Topical Review

Review on the characteristics of radiation detectors for dosimetry and imaging

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Received 22 January 2013, revised 14 July 2014
Accepted for publication 23 July 2014
Published 17 September 2014

Abstract

The enormous advances in the understanding of human anatomy, physiology and pathology in recent decades have led to ever-improving methods of disease prevention, diagnosis and treatment. Many of these achievements have been enabled, at least in part, by advances in ionizing radiation detectors. Radiology has been transformed by the implementation of multi-slice CT and digital x-ray imaging systems, with silver halide films now largely obsolete for many applications. Nuclear medicine has benefited from more sensitive, faster and higher-resolution detectors delivering ever-higher SPECT and PET image quality. PET/MR systems have been enabled by the development of gamma ray detectors that can operate in high magnetic fields. These huge advances in imaging have enabled equally impressive steps forward in radiotherapy delivery accuracy, with 4DCT, PET and MRI routinely used in treatment planning and online image guidance provided by cone-beam CT.

The challenge of ensuring safe, accurate and precise delivery of highly complex radiation fields has also both driven and benefited from advances in radiation detectors. Detector systems have been developed for the measurement of electron, intensity-modulated and modulated arc x-ray, proton and ion beams, and around brachytherapy sources based on a very wide range of technologies. The types of measurement performed are equally wide, encompassing commissioning and quality assurance, reference dosimetry, in vivo dosimetry and personal and environmental monitoring.

In this article, we briefly introduce the general physical characteristics and properties that are commonly used to describe the behaviour and performance of both discrete and imaging detectors. The physical principles of operation of calorimeters; ionization and charge detectors; semiconductor, luminescent, scintillating and chemical detectors; and radiochromic and radiographic films are then reviewed and their principle applications discussed. Finally, a general
discussion of the application of detectors for x-ray nuclear medicine and ion beam imaging and dosimetry is presented.

Keywords: radiation detectors, radiotherapy, medical imaging, dosimetry, radiation protection

(Some figures may appear in colour only in the online journal)

Part I.–Basic Properties of Detectors

1. Introduction

Since the discovery of x-rays in 1895, ionizing radiation has played a crucially important role in medicine. Today a range of different particle types, from photons and electrons to protons and carbon ions, and a range of energies from 10’s of keV to 100’s of MeV are used in imaging or therapy. Methods for detecting this radiation are therefore essential, not only for the successful operation of diagnostic imaging equipment, where the radiation sensor is the heart of the machine, but also to ensure reproducible operation of all such equipment and for the personal safety of staff and patients. To ensure safety and consistency between machines, centres and countries, calibration of radiation detectors relative to primary standards is also essential.

Accurate radiation detection is not easy: absorbed dose to water can be measured using calorimetry, probably the most accurate direct physical measurement of energy deposited per unit mass, with an uncertainty of roughly 0.5% (contrast this with measurement of time or distance, where measurement with uncertainties of one part in 10^{10} are possible). However, calorimetry is impractical for most routine clinical uses, so radiation detectors are employed which use a wide range of different physical and chemical interactions to convert dose to a directly measurable quantity, such as electronic charge collected from air ionisation or colour change arising from changes in atomic electronic states. The range and complexity of the different physical and chemical effects that are routinely used for radiation detection make this subject both a delight and a challenge.

The field of radiation detection is huge, and an exhaustive systematic review would be impractical and beyond current space constraints. In this review we concentrate on three subject areas: in Part I the basic properties used to characterise the behaviour and performance of both discrete detectors and imaging systems are briefly introduced. The physical principles of operation of the major types of radiation detector associated with the medical field are then detailed. Finally the practical applications of detectors, either as an integral part of an imaging system, or as a discrete devices for dosimetry are discussed. An attempt has been made to cover the majority of detector types commonly used in the medical field and in radiation protection for medical workers. Some detectors, such as radiographic film has been almost completely replaced by digital imaging detectors or radiochromic film in many centres, but are included for historical importance. Other technologies, like the Faraday cup and proton radiography, are just beginning to see more widespread use with the growing implementation of proton therapy.

2. General properties of radiation detectors

There are three general modes of operation of radiation detectors (1) pulse mode, (2) current mode and (3) mean square voltage mode. The most commonly used mode for radiation detectors is the pulse mode, where the detector is designed to record each individual quantum of radiation that interacts within it. For very high event rates the pulse mode becomes impractical...
and even impossible. In this case subsequent events become too close to distinguish and this leads to either the current mode or mean square voltage mode.

### 2.1. Temporal properties

The detectors are usually operated in either current mode or pulse mode. If timing information of the incident particle is not relevant, then the output of the detector can be recorded by measuring the average dc-current that’s produced. This provides a measurement of the incident flux, when the charge liberated per particle is the same. If however, the free charge produced within the detector by the incident particle is proportional to its energy, then the detector measures the energy deposited and the detector is operating in current mode, which is commonly used in radiation dosimetry. In many radiation detector applications, the properties of a single particle or the exact number of interacting particles are an important characteristic that must be measured. In these cases the detector is operating in pulse mode, where the detector output current changes after the arrival of each particle or particle bunch. In pulse mode the detector and recording circuit produces an ac-current, with a time constant $\tau = RC$. In the case of particle counting, the maximum frequency of the detector is more relevant than determining the exact charge. For this RC-circuits with $\tau \ll t_c$ ($t_c$ is collection time) are used, where the maximum frequency is then limited by the time $t_c$ it takes for charge carriers to leave the active area. A measure commonly used to characterize all these effects is the dead time, which corresponds to the time interval after the arrival of the particle during which the detector is unable to process another subsequent particle.

### 2.2. Resolution and statistical characteristics

An important property of any detector is its ability to resolve a certain quantity $X$. We define the resolution of a detector as the standard deviation $\sigma$ or the full width half-maximum, $\text{FWHM}$, of the distribution $D(X)$ of the measured quantity $X$, for a monochromatic input distribution $\delta(X - \bar{X})$, where the mean value of the measured quantity is $\bar{X} = \int X \cdot D(X) dX$ and the variance is $\sigma^2 = \int (X - \bar{X})^2 dX$. The detector relative resolution is defined as the dimensionless ratio $R = \sigma / \bar{X}$ (Bushberg et al 2002, Del Guerra et al 2010, Webb and Flower 2012).

### 2.3. Detector linearity

In general for any quantity $X$, the measured quantity is given by $\bar{X} = f(X)$. If this relationship is of the type $\bar{X} = cX$, then the detector is linear in respect of that quantity. Thus any linear increase in the quantity $X$, is followed by a linear increase in $\bar{X}$. There is usually a region of linearity outside which the response is distorted due to changes in efficiency, saturation, etc. If for example the incident particle energy $E_o$ increases, so does the mean value $\bar{E}$ of the measured distribution increase.

### 2.4. Detector efficiency

For an isotropic direction of emission, the detection efficiency, $\varepsilon$, is defined as the probability of detecting the emitted particle. The detection efficiency is composed of two factors: geometric efficiency, and intrinsic efficiency.

#### 2.4.1. Geometric or solid angle efficiency. The geometric efficiency represents the solid angle from which the particles can be collected and is defined as $\Omega / 4\pi$. Thus for example a
perpendicular detector surface with area $A$, at a distance $R$, away from the point source gives

$$\Omega = \frac{A}{R^2}, \text{if } A < R^2.$$  

2.4.2. Intrinsic efficiency, $\varepsilon_i$. The intrinsic efficiency of a detector $\varepsilon_i$, is the fraction of impinging particles that produced a measureable pulse in the detector, where $\varepsilon_i = 1$ represents a detector that converts all the incident particles into a measureable pulse.

The total detection efficiency is then defined as

$$\varepsilon = \frac{\Omega}{4\pi \varepsilon_i}.$$  

(1)

2.5. Response time and dead time

The detector intrinsic efficiency can also be influenced by the amount of time the detector takes to deal with the present event, and thus cannot deal with any new events. This phenomenon is called dead time and is usually an issue for high count rates, where not all the events are processed. There are two types of behaviour: (a) non-paralyzable and (b) paralyzable. For the non-paralyzable case, each recorded event is followed by a time interval $\tau$ needed to process the true event, during which no other true event is accepted. If the rate of recorded events is $R'$ and true events is $R$ then the fraction of time the detector is dead is $R' \tau$. The rate of true events lost by this dead time is $R R' \tau$, and since this is equal to $R - R'$, we obtain the true event rate as

$$R' = R \frac{1}{1 - R' \tau}.$$  

If the true event happens during the dead time period of $\tau$, it is desirable to start a new cycle. This is called paralyzable mode of operation. In this case the dead time intervals are of variable length and the rate of recorded events $R'$ as the rate at which time intervals larger than $\tau$ occur in the true sequence. Using Poisson statistics, the probability of obtaining an interval larger than $\tau$ is $e^{-R \tau}$, and thus the rate at which this occurs is $R' = R e^{-R \tau}$. For low events rates i.e. $R < < 1/\tau$, both equations give approximately the same solution $R' \approx R (1 - R \tau)$.

2.6. Quantum efficiency, QE

Quantum efficiency is the probability that the primary ionization process takes place after the particle enters the detector, for example that a photon absorbs and emits a photo-electron from a metal surface or that the first charge-pair is created within a gas. In photon detectors, QE is a function of the photon energy and detector effective atomic number.

2.7. Gain

In general, gain is a measure of the ability of an electronic circuit to increase the amplitude or power of a signal. Most detectors are amplifiers, the amplification process being the creation of secondary charge-pairs after the primary interaction has occurred. Gain is the final number of charge-pairs per one primary interaction. The values of gain can be very large, for example in a photomultiplier tube, PMT, the gain can exceed $10^6$ and in a Geiger-Muller counters around $10^{11}$.

3. General properties of imaging systems

3.1. Definition of image quality

The evaluation of imaging detector systems requires the use of ‘quantitative parameters’, which assess the image quality (Bushberg et al 2002, Prince and Links 2005). Image transfer
theory has been widely used in the field to assess signal and noise transfer characteristics. In its most generic form, the input signal $F(x_1, y_1)$ is transformed by the ‘imaging detector’ into an output signal $G(x, y)$, where the system operator $S$ represents the detector effect on the input signal:

$$G(x, y) = S[F(x_1, y_1)]$$

(2)

The system operator $S$ characterizes the imaging device used in the process to ‘visualize’ the signal $F(x_1, y_1)$. In the case of a point source (represented by a Dirac delta function), the output signal $G(x, y) = h(x, y; x_1, y_1)$ is known as the point spread function (PSF). The measurement of the PSF can be very cumbersome and a complex procedure, and therefore line spread function (LSF) is generally measured.

3.2. MTF, NPS and DQE imaging parameters

The modulation transfer function (MTF) provides the ratio of input to output amplitude for a sinusoidal input signal. An interpretation of the MTF is that it indicates how the imaging device alters the input spatial frequency spectrum, which is well known because of the input sinusoidal signal. The noise power spectrum (NPS) or the Wigner spectrum of fluctuations of a random-ergodic process is used to assess the structure and spatial correlations within noise (Prince and Links 2005). The NPS uses the well-known auto-correlation function to evaluate correlations between various points in space. A well-known application of NPS is to screen-film combination systems (as discussed in Del Guerra 2004). The detective quantum efficiency (DQE) defines the transfer of signal-to-noise ratio (SNR) that occurs when converting an input signal into an image with the use a measuring device.

$$DQE = \frac{SNR_{out}^2}{SNR_{in}^2}$$

(3)

The DQE represents the efficiency of an x-ray counter, where if, for example, we assume $N$ incident x-ray per unit area ($\sigma^2 = N$, using Poisson statistics) $SNR_{in}^2 = N$. If there are $N'$ exiting x-rays, then $SNR_{out}^2 = N'$, and the $DQE = \frac{N'}{N}$, representing the output to input number of photons.

3.3. Effective sampling aperture

An imaging device converts a point into a ‘blurred point’, smearing the edges and thus equivalently degrading any image. The effective sampling aperture represents the average blur size of the ‘smeared point’. The effective sampling aperture approach can be used in a cascade imaging system to obtain the overall effective aperture by the sum of the apertures of each component. Therefore, by imaging an object of equivalent sampling area $a_0$ with a system of equivalent sampling area $a_s$, the resulting object will have a resulting equivalent sampling aperture of $a_{final} = a_0 + a_s$.

4. Types of detector

4.1. Calorimetry

Calorimetry is perhaps the most direct physical measurement of absorbed dose, and works on the assumption that all energy imparted by ionizing radiation ultimately leads to temperature rise:

$$D_m = c_m \Delta T_m$$

(4)
where \( c_m \) is the specific heat capacity of the absorbing medium and \( \Delta T_m \) is the temperature rise resulting from the absorbed dose \( D_m \). A clear advantage of calorimetry over other measurement techniques is the fact that calorimeter calibration can be carried out based entirely on quantities that do not require a reference ionizing radiation field (e.g. electrical power and temperature). However, measurement of the extremely small temperature rises required (typically \( \mu K \)) is challenging, requiring highly stable measurement conditions. Calorimetry is therefore not practical for routine measurement in a clinical environment, but is widely used in standards laboratories, where the direct physical connection between absorbed dose and temperature rise make it an obvious choice for a primary standard. Primary absorbed dose standards for dosimetry of high-energy photon beams are reviewed in Seuntjens and Duane (2009), and include ionization chambers, ferrous sulphate solutions (Fricke gel) and calorimetry. Two types of calorimeter operate in standards laboratories based on either graphite or water (Guerra et al 1996, Palmans et al 2004, Baumgartner et al 2011).

4.1.1. Water calorimeters. Water calorimeters typically measure the temperature rise inside a sealed small volume of highly purified still water and rely on the fact that the thermal diffusivity of water is low, so the temperature distribution remains in place long enough to allow accurate measurement. The water is kept at a stable 4°C to avoid convective heat loss, and loss by conduction is typically carefully modelled. Absolute temperature rise is measured using a thermistor

\[
R = R_0 \exp \left( \beta \left( \frac{1}{T} - \frac{1}{T_0} \right) \right),
\]

where \( R_0 \) is the thermistor bead resistance at the reference temperature \( T_0 \). The thermistor sensitivity \( S = \beta T^{-2} \) is determined by prior calibration outside of the calorimeter, allowing a temperature rise to be determined from the change in resistance \( \Delta T = S^{-1} \left( \frac{\Delta R}{R} \right) \). The heat defect of a calorimeter quantifies the difference between energy absorbed and energy appearing as a temperature difference. For dose measurements in a water calorimeter permanently sealed in glass, the heat defect uncertainty has been shown to be about 0.3%, and the overall accuracy for \(^{60}\text{Co} \) photon beams to be between 0.2% and 0.4% (Seuntjens and Duane 2009).

4.1.2. Graphite calorimeters. Graphite calorimeters aim to measure the absorbed dose to graphite under standard conditions, and then convert this to absorbed dose to water. Heat flow is rapid in graphite, so the inner core of the calorimeter, typically a 20 mm diameter and 1 mm thick disc, is separated from the surrounding jacket with an evacuated core to reduce heat transport (see figure 1). The rapid heat flow makes it possible to use electrical heating as an integral part of the absorbed dose measurement. The change in total energy in the core can be expressed by

\[
\Delta E_{\text{tot, thermal}} = E_{\text{rad}} + \Delta E_{\text{elec}} + \Delta E_{\text{transfer}}
\]

showing the separate contributions of energy deposition by ionising radiation, electrical heating and energy loss by heat transfer. The change in core temperature \( \Delta T_{\text{core}} \) is related to \( \Delta E_{\text{tot, thermal}} \) by the mass and specific heat capacity of the core \( m_{\text{core}}c_p \); thus

\[
\Delta E_{\text{tot, thermal}} = m_{\text{core}}c_p \Delta T_{\text{core}}.
\]
Multiplying the change in absolute temperature by the specific heat capacity gives the absorbed dose to graphite. In isothermal mode, a measurement by substitution is made: a constant temperature is maintained by adjusting the electrical heating power with the radiation beam both on and off. The difference in electrical energy required to achieve a constant temperature is equal to the energy deposited by the radiation beam. The temperature changes that need to be measured in the core are of the order of μK, so isolation from room temperature changes is essential. This is often achieved in graphite calorimeters by electrically heating the evacuated core to a higher temperature than the surroundings to damp out changes in room temperature. Dose to water can be measured in a graphite calorimeter to an uncertainty of between 0.41% and 0.46% for high-energy photon beams (Seuntjens and Duane 2009).

4.2. Ionization chambers and charge detectors

In ionization chambers the energy deposited by the ionizing radiation produces ion pairs. If the medium within the chamber is gas, charged electrodes are used to collect the ion pairs. A voltage is applied across the electrodes in order to guarantee the motion of the ion pairs towards an electrode. The voltage is high enough to guarantee that all ion pairs produced by the incident radiation are collected and low enough to avoid any secondary ion pair production by the motion of the primary ion pairs. (Voltages are usually of the order a few hundred volts.)

Pulse-type ionization chambers are designed to produce voltage pulses by rapid collection of electrons. Current-type ionization chambers are designed to collect electrons by the anode, producing a direct current that may be amplified and measured with a conventional dc meter. Small currents may be measured with the aid of electrometers, which convert the signal from the ionization chamber into an alternating current that can be amplified with an amplifier.

4.2.1. Cylindrical and parallel plate ionization chambers. Cylindrical and parallel plate ionization chambers have electrodes of low atomic number material, usually aluminium, carbon, or conductive plastic, enclosing an air volume. The voltage between the electrodes is large enough to collect most of the charges within the sensitive volume, usually between 95% and 100%, and small enough that operation is below the threshold for the proportional amplification region. A guard electrode protects the signal electrode from leakage currents and
ionization signals arising from regions with low or distorted electric fields. A fully guarded chamber has the guard electrode protruding into the air volume, as opposed to being flush with or behind the insulator in the detector housing. Examples of cylindrical and parallel plate ionization chamber configurations are shown in figure 2. Other examples of the parallel plate chamber include the free air ionization chamber and extrapolation chamber (Khan et al 2010).

The mean dose deposited, $D$, within the air of volume, $v$, is related to the charge collected, $Q$, by

$$Q = \frac{P_{\text{ion}} e \rho v D}{W},$$

where $P_{\text{ion}}$ is the correction factor for ion recombination, $e$ is the electron charge, $\rho$ is the air density, and $W$ is the mean energy to produce an electron-ion pair in air.

As electrons and ions drift toward the anode and cathode, respectively, they induce mirror charges in the electrodes. The amount of charge induced, $dQ$, by a single electron or ion traversing a distance $dl$ in electric field $E$ can be calculated by relating the change in energy in a capacitor to the work done on the charge

$$\frac{1}{2} V_0^2 dQ = qEdl,$$

where $V_0$ is the potential difference between the electrodes and $q$ is the charge of the electron or ion. The induced mirror charges on the signal electrode are detected by an electrometer.

Recombination occurs whenever electrons and ions liberated by the radiation field combine within the charge collection volume. It reduces the amount of charge collected depending on the electric field and distance travelled by the charged particles between production and recombination. Recombination can be either intra- or inter-track. The former is called initial or columnar recombination, and is due to electrons and ions combining within a single track. The latter is called volume or general recombination, and is due to electrons and ions from different tracks combining. Increasing $V_0$ reduces the effect of recombination by increasing the force pulling the electron-ion pairs apart at the point of production and increasing the drift velocity.

Correction factors for recombination can be determined by varying $V_0$ and plotting the collected charge versus $1/V_0$ or $1/V_0^2$, whichever produces a relationship closest to linear. The choice will depend on the pulse structure of the beam, voltage, gap size, and fill gas (see figure 3). Recombination can be adversely affected by electronegative contaminants since free electrons can attach themselves

Figure 2. Schematics for (a) cylindrical and (b) parallel plate ionization chambers. C552 is an air-equivalent conductive plastic. [Figure (a) is reproduced from Erazo and Lallena (2013)] [Figure (b) is reproduced from Mattson et al (1981) both with permission from Elsevier.]
to these contaminants and produce more slowly moving ions. Recombination effects should be checked during operation in high humidity or whenever aerosols are used nearby.

4.2.2. Proportional counters. The usually very small signals produced by an ionization chamber must be amplified before they are measured, and thus the signal is very susceptible to electronic noise coming from the electrical circuit used for amplification. An alternative method of signal amplification is increasing the electrode voltage. If the voltage between the electrodes is raised beyond a certain value, electrons liberated by the incident radiation are accelerated sufficiently to produce additional ionization. Most of the additional ionization occurs near the anode of the chamber. This process is known as the gas amplification process, and this region is commonly designated as the proportional region for gas chambers. The amount of charge collected by the electrodes is proportional to the voltage applied across the electrodes. Therefore, the voltage must be very carefully regulated because the amplification factor is affected greatly by small changes in the voltage.

4.2.3. Geiger-mueller counters. If the voltage difference between the electrodes exceeds a certain value, the interaction of ionizing radiation within the chamber now produces an avalanche of ionization, which represents almost complete ionization of the counting gas in the vicinity of the anode. As a consequence of this avalanche process, the number of ion pairs collected by the electrodes is independent of the amount of ionization produced directly by the impinging radiation. This region of operation is known as the Geiger-Muller region. For detectors operating in the Geiger-Muller region, the amplification factor is 10⁶ to 10⁸.

4.2.4. Faraday cup. Faraday cups measure the current or charge carried by a beam of particles and consist of a metallic beam stop connected by wire to an electrometer (figure 4(a)). The dimensions of the beam stop are wider than the lateral extent and longer than the distance required to stop the entire beam, respectively. Faraday cup measurements have the advantage that no beam quality or ion recombination correction factors are applied. They are designed carefully, however, to reduce the influence of scattered charges on the electrometer signal (figure 4(b)).

Traditional Faraday cups (figure 5) (Verhay et al 1979, Ziegler et al 1996, Lin et al 2009) are complex devices designed to reduce the effects of scattered charged particles. They have
the following general features: the region before the cup is enclosed in vacuum with a thin window to reduce the number of scattered charges toward the collecting cup; the shape of the cup is highly concave to reduce the number of scattered charges backscattered from the cup surface; electromagnetic fields are applied to deflect and trap the trajectories of low-energy charged particles; and a guard ring is located close to the vacuum window.

Vacuumless Faraday cups (also called ‘Poor man’s Faraday cups’) (Gottschalk 2011) have a smaller and simpler design compared to the traditional cups. They consist of a metallic beam stop wrapped in a thin layer of insulating plastic, followed by a thin conducting layer connected to ground. The insulating layer traps scattered charges and restricts the net flow of charges from the grounded conducting layer to the cup after a few seconds of beam. The signal from vacuumless cups agree with traditional cups within 1–5% (Cascio and Gottschalk 2009).

4.3. Semiconductor detectors

Semiconductor detectors exhibit many desirable properties, including:

- linear response with deposited energy,
- negligible absorption of energy in the entrance window of the detector,
- excellent energy resolution,
- formation of pulses with fast rise times and
- small detector sizes.
The mechanism of signal formation in a semiconductor very much resembles that of an ionization chamber. Ionization produced within the sensitive volume of the detector is converted to a voltage pulse, which is then amplified and counted. In this section the use of semiconductor devices as both radiation detectors and imaging sensors is reviewed. For dosimetry, detectors are typically used as discrete single elements, although 1D and 2D detector arrays have been developed for specific applications, such as the verification of modulated beam in radiotherapy. For imaging, active-matrix arrays working in either direct or indirect detection mode are discussed. The advantages of semiconductor detectors for measuring ionizing radiation have long been known, with detectors being very small with excellent energy resolution, fast response and linear with energy (Parker 1970). In semiconductor detectors an electric field exists across a volume of material with a low free-carrier concentration (a depletion zone). When an ionizing particle releases energy $E$ in the detector, $N$ electron-hole pairs are formed:

$$N = \frac{E}{w},$$

where $w$ is the mean energy required to produce an electron-hole pair. These charge carriers are then swept apart in the electric field and the charge is collected at electrodes on either side of the material. The amount of charge collected is directly proportional to the energy lost by the nuclear particle in the detector, allowing straightforward and instantaneous readout, without the need for the intermediate optical stages found in scintillation-based devices. Semiconductor detectors applied primarily for scintillator readout are discussed in section 4.4.6.

4.3.1. Silicon diodes. For silicon, which is in group IV, doping with a Group V element such as phosphorus donates valence electrons producing a conductive n-type semiconductor. Conversely, doping with a group III element creates ‘holes’ in the silicon lattice, resulting in a p-type semiconductor. If a p–n junction is created where both semiconductor types meet, charge carriers flow away from the junction, generating a carrier-free region typically a few 10s of microns wide, with an intrinsic potential of 0.7V; this depletion layer acts as the sensitive volume of the detector.

For a typical p-type diode, a p-type bulk material (Si doped with a group III material) is counter-doped using a thin layer of an n-type (group V) material to form the p-n junction. Silicon diodes are typically constructed as either ‘p-type’ (a p-type bulk material coated with a thin layer of an n-type material) or ‘n-type’ (vice versa). A proportion of carriers are trapped by defects in the silicon and, since holes are trapped more easily than electrons, this effect is larger in n-type devices. Recombination is also affected by dose rate; at high dose rates the recombination centres can become saturated and lead to a proportionally higher signal. Again, this effect is more pronounced in n-type devices, which have lower doping levels and therefore lower numbers of recombination centres. Radiation damage to the Si crystal lattice also forms recombination centres, thus reducing the detected signal gradually with use. Typical values for sensitivity loss as a function of radiation damage are about 1% per kGy, although this depends on the beam energy and previous irradiation history. Again, n-type devices are affected by this more than p-type, making p-type diodes preferable for dosimetry (Essers and Mijnheer 1999). Sensitivity tends to fall off faster during initial exposure, so commercially available diode detectors are often available pre-irradiated to minimize this effect. Regular recalibration of any diode detector is, however, necessary if it is to be used for accurate, clinical dosimetry.

The value of $w$ in silicon is around 3 eV (10 times less than that in air), and the density of Si is 1,000 times higher than air, which gives a relative efficiency per unit volume for silicon diodes $10^4$ times greater than an air ionization chamber. Si diodes can therefore be made with much smaller sensitive volumes. The shape of the sensitive volume and surrounding package...
cause a significant angular dependence in the response, with up to 15% variation with angles being common. Care must therefore be taken in checking that the same orientations are used for calibration and measurement.

The materials the diode is comprised of are not tissue equivalent, so significant energy dependence is seen, with lower energy radiation giving a disproportionately high signal. Compensated diodes are produced where the diode encapsulation (stainless steel and epoxy resin) or build-up cap is designed to preferentially absorb the lower-energy radiation to give a more water-equivalent behaviour. These ensure that photon scatter conditions are as similar as possible to those at the D\text{max} in water to minimise the correction factors, although large build-up caps can cause significant perturbations to the radiation field and are more difficult to attach to the patient’s skin. The ease of use, small size and relatively high accuracy achievable with diodes makes them very attractive for both \textit{in vivo} dosimetry, providing an independent real-time measurement of dose delivered to a patient, and for relative measurements of dose distributions for quality assurance. Meijer \textit{et al} (2001) demonstrate the use of \textit{in vivo} diode dosimetry to ensure that dose delivery in a randomized clinical trial is guaranteed to within 2.5%, and Kadesjö (2011) describes the \textit{in vivo} diode dosimetry for IMRT capable of showing clinically relevant dose deviations >5%. Guidelines for \textit{in vivo} dosimetry are published by ESTRO (Huyskens 2001), AAPM Report 87 (2005) and Mijnheer \textit{et al} (2013), and the use of diodes for small-field dosimetry is described in the IPEM guide by Aspradakis (2010). For clinical \textit{in vivo} measurements, regular calibration against an air ionization chamber under standard reference conditions is recommended. The diode response should be calibrated as a function of the beam energy, with suitable correction factors measured for beam energy, field size, wedges, angle of incidence and source-to-surface distance. In general it is good practice to make the calibration conditions as close to those used in clinical practice as possible. For example, in total body irradiation (TBI), where large source-to-surface distances and low dose rates are used, diodes should be calibrated under similar conditions. Diode response is also affected by temperature, with fluctuations in response of 0.1% / °C seen, so diodes used in contact with skin should be given 2–3 min to reach thermal equilibrium and suitable corrections should be applied if they were calibrated at room temperature.

### 4.3.2. Diamond detectors.

An alternative semiconductor to the silicon diode is bulk diamond (Krammer \textit{et al} 1998). Diamond has a much larger band gap than silicon (5.54 eV for natural diamond, compared to 1.12 eV for silicon), which means there are very few free charge carriers present at room temperature, leading to a very high resistivity and correspondingly low leakage currents (see table 1). The very low density of charge carriers means that a diamond detector does not therefore have to be depleted, so no diode structure is necessary. The relatively high energy required to create an electron/hole pair in diamond compared to silicon is partially made up for by the good electron and hole mobilities, low dielectric constant, high saturation velocity and very good radiation hardness.

<table>
<thead>
<tr>
<th>Property</th>
<th>Silicon</th>
<th>Diamond</th>
</tr>
</thead>
<tbody>
<tr>
<td>Band gap [eV]</td>
<td>1.12</td>
<td>5.47 (5.54)†</td>
</tr>
<tr>
<td>Resistivity [$\Omega$ cm]</td>
<td>$2.3 \times 10^5$</td>
<td>$&gt;10^{11}$</td>
</tr>
<tr>
<td>Energy to form e/h pair [eV]</td>
<td>3.6</td>
<td>13</td>
</tr>
<tr>
<td>Electron mobility [cm$^2$ V$^{-1}$ s$^{-1}$]</td>
<td>1350</td>
<td>1800</td>
</tr>
<tr>
<td>Hole mobility [cm$^2$ V$^{-1}$ s$^{-1}$]</td>
<td>480</td>
<td>1200</td>
</tr>
</tbody>
</table>
The average distance that electron/hole pairs drift apart before they are trapped is called the collection distance. In diamond, this is typically smaller than the detector thickness; the collection distance is given by

\[ d_c = \mu \tau E, \]  

(11)

where \( \mu \) is the mobility of the charge carriers, \( \tau \) is the carrier mean lifetime and \( E \) is the applied electric field. Charge traps in diamond are filled—and therefore neutralized—by the initial irradiation, causing the charge collection distance to increase rapidly at low dose, then reaching a plateau that can be as low as 50% of its pre-irradiated value. This saturation state is referred to as the ‘pumped state’, and can be stable for several months if the detector is kept at room temperature and dark in CVD diamond. To avoid the possibility of serious dosimetry errors arising from lack of equilibrium, manufacturers of diamond detectors for radiotherapy recommend exposure of the detector to pre-measurement dose before each usage.

Another effect that must be taken into account is a decreasing response with increasing dose rate, which arises because of the very short electron/hole recombination time. If the detector is to be calibrated at one dose rate and then used at another, a correction should be applied. It has been shown that the following simple empirical formula can be used to relate the detector reading \( M \) to the dose per pulse (measured in water) \( D_w \):

\[ M = \alpha D_w^\Delta, \]  

(12)

where \( \alpha \) is a constant and the correction factor \( \Delta \) has a value close to 1.00. These dose-rate correction factors have been shown to be energy independent.

Detectors can be made from natural or synthetic (CVD) diamond (Lansley et al 2010). A disadvantage of natural diamond detectors is that careful selection and individual characterization of each crystal is required to ensure suitably high-quality material properties, which makes the detectors very expensive. However, the detectors have the advantage of a very small measurement volume (a few cubic mm), which gives excellent spatial resolution and almost no directional dependence or temperature dependence. Diamond is also quasi-water equivalent, at least in terms of atomic number, so it is particularly attractive for measurements in radiation fields where electron equilibrium cannot be assumed, such as in small fields. Diamond detectors have been shown to be independent of energy for photon beams from 4 to 25 MV, and electron beams from 5 to 20 MeV (Fidanzio et al 2000).

### 4.3.3. MOSFETs

Metal oxide semiconductor field effect transistors (MOSFETs) were developed for radiation monitoring in earth-orbiting satellites in the late 1970s (Rosenfeld 2002). A MOSFET device is comprised of a semiconductor silicon substrate separated from a metal gate by an insulating oxide layer. When a negative bias voltage is applied to the gate, a positive ‘mirror’ charge builds up in the silicon, allowing a current to pass. The gate voltage required to allow conduction through the MOSFET is termed the threshold voltage \( V_{\text{th}} \).

When a MOSFET is irradiated, electron-hole pairs are generated within the oxide layer; the electrons move rapidly out of the gate electrode, while the holes (whose mobility at room temperature is ~4 orders of magnitude lower than that of the electrons) move in a stochastic fashion towards the Si/SiO2 interface, where they become trapped in long-term sites, which can persist for years. (See figure 6) This build-up causes a negative drift in the threshold voltage.

Irradiation of MOSFETs can be active or passive (with or without a gate voltage applied, respectively), although application of the gate voltage reduces electron-hole pair recombination, and therefore makes active devices more sensitive and linear. In active mode, the dose-response is linear over a wide range, depending on the oxide thickness and the applied bias.
The accumulated charge can be annealed under ~150°C, making the MOSFET reusable, although recalibration will be required following annealing (Rosenfeld 2002).

Non-water equivalence causes some energy dependence, with variations in calibration factor (converting change is $V_{th}$ in mV to absorbed dose in Gy) varying by approximately 3%, for electron energies from 5 to 19 MeV and photon energies from 6 to 18 MV for some types of device (Ramani et al. 1997). MOSFETs also demonstrate ‘creep-up’ behaviour, where the threshold voltage increases with consecutive reading due to charge injection by the readout circuitry; this can be prevented by leaving >1 min between readouts. The angular dependence of the readout depends on device fabrication, and is generally small (within 1% at 0°, 45° and 90°); however, a discrepancy of up to 12% can be seen where the device is irradiated at 180° (Lonsdale 2012). Temperature dependency of 2–3% in the range 20–37°C has been observed, although since this is generally within the reproducibility of the device, it is often neglected for clinical measurements. A method of reducing the temperature dependency is to construct a two-channel device, where two MOSFETs are fabricated on the same silicon substrate and held at different bias voltages. The difference in the $V_{th}$ shift between the two channels is then read out. A different form of two-channel readout has also been reported, with two discrete MOSFETs being mounted in different orientations to reduce angular dependence (Hardcastle et al. 2010).

The small size of the dosimetric volume, ability to permanently store accumulated dose, dose-rate independence and ease of readout make them ideal for many applications in radiotherapy, particularly for in vivo dosimetry. Applications of MOSFETs have been described for electron beams, with overall deviations between MOSFET measurements and prescribed dose of ~3% (1 SD) (Gurp et al. 2006); IMRT dose verification, comparing treatment planning system calculation to phantom measurement within 5% (Chuang et al. 2002); and brachytherapy with $^{192}$Ir (Kinjikar 2006) and for intraoperative (Consorti et al. 2005), intra-cavity (Hardcastle 2010) and implantable (Black et al. 2005) use1. In proton therapy, the use of MOSFETs is currently limited by a large LET dependence; the Bragg peak height estimated using MOSFET detectors was shown to be almost 40% lower than with an ionization chamber when normalized to the proximal plateau (Kohno et al. 2006).

4.3.4. Flat panel detectors. To replace radiographic film in medical imaging, a detector system is required that is ~40 cm wide, and has a spatial resolution of 100–150 µm and a noise level of 1–5 x-ray quanta per pixel, a dynamic range exceeding 1:1000, and a sensitivity that allows quantum-noise-limited operation at doses of order 1 $\mu$Gy (Moy 2000, Neitzel 2005). There are essentially two types of flat-panel imager that seek to meet this design brief: indirect

1 Not all of the systems described are currently commercially available.
detection systems in which the incident x-rays interact principally with a phosphor screen, such as Gd\textsubscript{2}O\textsubscript{2}S:Tb, and the visible light emitted by the phosphor is then detected by a two-dimensional array of photodiodes (Yaffe and Rowlands 1997); and direct detection systems in which a photoconductor such as amorphous selenium (a-Se) is used as the principal detecting element and absorbs the x-ray photons, converting the energy deposited directly to electronic charge (Darambara 2006, Kasap et al 2011, see figure 7). Both types of detector are designed around an active matrix comprised of a 2D array of pixels containing a hydrogenated amorphous silicon (a-Si:H) thin-film transistor (TFT). The pixel has three functions: to convert x-rays to an electronic charge, temporarily store the charge and transfer charges to a readout amplifier.

4.3.5. Indirect detection systems. Many x-ray imaging systems use a scintillator or phosphor screen for the initial x-ray conversion. A substantial portion of the initial x-ray energy is typically transferred to the photoelectron produced in screen, and this energy is typically much larger than the bandgap in the crystal. Large numbers of optical photons are therefore typically produced per individual x-ray interaction, termed quantum amplification or conversion gain $g$. For example, in Gd\textsubscript{2}O\textsubscript{2}S:Tb, the energy carried by a 60 keV x-ray photon is equivalent to about 25,000 quanta of green light ($E_g = 2.4$ eV). However, the conversion efficiency is about 15%, requiring $\sim 13$ eV per optical photon. The overall conversion gain is therefore $4,500$ visible quanta per x-ray photon (Yaffe and Rowlands 1997). The efficiency of the conversion gain $g$ is stochastic, which introduces a standard deviation $\sigma_g$ that depends on the incident x-ray energy, adding noise to the image that is related to the shape of the distribution of $g$ (called the ‘Swank factor’).

Phosphor screens are typically produced by binding 5–10 $\mu$m phosphor particles in a transparent plastic. As visible light propagates through the phosphor screen, it is also scattered and absorbed, leading to a trade-off between detection quantum efficiency and both spatial resolution (due to dispersion) and noise (due to signal loss). One way of improving the imaging characteristics of thick screens is to use a vapour-grown CsI:Tl with 5 $\mu$m diameter columnar crystals, which channels the optical photons along the crystals. Screen thicknesses of 500–600 $\mu$m are typically used, grown or mounted directly above the a-Si:H layer (Cowen et al 2008). Photodiodes are fabricated on the a-Si substrate, and are typically a 0.5–0.8 $\mu$m thick n-i-p design (see figure 7) with quantum efficiencies of up to 85%, since the 550 nm spectral output of the commonly used Gd\textsubscript{2}O\textsubscript{2}S:Tb and CsI:Tl phosphor screens are well-matched to the response of a-Si:H.

4.3.6. Direct detection systems. An excellent review of the principles of operation and current status of direct detection flat panel imagers was presented by Kasap (2011). Briefly, the active matrix array is coated with a photoconductor material, such as stabilized amorphous selenium, and a surface electrode is added to supply a bias voltage. Charge carriers released by x-ray interactions drift along the electric field, generated by the bias voltage, and are stored as charge in a storage capacitor within the pixel. The amount of charge collected is determined by the pixel integration time and the number of electron-hole pairs created, which in turn is related to the x-ray flux and energy. The lower pixel electrode does not cover the entire pixel...
area, since some space is needed for the switching TFT, for example, reducing the ‘fill factor’ to 75–85%. However, in a direct detection system, the electric field bends towards the pixel electrode on the upper surface of the photoconductor covering this ‘dead zone’, effectively increasing the fill factor to nearly 100%. A thin blocking layer is added between the upper electrode and the photoconductor to prevent charge injection from the bias supply and reduce dark current.

The quantum efficiency of a direct detection system $A_Q$ is simply given by fraction of photons attenuated by the detector

$$A_Q = 1 - \exp(-\alpha t)$$

for a detector with photoconductor thickness $t$ and an effective linear attenuation coefficient $\alpha$ for the applied energy spectrum and detector atomic composition and density. For a-Se at 20keV (within the range used for mammography), a typical 200µm thick a-Se layer has an $A_Q$ of 98.3%. For narrow band-gap semiconductors such as a-Se the ‘classical’ formula that relates the energy required to form an electron-hole pair $W_{\pm}$ is not simply proportional to the band gap, but depends on the applied electric field and the x-ray energy:

$$W_{\pm} = W_{\pm}^0 + B F$$

where $F$ is the applied field and $B$ is a function that depends weakly on x-ray energy. For the 20–40keV range, $W_{\pm}^0 \approx 6$ eV and $B \approx 4.4 \times 10^2$ eV·V·µm$^{-1}$, which gives a value of $W_{\pm}$ at $F = 10$V·µm$^{-1}$ of about 10 eV (Kasap 2011). Once charges have been produced, they need to be collected before recombination. For an applied field $F$, assuming a mean carrier lifetime of $\tau$ and a drift mobility of $\mu$, the mean distance travelled before recombination is $\mu F \tau$, which is typically much larger than the 200µm photoconductor thickness in currently produced systems.

### 4.3.7 Alternative detector materials

Although the majority of current commercially available direct detection systems are a-Se based, other photoconductors have been studied, including CdZnTe, PbI$_2$ and HgI$_2$, which have high atomic numbers and can be deposited over wide areas in polycrystalline form. Early CdTe detectors were limited by low charge collection efficiency and poor ohmic contacts, which limited their application to medical imaging. By alloying CdTe with Zn, sensors with a wide band gap and high mass density could be produced that also exhibited low leakage current, low noise (Scheiber and Giakos 2001) and good energy resolution. Single detector CdZnTe inter-operative probes are currently commercially available, and small-area imaging detectors are available in a research environment. As technology advances to produce affordable large-area arrays, the physical properties of CdZnTe make this a very attractive technology for hybrid medical-imaging detectors (PET/CT, SPECT/CT, PET/MR); there is specific interest in CdZnTe and CdTe due to their high spatial and energy resolutions over a wide range of energies (Guerra et al 2008), see Table 2.

### Table 2. Material properties of a range of semiconductor detector materials (Darambara 2006).

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>a-Se</th>
<th>HgI$_2$</th>
<th>PbI$_2$</th>
<th>CdZnTe</th>
</tr>
</thead>
<tbody>
<tr>
<td>Atomic number</td>
<td>34</td>
<td>80, 53</td>
<td>82, 53</td>
<td>48, 30, 52</td>
</tr>
<tr>
<td>Relative density</td>
<td>4.3</td>
<td>6.4</td>
<td>6.2</td>
<td>6.2</td>
</tr>
<tr>
<td>Bandgap [eV]</td>
<td>2.2</td>
<td>2.1</td>
<td>2.3</td>
<td>1.5</td>
</tr>
<tr>
<td>Pair creation $W_{\pm}$ [eV]</td>
<td>50</td>
<td>5.5</td>
<td>6.6</td>
<td>4.5</td>
</tr>
<tr>
<td>Mobility lifetime $\mu \tau$ [cm$^2$/V]</td>
<td>$10^{-6}$–$10^{-5}$</td>
<td>$10^{-5}$</td>
<td>$10^{-5}$–$10^{-6}$</td>
<td>$10^{-5}$</td>
</tr>
</tbody>
</table>
Polycrystalline lead iodide (PbI$_2$) has been demonstrated to give a ~15-fold increase in signal compared to a Gd$_2$O$_2$S phosphor screen when deposited as a 100µm thick photoconductive layer over a-Si. Prototype devices with 5 cm x 5 cm arrays of 100µm pixels have been demonstrated, showing good linearity and over a range of x-ray energies suitable for mammography (Shah et al 2001). Experimental polycrystalline HgI$_2$ detectors have also been fabricated and demonstrated to be radiation hard. Their sensitivity, although similar to PbI$_2$, was found to be strongly dependent on the fabrication technology, being related to grain size and quality of electrical contacts, although it is possible to deposit polycrystalline PbI$_2$ films over large areas, which makes them potentially useful for future direct detection x-ray systems.

Virtually all large-area flat-panel detectors available to date use hydrogenated amorphous silicon (a-Si:H) active matrix readout. However, the electronic properties of a-Si:H are not ideal and limit the minimum size of structures that can be fabricated. An alternative is to use CMOS technology on crystalline silicon wafers, which allows integration of much more on-chip electronics. Active pixel sensors have been demonstrated using CMOS technology, which includes ‘on-chip intelligence’, allowing fixed pattern noise reduction, an increase in dynamic range, flexible region of interest readout, ADC and storage (Allinson et al 2009). These devices have been shown to have potential applications in radiotherapy in the verification of complex dose distribution and quality assurance of beam collimation systems (Osmond et al 2008). CMOS array size is limited by the currently available Si wafer dimensions (typically 8 inch).

4.4. Thermoluminescent and scintillation detectors

4.4.1. Thermoluminescent detectors. Thermoluminescent dosimetry uses materials that trap the charge-formed ionization electron/hole pairs in metastable states within the crystal. A very common material used is lithium fluoride doped with Mg and Ti (LiF:Mg,Ti), which is supplied in a range of shapes and sizes, including 1 mm diameter rods and 3.2 mm x 3.2 mm ‘chips’ with a range of thicknesses. Alternative materials have also been used, including CaF$_2$:Mn and LiF:Mg,Cu,P, the latter being preferred by some laboratories for environmental and personal dosimetry due to its improved linearity and reproducibility compared with LiF:Mg,Ti (Kron 1994, Moscovitch et al 2006). The metastable states introduced by the doping materials are deep enough to prevent escape of the majority of electrons at room temperature; however, if the material is then deliberately heated, enough energy is given to promote electrons back to the conduction band, where they will subsequently recombine with the holes. If a suitable material is selected, the energy released by recombination can be emitted as optical photons, which can then be detected see figure 8.
For a given temperature $T$, the probability of escape from a trap of depth $E$ is given by

$$p = \alpha \exp \left( -\frac{E}{kT} \right).$$

(15)

where $k$ is the Boltzman constant and $\alpha$ is a constant called the frequency factor ($\sim 10^9$ s$^{-1}$).

If the temperature is gradually increased, then the number of trapped electrons will also increase, reaching a maximum as the majority of the trapped electrons are released. This temperature for maximum release $T_m$ is related to the rate of heating $q$ by the following:

$$\frac{E}{T_m} = \frac{\alpha}{E/kT_m} \exp \left( -\frac{E}{kT_m} \right).$$

(16)

$T_m$ is 216 °C for a trap depth $E$ of 1 eV and $q$ of 1 K s$^{-1}$. LiF:Mg,Cu,P TLDs have a useful dose range from 0.5 µGy to 12 Gy and LiF:Mg,Ti from 0.05 mGy to 500 Gy (PTW Freiburg, Lörracher Strasse 7, 79115 Freiburg, Germany).

Practical measurements are made using automated TLD reading machines that heat each sample individually in a light-tight enclosure and measure the light emitted as a function of temperature, or glow curve, using a photomultiplier tube. The exposure is roughly linearly related to the light output (Horowitz and Moscovitch 2013). Heating can be performed in an atmosphere of dry nitrogen to prevent surface oxidation effects, and great care should be taken to ensure that the TLD surfaces remain clean to prevent spurious signals. TLDs are usually heated to 300 or 400 °C during readout, and then annealed, typically by heating to 300 or 400 °C for 1 h, followed by controlled cooling with at least 20 h at 80 °C before reuse, although protocols may vary between laboratories depending on absorbed dose and for different TLD materials. Since the signal fades with time (since the electrons will have a finite probability of escape from the trapped states), the same length of time should be left between irradiation and readout for calibration and clinical measurements. One method of ensuring this is to perform calibration experiments immediately before or after the clinical irradiation and then read out both sets of TLDs together (although in practice, dose-response non-linearity can be more significant than fading, depending on the TLD material, dose and time delay before readout). To compensate for background radiation effects, a control batch of unirradiated TLDs can also be read out at the same time. To improve statistical accuracy, 2 or 3 TLDs are often used for each measurement. In normal clinical use the expected accuracy is between 2 and 3%.

TLDs are frequently used for in vivo dosimetry (Mijnheer et al 2013), particularly for treatments where accurate dosimetry is difficult, such as total body irradiation. They can also be used internally in catheters, or easily placed inside phantoms for quality assurance measurements (Venables et al 2004, Hsi et al 2013). They offer the advantage that they are easy to use, since they can be simply taped to the skin, but have the disadvantage that they can only be read out after irradiation and careful calibration, and handling is required to get reasonably accurate results. Mailed TLDs are used for quality assurance checks or dosimetry audits.

4.4.2. Optically stimulated luminescence. An alternative to TLDs are optically stimulated luminescence dosimeters (OSLDs), being highly sensitive (giving large amounts of light output for relatively small absorbed doses) and well-suited to personal dosimetry, where carbon-doped aluminium oxide ($\text{Al}_2\text{O}_3:\text{C}$) has been used for some 15 years. Their principle of operation is very similar to TLDs, except that trapped charge is released by controlled illumination instead of heating (Yukihara and McKeever 2008). A limitation of $\text{Al}_2\text{O}_3:\text{C}$ is the slow trap-emptying time (typically 10s of seconds). In contrast, the short trap emptying time (~25 ms) and short luminescence lifetime (1.1 µs) of the europium-doped alkali halides lend
themselves to ‘real-time’ dosimetry, where an OSL probe is repeatedly read out during continuous exposure. Potassium bromide doped with europium (KBr:Eu) exhibits OSL emission at about 420 nm when excited at 620 nm. It is thought that electrons are selectively trapped at neutral Br-vacancies, whereas holes are trapped at by Eu\(^{2+}\) ions, forming Eu\(^{3+}\). Stimulation of the Br-vacancies at 620 nm releases the trapped electron, which recombines with an Eu\(^{3+}\) ion to produce Eu\(^{2+}\) and a 420 nm emission (McKeever 2011).

\(\text{Al}_2\text{O}_3:\text{C}\) OSL devices have been shown to be capable of precision of ~0.7% (single readout) in high-energy photon and electron beams with appropriate protocols and readout, although significant fading (>5%) can be seen in the first 5–15 min after irradiation (Yukihara et al 2010). X-ray CT dose profiles have also been demonstrated using \(\text{Al}_2\text{O}_3:\text{C}\) OSL strips, although the high effective atomic number resulted in some low-energy over-response; for proton beams, however, the energy dependence is small (< ±2%) above 100 MeV (Kerns et al 2012). KBr:Eu OSL probes have been demonstrated in ‘real-time’ mode during both proton therapy and x-ray CT (McKeever 2011). The clinical applications of OSLDs are similar to those for TLDs, with uses demonstrated for in vivo dosimetry (IAEA Human Health Reports, No. 8, 2013, Sharma and Jursinic 2013) and dosimetry audits (Lye 2014).

4.4.3. Radiophotoluminescent glass dosimeters. There is renewed current interest in the use of radiophotoluminescent glass dosimeters (RPLGDs), which were used by a few centres for radiotherapy applications in the 1950s and 60s (Roswit et al 1970), but not very widely since. Silver-activated phosphate glass (comprised of Ag\(^+\) and PO\(_4^{3-}\) ions), when irradiated, forms stable luminescence centres (Ag\(^0\), Ag\(^{2+}\)), which are able to absorb and release energy. By illuminating the RPLGD with an ultraviolet laser, orange luminescent light is produced. Unlike the case in TLDs, the luminescence centres are not destroyed by the readout process, so the devices can be read out multiple times (typically 50–60 times in a few seconds) to reduce random error. Heating to 400 °C for 60 min following readout allows stable luminescence centres to anneal and the devices to be reused. Careful handling and cleaning was necessary in some early devices to prevent surface contamination from affecting results. Current systems use the difference in fluorescence decay times between surface contamination (0.3 \(\mu\)s) and radiophotoluminescence (3.0 \(\mu\)s) to discriminate signal from contamination noise, making handling easier.

One complication with the use of RPLGDs is the fact that following irradiation, some electrons require additional energy to correctly enter luminescence centres, which can be supplied by heating, thus increasing the luminescence signal by up to 50% (sometimes termed the ‘build-up effect’). Heating to 70° C for 30 to 60 min is recommended by some manufacturers, but heating to higher temperatures for a shorter period of time (100° C for 15 min) may give a better response (Manninen et al 2012). Preheating before measurement is also recommended to avoid the risk of significant uncertainty. The high atomic number (\(Z_{\text{eff}} = 12.0\)) of silver-activated phosphate FD-7 type devices relative to water makes careful calibration necessary in the diagnostic energy range to account for photoelectric over-response; some users employ a tin filter to improve response (although this dramatically reduces response below 30 keV). In the therapeutic range (>0.3 MeV), very weak energy dependence (energy correction factors of 1.014 ± 0.009 and 1.026 ± 0.007) has been reported for 6MV and 10MV measurement relative to calibration with \(^{60}\)Co (Mizuno et al 2008). The (high) photon energy independence, high sensitivity and good reproducibility make RPLGDs attractive for a range of dosimetry applications, with useful readout ranges from 10 \(\mu\)Gy to 10 Gy, extendable up to 500 Gy, especially since they are rugged and easy to handle and can be read out multiple times.
Typical geometries used are rods with readout volumes 1.5 mm in diameter and 6.0 mm in length; this geometry leads to a small dependence on the orientation at irradiation (up to 1.5%), which should be taken into account. A slow fading observed between irradiation and readout (1.6% over 133 days) (Rah et al 2009) has also been observed, but may be negligible if the time between irradiation and readout is small.

In vivo use was reported with RPLGDs applied to the tumour surface and surrounding normal tissues in a group of head and neck cancer patients (Nose et al 2005), which demonstrated the feasibility of clinical use. A feasibility study into the use of RPLGDs for postal dose intercomparison for high-energy photon beams was also reported, finding a reproducibility of 0.7% (1 SD) and combined standard uncertainty of 1.9% (Rah et al 2009).

4.4.4. Organic and plastic scintillators. Organic scintillators are pure crystals composed of aromatic hydrocarbon compounds containing benzene ring structures and with decay times of a few nanoseconds. The most common scintillators are (1) anthracene (C\(_{14}\)H\(_{10}\), decay time 30 ns), (2) stilbene (C\(_{14}\)H\(_{12}\), decay time 4.5 ns) and (3) naphthalene (C\(_{10}\)H\(_{8}\), decay time of a few ns). Anthracene has the highest light output of all organic scintillators, and thus is commonly chosen as the reference for light output when comparing scintillators. Although organic scintillators have fast decay times and are very durable, their response is usually anisotropic; they cannot easily be machined, and they cannot be easily grown into large sizes. Organic scintillators thus have a limited range of applications.

However, because of the ease in shaping and fabricating plastics, these have become an extremely useful medium in which to place organic scintillators. Plastic scintillators are produced by dissolving an organic scintillator in a solution containing the monomer precursors prior to polymerization. Plastic scintillators are available in rods, cylinders and flat sheets, and have the advantage of a fairly high light output and a relatively quick signal with decay time of 2–4 ns.

4.4.5. Liquid scintillators. Liquid scintillators are produced by placing the organic scintillating crystals in an appropriate solvent. In some cases, a third constituent is added to the liquid scintillator as a wavelength shifter to tailor the emission spectrum to match the spectral response of the photomultiplier tube. Because of their lack of solid structure, liquid scintillators can be more resistant to radiation damage than plastic scintillators, which makes them radiation hard to doses of 10\(^5\) Gy. Liquid scintillators are sold in closed glass containers, where contact with air is avoided to reduce the quenching effect of dissolved oxygen, which reduces the fluorescence efficiency. Liquid scintillators are effective for measuring beta particles or any radioactive material that can be dissolved as part of the scintillator solution, where the counting efficiency is close to 100%.

4.4.6. Scintillator readout. The photomultiplier tube (PMT) remains the most commonly used detector for converting scintillation photons to electronic signals because of its combination of very high gain (>10\(^6\)), low noise and fast response time (Lecomte 2009). One limitation of the PMT is tube size, which is typically much greater than crystal size, making one-to-one coupling unfeasible and requiring a crystal-encoding scheme to be employed. The two main schemes used are Anger readout, with discrete crystals optically coupled to the PMTs through a continuous light guide (to allow for optical dispersion between a group of PMTs), and the block detector principle, where a 64-or-more-element slotted pseudo-continuous crystal is read out with a quadrant PMT. One significant drawback with PMTs is their high sensitivity to magnetic fields, making them unsuitable for use in hybrid PET/MR systems; significant gain variations are seen at ~10 mT.
Solid state detectors are therefore an attractive alternative, offering high quantum efficiency, insensitivity to magnetic fields, compact shape and the potential to allow a one-to-one coupling with each scintillation crystal. However, standard PIN photodiodes do not produce a high enough signal-to-noise ratio to be used for coincidence detection, so alternative designs must be considered. Avalanche photodiodes (APDs) have a depletion layer where visible light photons can create electron-hole pairs via the photoelectric effect. Application of a high field (~2 × 10^5 V/m) across this narrow region causes electrons to be multiplied by impact ionization (holes are typically collected without multiplication), although this amplification process does introduce excess statistical noise. Gains of up to 1,000 can be achieved with APDs, but they are more typically operated with multiplications of 50–150 to keep noise within acceptable limits. The fast and low-noise front-end electronic necessary for large-scale implementation of APDs in PET is challenging and currently limits their use.

If the bias voltage of an APD is increased to above the avalanche breakdown limit (i.e. in Geiger mode), breakdown discharges can be caused by incoming photons or thermally generated free electrons. Arrays of this type of APD are often referred to as a silicon photomultiplier, or SiPMs. High gains of 10^5 to 10^6 can be achieved, although their implementation is challenging because of non-linear response limiting their dynamic range (significant deviations from linearity are typically seen when the number of photoelectrons per cell exceeds 50%), high dark count due to thermally generated carriers, strong temperature dependence of both gain and bias, and crosstalk between elements. Dark rates of SiPMs are typically in the range of 1–3 MHz mm^{-2} at the 1–2 photoelectron level; however, for 511 keV photons, larger numbers of photoelectrons are typically produced, and the dark count rate at 3–4 photoelectrons is down to ~kHz, making their use for PET feasible (Del Guerra et al 2010). Also, the high gain of SiPMs compared to APDs reduces the need for low-noise preamplifiers and preserves the intrinsically fast response, and makes these detectors very promising for time-of-flight PET systems with time resolutions of less than 100 ps (Ambrosi et al 2010).

4.5. Chemical detectors

4.5.1. Fricke dosimeters. The fact that ionizing radiation causes ferrous ion in aqueous solutions of ferrous (Fe^{2+}) sulphate solution to oxidize to the ferric (Fe^{3+}) state has been known since the 1920s (Fricke and Morse 1929). In this early work, radiation dose was quantified by titration of the solutions and measurement of oxidation reduction potentials. A method for quantifying dose that is currently used by standards laboratories is to measure optical density of a Fricke solution containing 10^{-3} M ferrous ammonium sulphate and 10^{-3} M sodium chloride in 0.4 M sulphuric acid. The average dose to a Fricke solution $D_F$ is given by the change in optical density at 303 nm, $\Delta OD$ thus

$$D_F = \frac{\Delta OD}{\epsilon G \rho L}$$  \hspace{1cm} (17)

where $\epsilon$ is the molar extinction coefficient (21741 mol^{-1} cm^{-1} at 25 °C), $G$ is the yield of ferric ions (1.617 × 10^{-6} mol J^{-1} at 25 °C), $\rho$ is the density of the Fricke solution (1.023 kg l^{-1} at 25 °C) and $L$ is the path length over which the optical signal is read (typically 2 to 4 cm). Dose to water is then calculated by correcting for the differences in mass-energy absorption coefficients between the Fricke solution and water $R_F^W$, and a correction factor for the effect of the walls of the measurement vial $k_{vial}$, thus: $D_w = D_F R_F^W k_{vial}$. Comparisons with water calorimetry showed that, for ^{60}Co, the optical readout of the Fricke system
agrees to within 0.6(±0.4)% (Ross et al 1989). Very slight energy dependence has been observed in the value of \( G(\text{Fe}^{3+}) \), with a measured ratio between 20 and 30 MV x-rays and \( ^{60}\text{Co} \), \( G_{\text{Fe}^{3+}} \), of 0.007 ± 0.003 (Klassen et al 1999). This near-energy independence, accuracy, reproducibility and linearity make Fricke dosimetry an attractive choice for standards laboratories.

In addition to optical absorption, Fricke systems also show differences in magnetic properties, with the strongly paramagnetic \( \text{Fe}^{3+} \) showing a very different molar relaxivity to \( \text{Fe}^{2+} \). Linear relationships are seen between ion concentration and \( 1/T_1 \) or \( 1/T_2 \) relaxation rates measured by magnetic resonance. Three-dimensional dose readouts have been demonstrated using Fricke gels (aqueous ferrous sulphate fixed in a gelatine matrix); however, the \( \text{Fe}^{3+} \) ions diffuse relatively rapidly through the gel, significantly blurring the observed dose distributions, so they need to be read out using MRI within hours of exposure.

4.5.2. Polymer gels. A gelatine-matrix-based dosimetry system that avoids the diffusion problem is the polymer gel (Maryanski et al 1993). In these inherently three-dimensional dosimetry systems, the gelatine matrix contains monomers that polymerize by free-radical induced chain reactions to form spatially fixed cross-linked networks. Radiation-induced polymerization has been investigated since the 1930s (Hopwood and Phillips 1939), but it was not until the demonstration that \( T_2 \) relaxation rates measured by MRI reduced with an increasing degree of polymerization and cross-linking that polymer gels were developed for radiation dosimetry by Maryanski in the 1990s (Maryanski et al 1993).

The gel formulation originally proposed by Maryanski was based on copolymerization of acrylamide and \( N,N’ \)methylene-bis-acrylamide (bis) monomers suspended in an aqueous agarose matrix. Agarose was later substituted for gelatine, improving the dynamic range due to the lower \( R_2 \) relaxation rate of water in gelatine. Various formulations have been demonstrated using acrylamide, acrylic acid or methacrylic acid. A drawback of many of the early formulations was sensitivity of oxygen, which inhibited polymerization by free radical quenching. Gels had to be manufactured in anoxic conditions and housed in impermeable containers. A useful development was the addition of ascorbic acid and copper II sulphate to a methacrylic acid homopolymerization system (Fong et al 2001). The ascorbic acid acted as a molecular oxygen scavenger, binding oxygen in a metallo-organic complex catalyzed by the copper II sulphate. This allows gel manufacture under normal atmospheric conditions. Methacrylic acid auto-polymerization is inhibited by the addition of hydroquinone.

An extensive review of polymer gel dosimetry was recently presented by Baldock et al (2010); in brief, ionising radiation causes the radiolysis of water \( \text{H}_2 \text{O} \rightarrow \text{H}^* \text{OH}^* \), where \( \text{R}^* \) represents the primary free radicals \( \text{H}^* \) and \( \text{OH}^* \) with rate of production \( k_D \)

\[
k_D = \frac{dD}{d\text{r}} \frac{\rho G(\text{R}^*)}{100eN_A}, \quad (18)
\]

where \( dD/d\text{r} \) is the absorbed dose rate, \( \rho \) is the number of free radicals produced per 100 eV absorbed, \( e \) is the elemental charge, and \( N_A \) is Avagadro’s number. The water free radicals then undergo an \textit{initiation} reaction with a monomer unit M, breaking a carbon-carbon double bond and forming a new bond with the radical, transferring the unpaired electron to the polymeric radical,

\[
\text{R}^* + \text{M} \rightarrow \text{RM}^* \quad (19)
\]

Propagation reactions can then occur, adding repeated monomer units,

\[
+ \text{M} \rightarrow \text{RM}_{n+1}^* \quad (20)
\]
until a termination reaction, typically a radical-radical reaction terminates the chain

$$R' + R'' \rightarrow RR.$$  \hspace{1cm} (21)

In addition, MRI, x-ray and optical tomography have been investigated for dose readout. X-ray CT has the obvious disadvantage of further irradiating the gel, and is very challenging since the difference in mass attenuation coefficient of the monomer in solution and the polymer are very small, yielding very low contrast. Optical readout is promising, but has the disadvantage of requiring optically transparent containers (reviewed in Doran 2009). Some experimental work has been done investigating the use of ultrasound to read out gels (Crescenti et al 2007), although this remains challenging.

Polymer gel dosimetry remains one of the only methods of obtaining fully three-dimensional readout of dose, and is useful for research and commissioning of complex new clinical radiation delivery techniques. Significant care is still required to obtain accurate results with batch-to-batch variations in sensitivity affecting most of the currently used systems. Factors that are known to influence results are the reagent batch, thermal history of the gels, container size and post-irradiation storage conditions. For these reasons gel dosimetry is largely limited to relative dosimetry and is not recommended for reference dosimetry use (MacDougall et al 2002).

4.5.3. Alanine dosimetry. Alanine is an amino acid that forms stable free radicals when irradiated. The concentration of the radicals can be measured by electron paramagnetic resonance (EPR) spectroscopy, and is proportional to absorbed dose.

In a magnetic field $B$, unpaired electrons are split into two discrete energy levels, where the separation between the levels $\Delta E$ is given by the electron-spin factor $g$:

$$\Delta E = \frac{eh}{2mc} gB.$$  

For a 300 mT magnetic field, transitions between these levels can be probed using 9 GHz microwave radiation. Alanine is typically used in pellet form, 5 mm diameter and ~2.5 mm thick, and comprised of 90% alanine, 10% paraffin wax. These pellets are sensitive enough for measurements in small radiotherapy fields for doses ≥10 Gy and close to water equivalent, although correction for volume-averaging effects must be applied for fields < 2 × 2 cm$^2$. They have negligible energy-dependence and isotropic response, which makes them ideal for calibration measurements in small or non-standard fields from brachytherapy (Anton et al 2009) or tomotherapy (De Ost et al 2011; Perichon et al 2011). The disadvantages are that readout can only be performed using a specialist EPR spectrometer, and pellets must be left for at least 24 h between irradiation and readout to allow the signal to stabilize.

This does, however, mean that alanine is an attractive material for a mailed calibration service run by a standards laboratory or dosimetry service. The relatively low sensitivity of the alanine-EPR system compared to other dosimetry methods is a limitation, although with measurement conditions, relative uncertainties of less than 0.5% have been reported for doses in the 5–25 Gy range with $^{60}$Co (Anton 2006).
4.6. Radiographic and radiochromic films

Photographic film was the medium used by Roentgen in 1895 to discover the x-ray, and has been extensively for x-ray imaging and dosimetry ever since (Webb and Flower 2012). Modern x-ray film is usually comprised of a suspension of silver bromide grains, with up to 10% silver iodide suspended in a gelatine matrix. The grain sizes range from 0.2 to 10 μm depending on the application. The photographic emulsion is bonded to either one or both sides of a synthetic polyester base material; a thin protective layer can then cover the emulsion. (See figure 9).

Photons interact photoelectrically with the halide, and the electrons produced reduce the silver ions forming a ‘latent image’ of metallic silver atoms. When the film is developed, the film grains containing more than a critical number of non-ionized silver atoms are then completely reduced to metallic silver. Grains with less than this critical number are removed by the fixing process. X-ray film images, on a microscopic scale, are therefore binary images formed by a noisy amplification process. On a macroscopic scale, however, the average optical transmittance varies smoothly over a wide range. The transmittance \( T \) of a film illuminated by a source with irradiance \( I_0 \) is given by

\[
T = \frac{I}{I_0},
\]

where \( I \) is the transmitted irradiance. It is conventional, however, to express film appearance in terms of optical density (OD), which is simply the logarithm of transmittance,

\[
\text{OD} = \log_{10} \left( \frac{I}{I_0} \right).
\]

A simple physical model can be derived to relate the optical density of a film to the exposure \( X \), giving the following expression:

\[
\text{OD} = \text{OD}_{\text{max}} \left( 1 - e^{-kX} \right).
\]

The maximum optical density of the film is given by the transmittance when every silver halide grain is reduced (i.e. saturation of the film) and the rate constant \( k \) describes the sensitivity. In practice, as well as the limited maximum film density, there is also a minimum density caused by the fact that some unexposed silver ions are still reduced during developing, giving rise to a background ‘fog’ or base density level. Figure 10 shows a typical film dose-response curve, illustrating the limitations of a non-zero film base density. This standard form of curve is called an H-D curve, after Hurter and Driffield, who presented the first film sensitivity curves in 1890. The film characteristics over the region where the optical density varies roughly linearly with the logarithm of exposure is called the \textit{gamma} of the film, and is defined thus:
where $X_0$ is the apparent exposure indicated by unexposed processed film (i.e. the base density level). For a small contrast $\Delta X/X$, the logarithm can be expanded and the change in optical density $\Delta D$ expressed as

$$T\Delta D = 0.434\Gamma \left( \frac{\Delta X}{X} \right).$$

(26)

For the dose range where this equation holds, the difference in optical density is simply proportional to the product of the film gamma and input contrast. Even outside of this range, we can still use this equation to describe the change in optical density for small contrasts using the definition

$$\Gamma = \frac{dD}{d\log_{10}X}.$$  

(27)

We can see from figure 10 that contrast will be low for exposures that result in very low or very high optical density, which limits the useful dynamic range of this type of film. It is important to expose film such that the regions of interest within the patient produce optical densities where gamma is large. The range of exposures where this is possible is known as the film latitude. High-contrast films with a large gamma have narrow latitude, whereas low-contrast films with a small gamma have a wide latitude.

4.6.1 Film-screen systems. X-ray films can be used in direct exposure mode, where x-rays interact directly with the film emulsion or base, but for higher-energy x-rays the low interaction cross-section makes this process inefficient. Film-screen systems are therefore used to increase sensitivity and work in a multi-step process. An x-ray is absorbed by the fluorescent screen (typically 70–200 mg cm$^{-2}$ gadolinium oxysulphide), and energy is re-emitted as visible light.

The visible light photons interact with the photographic emulsion, which is chemically developed in the usual way. Film cassettes with either single (one side only) or double
screens (both sides) can be used (see figure 11). Single screens are typically used for applications where high spatial resolution is important, since they exhibit less optical dispersion.

Two sources of noise typically dominate in film-screen systems (Webb and Flower 2012); the first is quantum mottle, which is the uncertainty in the number of x-ray photons absorbed per unit area. The noise due to quantum mottle can be described as

$$\Delta D_Q = 0.434 \Gamma (A \varepsilon N)^{-\frac{1}{2}},$$

where, averaged over an area $A$, $\varepsilon$ is the probability of an x-ray interacting with the screen and emitting $N$ visible light photons. The second principle noise source is the film granularity, which leads to fluctuations in the number of silver halide grains per unit area.

$$\Delta D_G = \left(0.434 D \frac{\sigma}{A}\right)^{\frac{1}{2}},$$

where $\sigma$ is the average cross-section of a developed silver grain in area $A$. Film granularity is proportional to the square root of optical density, and therefore increases with increasing density. As a consequence of this, quantum mottle dominates for a correctly exposed film, whereas film granularity dominates where a film is under or over-exposed.

4.6.2. Radiochromic films. A major change occurred in film dosimetry starting in the early 1990s, with the introduction of a range of poly-diacetylene-based radiochromic films (GAFCHROMIC®Tm, International Specialty Products, Wayne, NY, USA). Radiochromic films show a weak energy dependence, high spatial resolution, near tissue equivalence which makes them very attractive for radiotherapy dosimetry and quality assurance. However, perhaps the most convenient feature is that these films are relatively insensitive to visible light, with exposure to direct sunlight resulting in film darkening equivalent to a dose of a few Gy. They undergo colour change by polymerization (leading to dye creation within the film), and so remove the need for a separate chemical development process. Three products have been designed specifically for the external beam radiotherapy market: EBT, EBT2 and EBT3; reviewed in Devic (2011).

The original ‘EBT’ film had a useful range up to 8 Gy (although it has been shown that this can be extended up to 100 Gy by using RBG channels (Devic 2011)). A practical limitation of EBT was that non-uniformity of the sensitive layer thickness could cause response variations of up to 2%, making careful calibration of each batch essential. This problem has been
partially addressed in EBT2 by the addition of a yellow dye to the sensitive layer. If the film is read out on a flatbed scanner, measurement in the blue channel (where the yellow dye absorption is strongest) can be used to compensate for non-uniformities in the sensitive layer thickness and applied to the radiation-sensitive absorption values measured from the other colour channels (Mayer et al 2012). A further refinement seen in EBT3 is to prepare the film with a matte finish, eliminating problems with Newton’s rings sometimes seen due to imperfect contact between the film and glass surface of the flatbed scanner (see figure 12).

It has been shown (Devic 2012) that the logarithmic film response can be described using a calibration equation containing a linear and single power term

\[ D = b \cdot \text{netOD} + c \cdot \text{netOD}^n. \]  

(30)

By inspecting a number of calibration curves, it was further shown that the following functional form provided a good description of observed behaviour.

\[ \zeta = (-1) \cdot \frac{\text{netOD}^2}{\ln(\text{netOD})}. \]  

(31)

By using this form of calibration curve, the parameter \( \zeta \) shows a linear relationship with dose to within \( \pm 2\% \).

It should be noted that because radiochromic film is still essentially a chemical readout, care is needed to standardize the time that radiochromic films are read out and temperature they are exposed to and stored at. The polymerization reaction is relatively slow and continues to ‘develop’ for at least 8 h, leading to the recommendation that all films should be read out 8 h post-irradiation. However, the polymerization does not stop at this point, and films continue to darken, so should be read out within 24 h and cannot be retrospectively re-read without reference to the calibration films. They are also temperature dependent, so should be stored...
at a constant temperature between irradiation and readout. It has been shown that if a careful protocol is followed using the same time between irradiation and readout for calibration and measurements, faster readout is possible with reasonable accuracy (30 min +/- 5 min gives less than 1% dose error). It should also be noted that the films are polarization sensitive, so the orientation of the film calibration and readout must be kept constant, or errors of up to 10% can be seen (Soares 2006). Because of the high spatial resolution, good energy sensitivity and experimental ease of use, radiochromic films have been widely used for a number of applications, including: IMRT verification, small field dosimetry, microdosimetry and dosimetry for radiobiology experiments.

Part II – Applications of Detectors

5. Imaging

5.1. Computed and digital radiography, mammography

Conventional film systems use intensifying screens to capture x-rays and reduce radiation dose. X-rays that pass through tissue are collected by phosphor screens. When an x-ray is absorbed, the resultant scintillation creates a number of optical photons that spread and illuminate the film in a distribution cloud. An important parameter to understand is the thickness of the intensifying screen. Thicker screens capture more x-rays and are therefore more dose efficient and higher speed. However, thicker screens also create more light scatter and blurring of the image. Therefore, it is impossible to offer a screen-film system while simultaneously offering the highest possible resolution and lowest possible radiation dose. This trade-off between radiation dose and image quality must be optimized for the specific clinical application, such as computer and digital radiography or mammography.

Computed radiography has gone through various changes over the past 35 years, where the most important transition was from cassette-based imaging to cassetteless based imaging. Cassette-based radiography is commonly known as ‘computed radiography’, or CR, while the cassetteless version of radiography is known as ‘digital radiography’ or DR (also known as direct radiography). The difference between the two methods is how the image is created and how it is processed. In addition, radiography systems are also split into two broadly defined categories, depending on the conversion process of x-ray energy to electric charge: i) indirect and ii) direct radiography systems (Samei 2003).

In CR the imaging cycle has three steps (1) expose, (2) readout and (3) erase, where photosensitive storage phosphors (PSP) are used to store the latent image in the form of electrons in semi-stable traps within the phosphor structure. Since the storage-phosphor image is separate from the readout process, the CR is an indirect conversion process. Phosphors are crystalline host materials, almost always containing a trace amount of an activator element. The activator molecules can be incorporated into the matrix material by diffusion at a temperature close to the matrix melting temperature. The materials used to store the latent image are either BaFBr:Eu<sup>2+</sup> or CsBr:Eu<sup>2+</sup>, where the energy needed to create a trapped electron and hole are respectively 100 eV and 67 eV. The BaFX matrix is a layered material with a tetragonal PbFCl or Matlockite structure (Nicklaus and Fischer 1972). The image is extracted using a laser that scans point by point. The CR phosphor emission matches the sensitivity spectrum of the photomultiplier tube (PMT) light detector, which is predominantly below 500 nm.

In the case of DR, both direct and indirect processes are used to convert x-rays into electric charges (Chotas et al 1999, Chotas and Ravin 2001), as discussed in section 4.3.4. Recent developments for a novel pixel-structured scintillation screen with nanocrystalline Gd<sub>2</sub>O<sub>3</sub>:Eu
particle sizes for high-spatial-resolution x-ray imaging detectors are being made for indirect x-ray imaging sensors with high sensitivity and high spatial resolution (Cha et al 2011a, 2011b).

In conventional mammography, intensifying screens are used in conjunction with films to capture x-rays and reduce radiation dose, as discussed in section 4.6. While screen-film combinations offer various advantages, there are significant limitations with them, such as non-linear sensitivity to photoflux, film granularity that affects detective quantum efficiency (DQE), processing time and chemical processing procedure. Digital mammography detectors allow the capture of the image as a digital signal, eliminating any storage requirements. In addition, digital systems offer larger dynamic range, improving contrast of tissues with very similar density in the breast. Digital mammography captures images based on either (i) indirect or (ii) direct conversion methods. Examples of indirect conversion methods are cesium iodide doped with Thallium (CsI(Tl)), coupled to an array of thin-film diodes, or CsI(Tl) coupled with charge-coupled devices (CCDs), where the electronic signal then impinges on the thin-film transistor (TFT). For direct conversion systems, amorphous selenium is used as a photoconductor that produces the electronic signal that impinges on the TFT. Digital mammography offers the potential of clinical applications that are not feasible with screen-film, such as stereotactic breast biopsy, tomosynthesis and 3D imaging and computer-aided diagnosis.

5.2. CT

Since the invention of x-ray-computed tomography by Hounsfield and Cormack in 1971, there have been a series of major technical improvements in scanner design, increasing the number of detectors, contrast and spatial resolution and decreasing acquisition time and patient exposure. Helical scanning with multi-element detectors covering at least 16 (and up to 256) slices simultaneously are now common (Grignon et al 2012), with the majority of these multi-slice helical systems based on ceramic scintillator/photodiode detectors (Kalender 2008). Cone-beam scanning geometries are also increasingly common, made possible by large-area amorphous silicon flat-panel detectors, and are clearly advantageous where placing the patient inside a toroidal gantry is not possible – for example, in interventional radiology, where C-arm systems offer much greater access to the patient (Orth et al 2008). Cone-beam kV imaging systems are also increasingly commonly integrated with radiotherapy linear accelerators, providing high-quality anatomical information for online guidance (Jaffray 2012). Development of ultra-low dose scanners based on multi-wire proportional chamber detectors (Charpak 1993) have been demonstrated, showing potential dose reduction by factors of 13.1 in the spine and 18.8 in the pelvis for paediatric patients (Kalifa et al 1998). Flat panel detectors are also finding application in digital breast tomosynthesis systems (Helvie 2010).

The rapid data acquisition capabilities of modern CT scanners make respiratory-correlated or ‘4D’ CT acquisition possible, where a surrogate signal of the patient’s breathing trace is encoded during acquisition and data is retrospectively sorted into bins of either breathing phase or amplitude for reconstruction. A wide range of surrogates are currently used, including reflective markers placed on patients’ skin or surface tracking, spirometry, strain measurement with a belt, and implanted markers localized either with x-ray imaging or electromagnetically (Moorrees and Bezak 2012). Within this wide range of options there are clear tradeoffs between invasiveness (for implanted markers) and accuracy (for external optical markers, which rely on a motion model to predict internal anatomical motion). There has also been some success in extracting respiratory phase from the projection data itself (particularly in cone-beam CT).
5.3. SPECT

In single-photon emission computed tomography (SPECT), recent improvements in the reliability and performance of cadmium zinc telluride (CZT, see section 4.3.4) detectors have not only allowed improvements in energy resolution (CZT energy resolution typically 5–6%), but also the construction of compact and novel detector geometries (Hutton 2010). The improved energy resolution can not only improve image quality with better scatter rejection, but also allow dual radionuclide imaging – for example, the use of $^{99m}$Tc and $^{201}$Tl for simultaneous rest/stress myocardial perfusion imaging (Ben-Haim et al 2010). Novel camera designs for specific applications such as nuclear cardiology are now available, featuring stationary focused arrays of multi-pinhole collimators and CZT detectors. With this design, all views of a specific body segment can be acquired simultaneously, increasing acquisition speed and reducing motion artifacts.

5.4. PET

Some excellent and comprehensive reviews of recent advances in detector technology for positron emission tomography have been presented recently (Lewellen 2008, Lecomte 2009). The main detector technology employed by the current generation of clinical PET scanners is the scintillator/photomultiplier combination. There has been a recent focus on improving the detector timing resolution to allow time-of-flight measurement; this has led to a move away from the widely used BGO $^2$ and GSO $^3$ to LYSO $^4$ because of the much shorter scintillation light decay time (~40 ns for LYSO compared to ~300 ns for BGO). The current generation of time-of-flight (ToF) scanners shows an effective system timing resolution of ~600 ps, giving an approximately twofold improvement in signal-to-noise ratio (SNR) compared to non-ToF systems. This improvement in SNR has been shown to have a number of clinically useful applications, including bringing image quality for larger patients closer to that for patients of average size, improvements in image quality and lesion detectability, reduction in examination time, and reduction in administered activity to the patient (Conti et al 2011). In parallel with implementation of ToF capability, resolution recovery techniques have also been developed, including spatial variations in the system point-spread function (PSF) within the field of view of the scanner during reconstruction. PSF modeling during reconstruction has been shown to give a similar SNR improvement to the use of ToF information, with added benefits if both are used together (Akamatsu et al 2012).

The desire for faster detector response to allow further improvements in ToF resolution has led to significant research efforts investigating alternative scintillator materials, such as LaBr$_3$:Ce and solid-state visible light detectors, including avalanche photodiodes (APDs) and silicon photomultipliers (SiPMs). APDs have the advantage of a ~10$^2$–10$^3$ higher gain than conventional silicon PIN photodiodes, but their currently limited timing resolution (~1 ns), high temperature sensitivity and high multiplicative noise limit potential application. Alternatively, APDs operated in avalanche breakdown, or Gieger-mode (termed SiPMs), show gains of 10$^5$ to 10$^7$ and very fast response times: combined LaBr$_3$:Ce and SiPM detectors have been shown to be capable of 100 ps timing resolution, corresponding to a ToF position resolution of ~15 mm (Schaart et al 2010), see table 3. Another advantage of solid-state APD and SiPM detectors is that they can be used inside the high magnetic field of an MR system, where the use of traditional photomultiplier tubes is impossible, making it possible to design hybrid PET/MR systems.

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2 BGO, bismuth germinate (Bi$_4$Ge$_3$O$_{12}$).
3 GSO, cerium-doped gadolinium orthosilicate (Gd$_2$SiO$_5$:Ce).
4 LYSO, cerium-doped lutetium-yttrium oxyorthosilicate (La$_{1.5}$Y$_{0.1}$SiO$_5$:Ce).
A physical limitation of discrete crystal/detector PET systems is that, to ensure good detection efficiency, relatively thick crystals (10–30 mm) are used, and the depth of the primary photon interaction within this crystal is not recoverable, leading to degradation in spatial resolution away from the central axis of the scanner. A number of methods have been presented for recovering this depth-of-interaction (DOE) information, including: the use of scintillation light readout from both ends of each crystal element (Yang et al. 2008), the use of pulse shape analysis from dual layer ‘phoswich’ detectors employing two different scintillator materials (Chung et al. 2004) and the use of APD arrays to measure light dispersion in either single continuous crystal detectors (Ling et al. 2007) or multiple layers of continuous crystal (McCallum et al. 2005), although these technologies are not yet in routine clinical use.

### 5.5. Proton beam imaging

Proton beam imaging devices (pCT) and proton radiography are being developed at the various proton therapy centres (Wilson 1946, Cormack 1963, Koehler 1968, Koehler and Steward 1973, Kramer et al. 1980, Hanson et al. 1982, Schneider and Pedroni 1994, Sauli 1997, Pemler et al. 1999, Zygmanski et al. 2000, Johnson et al. 2003, Schneider et al. 2004, Schulte et al. 2005, Li et al. 2006, Seco and Depauw 2011) for improved proton stopping power estimates for daily patient set-up and tumour tracking. Proton-imaging devices have been designed using (a) gas electron multipliers (GEM) detectors, (b) scintillating tiles (hodoscopes), (c) silicon strip detectors (d) CMOS detectors and (e) scintillating fibres. The principle of pCT and proton radiography is to measure at the entrance and exit plane the position, direction of motion of the proton, and the energy of each individual proton. The curved trajectories of the protons through the medium are significantly better estimated once the information of the entrance and exit planes are obtained.

#### 5.5.1. Gas electron multipliers (GEM)

GEM detectors were developed by Sauli (1997) and are composed of a gaseous scintillation material Ar+CF₄ filling and a grid of two metal layers separated by a thin insulator that serves for charge amplification. GEMs are used to localize protons in a plane perpendicular to the proton motion. In the case of proton radiography, the GEM detector localizes the XY coordinates of each proton before and after it exits the patient, where the two signals are obtained in coincidence. GEMs are built as 2D positioning devices with a CCD camera that readouts the light output. For proton imaging, the GEM would need to be redesigned to allow light output measurement in parallel with the proton energy discrimination after passing through the GEM detector. At present no GEM proton-imaging device exists that measures both position and energy of exit protons.

#### 5.5.2. Scintillating tiles

A proton imaging device with scintillating tiles, or hodoscopes, was developed at the Paul Scherrer Institute (PSI) (Pemler 1999, Schneider et al. 2004). A hodoscope is approximately 2 × 2 mm² of Bicron BCF scintillating material. A groove is cut in the scintillating block, where a fibre is placed. Signal blurring will occur due to the fact that...
protons will multiple scatter within the scintillator, producing a shower of optical photons, reducing their spatial resolution. Fibres are coated with an extramural absorber to reduce optical crosstalk. The hodoscopes are arranged in two layers, where they are displaced by half a fibre width against each other, allowing 1 mm spatial resolution.

The proton range telescope was built with 64 closely packed and optically isolated scintillating tiles. The scintillation light is collected by a wavelength shifting fibre and transmitted to a receiving photo-multiplier (PM) tube. The light is converted into an image, where the first PR of a dog is given in figure 13 (Schneider et al 2004). The image produced had a lot of noise and was produced with $10^7$ protons of 214 MeV over a period of 10 s. The head of the dog can be depicted, where a total dose of 0.03 Gy was given to produce the image, which is a factor of 50–100 smaller than for an x-ray image of the skull. However, the spatial resolution is significantly poorer in the proton radiography than in a standard x-ray image. Bony anatomy contrast is also much lower than in a standard x-ray image due to significant image noise and possibly poor coincidence recording of protons pre- and post-dog patient.

5.5.3. Silicon microstrip detectors. A preliminary study was performed (Johnson et al 2003, Schulte et al 2005, Li et al 2006) using silicon strip detectors for proton radiography. The strip detector was manufactured from a high-resistance wafer of 400 µm thickness, with a pitch of 194 µm and outer dimensions of 6.4 cm x 6.4 cm. A total of 320 strips were read out per plane, and a total of two planes, placed perpendicularly to each other, were used to generate X and Y coordinates. The proton position and energy in the 20–300 MeV range can be derived using the silicon strip detectors (SSD). Spatial resolution is ~1 mm, and temporal resolution is 1.3 ns. The proton energy is derived from the specific energy deposition in each SSD, using the time over threshold (TOT) signal. There is less energy discrimination for higher proton energies due to the reduced dependence of the energy loss on primary proton energy. In addition, at ms readout times, single-hit proton events become harder to achieve, decreasing the spatial resolution of the detector. In addition, long irradiation times are required with subsequently higher proton doses in order to improve image quality. The TOT value will also be extremely susceptible to any electronic noise in the connecting channels, possibly affecting the energy prediction of the proton.

5.5.4. CMOS detectors. The CMOS active pixel sensors (APSs) were evaluated as proton radiographic imaging detectors by Seco and Depauw (2011) at the proton facility at MGH. CMOS APS detectors have 40 × 40 µm$^2$ pixels and readout rates of 1–100 frames/s. They were shown...
to have excellent spatial resolution, but poor temporal resolution. Although promising, full-frame readout can be slow, and they are not currently sufficiently radiation hard for daily use in a proton therapy facility. Future development of CMOS APS detector proton imaging requires improved readout speed, potentially larger sizes of the detector, and radiation resistance to large doses.

5.5.5. Scintillating fibres. The use of scintillating fibres as detectors for proton imaging was initially proposed by Seco and Depauw (2010), where a X/Y mesh of commercially available fibres would detect the entrance and exit of individual protons. Further research into the use of scintillating fibres for proton imaging has been pursued because fibres offer many advantageous relative other detectors, such as (i) fast read-out in the few to tens of nano-seconds, (ii) cheap and easily available, (iii) radiation hard, etc (Sadrozinski et al 2011, Koybasi et al 2012).

5.6. Ion beam imaging

One of the potential advantages of the use of carbon ion beams over proton or photon beams for therapy is their improved lateral penumbrae. The larger mass makes carbon ions inherently harder to deflect from their path by elastic scattering of nuclei in the medium. To first order, carbon ion imaging (radiographic or computed tomography) is approximately similar to x-ray imaging because the carbon ions have an approximately linear trajectory through the patient/phantom. The linear trajectories simplify the requirements on the imaging detector relative to proton beam imaging, where, in the case of ion beams, only the exit information of the ion is required for imaging purposes. One of the first carbon ion radiographic detectors was composed of an amorphous silicon flat-panel detector that was commonly used in photon imaging. The commercially available detector from PerkinElmer Optoelectronics GmbH &Co (RID 256-L) was employed to measured 2 D water equivalent thickness and water-equivalent path length maps (Telsemeyer et al 2012). The detector had an active area consisting of matrix with pixel size 800 × 800 μm² and consists of Lanex fast phosphor screen bonded to an amorphous silicon layer deposited on glass substrate. The detector had to be corrected for sensitivity differences between separate pixels, and a signal-to-particle energy calibration had to be performed in order to obtain the true particle energy from measured images.

An independent carbon ion radiographic imaging device was developed, composed of a stack of large-area parallel-plate ionization chambers (IC) interleaved with removable absorber plates of homogeneous thickness that served as range degradation (3 mm slabs of polymethyl methacrylate, PMMA) (Rinaldi et al 2013). The active cross-section of each IC was 300 × 300 mm², and the collecting gas was 6 mm of air. An energy calibration and a Bragg peak evaluation on the target thickness study were performed on the detector in order convert the radiographic image into water-equivalent target thickness.

6. Dosimetry

The measurement of dose is clearly one of the key application areas for detectors. Measurements typically fall into two categories: absolute and relative. For the absolute determination of dose, termed reference dosimetry, measurements are performed under carefully controlled reference conditions, which are chosen to ensure transient electron equilibrium and hence stable and reproducible measurement conditions. It is very important to follow published guidelines, which carefully specify the measurement geometry, choice of detector and corrections factors to be applied. The aim of reference dosimetry is to produce accurate and consistent values relative to primary standards (typically calorimetry); this both safeguards patients by ensuring consistent delivery between hospitals and assists practice by allowing comparison of
outcomes between centres. However, to ensure safe operation of equipment, in order to check dose calculations and verify delivered dose in vivo, it is necessary to make measurements over a wide range of conditions that cover all aspects of clinical use, not just reference conditions. Such measurements are typically normalized to a reference condition (such as defining the detector response at $D_{\text{max}}$ to be 100%), and hence provide useful relative information.

### 6.1. X-rays

Extensive dosimetric measurements are needed for the acceptance and commissioning of new radiation-producing equipment, such as linear accelerators (linac) and other x-ray producing machines. For a linac, the measurement of the machine output is required to define the monitor unit – that is, the charge recorded in the machine’s primary dosimetry system, which corresponds to delivery of 1 cGy under reference conditions (typically at the isocentre in a 10 cm × 10 cm field at a well-defined depth in water and using a cylindrical ionization chamber). Sets of reference conditions are specified by IAEA report 398 (2000), the AAPM’s TG-51 report (Almond et al. 1999), Addendum to AAPM TG-51 (McEwen et al. 2014) and NPL code of practice for high-energy photon therapy dosimetry (Lillicrap et al. 1990). For determination of absorbed dose for x-rays below 300 kV, details are given in the IPEMB code of practice (IPEMB 1996). Accurate dosimetry is also essential for protecting patients against unnecessary radiation exposure in diagnostic radiology (Meghzifene et al. 2010), with well-defined codes of practice provided by IAEA (2007). Although the methods described in the codes of practice can differ slightly from one to another, a comparison of results of measurements made using a range of different codes of practice for photon beams in the range 6 to 18 MV have shown no statistically significant difference between the absorbed doses produced under reference conditions (Al-Ahbabi et al. 2012), which demonstrates that the aim of producing uniform results is met. For example, the TG-51 protocol adopted the $k_0$ formalism where linac calibration was based on $^{60}$Co absorbed dose to water. The $k_0$ factors were presented in TG-51 for 18 cylindrical chamber types used for reference dosimetry. The recent publication of the addendum to TG-51 has expanded the number of published $k_0$ factors to new reference ionization chambers developed over the last 15 years, while revising old values.

Relative measurement of beam output factors for a wide range of field sizes and shapes, beam flatness and symmetry are typically made in a water tank or water-equivalent phantom with a fixed source-to-phantom distance to ensure adequate build-up and electron equilibrium. The most commonly used detector is an air ionization chamber due to its long-term stability, high sensitivity and low energy dependence. For relative output factors, the 0.6 cc Farmer-type cylindrical ionization chamber is well-suited, with the 0.1 cc thimble chamber used when higher spatial resolution is required (such as for small fields for beam penumbra measurements). The output spectrum, or beam quality, cannot be directly measured, but relative measurement of the dose with depth from the water surface is sufficient to characterize the beam. For near-surface measurements, such as depth-dose profiles, a small active area parallel plate chamber is a common choice. Consistency in beam quality can be assessed by comparing the ratio of doses in a reference field at two depths (for example, 10 cm and 20 cm deep on the central axis of the 10 cm × 10 cm field, the TPR$_{20,10}$.) For intensity-modulated beams, where finding stable measurement conditions for taking point measurements is challenging, a range of methods have been developed that employ a mixture of point measurement, 1D and 2D detector arrays (or ionization chamber or diodes), film and electronic portal imaging; these methods are well-described in the AAPM TG-120 report (Low et al. 2011). The codes of practice specify a number of correction factors that need to be made to correct for perturbation of the dose distribution by the detector, the physical location of the effective point of measurement and the response of the detector. Insertion of an air cavity in a water medium has two
effects: reduced attenuation of the primary beam and a reduction in the scattered radiation contribution due to the ‘missing’ water. These two effects partially cancel out, since the first will cause an over-response the second under-response of the detector, with respect to the ‘true’ unperturbed dose to water. The net effect is typically an increase in signal, and is summarized by the displacement factor $p_{\text{dis}}$, which is dependent on the beam quality and detector geometry (for a Farmer chamber in $^{60}\text{Co}$ beam, the value of $p_{\text{dis}}$ is close to 0.988 for depths $d > d_{\text{max}}$) (Mayles et al 2007). Another effect of the chamber geometry is a shift in the effective point of measurement $P_{\text{eff}}$ away from the geometric centre of the detector. For an air ionization chamber the electron fluence in the chamber is equivalent to that of a point in a uniform medium shifted slightly upstream of the chamber centre. A simple geometric treatment gives a theoretical $P_{\text{eff}}$ of $8r/3\pi$ for a cylindrical cavity of radius $r$. Finally, the mass of air in the chamber itself depends on temperature and pressure, so correction for the difference in ambient temperature and atmospheric pressure between measurement and the detector calibration also needs to be applied, as specified in the relevant protocol. Relative humidity can also have a small effect on chamber reading, but is second-order, so typically ignored.

The uncertainties in the dose actually delivered to a patient in external beam radiotherapy, in addition to just the primary machine calibration, can be divided into four main groups (Mayles et al 2007):

- Uncertainty in dose determination using a calibrated ionization chamber under reference conditions following a code of practice (discussed above);
- Uncertainty arising from relative measurements, such as output factor and collimation;
- Uncertainty of dose delivered to the patients, including treatment planning system accuracy and accuracy of delivery system; and
- Uncertainty in the daily set-up of the patient and long-term machine stability.

The overall accuracy of dosimetry depends on many factors, including chamber calibration, intrinsic uncertainty in stopping power ratios, application of correction factor variations in experimental set-up. Combined standard uncertainties in determination of absorbed dose to water for megavoltage photon beams have been estimated to be about 1.5%. For a detailed discussion of uncertainties in dosimetry, readers are directed to Mayles et al (2007).

6.2. Electrons

Electron beam therapy is commonly used for treatment of skin or shallow cancers, with the use of a relatively uniform dose distribution that falls off quickly with depth. Electron dosimetry protocols (AAPM TG-51 [Almond et al 1991], Addendum to TG-51 [McEwen et al 2014], IAEA TRS-398 [IAEA 2000], IPEM [Thwaites et al 2003]) have significantly improved the dosimetry accuracy of electron beam therapy in clinics, where at standard laboratories electron dosimetry has changed from exposure/air kerma to absorbed dose in water. Further recommendations are provided by Gerbi et al (2009) for clinical electron beam dosimetry, where the use of radiochromic films for electrons is addressed.

In the case of electron beam dosimetry, the detectors used are (1) well-guarded plane-parallel ionization or (2) cylindrical chambers.

For well-guarded ionization chambers the collecting electrode is surrounded by a guard electrode having a width not smaller than 1.5 times the cavity height. Plane-parallel ionization chambers offer several advantages over cylindrical chambers when performing percent depth dose (PDD) and output factor measurements such as (1) replacement perturbation correction factor, $P_{\text{rep}}$, is unity, (2) they are built to minimize scattering perturbation effects and (3) the effective point of measurement is the same as the point of measurement. These chambers are recommended as reference dosimeters for beam quality evaluations. However, great care
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is required when using plane-parallel ionization chambers in a water phantom, and special attention should be paid to polarity effects because of its dependence on energy and angular distribution of incident beam. The measurement of central-axis depth doses is commonly performed using the cylindrical ionization chambers. Positioning the cylindrical requires special attention because the effective point of measurement does not correspond to the point of measurement (effective point is 0.5 \( r_{cav} \) upstream of the measurement point, where \( r_{cav} \) is the chamber radius). At depths greater than 0.5 cm, cylindrical chambers are as accurate as plane-parallel ionization chambers in dose measurements.

For relative electron beam dosimetry (1) film dosimetry, (2) diodes, (3) cylindrical and (4) plane-parallel ionization, chambers are commonly used. In addition to cylindrical and plane-parallel ionization chambers (discussed previously), radiographic films and diodes can also be used to obtain relative dose distributions (TG 25, Khan 1991, TG 69, Pai et al 2007). Radiographic films are commonly used in non-water phantoms to measure central-axis depth doses and isodose distributions of regular and irregular fields, to identify \( d_{\text{max}} \) and the depths of 90%, 80% and 50% dose levels. The depth doses are measured by placing the radiographic films parallel to the central axis of the beam under the required treatment setup. In addition, the relative output factor of the electron field under evaluation can be obtained with film by performing a film measurement for the reference cone and field. Kodak EDR, Kodak XV-2 and XTL films are commonly used to measure uniformity and width of isodose distributions at various depths in the electron beam (such as \( d_{\text{max}} \), \( R_{50} \) or \( R_{90} \), depth at which the absorbed dose falls to respectively 50% or 90% of the maximum dose). Diodes can also be used for depth dose measurements, where commonly the effective point of measure is the die, the active component of the diode. If the effective point of measurement is not known, TG-25 (Khan 1991) should be used to identify it. Diodes used for photon beam therapy are not suitable for electron beam therapy, because of the additional material placed on them to increase their sensitivity to photons. So care is needed when selecting diodes for electron beam therapy. In addition, diodes should always be initially tested relative to ionization chamber measurements.

6.3. Protons

This section describes how detectors are used in proton therapy for measuring the following beam properties: longitudinal and lateral beam profiles, fluence or number of protons, and absorbed dose. The proton relative biological effectiveness (RBE) is not discussed here, and an overview of this topic can be found in Paganetti (2012).

6.3.1. Longitudinal measurements. Parallel plate ionization chambers (PPIC) are commonly used to measure dose to water versus depth. These detectors offer simplicity, minor corrections for beam quality vs. depth (IAEA 2004), and high longitudinal resolution with the detector face perpendicular to the beam. Cylindrical chambers are normally avoided for longitudinal measurements because of the rapid increase and decrease of dose with depth at the Bragg peak. The signal obtained from PPIC versus depth are strictly depth-ionization curves and energy-dependent corrections for stopping power, mean energy to produce an ion pair, and wall perturbation should be applied to obtain depth-dose distributions (Paganetti 2012).

Important properties of the beam extracted from depth-dose distributions include range, entrance dose, peak width, and distal fall-off. Proton range is the depth of water at the deepest point on the depth dose curve that crosses a specified percentage of the maximum dose. Range specified at 80% of the maximum dose is useful because it is almost independent of the energy spread of the incident protons. Range specified at 90% of the maximum dose is clinically useful because the target volume and margins are covered with the 90% isodose line.
Detector configurations for longitudinal measurements usually have the detector-sensitive volume in regions that have lateral proton scatter equilibrium or with the detector sensitive volume, encompassing the entire lateral extent of the beam. The first configuration uses a small detector radius in a wide radiation field, and the second uses a wide detector radius in a narrow radiation field. The measured result in the first configuration gives dose [Gy] from the detector as a function of depth per incident proton fluence (protons per square cm) from the machine. The measured result in the second configuration gives dose integrated over area [Gy.cm²] from the detector as a function of depth per incident number of protons from the machine. Both configurations therefore yield depth-dose distributions with the same units and equivalent results in the case of infinite source-to-axis distance. Gottschalk (2004) highlights the complications that can arise when using an intermediate-sized detector for longitudinal measurements.

Multi-layered devices such as ion chambers (Shimbo et al 2000, Cirio et al 2004, Mumot et al 2010, Dhanesar et al 2012) and Faraday cups (Gottschalk et al 1999, Paganetti and Gottschalk 2003, Gottschalk 2004, Hsi et al 2009) are used to measure longitudinal properties of the beam with a single acquisition, significantly reducing the time needed for measurements. Multi-layer ionization chambers have alternating layers of water-like material and ionization chamber volumes, and give dose versus depth. Multi-layer Faraday cups have alternating layers of conductive metal plates and electrical insulators. The signals from the metal plates produce are proportional to the depth where protons stop, which can be used to extract the proton range (figure 14).

6.3.2. Lateral dose measurements. Ionization chambers on linear translation stages or ionization chamber arrays (Nichiporov et al 2007, Arjomandy et al 2008) can be used to measure lateral dose profiles, with the caveat that corrections for proton energy should be applied for cases where proton energy is changing with lateral position in a phantom (Paganetti 2012). These corrections are generally small for ionization chambers (<1% for proton energies between 15 and 300 MeV (Paganetti 2012)). Profiles can also be measured with high resolution with film (Niroomand-Rad et al 1998, Karger et al 2010), polymer gel (Jirasek and Duzenli 2002, Heufelder et al 2003, Baldock et al 2010, Zeiden 2010), and scintillating material coupled with a camera (Boon et al 1998, Boon et al 2000, Safai et al 2004). There has been recent development to reduce the

Figure 14. Simulated response of a Multilayer Faraday cup in a proton beam using GEANT4 (Agostinelli et al 2003). Although this is not a Bragg peak, the signal in each plate indicates where the protons stop. The signal is almost zero in the entrance region because most protons pass through the first plates without stopping. The peak in the signal indicates the plate where most of the protons stop.
energy-dependent response of these detectors with uncertainties of 5% for gel (Zeiden 2010), 10% for GafChromic™ film (ICRU 2007, Zhao and Das 2010), and no observable uncertainty demonstrated at a single proton energy for scintillating material (Safai et al 2004).

Detector systems that combine lateral and longitudinal measurements are film stacks (Troja et al 2000, Kim et al 2012), polymer gels, and ionization chambers (Karger 1999, Cirio et al 2004).

6.3.3. Fluence. Faraday cups are used to measure the number of protons in a beam and are useful during commissioning and monitoring of beam currents. Faraday cups have also been used to determine absorbed dose, but this is not recommended (ICRU 2007), and calorimeters and ionization chambers are used for this purpose (see figure 14).

6.3.4. Absorbed dose in water. Water calorimeters are the gold standard for absolute dosimetry in proton beams. These detectors provide dose to water with the smallest uncertainty, typically 0.4 to 1.4% (Seuntjens and DuSautoy 2003, Brede 2006, Sarfelnia et al 2010). On the other hand, these detectors are rather large and complex, and are not practical for routine measurements; ionization chambers are more suitable for this purpose. Ionization chambers provide fast and precise measurements of absorbed dose, but one should follow an established protocol such as IAEA TRS-398 (IAEA 2004). ICRU 2007 (ICRU 2007) recommends the adoption of IAEA TRS-398 rather than ICRU 59 (ICRU 1998) due, in part, to the simplicity, better accuracy, and the already widespread use of this protocol.

The TRS-398 (IAEA 2004) protocol uses ionization chambers calibrated at a standards laboratory in 60-Co radiation to obtain absolute dose in water. The protocol recommends the use of cylindrical or parallel plate chambers, although the uncertainty is larger for parallel plate chambers due to the non-water equivalence of the chamber walls. The standards laboratory provides the absorbed dose-to-water calibration factor, \( N_{D_w,Q} \), for a particular chamber, and the absorbed dose is given by

\[
D = M_Q N_{D_w,Q} Q_{D_w,Q} \tag{32}
\]

where \( M_Q = M_{raw} k_{TP} k_{elec} k_{pol} k_s \) is the raw meter reading \((M_{raw})\) corrected for influence quantities. The influence quantities are: \( k_{TP} \) the pressure and temperature correction factor, \( k_{elec} \) the electrometer calibration factor, \( k_{pol} \) the ion chamber polarity correction factor, and \( k_s \) the recombination correction factor. \( Q_{D_w,Q} \) is the beam quality correction factor between protons and 60-Co, and is provided in the protocol.

6.4. Carbon ions

Prescriptions in carbon ion therapy are specified by biological effective dose, which is the product of the relative biological effectiveness (RBE) and absorbed dose. The RBE describes the response of cells relative to the reference photon radiation and can be as large as 2 to 4 (Kanai et al 1997, Krämer and Scholz 2000, Kempe et al 2007, Wilkens and Oelfke 2008, Elsässer 2008). While knowledge of the RBE and absorbed dose are both important for carbon ion therapy, this section only describes the detectors used in the measurement related to absorbed dose and dose profiles.

Calorimeters provide the most direct approach to measure the absorbed dose (Karger et al 2010). Corrections for the heat defect, which depend on linear energy transfer (LET), become more difficult to apply to carbon ions because of the varying LET in these beams. Brede et al (2006) reports a standard uncertainty of 1.8% in the measurement of absorbed dose for a water calorimeter.

The IAEA TRS-398 protocol (IAEA 2004) has been widely adopted for measuring absorbed dose with ionization chambers (Karger et al 2010). The preferred set-up in this protocol is a graphite walled cylindrical chamber for spread-out Bragg peak (SOBP) widths \( \geq 2 \text{ g cm}^{-2} \).
The carbon ion therapeutic depth-dose distribution is not flat in the high-dose region, so a gradient correction should be applied to cylindrical chamber measurements corresponding to a shift of the effective point of measurement by 0.75 times the inner radius of the ionization chamber cavity. The protocol recommends parallel plate ionization chambers for SOBP widths < 2 g cm$^{-2}$. As in proton therapy, the absorbed dose follows from

$$D = M_0N_{0,w,Q_{0}}K_{Q_{0}},$$  \hspace{1cm} (33)

where values for $k_{Q_{0},0}$ are given in the TRS-398 protocol and calculated due to the lack of experimental data. The total standard uncertainty is 3.0% and 3.4% for cylindrical and parallel plate ionization chambers, respectively, with the largest uncertainty arising from $k_{Q_{0},0}$.

The measurements of longitudinal and lateral dose profiles for carbon ions uses detectors similar to those used for proton therapy. Parallel plate ionization chambers can be used to measure longitudinal relative dose distributions (Karger et al 2010). Lateral relative dose profiles are measured with ionization chambers or, with higher resolution, by radiographic or radiochromic films, scintillating screens or silicon detectors. More information on the commissioning of heavy ion systems with these detectors can be found in (Ma and Lomax 2013).

7. Conclusions

This review has focused on describing the range of technologies used for detecting ionizing radiation in the medial field. The metrics used to characterize performance of detectors and imaging systems have been briefly presented, followed by a description of the physical principles of operation of the major types of detector. The practical application and typical uses of these detectors, either as part of an imaging system or discretely, were then discussed.

Since Röntgen’s discovery (initially by fluorescence of a barium platinocyanide screen and, shortly after, by film exposure), x-rays – and therefore x-ray detectors – have been central to diagnostic medicine. Over the past 20 years, radiographic film has largely been replaced with digital x-ray imaging systems, and x-ray CT has been complemented more and more with MRI and PET. Advances of these systems have driven research into ever larger and more sensitive multi-slice CT detectors and faster and more sensitive gamma ray detectors for PET with time-of-flight and depth-of-interaction capabilities continuing to be developed. The combination of modalities into hybrid detectors is also an important driver of detector technology, such as the development of PET/MR, where the PET detector has to work inside the high magnetic field of the MR system, making use of conventional photomultiplier tubes impossible. Future advances in hybrid imaging systems therefore depend on advances in detector technology.

Advances in radiotherapy in recent years have also challenged detector technology. The use of online image guidance using megavoltage portal imaging of kV cone beam CT devices is now routine, but keeping the radiation exposure to the patient as low as reasonably practicable with increasing imaging frequency demands higher-sensitivity imaging detectors. The radiation beams typically applied during modern intensity-modulated or arc therapies are now typically highly modulated and can be comprised of quite small field segments. The need for accurate dose verification of these complex and sometimes small-area beams has led to the development of a range of dosimetry systems. The increasing development of proton therapy will continue to challenge detector developers for dosimetry and imaging.

It is probably safe to say that the use of ionizing radiation will continue to play a central role in medicine for years to come. Further advances in imaging and therapy will both be driven by advances in detector technology and will also drive advances in detector technology. This will enable new types of measurement or improvements in imaging and will continue to ensure the safety of staff and patients.
Acknowledgements

MP gratefully acknowledges funding from Cancer Research UK, grant reference C5255/A15935.

References

Akamatsu G et al 2012 Improvement in PET/CT image quality with a combination of point-spread function and time-of-flight in relation to reconstruction parameters J. Nucl. Med. 53 1716–22
Almond P R et al 1999 AAPM’s TG-51 protocol for clinical reference dosimetry of high-energy photon and electron beams Med. Phys. 26 1847–70
Aspradakis M M (ed) 2010 Small Field MV Photon Dosimetry (UK: IPEM)
Attix F H 1986 Introduction to Radiological Physics and Radiation Dosimetry (New York: John Wiley and Sons)
Baumgartner A et al 2011 Re-evaluation of correction factors of a primary standard graphite calorimeter in 60Co gamma ray beams as a basis for the appointment of the BEV absorbed dose rate to water reference value Radiat. Prot. Dosim. 145 3–12
Cascio E and Gottschalk E 2009 A simplified vacuumless faraday cup for the experimental beamline at the Francis H Burr Proton therapy center Radiation Effects Data Workshop IEEE pp 161–5


Cormack A M 1963 Representation of a function by its line integrals, with some radiological applications *J. Appl. Phys.* **34** 2722–27.


Fricke H and Morse S 1929 The action of x-rays on ferrous sulphate solutions *Phil. Mag.* **7** 129–41.


Gottschalk B 2004 Techniques of Proton Radiotherapy internet URL address is the following http://users.physics.harvard.edu/~gottschalk/BGtalks.zip

Hopwood F L and Phillips J T 1939 Polymerization of liquids by irradiation with neutrons and other rays Nature 143 640
Horowitz Y S and Moscovitch M 2013 Highlights and pitfalls of 20 years of application of computerised glow curve analysis to thermoluminescence research and dosimetry Radiat. Prot. Dosim. 153 1–22
Huyskens D P 2001 Practical guidelines for the implementation of in vivo dosimetry with diodes in external radiotherapy with photon beams (entrance dose) (Belgium: ESTRO Mounierlaan 83/12 – 1200 Brussels)
IAEA Human Health Reports, No. 8 2013 Development of procedures for in vivo dosimetry in radiotherapy (Vienna: International Atomic Energy Agency)
International Commission on Radiation Units and Measurements 2007 Prescribing, recording, and reporting proton-beam therapy ICRU report 78
Jirasek A and Duzenli C 2002 Relative effectiveness of polyacrylamide gel dosimeters applied to proton beams: Fourier transform Raman observations and track structure calculations Med. Phys. 29 569–77
Kalender W 2000 Computed Tomography (Munich: Publicis MCD Verlag)
Kanai T et al 1997 Irradiation of mixed beam and design of spread-out Bragg peak for heavy-ion radiotherapy Radiat. Res. 147 78–85
Kasap S et al 2011 Amorphous and polycrystalline photoconductors for direct conversion flat panel x-ray image sensors Sensors (Basel) 11 5112–57
Kempe J, Gudowska I and Brahme A 2006 Depth absorbed dose and LET distributions of therapeutic 1H, 4He, 7Li, and 12C beams Med. Phys. 34 183–192
Khan M 2010 The Physics of Radiation Therapy 4th edn (Baltimore, MD: Lippincott-Williams & Williams)
Klassen N V et al 1999 Fricke dosimetry: the difference between G(Fe3+) for 60Co gamma-rays and high-energy x-rays Phys. Med. Biol. 44 1609–24
Koehler A M 1968 Proton radiography Nature 245 38–40
Koehler A M and Steward V W 1973 Proton beam radiography in tumor detection Science 179 913–4
Li T et al 2006 Reconstruction for proton computed tomography by tracing proton trajectories—a Monte Carlo study Med. Phys. 33 699–706
Lonsdale A P 2012 Multistage evaluation and commissioning of a pre-calibrated, single-use OneDosePlus MOSFET system for in vivo dosimetry in a radiotherapy department Br. J. Radiol. 85 451–7
Ma C M C and Lomax T ed 2013 Proton and Carbon Ion Therapy (Boca Raton: CRC)
McEwen M et al 2014 Addendum to the AAPM’s TG-51 protocol for clinical reference dosimetry of high-energy photon beams Med. Phys. 41 041501
Mehgzieneh A et al 2010 Dosimetry in diagnostic radiology Eur. J. Radiol. 76 11–4
Mizuno H et al 2008 Feasibility study of glass dosimeter postal dosimetry audit of high-energy radiotherapy photon beams Radiother. Oncol. 86 258–63
Sadrozinski H F et al 2011 Detector development for proton computed tomography (pCT) IEEE Nuclear Science Symp. and Medical Imaging Conf. (NSS/MIC) 4457–61
Schneider U et al 2004 First proton radiography of an animal patient Med. Phys. 31 1046–51
Seco J and Depauw N 2010 Proof of principle study of the use of a CMOS active pixel sensor proton radiography Med. Phys. 38 622–4
1996 The IPEMB code of practice for the determination of absorbed dose for x-rays below 300kV generating potential (0.035 mm Al-4 mm Cu HVL; 10–300kV generating potential). Institution of Physics and Engineering in Medicine and Biology Phys. Med. Biol. 41 2605–25
Venables K et al 2004 The use of in vivo thermoluminescent dosimeters in the quality assurance programme for the START breast fractionation trial Radiat. Oncol. 71 303–10
Webb S and Flower M A 2012 Webb’s Physics of Medical Imaging, 2nd edn Series in medical physics and biomedical engineering (Boca Raton: Taylor & Francis)
Wilson R R 1946 Radiological use of fast protons Radiology 47 487–91
Yang Y et al 2008 A prototype PET scanner with DOI-encoding detectors J. Nucl. Med. 49 1132–40