W/e)_{air}$ and the value of $\left(\overline{L}/\rho\right)_{air}^{water}$ which is appropriate for the radiation beam being used. Determining this latter quantity is a somewhat complex task, but clinical dosimetry protocols recommend the appropriate procedures and data.

Equation(6) is conceptually simple, but only applies for the ideal case. For example, it assumes all the charge released is measured, and it ignores the fact that the mass of the air varies with temperature and pressure. Both of these effects can be handled easily and directly but there are more complex problems. Equation(6) also assumes that the cavity is small enough and the water is uniformly irradiated so that the Bragg-Gray cavity theory applies and it ignores the fact that the ion chamber has a wall which is graphite, not water, as required by cavity theory. Corrections accounting for these assumptions are handled by dosimetry protocols and for high-quality clinical chambers the required corrections are typically 2% or less (see, *e.g.* ref[4]).

The real complication of clinical dosimetry comes in determining the mass of the air in the ion chamber. If this were strictly a mechanical measurement, the ion chambers could be made with sufficiently well-known dimensions to specify the volume and hence the mass of air. Since the electric fields in these small chambers are not uniform, there are small regions from which the charge is not completely collected. Thus one really requires the effective mass of air in the chamber. In practice this is determined from calibrations at the primary standards labs (as discussed below). This calibration procedure has the benefit of ensuring that the calibrated clinical ion chambers are working properly and hence this "first step in the chain" is considered essential to guarantee that the clinical dosimeters are accurate.

The present clinical dosimetry chain and ^{60}Co exposure standards

As discussed above, the goal of clinical dosimetry is to determine the absorbed dose to water in a radiotherapy beam. Primary standards of absorbed dose in a ^{60}Co beam had been

developed in many laboratories in the seventies. In 1977, the NRC, under Bill Henry, was the first standards lab to offer an absorbed-dose calibration service. However, since the current generation of clinical dosimetry protocols was being developed in the late seventies, they were based on ^{60}Co exposure calibrations since these were widely available.

Historically, exposure played a central role because it could be directly measured in low-energy x-ray beams using free air chambers (FACs). However, these FACs do not work in ^{60}Co beams and primary standards became based on ion chambers for which the effective mass of air in the chamber could be determined mechanically. In this case the ion chamber has "thick" walls of graphite and eqn(6) is used to determine the absorbed dose in the graphite walls. Taking into account effects of photon attenuation, scatter, and electron transport in the walls by multiplying by a factor called K_{wall} (typically a 1% correction), the absorbed dose to graphite is transformed into the graphite kerma. This is easily related to air kerma because, for ^{60}Co beams the energy per unit mass transferred to electrons in air and graphite is the same to one part in a thousand (the Compton interaction predominates and the number of electrons per unit mass in air and graphite are close to the same). Given the air kerma, eqn(4) can be used to give the exposure (note that the value of $(W/e)_{air}$ drops out):

$$X = \frac{Q_{air}}{m}\left(\frac{\bar{L}}{\rho}\right)^{gr}_{air} K_{wall} \qquad [C/kg]. \tag{7}$$

This equation, used with many other small correction factors, provides the basis of exposure standards in ^{60}Co beams around the world. Experimentally one need measure accurately only the charge and the volume of the chamber (to determine the mass of gas it contains). However, exposure, which was originally defined because it could be measured directly, now requires a very complex procedure to measure it. In fact, it is quite unsatisfactory because it requires the stopping-power ratio, $\left(\bar{L}/\rho\right)^{gr}_{air}$ which turns out to be one of the least well known stopping-power ratio, and because recent theoretical work at NRC has shown that the technique used for determining K_{wall} is wrong by up to 1% which is roughly 5 times the previously assumed systematic uncertainty[5].

Despite these problems, in 1995, exposure calibration of clinical ion chambers is currently the basis for clinical dosimetry. By calibrating the ion chamber in the standards lab's ^{60}Co beam of known exposure, one basically uses eqn(7) (where the exposure calibration factor, $N_x = X/Q_{air}$) to determine m, the effective mass of air in the clinical ion chamber. Following this, eqn(6) is used to establish the absorbed dose to water by placing the ion chamber in a water tank. Equation(6) introduces several systematic uncertainties into the dose determination since it requires knowledge of $(W/e)_{air}$ and $\left(\bar{L}/\rho\right)^{water}_{air}$. To obtain these factors requires measurements in radiation fields and considerable theoretical analysis.

Overall the present system of clinical dosimetry has an uncertainty of 3 to 4% in accelerator radiotherapy beams under reference conditions. The system also requires external radiation data - external in the sense that they are not measured in standards labs. This latter drawback can be minimized by realizing that the final dose really depends on the product $(W/e)_{air}\left(\bar{L}/\rho\right)^{gr}_{air}$ and this product can be measured very accurately in a standards lab using an absorbed-dose calorimeter. Thus the exposure-based dosimetry system is actually based on absorbed-dose calorimeters, which are themselves standards for absorbed dose. While this approach is valid, it certainly becomes exceptionally convoluted[6].

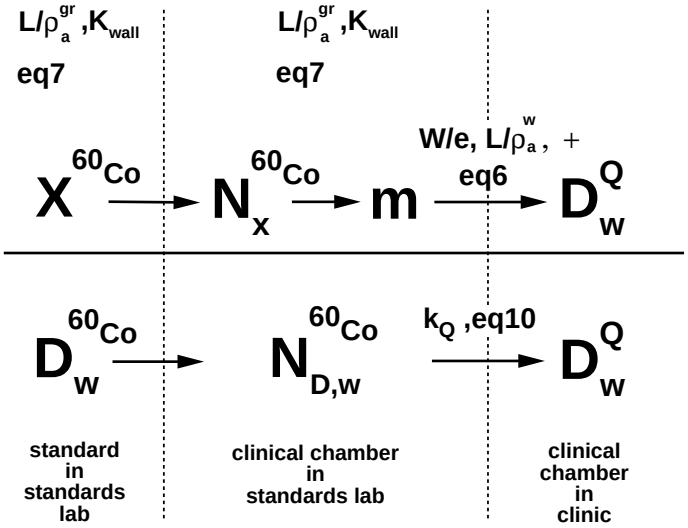

Figure 3: Protocol chains based on exposure calibrations (upper) and absorbed-dose calibrations (lower).

To simplify the overall system, standards labs have been developing new standards for absorbed dose to water in clinical beams.

A system based on absorbed-dose standards

As mentioned above, the most direct way to measure absorbed dose is with an absorbed-dose calorimeter. The first generation of primary standards for absorbed dose were based on graphite calorimeters in which a small disk of graphite was thermally isolated from the surrounding graphite phantom and its temperature rise was measured when irradiated. The measuring system is calibrated by injecting a known amount of electrical energy into the same graphite disk and measuring the temperature rise. This system works very well because all of the energy absorbed in graphite is transformed into heat and hence into temperature rise. Also the heat diffuses very quickly so that there are no hot spots and one or two thermistors can get an accurate estimate of the average temperature rise. The only problem with this system is that the measured absorbed dose to graphite must be converted to an absorbed dose to water, but this can be done in several ways with an accuracy of better than 1%.

In 1980, Steve Domen of NIST (the US standards lab) proved that calorimetry can be done directly in a water tank[7]. Water has a very low thermal diffusivity, which means that the heat does not rapidly flow away from where it is created. Thus, absorbed dose to water measurements can be done at a point in a water tank without providing thermal isolation. This system has the obvious advantage of measuring the quantity of interest. There is one significant problem: irradiated water undergoes chemical reactions which can be endothermic or exothermic. This leads to what is called a thermal heat defect, *i.e.* a difference between

the energy absorbed by the material from the radiation and the thermal energy released. The magnitude of the defect depends sensitively on the (sometimes) trace impurities in the system and on the accumulated dose and dose rate. However, in the last 15 years the understanding of the heat defect of irradiated water has improved dramatically and the heat defect in water-based systems is known to within a few tenths of a percent. This has allowed several primary standards based on water calorimetry to be developed at NRC and elsewhere[8].

To summarize, there are several major approaches to primary standards for absorbed dose to water (besides the two described) and various on-going comparisons between national primary standards labs suggest that they are in agreement at the 1% level or better[9, 10]. These standards apply in ^{60}Co beams and accelerator beams as used in radiotherapy.

Thus, in principle a clinical physicist will send an ion chamber to NRC and have it directly calibrated in the type of beam in which it will be used. Then:

$$D_{water} = Q_{air} N_{D,w} \qquad [Gy] \tag{8}$$

where $N_{D,w}$ is the absorbed dose to water calibration factor. The only problem is that clinically there may be many beams and each accelerator calibration factor is very expensive to determine (likely $3,000 or more).

For this reason, the AAPM is developing a new dosimetry protocol which starts from an absorbed-dose calibration factor in a ^{60}Co beam and introduces a factor, called k_Q, which takes into account the variation of $N_{D,w}$ with beam quality, Q, so that the calibration factor in a beam of quality Q is given by:

$$N^{Q}_{D,w} = k_Q N^{^{60}Co}_{D,w} \qquad [Gy/C] \tag{9}$$

While equations for k_Q can be derived in terms of the various quantities introduced in previous protocols, it is conceptually simpler to recognize it as a quantity which can be experimentally measured using primary standards for beams of different quality Q. It varies by less than 5% for different photon beam qualities and all clinical ion chambers made of a given wall material are predicted to have the same (within 0.4%) values of k_Q as a function of beam quality[6].

Thus the new generation of protocols will consist of assigning dose based on:

$$D_{water} = Q_{air} k_Q N^{^{60}Co}_{D,w} \qquad [Gy] \tag{10}$$

where $N^{^{60}Co}_{D,w}$ is measured at the primary standards laboratory for each clinical chamber, Q_{air} is measured in the clinic and k_Q is taken from a protocol but is based on measurements with primary standards for absorbed dose. There are no other "corrections" needed. This is much simpler than the current procedures based on eqn(6) and (7) which actually include between four and six further "corrections" which were not included here to keep the presentation "simple". The new approach is more coherent because it uses only the quantity absorbed dose and ties the clinically-assigned absorbed dose to primary standards in a corresponding accelerator beam. In contrast, the exposure-based system is very complex because: it requires conversion from exposure to absorbed dose; it uses a protocol based on theory which uses many factors which are not rigorously understood; and it requires externally-determined values of parameters such as $(W/e)_{air}$ and $\left(\overline{L}/\rho\right)$.

While the new approach appears to be giving nearly the same dose estimates in clinical photon beams as the old system, the added clarity and simplicity of the new absorbed-dose-based system should lead to fewer mistakes in the clinic, and should also allow more time to be spent developing innovative solutions to the myriad other problems faced by physicists delivering cancer radiotherapy.

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