Overview of Dosimetry

1. Introduction

It would seem self-evident that the meaning of dosimetry is the measurement of dose. In radiation measurement the only proper use of the term dose is as an abbreviation for absorbed dose. However, in a wider sense the term dosimetry is used to refer to measurement of various quantities related to the effects of radiation on matter: energy imparted per unit mass (absorbed dose), kinetic energy released per unit mass (kerma), numbers of particles (e.g. fluence) or a function of the above quantities such as the product of absorbed dose and a biological radiation quality factor.

Many different detectors have been developed for the measurement of ionising radiation. Its fundamental property, the ability to split neutral atoms and molecules into charged ions and free radicals is the cause of the harmful effects in biological systems on which radiotherapy depends. The amount of damage done is related to the energy absorbed from the radiation. It turns out that this relation is proportional for high-energy photon and electron beams.

The energy $W$ required to create an ion pair is only of the order a few electron volts (the value depends on the material), but an electron with a kinetic energy of thousands or millions of electron volts will create a large number of ion pairs and do damage in proportion to their number. To a good approximation, $W$ is independent of the energy of the primary radiation and takes the same value for high-energy photons and electrons.

At low intensities (doses), most of the immediate effects of ionising radiation on biological systems are reversed within a few hours by the highly effective repair mechanisms operating inside cells. What remains are random (stochastic) effects in which the probability of occurrence is proportional to the dose received.

This is the basis of the current view, which is that the most useful quantity for the measurement of ionising radiation is absorbed dose. Water has been chosen as the preferred absorber, because it is similar enough to tissue.

An overview is given of various detectors that are used for dosimetry with emphasis on those that are used for absolute or reference measurement of absorbed dose or air kerma and those that from the basis of primary standards. They can be divided in three categories: detectors that directly measure the quantity absorbed dose, detectors that measure ionisation and detectors that quantify in a direct or indirect way the number of radicals formed in a medium. The lecture ends with a brief discussion on the relation between the energy dependence of the detector response and the beam quality specification.
2. Detectors

2.1. Dosimeters directly measuring the energy imparted: calorimeters

A charge liberated in the medium by an energetic ionising particle results in an energy cascade in which the energy of the initial particle is shared among many less energetic secondary particles. Eventually charges and ions recombine and the liberated energy ends up as heat, which can be measured as a temperature rise. However, even though a typical radiotherapy treatment fraction creates enough ionisation in the target volume to cause cell death through biochemical processes, the amount of energy involved is tiny. The technique of deriving absorbed dose via a measurement of the temperature rise is called calorimetry.

The temperature rise in the calorimeter medium is proportional to absorbed dose according to the specific heat capacity of the absorbing medium under the condition that no physico-chemical changes of the state of the medium take place as a result of the irradiation. If such a change takes place the energy deficit is called the heat defect.

The expression to derive absorbed dose to the medium \( D_{\text{med}} \) from the measured temperature rise \( \Delta T \) is

\[
D_{\text{med}} = c_{\text{med}} \cdot \Delta T \cdot \frac{1}{1 - h}
\]

where \( c_{\text{med}} \) is the specific heat of the absorbing medium and \( h \) is the heat defect of the medium. The main challenges for calorimetry are the control of the heat defect and the prevention of heat exchange between the irradiated material and its environment. The heat defect of different materials has been measured by different authors and a review on this can be found in the ICRU report 64.

<table>
<thead>
<tr>
<th>Material</th>
<th>( \rho ) (kg m(^{-3}))</th>
<th>( c ) (J kg(^{-1}) K(^{-1}))</th>
<th>( k ) (J s(^{-1}) m(^{-1}) K(^{-1}))</th>
<th>( \alpha ) (m(^2) s(^{-1}))</th>
<th>( \Delta T/D ) (mK Gy(^{-1}))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water</td>
<td>998</td>
<td>4180</td>
<td>0.602</td>
<td>1.44\times10(^{-7})</td>
<td>0.24</td>
</tr>
<tr>
<td>Graphite</td>
<td>1770</td>
<td>725</td>
<td>135</td>
<td>1.05\times10(^{-4})</td>
<td>1.4</td>
</tr>
</tbody>
</table>

*Table 1: Thermal properties of water and graphite: the mass density \( \rho \), the specific heat capacity \( c \), the thermal conductivity \( k \) and the thermal diffusivity \( \alpha \). The last column gives the temperature rise due to an absorbed dose of 1 Gy.*

Table 1 gives some thermal properties of water and graphite as well as the temperature rise that is the result of an absorbed dose of 1 Gy. It is obvious that these temperature increases are very small and require sensitive technology for temperature measurement as well as excellent temperature control. Figure 1 gives a schematic overview of how calorimetry is implemented. The devices to measure the temperature rise are thermistors in contact with the calorimeter medium and coupled into a Wheatstone bridge.
The measurement typically involves the recording of temperature drift curves before and after the irradiation and extrapolations to mid-run to derive the ionising radiation induced temperature rise, although a variety of alternative operational and analysis techniques are in use.

![Diagram of calorimeter set-up]

**Figure 1:** Schematic representation of a calorimeter set-up including a calorimeter phantom, thermally isolated from its environment, a thermistor coupled into a Wheatstone bridge and a typical bridge output signal.

### 2.1.1. Graphite calorimeters

Table 1 shows that for the same dose, the temperature rise in graphite will be six times the temperature rise in water. This puts less stringent requirements on the sensitivity of the temperature probe and is one of the reasons why graphite calorimetry as a dosimetry technique has been developed much earlier than water calorimetry. NPL’s absorbed dose standards are at present based on graphite calorimeters.

In graphite calorimetry, lattice impurities and chemical reactions with dissolved oxygen have been proposed as mechanisms for a potential heat defect, but they are in general assumed to be negligible. Experiments have confirmed this within the achievable uncertainties.

As can be seen from table 1, the thermal diffusivity in graphite is much larger than in water and its magnitude is such that heat deposited in a graphite volume distributes itself within that volume in a timescale of milliseconds, far too short to allow an accurate measurement of temperature rise at the point of measurement inside the phantom. Hence, in graphite calorimeters a core needs to be thermally isolated from the surrounding graphite by one or more air or vacuum gaps as shown in figure 1.

Since we use graphite calorimeters, the primary standards do not, after all, quite realise the quantity that we disseminate in our calibration services. For both photon and electron beams, we maintain sets of reference standard...
ionisation chambers which are calibrated in terms of absorbed dose to graphite by direct comparison against the relevant primary standard. By a combination of theoretical analysis, experimental measurement and Monte Carlo simulation, the calibrations of these chambers are converted from dose to graphite to dose to water. These reference chambers are then used, in water, to calibrate our own and other secondary standard chambers.

The conversion to water introduces several factors, of which the most significant are the ratios of photon mass-energy absorption coefficients (for photon beams) and electron mass stopping powers (for electron beams), for water and graphite. These are reasonably well known (i.e. with only small uncertainty), and vary by less than 1.5% over the range of energies available from the NPL linac.

![Diagram](image)

**Figure 2:** (a) Schematic drawing of a simplified graphite calorimeter and (b) sketch of the graphite calorimeter that forms the basis of NPL's primary absorbed dose to water standard for high-energy photon beams.

Overall, the variation of the correction factors that apply to the primary standard calorimeters is less than 1%. They also take account of:

- The effect of gaps in the nearly homogeneous graphite calorimeter
- Beam uniformity effects due to the finite size of the calorimeter core

### 2.1.2. Water calorimeters

Water calorimeters have been developed since about 1980, when Domen (1980) realized that it is not required to thermally isolate a sample of water to measure the dose. Due to the low heat diffusivity of water, the spatial dose
distribution due to the irradiation remains stable for a period of time, which is long enough to allow an accurate measurement of the temperature rise. For water calorimetry, correction factors have to be applied to the basic calorimetry equation for scatter perturbations, conductive heat flows, convective heat flows, non-uniformity of dose distributions and radiation induced chemical reactions.

For water calorimetry, a chemical heat defect results from the chemical reactions that take place between radiation induced radicals and chemicals in the aqueous environment. Especially when impurities are present these reactions can result in net endothermic or exothermic effects. The measured heating of the medium is then not equal to the energy that has been dissipated locally by radiation. The proper control of the chemical heat defect requires the use of very high purity water and good control of solved gasses.

Water calorimetry should, in principle, be regarded as the preferred method for reference dosimetry as it is the most direct way to determine the quantity of interest, absorbed dose-to-water. However, it is a time consuming method and not very suitable for periodic dosimetry in a clinical environment.

![Figure 3: (a) Sealed water calorimeter phantom with is enclosure, (b) cylindrical glass vessel with thermistor probes positioned and (c) detail of the thermistor probes.](image)

2.2. Dosimeters measuring ionisation

Ionisation is the physical process at the heart of most precise measurements in dosimetry though the ionisation is not in a biological system, but usually in air. An electric field is applied to the sensitive volume, which sweeps the ionisation onto electrodes of opposite polarity. The electric current produced is...
usually tiny (a fraction of a nanoamp) but it is at least an order of magnitude easier to measure this than to measure the millidegree temperature rise in our calorimeters.

The amount of ionisation is proportional to the mass of the sensitive volume – when this is air at ambient pressure, account must be taken of density variations, by correcting to standard temperature and pressure.

So in principle we can derive dose to air \( D_{\text{air}} \) in the cavity from the charge \( Q_{\text{air}} \) collected on the collecting electrode from the accurate knowledge of the volume, the air density \( \rho_{\text{air}} \) and the fundamental quantity \( W_{\text{air}} \), which is the mean energy required to produce an ion pair in dry air (\( W_{\text{air}} = 33.97 \) eV for radiotherapeutic photon and electron beams).

\[
D_{\text{air}} = \frac{Q_{\text{air}}}{\rho_{\text{air}} \cdot V} \cdot \frac{W_{\text{air}}}{e}
\]

Note that if the air density \( \rho_{\text{air}} \) in this expression refers to normal conditions of pressure and temperature, the charge should be corrected for any deviation from normal conditions at the time and location of the measurement, using the temperature and pressure correction factor

\[
f_{\rho T} = \frac{1013.25}{p} \cdot \frac{(T + 273.15)}{293.15}
\]

If it is an air volume surrounded by a medium then Bragg-Gray cavity theory is used to relate absorbed dose to air in the cavity to absorbed dose in the medium:

\[
D_{\text{med}} = D_{\text{air}} \cdot \left( \frac{S^{SA}}{\rho} \right)_{\text{med,air}}
\]

where \((S^{SA}/\rho)_{\text{med,air}}\) is the medium to air Spencer-Attix stopping power ratio for the charged particle fluence spectrum present in the undisturbed medium.

The ionisation in air depends weakly on the relative humidity (RH), but this variation is negligible (less than 0.1%) provided RH lies between 20% and 70%.

Finally, some positive and negative ions may recombine before they reach the electrodes: the smaller the electric field applied, the slower the ions drift, and the more time they have to recombine. Recombination also increases with ion density and hence with (instantaneous) dose rate. For this reason, pulsed beams show greater and variable amounts of recombination, and care must be taken when transferring a calibration between one pulsed beam and another, or between a continuous and a pulsed beam.

2.2.1. Free-air ionisation chambers

A direct measurement of the kinetic energy released by photons interacting in air is unfortunately not possible. Instead, we collect the ionisation produced as this kinetic energy is spent breaking up the atoms in air. The quantity
exposure (total charge of one sign produced per unit mass of air) is much more closely related to free-air chamber operation. The required conversion from exposure to air kerma introduces more factors, of which the energy required to produce an ion pair in dry air, $W_{\text{air}}$, is the key ingredient.

![Figure 4: Schematic representation of a free air ionisation chamber.](image)

The key to understanding the principle of a free-air chamber design is the idea of charged particle (electron) equilibrium, CPE. Every photon passing through the aperture at the front of the chamber goes on to interact at some point downstream. The chamber must be large enough that, if the interaction occurs in the plane of the aperture, the resulting electron cannot reach the collecting volume. In that case, every electron which does reach the collecting volume must have originated somewhere inside the air volume enclosed by the chamber. Neglecting the effects of photon attenuation, and considering all possible electron tracks, one can see that for every electron track leaving one end of the collecting volume, there will be another track entering the collecting volume at the opposite end. A reciprocity theorem states that the total length of those parts of all electron tracks that lie within the collecting volume is the same as the total length of the complete tracks of only those electrons that start in the collecting volume.

In this way we can relate the quantity of interest (kerma) which is associated with the point at which photons first interact, to the quantity that the free-air chamber responds to (ionisation) which is associated with the points at which the photons’ energy is finally absorbed.

The sensitive volume $V$ of a free-air chamber is the product of the aperture area $A$ and the effective length $L$ of the collecting volume (see figure 4).

Apart from the factors required to convert exposure into air kerma, the corrections applied to the free-air chamber reading take account of:

- Attenuation of the primary X-ray beam between the aperture and the collecting volume ($K_{\text{air}}$)
- The extra ionisation collected from electrons produced by photons scattered within the chamber \((K_{sc})\)
- Ionisation lost when electrons strike the collecting electrode (non-zero only when the free-air chamber is smaller than optimal) \((K_{e})\)

The overall expression to derive air kerma from the measured charge \(Q_{air}\) becomes

\[
K_{air} = \frac{Q_{air}}{\rho_{air} \cdot V} \cdot \left(\frac{W_{air}/e}{1-g}\right) \cdot K_{att} \cdot K_{e} \cdot K_{sc} \cdot K_{hum} \cdot K_{pol} \cdot K_{ion}
\]

where, besides the quantities defined above, \(g\) is the fraction of kinetic energy of charged particles that is lost in radiative energy (such as bremsstrahlung), and \(K_{hum}\) the humidity correction factor, \(K_{pol}\) the correction for the difference between the charge measured when operating at polarising voltages that are equal in magnitude but opposite in sign and \(K_{ion}\) the correction factor for ionisation losses due to recombination.

### 2.2.2. Cavity ionisation chambers

The size of a free-air chamber should be twice the range of the maximum energy electron generated by the photon beam being measured. For \(^{60}\text{Co}\) γ-radiation, this would be several metres, and quite impractical. Even if the chamber were operated at elevated air pressure, the attenuation and scatter corrections would be uncomfortably large. Instead, standards laboratories have adopted graphite-walled cavity chambers (graphite is reasonably air-equivalent) in which the sensitive volume can be measured mechanically. These chambers are mostly intended to realise air kerma (originally exposure). The steps required to derive air kerma from the measured charge are explained below and are illustrated in figure 5.

**Figure 5: Steps in the derivation of air kerma free in air from the ionisation in a graphite walled cavity ionisation chamber.**
As seen before, dose to air for the electron fluence spectrum present in the air cavity can be derived provided the volume of the cavity is accurately known:

\[ D_{air} = \frac{Q_{air}}{\rho_{air}} \cdot \frac{W_{air}}{e} \cdot V \]

Applying Bragg-Gray cavity theory using Spencer-Attix stopping power ratios yields absorbed dose in graphite (figure 5a):

\[ D_g = D_{air} \cdot \left( \frac{S_{SA}}{\rho} \right)_{g,air} \]

Absorbed dose in graphite equals collision kerma in graphite apart from a displacement due to the finite range of the electrons produced in graphite (figure 5b). This can be accounted for by a correction factor \( K_{cep} \):

\[ K_{col,g} = D_g \cdot K_{cep} \]

Collision kerma in graphite is related to collision kerma in the surrounding air through the ratio of mass energy absorption coefficients. Correction factors for attenuation and scatter in the wall thickness need to be applied (figure 5c):

\[ K_{col,air} = K_{col,g} \cdot \left( \frac{\mu_{en}}{\rho} \right)_{air,g} \cdot K_{att} \cdot K_{sc} \]

Correction for radiative losses yields kerma in air:

\[ K_{air} = \frac{K_{col,air}}{(1 - g)} \]

The overall expression becomes

\[ K_{air} = \frac{Q_{air}}{\rho_{air}} \cdot \frac{(W_{air} / e)}{(1 - g)} \cdot \left( \frac{S_{SA}}{\rho} \right)_{g,air} \cdot \left( \frac{\mu_{en}}{\rho} \right)_{air,g} \cdot \prod K_i \]

where

\[ \prod K_i = K_{att} \cdot K_{cep} \cdot K_{sc} \cdot K_{stem} \cdot K_{hum} \cdot K_{pol} \cdot K_{ion} \]

The product \( K_{att} \cdot K_{sc} \cdot K_{cep} \) is summarised as \( K_{wall} \). In the past, there has been considerable debate about this factor since part of it, \( K_{att} \cdot K_{sc} \), used to be measured in many PSDLS by a linear extrapolation method, which has been proven to be wrong. Other PSDL’s, including NPL, based wall perturbation correction factor on a full Monte Carlo simulation of the response of the ionisation chamber. It is now generally accepted that \( K_{wall} \) should be obtained from such a simulation.

One correction factor that has not been mentioned before is \( K_{stem} \), which corrects for the perturbation of the radiation field due to the stem. It is usually measured by putting a dummy stem on top of the chamber.
NPL maintains a set of three primary standard cavity ionisation chambers, which are now over 40 years old, and their replacement with new chambers is ongoing in the present programme.

It is also possible to realise absorbed dose to water using an ionisation chamber. This is the procedure adopted by the BIPM and their standard is (arbitrarily) selected to provide the reference value in the worldwide key comparison database for absorbed dose to water. The method also requires the ratio of photon mass-energy absorption coefficients water to graphite, the ratio of electron mass-stopping powers graphite to air, the \( W_{air} \) value, the fraction of energy, \( g \), going into bremsstrahlung and correction factors for photon scatter and attenuation in the chamber wall. The steps required in this procedure are illustrated in figure 6.

\[
\text{Beam} \quad \frac{E}{m} \quad (E/m)_0 \quad k_{cav} \quad (E/m)_{cav} \quad (\frac{\mu}{\rho})_{w,c} \cdot \psi_{w,c} \cdot (1-\epsilon)_{w,c} \quad D_c \quad D_w \quad P
\]

Figure 6: The successive steps considered in the derivation of absorbed dose to water air from the ionisation in a graphite walled cavity ionisation chamber positioned in a water phantom.

### 2.2.3. Extrapolation chambers

Extrapolation chambers are plane-parallel chambers of which the distance between the electrodes can be varied and accurately set. This allows one to derive the gradient of the ionisation with respect to the volume (proportional to the plate separation) and thus to avoid the need of calibrating the chamber as a means to determine its volume. Extrapolation chambers are
the basis of standards of absorbed dose in low and medium-energy x-rays as well as for beta emitting sources.

In an extrapolation chamber, not the charge per unit volume is measured, but the change of the measured charge when the volume is changed:

\[
D_{\text{air}} = \frac{Q_{\text{air}}}{\rho_{\text{air}} \cdot V} \cdot \frac{W_{\text{air}}}{e} = \frac{1}{\rho_{\text{air}}} \cdot \frac{dQ_{\text{air}}}{dV} \cdot \frac{W_{\text{air}}}{e} = \frac{1}{\rho_{\text{air}} \cdot A} \cdot \frac{dQ_{\text{air}}}{dx} \cdot \frac{W_{\text{air}}}{e}
\]

2.2.4. Ionisation in solid state detectors

Ionisation in solid semiconductors (diodes, diamond, germanium) provides the basis of extremely sensitive detectors which can either be made very small or which can count individual ionising particles with high efficiency. A disadvantage of this type of detectors is that they often exhibit a significant dose rate dependence.

In a TLD crystal, on the other hand, the free electrons created by radiation become trapped in a long-lived metastable state, and the dose received is inferred from the light given off when the crystal is subsequently annealed by heating. On the other hand, TLD shows a much more marked energy-dependence in its sensitivity when calibrated in terms of absorbed dose.

Solid-state detectors are usually not deemed to be appropriate for reference dosimetry but are commonly used as relative dosimeters.

2.3. Dosimeters measuring radical formation

Some radiation detectors make use of the fact that ionisation can leave radicals in the medium for a short or longer time, which can be quantified by direct or indirect methods.

2.3.1. Direct measurement of radical formation: ESR

In the case that permanent radicals are formed, the spin of the unpaired electron can be separated in two quantum mechanical energy states in a magnetic field. Excitation of these states shows a resonance at a certain radio frequency and an energy absorption spectrum can be measured around this resonance frequency. Figure 8 gives a schematic representation of this process.

Alanine, an amino acid, in crystalline form has this property of forming a stable radical and is used at NPL for dosimetry. One reason for interest in
alanine ESR is that the dosimeter readout is a non-destructive process, and the dosimeter may be kept for some years as an archival record of the dose actually delivered.

Over the whole range of megavoltage X-ray beam energies available at NPL, the alanine ESR signal really is proportional to the absorbed dose to water where the dosimeter was irradiated (the variation of alanine sensitivity with X-ray energy is within the range ± 0.5 %, and apparently random).

2.3.2. Indirect measurement of radical formation: chemical dosimetry

In chemical dosimetry systems the dose is determined by measuring the chemical yield produced in the medium of the dosimeter due to irradiation. Various chemical systems are used but for standard dosimetry purposes the most commonly used chemical dosimeter is the ferrous sulphate dosimeter in which the change in optical absorbance due to the oxidation of Fe$^{2+}$ ions to Fe$^{3+}$ is measured with a spectrophotometer making use of the strong absorption peak of the latter ions at $\lambda = 304$ nm. The ferrous sulphate dosimeter response is expressed in terms of its sensitivity, known as the radiation chemical yield, G-value, and defined as the number of moles of ferric ions produced per joule of the energy absorbed in the solution. An advantage of this chemical dosimeter is that it is almost water equivalent.

In the use of the ferrous sulphate dosimeter as a primary standard, its sensitivity is determined by a total absorption experiment in which a known amount of charged particles with a known energy are totally absorbed in the ferrous sulphate solution. From this experiment the chemical yield is derived after correction for bremsstrahlung losses and scatter perturbation. The solution is then used in small quartz vials to determine dose to water in a water phantom, requiring a conversion from dose to ferrous sulphate solution and corrections for perturbations of the radiation field by the quartz walls of the vial.
Until recently it was assumed that the chemical yield of the ferrous sulphate dosimeter does not change with energy for high-energy photon and electron beams, but it has been shown that there is a variation of about 0.7% between the chemical yield in $^{60}$Co and in high-energy photon beams.

Chemical dosimetry could in principle be used as a primary method for the measurement of absorbed dose provided that the chemical yields (amount of molecules formed per 100 eV of absorbed energy) of the reactive species in the system are known. In practice this is not the case but the ferrous sulphate dosimeter has been used as a standard for establishing absorbed dose to water by totally absorbing an electron beam with known energy and integrated charge in the solution.

Other detectors that make use of the indirect results that radicals have in the chemical environment are radiographic film and gel dosimeters.

3. **Energy dependence of dosimeters and beam quality**

Many detectors for ionising radiation show an energy dependence of one or more response characteristics. It is therefore important to characterise this energy dependence and in addition, to be able to uniquely specify the complex energy spectrum of radiotherapeutic beams in an appropriate way.

For example, air-filled ionisation chambers used in megavoltage beams show a variation in sensitivity of a few per cent. The requirement is simple enough – to find a parameter QI (quality index) which is reasonably easy to measure, and which can be used to provide an unambiguous calibration curve, when the calibration is plotted against QI.

For megavoltage photons, this parameter is the Tissue Phantom Ratio (TPR$_{20/10}$) obtained as the ratio of ionisation currents at depths 10 and 20 cm from the front face of a water phantom, for a fixed source to chamber distance, and a fixed 10 × 10 cm field at the chamber. This parameter is not strongly sensitive to the actual source to chamber distance, but 1m is the conventional distance to use. Actually the use of TPR$_{20/10}$ as a beam quality index leaves some small ambiguity in the calibration factor, which has been observed at NPL to be at most 0.5%, but TPR$_{20/10}$ takes account of over 80% of the variation with energy that is seen in the calibration factors of graphite-walled air-filled ion chambers.

For high-energy electrons, the beam quality parameter is essentially the range R$_{50}$ at which the dose has fallen to 50% of its peak value.

For medium energy X-rays, the calibration data are provided as a function of the half-value layer thickness (HVL) in aluminium.

**References**

Three textbooks cover radiation dosimetry for radiotherapy:


The most up-to-date survey of the calorimetry are the following proceedings:


Many countries including the UK maintain national protocols for dosimetry, but the recent IAEA code is currently the most up-to-date and review: