Topical Review

Advances in kilovoltage x-ray beam dosimetry

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Abstract

This topical review provides an up-to-date overview of the theoretical and practical aspects of therapeutic kilovoltage x-ray beam dosimetry. Kilovoltage x-ray beams have the property that the maximum dose occurs very close to the surface and thus, they are predominantly used in the treatment of skin cancers but also have applications for the treatment of other cancers. In addition, kilovoltage x-ray beams are used in intra operative units, within animal irradiators and in on-board imagers on linear accelerators and kilovoltage dosimetry is important in these applications as well. This review covers both reference and relative dosimetry of kilovoltage x-ray beams and provides recommendations for clinical measurements based on the literature to date. In particular, practical aspects for the selection of dosimeter and phantom material are reviewed to provide suitable advice for medical physicists. An overview is also presented of dosimeters other than ionization chambers which can be used for both relative and in vivo dosimetry. Finally, issues related to the treatment planning and the use of Monte Carlo codes for solving radiation transport problems in kilovoltage x-ray beams are presented.

Keywords: kilovoltage x-ray beams, dosimetry, ionization chambers, phantoms, Monte Carlo methods
1. Introduction

X-rays were first discovered by Wilhelm Roentgen in 1895 for which he was awarded the first Nobel prize for physics in 1901 (Roentgen 1895). In a relatively short time, it was discovered that ionizing radiation could be used to treat various benign and malignant medical conditions which prompted the rapid development of x-ray units. Therapeutic kilovoltage x-ray units
Figure 1. Percentage depth dose curves for kilovoltage x-ray beams with energies 50–280 kVp at an SSD of 30 cm.

operate with peak voltages in the range from 10–400 kVp with the most common x-ray units operating at energies 50–150 kVp (Klevenhagen et al. 2000).

The main dosimetric property of kilovoltage x-ray beams is that the maximum dose occurs close to the surface of the patient, within the first few mm, and the dose drops off rapidly with depth due to attenuation and scattering of the beam. This is evident from depth doses measured in water using a PTW Advanced Markus ionization chamber for x-ray beams in the energy range 50–280 kVp presented in figure 1.

Historically, the different types of kilovoltage x-ray beams were classified according to their peak potential as follows:

(i) Grenz rays—beam potentials from 10 to 20 kVp.
(ii) Contact therapy—beam potentials of up to 50 kVp.
(iii) Superficial therapy—beam potentials of 50 to 150 kVp.
(iv) Orthovoltage or deep therapy—beam potentials from 150 to 500 kVp.

Prior to the invention of Cobalt-60 treatment units and linear accelerators, kilovoltage x-ray beams were extensively used for the treatment of many forms of cancer including deep seated tumours (Biggs et al. 1999, Poen 1999). Due to the inadequate penetration of kilovoltage x-rays through the patient, multiple beams from different angles were required in order to maximize the tumour dose while trying to minimize high surface doses. However, the development of linear accelerators to produce megavoltage x-ray beams has eliminated the treatment of deep-seated tumours with kilovoltage x-ray beams.

Nowadays, the primary clinical use for kilovoltage x-rays is the treatment of skin cancers (Amdur et al. 1992, Caccialanza et al. 2009, Locke et al. 2001, Poen 1999). Radiotherapy is often delivered after surgery where the tumour has been surgically excised and the radiation is used to eliminate any cancer cells remaining within the tumour bed. Therapeutic kilovoltage x-rays are also used for the treatment of both AIDS-related and non-AIDS-related Kaposi’s sarcoma, rectal cancer treatments, electronic brachytherapy systems, intraoperative

Kilovoltage x-ray beam dosimetry provides a number of challenges which are not present for megavoltage x-ray beams (Nahum 1996). Firstly, as most dosimeters have a relatively large dimension in the depth direction, the rapid fall off of dose with depth means there can be a significant dose gradient over the measuring volume of the dosimeter. Another issue is that dosimeter response is sensitive to the materials used in its construction. For low energy x-ray beams, the photoelectric effect is a dominant interaction process and the photoelectric cross section has a strong dependence on the atomic number of the material. The third challenge is that ionization chambers do not act as Bragg–Gray cavities in the kilovoltage x-ray energy range and so cavity theory cannot be used for reference dosimetry (Ma and Nahum 1991).

In this review paper, we discuss a range of aspects of the dosimetry for therapeutic kilovoltage x-ray beams as used in the radiation oncology. It is noted that there is an overlap with dosimetry of imaging kilovoltage beams which have become more important given the wide availability of planar and cone beam CT imaging systems in the radiation therapy. However, the primary focus of this review paper is the dosimetry of therapeutic kilovoltage x-ray beams.

2. Beam quality specification

2.1. Half-value layer

Knowledge of the beam quality is a precursor to any reference dosimetry (or absolute dose calibration) of kilovoltage x-ray beams. Direct measurement of the x-ray beam spectrum is a difficult task and not readily achievable in the clinic (Ma et al 2001, Seuntjens et al 1987a). In current dosimetry protocols, the beam quality is specified in terms of the half-value layer (HVL) in combination with the peak generating voltage (kVp) (Andreo et al 2000, Klevenhagen et al 1996, Grimmergen et al 1997, Ma et al 2001). The HVL is defined in terms of the thickness of absorber (typically high purity aluminium or copper) which reduces the air-kerma rate by a factor of one half (Klevenhagen et al 1996). Many of the parameters used in the reference dosimetry calculations are defined as a function of the HVL.

A Farmer type ionization chamber is typically recommended for HVL measurement and absolute calibration except for lower energy x-ray beams (typically less than 50 kVp), where a low energy parallel-plate chamber is recommended (Klevenhagen et al 1996, Ma et al 2001). The air-kerma calibration of the ionization chamber used for the HVL measurement should vary smoothly and by less than 5% over the energy range concerned.

According to dosimetry protocols, the HVL should be measured under scatter free and narrow beam conditions. Scatter free conditions are achieved by placing the ionization chamber at a distance from the source, ensuring distances from detector to walls, floor or ceiling are maximized and placing the absorber at half the distance between the source and detector. The source-detector distance is typically 80–100 cm, and where possible should be consistent with the source-detector distance used by the primary or secondary standards dosimetry lab when they perform their calibrations. Narrow beam geometry can be achieved either by using a small size applicator or a lead sheet with a small cutout. The size of the collimated beam
should be such that it covers the ionization chamber with a small margin around the chamber of 5–10 mm, for instance.

Good practice in HVL measurement includes exposing a piece of film behind the detector to ensure the ionization chamber is correctly positioned in the radiation field and ensuring the chamber axis is perpendicular both to the filament-target direction of the x-ray tube and the beam central axis to avoid the heel effect. During beam commissioning, it is common to measure both the first and second HVLs for each radiation beam for quality assurance purposes.

2.2. Other beam quality specifiers

The beam quality for a kilovoltage x-ray beam cannot be completely described by the HVL only (Ma et al 2001). It has been demonstrated that for clinical x-ray beams with the same measured HVL, the beams had a wide range of beam potentials and vice versa. This limitation has been addressed in the NCS reference dosimetry protocol by listing parameters in terms of both HVL and kVp (Grimbergen et al 1997). However, to date there has not been any alternative beam specifier found to replace the HVL.

Several beam quality specifiers were investigated by Rosser (1998) as an alternative quality index that would correlate with the ratio of mass–energy absorption coefficients of water to air at 2 cm depth in water, \( \left( \frac{\mu_{\text{en}}}{\rho} \right)_{\text{water}} \left( \frac{\mu_{\text{en}}}{\rho} \right)_{\text{air}} \). The three parameters investigated were the HVL, the mean energy at a depth of 2 cm and the ratio of the doses at depths of 2 and 5 cm in water. The ratio of doses showed the best correlation but has not been widely implemented with the HVL still the dominant beam quality specifier.

More recently, Chica et al (2014) investigated the relationship between ionization chamber calibration factors and the ratios of absorbed doses at two depths in water by using Monte Carlo calculations. They found that the relationship was almost linear. Therefore, the determination of superior beam quality specifiers for kilovoltage x-ray beams requires additional investigation.

2.3. X-ray spectral determination

The direct measurement of the primary and/or scattered x-ray beam spectra is a difficult process and requires specialist equipment (Birch and Marshall 1979, Mainardi and Bonzi 2008, Seuntjens et al 1987a, Tucker et al 1991a). However, if beam spectra information is required, it is usually sufficient to use either published spectral data or an analytical calculation (Birch and Marshall 1979, Chica et al 2008, Ma and Seuntjens 1999, Poludniowski 2007, Poludniowski and Evans 2007, Poludniowski et al 2009, Tucker et al 1991b).

Alternatively, one can determine the primary x-ray spectra by using Monte Carlo calculations. However, this involves determining a complete model of the x-ray unit and requires a good knowledge of the geometry and materials in the x-ray tube (Ali and Rogers 2008a, 2008c, Mainegra-Hing and Kawakow 2006, Mesbahi and Zakariae 2013, Munck Af Rosenschold et al 2008, Omrane et al 2003).

A number of studies have also investigated methods to derive primary x-ray beam spectra from the measured transmission data (Archer and Wagner 1988a, 1988b, Delgado 1999, 2007, Du et al 2006, Mainardi and Bonzi 2008, Sidky et al 2005, Waggener et al 1999). The derivation of the spectra requires the use of mathematical techniques to solve what is a very difficult mathematical problem (Sidky et al 2005). Nevertheless, many of these studies gave results that were in a reasonably good agreement with either measured beam spectra, published data or Monte Carlo calculations.
3. Reference dosimetry of kilovoltage x-ray beams

3.1. Primary standards for low energy x-ray beams

For low energy x-ray beams, the free-air ionization chamber (FAC) is considered the reference instrument for the determination of air-kerma (Attix 2004, Burns and Büermann 2009, Lye et al 2010b, Mayles et al 2007, Rapp et al 2013, Snow et al 2013). It is known as a FAC because it relies on the principle that the walls of the chamber do not influence the measurement of charge. This requires that the FAC has a sufficiently large size for the air cavity and larger than the range of electrons in air. The FAC can be used to directly measure air-kerma (or exposure) as the size of the ionization chamber is such that charged particle equilibrium is achieved (Buhr et al 2012, Burns and Büermann 2009, Mayles et al 2007). Due to the large size and complex use of the FAC, they are predominantly found in primary standard dosimetry laboratories (PSDLs) but are too bulky and difficult to use in radiotherapy clinics.

Water calorimeters are used by a number of national standard laboratories as the absorbed dose primary standards for either Cobalt-60 or megavoltage x-ray beams (Seuntjens and Duane 2009). However, there has been recent work in developing water calorimeters for use with kilovoltage x-ray beams (de Prez and de Pooter 2008, Krauss et al 2012, Rapp et al 2013). It is expected that absorbed dose standards for kilovoltage x-ray beams will continue to be established in standards laboratories around the world.

3.2. Ionization chamber dosimetry

In the clinic, the Farmer type cylindrical ionization chamber is considered to be the gold standard for reference dosimetry of radiotherapy x-ray beams including kilovoltage x-ray beams (Attix 2004, Mayles et al 2007). The main reason for this is that the Farmer chamber has an energy response variation within 2–3% for beam energies ranging from 50–300 kVp (Andreo et al 2000, Mayles et al 2007). In addition, Farmer chambers are robust, easy to use, stable in their response and can be used for a wide range of dosimetry measurements (Mayles et al 2007). This is supported by the reference dosimetry protocols which state that the gold standard dosimeter for kilovoltage x-ray beam dosimetry is the Farmer chamber (Andreo et al 2000, Aukett et al 2005, IPSM 1991, Klevenhagen et al 1996, Ma et al 2001). However, for the very low energy x-ray beams, typically less than 50 kVp, the gold standard detector is a low energy x-ray parallel plate ionization chamber (Andreo et al 2000, Klevenhagen et al 1996).

For megavoltage x-ray beams, the absorbed dose to a medium such as water can be measured with an ionization chamber by the use of Bragg–Gray or Spencer-Attix cavity theory (Attix 2004, Mayles et al 2007). However, the Bragg–Gray principle does not apply for kilovoltage x-ray beams as the range of secondary electrons within the medium is very small (Mayles et al 2007). It was shown by Ma and Nahum (1991) that for beam energies less than 240 kV, up to 30% of the ionization within the air cavity was due to photon interactions that occurred within the air itself and not from the surrounding medium.

For these reasons, the determination of absorbed dose to water for kilovoltage x-ray beams is based on using an ionization chamber calibrated in terms of air-kerma, $K_{air}$, or exposure $X$. It should be noted that PSDLs no longer provide calibration factors in terms of exposure but rather in terms of air-kerma.

Since the absorbed dose is almost equal to the kerma in the medium for kilovoltage x-ray beams, one is able to assume charge particle equilibrium (CPE) and directly relate the kerma in air to the dose in air and finally convert this to the dose in the water. The different methods of calibration will now be discussed in more detail.
3.3. Air kerma calibrations


The two most commonly used air-kerma calibration protocols are those published in the Task Group 61 report from the American Association of Physicists in Medicine (AAPM) and the Code of Practice from the Institution of Physics and Engineering in Medicine and Biology (IPEMB) (Klevenhagen et al. 1996, Ma et al. 2001). Both of these protocols recommend air-kerma calibrations for kilovoltage x-ray beams in the energy range from 40–300 kVp and provide a list of parameters required for use in the calibration.

There are two methods for performing air-kerma calibrations being the in-air method and the in-phantom method. These methods will now be discussed in more detail.

3.3.1. The in-air calibration method. For kilovoltage x-ray beams, CPE exists at the surface of a water phantom and so one can determine the absorbed dose by taking in-air measurements and applying backscatter factors (BSFs) (Aukett et al. 2005, Grimbergen et al. 1997, Klevenhagen et al. 1996, Ma et al. 2001, Mayles et al. 2007). In this method, the absorbed dose is determined by placing the ionization chamber at the end of the applicator, free-in-air, to first measure $K_{air}$. This is converted to the dose at the surface of a water phantom by using two factors: the mass–energy absorption coefficient ratio of water to air.

The following equation is used to determine the dose at the surface of a water phantom $D_{w,z=0}$:

$$D_{w,z=0} = MN_K B_w \left[ \left( \frac{\nu_{en}}{\rho} \right)_{\text{water,air}} \right]_{\text{air}}$$

where

- $M$ is the reading of the ionization chamber converted to standard conditions for temperature and pressure,
- $N_K$ is the air-kerma calibration factor of the ionization chamber for the particular HVL of the radiation beam,
- $B_w$ is the BSF in water for the particular HVL, field size and source to surface distance (SSD) and
- $\left[ \left( \frac{\nu_{en}}{\rho} \right)_{\text{water,air}} \right]_{\text{air}}$ is the mass–energy absorption coefficient ratio of water to air for the primary x-ray spectrum only which is also known as the condition of free-in-air.

This calibration method can be used over a wide energy range from 40 to 300 kVp depending on the particular protocol being used (Aukett et al. 2005, Klevenhagen et al. 1996, DIN 1988, Ma et al. 2001). It should be noted that the earlier IAEA TRS-277 dosimetry protocol provided a similar methodology for kilovoltage x-ray beams with energies less than 100 kVp but uses slightly different nomenclature (IAEA 1987).

3.3.2. The in-phantom calibration method. The in-phantom method for reference dosimetry of kilovoltage x-ray beams involves the determination of the dose in a water phantom at a
reference depth of typically 2 cm (Ma et al. 2001, Klevenhagen et al. 1996). This method is recommended for x-ray beams with an HVL between 0.5 to 4 mm of Cu or peak potential ranging from 160 to 300 kVp depending on the dosimetry protocol (Klevenhagen et al. 1996, Ma et al. 2001). The ionization chamber is positioned at a depth of 2 cm within the water phantom and the dose, $D_{w,z=2}$, at this point is determined by the following equation:

$$D_{w,z=2} = MN_k k_{ch} \left( \frac{\mu_{en}}{\rho} \right)_{\text{water}, \text{air}} z=2, \phi$$

where

- $M$ is the reading of the ionization chamber converted to standard conditions for temperature and pressure,
- $N_k$ is the air-kerma calibration factor for the chamber at the particular HVL of the radiation beam,
- $k_{ch}$ is a correction factor that takes into account the change in response of the ionization chamber from the calibration in air to the measurements within the water phantom and
- $\left( \frac{\mu_{en}}{\rho} \right)_{\text{water}, \text{air}} z=2, \phi$ is the mass–energy absorption coefficient ratio of water to air, for the x-ray spectrum in water at a depth of 2 cm for the particular diameter, $\phi$, of the radiation beam.

The correction factor $k_{ch}$ combines a number of different effects for in-phantom measurements (Klevenhagen et al. 1996, Ma and Nahum 1995a, 1995b, Seuntjens et al. 1993, Seuntjens and Verhaegen 1996). These effects are: the displacement of the water by the ionization chamber; the stem effect of the ionization chamber; the changes in both the x-ray energy spectrum and angular distribution of the photons between the in-air and in-phantom setup and the conversion of air-kerma at the centre of the cavity to the air-kerma at the centre of the water that is replaced by the chamber.

The earlier IAEA TRS-277 dosimetry protocol provided a similar methodology for x-ray beams with energies between 100–280 kVp (IAEA 1987). However, this protocol used a perturbation correction factor $p_u$ which considered the replacement of the water by the ionization chamber and other factors. The values of $p_u$ in TRS-277 ranged in value from 1.01–1.10 for beam qualities with a HVL ranging from 0.17 to 3.37 mm Cu. However, these were subsequently amended to be close to 1.0 and hence the original values of $p_u$ should no longer be used (IAEA 1993, Seuntjens et al. 1988).

### 3.3.3. Very low energy x-rays

For very low energy x-ray beams, with a corresponding HVL of 0.035–1.0 mm Al (or 8–50 kVp), the dose calibration is performed using a low energy parallel plate ionization chamber. The calibration is performed in a phantom with the chamber positioned so that the front face is at the surface of the phantom. The absorbed dose to water at the surface of the phantom, $D_{w,z=0}$, is given by the following equation (Klevenhagen et al. 1996):

$$D_{w,z=0} = MN_k k_{ch} \left( \frac{\mu_{en}}{\rho} \right)_{\text{water}, \text{air}} z=0, \phi$$

where $\left( \frac{\mu_{en}}{\rho} \right)_{\text{water}, \text{air}} z=0, \phi$ is the mass–energy absorption coefficient ratio of water to air, for the x-ray spectrum at the surface of the water phantom and $\phi$ is the diameter of the radiation beam and the other factors are as previously defined.

The values of $k_{ch}$ for parallel plate chambers were initially taken to be unity due to the limited data at the time of publication (Klevenhagen et al. 1996). However, the published
addendum to the IPEMB protocol includes updated $k_{ch}$ values of up to 1.10, based on later calculations and these are now the recommended values to be used in such calibrations (Aukett et al 2005, Ipe et al 2001, Perrin et al 2001).

3.4. IAEA TRS-398 protocol

The IAEA published an updated reference dosimetry protocol, TRS-398, entitled ‘Absorbed Dose Determination in External Beam Radiotherapy: An International Code of Practice for Dosimetry based on Standards of Absorbed Dose to Water’ (Andreo et al 2000). This dosimetry protocol is based on a standard of absorbed dose to water and is applied to a wide range of radiation beams including kilovoltage x-ray beams (Seuntjens et al 1999).

The formalism in the TRS-398 protocol is such that the absorbed dose to water, $D_{w, Q}$, for a radiation beam of quality $Q$ is given by the following equation:

$$D_{w, Q} = M_Q N_{D_{w, Q_0}} k_{Q_0, Q}$$

(4)

where $M_Q$ is the dosimeter reading corrected for the influence quantities such as air temperature and pressure to their reference values, $N_{D_{w, Q_0}}$ is the calibration factor for the dosimeter in terms of absorbed dose to water for a reference beam quality $Q_0$ and $k_{Q_0, Q}$ is a correction factor which accounts for the effects of the difference between the reference beam quality $Q_0$ and the actual clinical beam quality $Q$ (Andreo et al 2000).

However, the main issue in adopting the TRS-398 protocol for kilovoltage x-ray beams is that absorbed dose to water calibrations are not generally available for low and medium energy x-ray beams (Andreo et al 2000, Mayles et al 2007). The number of standards laboratories that have implemented absorbed dose to water calibrations to date is limited (de Prez and de Pooter 2008, Seuntjens and Duane 2009, Krauss et al 2012). However, according to the TRS-398 protocol, it is possible to derive calibration factors in terms of absorbed dose to water using air-kerma calibration factors and one of the published air-kerma calibration protocols.

3.5. Comparisons of reference dosimetry protocols

A number of studies have investigated the different dosimetry protocols as well as the differences in dose when using them (Chica et al 2008, Jhala et al 2009, Munck af Rosenschold et al 2008, Nisbet et al 1999, Peixoto and Andreo 2000, Williams and Thwaites 2000, Yoo et al 2002).

Peixoto and Andreo (2000) compared four dosimetry protocols for kilovoltage x-ray beams (DIN, IAEA TRS-277, IPEMB and NCS) using a Farmer type ionization chamber. For the x-ray beam energies for which direct comparisons could be made, they found that the agreement was within ±1% except at the extremes for beam energy. This agreement was expected due to the consistency of the various correction factors used within the protocols.

However, greater differences in dose have been reported when changing to a newer calibration protocol. Yoo et al (2002) found differences of up to 5% when they implemented the AAPM TG-61 protocol over an older method, with the differences attributed to the use of more recent correction factors. Larger differences have also been reported in the adoption of the IAEA TRS-398 protocol and attributed to issues in determining depth doses and published perturbation factors (Jhala et al 2009, Munck af Rosenschold et al 2008).

It has been reported that using generic BSFs and mass–energy absorption coefficients leads to greater uncertainty in the absorbed dose of more than 5% (Chica et al 2008, Munck af Rosenschold et al 2008). Determination of machine specific values can reduce the uncertainty in dose but does require extensive calculations using Monte Carlo methods (Knight and Nahum 1994).
Based on these comparisons, the use of an air-kerma or absorbed dose to water calibration method should follow one of the more recently published dosimetry protocols. This will ensure reliability and accuracy in dosimetry and lead to consistency in the delivery of radiation doses to patients (Andreo et al 2000). The use of the older dosimetry protocols with the older data has the potential to cause inconsistent and inaccurate radiation dose delivery.

3.6. Quality audits

Quality audits are very useful as an independent check of dose calibrations of radiotherapy beams (Nisbet et al 1998, Thwaites et al 1995, Thwaites 1996, Van Dyk 1999). Dosimetry audits have been performed on megavoltage x-ray and electron beams for many years through organizations such as the IAEA (Van Dyk 1999). These audits have been useful for identifying gross errors with beam calibration and are often required prior to the participation in clinical trials.

There have been a number of audits of kilovoltage x-ray beams (Austerlitz et al 2008, Burton et al 2008, Eaton et al 2013, Nisbet et al 1998, Palmer et al 2011). Some audits did not find any significant dose calibration issues, with dose differences of less than 3% (Nisbet et al 1998, Palmer et al 2011). However, the study by Austerlitz et al (2008) found dose differences of up to 25% for some kilovoltage x-ray beams. The cause of these dose differences were incorrect dose calibrations, incorrect HVL measurements, out of date calibration factors for equipment and insufficient quality assurance testing on the x-ray units.

Therefore, it is recommended that the last step in the commissioning of a new kilovoltage therapy x-ray unit is an independent dosimetry audit of all x-ray beams. This process is also recommended after recommissioning an older x-ray unit. This could be achieved by an independent dose calibration using equipment with independent calibration factors.

4. Relative dosimetry of kilovoltage x-ray beams

In this section, a range of different dosimeters will be examined for their suitability for the dosimetry of kilovoltage x-ray beams. A summary of the different dosimeters available for use in radiotherapy is presented in table 1.

4.1. 1D dosimeters

4.1.1. Ionization chambers. Ionization chambers are commonly used to measure the radiation dose from an ionizing radiation beam (Attix 2004, Johns and Cunningham 1983, Khan 1994). There are a variety of different ionization chambers with different physical characteristics and applications.

(i) Thimble-shaped cylindrical chambers with a small air cavity used for reference and relative dosimetry.
(ii) Small cylindrical chambers with a small air cavity used for scanning radiation beams in water phantoms.
(iii) Pinpoint cylindrical chambers used for scanning very small radiation beams.
(iv) Parallel plate chambers designed primarily for electron beam dosimetry.
(v) Thin window parallel plate chambers designed primarily for dosimetry of very low energy x-ray beams.
Table 1. A summary of dosimeters used for kilovoltage x-ray beam dosimetry.

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<tr>
<th>Dosimeter</th>
<th>Applications</th>
<th>Advantages</th>
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<tbody>
<tr>
<td>Ionization</td>
<td>Relative dosimetry measurements,</td>
<td>High level of accuracy and reproducibility, some chambers are well characterized, ease of use,</td>
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<tr>
<td>chambers</td>
<td>reference dosimetry, quality assurance testing,</td>
<td>many types of chambers available</td>
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<tr>
<td>Diamond</td>
<td>Relative dosimetry measurements</td>
<td>High sensitivity, good spatial resolution</td>
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<td>detectors</td>
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<td>Diodes</td>
<td>Relative dosimetry measurements,</td>
<td>High sensitivity, good spatial resolution</td>
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<td>MOSFETS</td>
<td>Relative dosimetry measurements,</td>
<td>High sensitivity, good spatial resolution</td>
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<td>Optical fibres</td>
<td>Relative dosimetry measurements,</td>
<td>High sensitivity, good spatial resolution</td>
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<td>OSLs</td>
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<td>PSDs</td>
<td>Relative dosimetry measurements,</td>
<td>High sensitivity, good spatial resolution</td>
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<td>TLDs</td>
<td>Relative dosimetry measurements,</td>
<td>High sensitivity, good spatial resolution</td>
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<td>Radiochromic</td>
<td>Relative dosimetry measurements,</td>
<td>High sensitivity, good spatial resolution</td>
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<td>film</td>
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<td>Genipin gel</td>
<td>Relative dosimetry measurements</td>
<td>3D dosimetry and near tissue equivalence</td>
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<td>PAG gels</td>
<td>Relative dosimetry measurements</td>
<td>High sensitivity, good spatial resolution</td>
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<tr>
<td>PRESAGE</td>
<td>1D, 2D and 3D dosimetry measurements</td>
<td>High spatial resolution in 3D and can be read out with optical CT scanner</td>
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4.1.2. Diamond detectors. Diamond detectors have high spatial resolution and have been used for the dosimetry of small megavoltage x-ray fields such as those used in stereotactic radiosurgery beams (Heydarian et al. 1996, Hoban et al. 1994). For low energy x-ray beams, it has been shown that diamond detectors have a significant energy response due to the atomic number of the diamond and other materials within the detector (Planskoy 1980, Yin et al. 2004). Diamond detectors have been used for the relative dosimetry measurements of depth dose and depth kerma data for kilovoltage x-ray beams in the energy range from 45 to 200 kVp and compared with Monte Carlo calculations (Hugtenburg et al. 2001, Knöös et al. 2007, Seuntjens et al. 1999).

Correction factors for the nonlinearity in the dose rate response and for the energy dependence have been calculated and in some cases correction factors were found to be up to 12% for the lower energy beam energies (Hugtenburg et al. 2001, Seuntjens et al. 1999). However, it was found that the diamond detector could directly measure depth doses for the 100 kVp beam with differences of less than 2% (Seuntjens et al. 1999). In a more recent study, Knöös et al. (2007) found a good agreement for depth doses measured with a diamond detector and ionization chambers for 120 and 200 kVp x-ray beams. Measured depth doses agreed well with Monte Carlo calculations except in the first centimetre below the surface.

4.1.3. Diodes. Solid state diode detectors have been found to be suitable for some relative dosimetry measurements but for certain x-ray beam energies only. Li et al. (1997a) measured depth dose and profiles for kilovoltage x-ray beams using a Scanditronix photon diode and found the diode was suitable for depth doses for the 100 kVp x-ray beam but not the 300 kVp beam. This diode was not recommended for profile measurements as it incorrectly measured the profile tails which was attributed to a nonlinear dose response and angular response variation of the detector. In a similar study, Ma et al. (1998) determined perturbation correction factors for 100 to 300 kVp x-ray beams using Monte Carlo methods. The factors varied by up to 15% and were applied to the direct dose readings by the diode.
A stereotactic diode was used to measure total scatter factors in a small animal irradiator which used an x-ray beam operating at 250 kVp and with an HVL of 0.45 mm Cu (Pidikiti et al 2011). These output factors agreed with those measured using an ionization chamber and Gafchromic EBT2 to within 3% for collimators greater than 5 mm diameter.

4.1.4. MOSFETS. Metal oxide semiconductor field-effect transistors (MOSFETS) are used in radiation dosimetry and possess several desirable features such as small size, being able to provide real time dose rate information and instantaneous dose readouts (Rosenfeld 2006). Several studies have shown that MOSFETS do have a large variation in energy response for low energy x-ray beams (Kron et al 1998, Lian et al 2011). One solution is the determination of correction factors using Monte Carlo calculations (Lian et al 2011).

In one study, Cheung et al (2003) compared in vivo dosimetry measurements using MOSFETS against calculated doses and TLD measurements for a wide range of kilovoltage x-ray beams. The MOSFETS gave an average dose difference of 5.6% as compared to calculated doses. In a similar study, Ehrlingfeld et al (2005) examined the dosimetric characteristics of MOSFETS for x-ray beams in the energy range 80–220 kVp. They found the agreement between TLD and MOSFET measured doses were generally within 3% but indicated that energy correction factors may be required.

4.1.5. Optical fibres. Issa et al (2011) investigated the thermoluminescence (TL) properties of Ge-doped silica optical fibres for the dosimetry of kilovoltage x-ray beams. These fibres provide good spatial resolution for radiation dosimetry measurements. The maximum difference in depth doses obtained from the optical fibres and a Farmer ionization chamber was 2.1% for the 90 kVp x-ray beam and 1.5% for 300 kVp x-ray beam. Similar agreement was also found in the measured depth doses as compared to the BJR25 reported data. This indicates a good energy response for the optical fibres and potential use in kilovoltage x-ray beam dosimetry.

4.1.6. Optically stimulated luminescence dosimeters. Optically stimulated luminescence (OSL) dosimeters are now commonly used in radiation dosimetry particularly for personnel monitoring and in vivo dosimetry (Akselrod et al 2006, Yukihara and McKeever 2008). OSLs have an easy to read dosimetry system and are quite small in size. The most commonly used material for OSLs is Al₂O₃.

Reft (2009) performed a study on the energy and dose response of an OSL system for radiation beams including kilovoltage x-rays. He found that the response of Al₂O₃ dosimeters was up to 3.5 times greater for the 125 kVp x-ray beam as compared to the 6MV x-ray beam and attributed this predominantly to the relatively high atomic number of the OSL compared to water. This means the use of OSL dosimeters requires calibration in similar energy x-ray beams as those used for measurement.

4.1.7. Plastic scintillator detectors. Plastic scintillator detectors (PSDs) have found an application in the radiation therapy due to their small size and other dosimetric properties (Beddar 2006). One study has investigated the suitability of PSDs for relative dosimetry measurements in kilovoltage x-ray beams for beams with energies ranging from 80–150 kVp (Lessard et al 2012). They found that that the PSDs had a high-energy dependence which was corrected using mass–energy-absorption coefficients determined using Monte Carlo calculations with measured x-ray spectra information. An over-response of up to 3% was found in the measured PDDs at depth due to beam hardening even with correction factors applied.

4.1.8. TLDs. Thermoluminescent dosimeters (TLDs) are widely used in radiation dosimetry measurements and have the properties of being small in size, reusable and near tissue
equivalence for most beam energies (Bos 2006, Kron 1994, 1995, Mayles et al 2007, Metcalfe et al 1997, Moscovitch and Horowitz 2006). There are a variety of TLD materials available with the most commonly used being the LiF : Mg, Ti (TLD100), LiF : Mg, Cu, P (TLD100H) and Li2B4O7 : Mn (TLD800).

However, it has been shown that many TLDs have a large energy response for low energy x-ray beams (Ferguson et al 1995, Kron et al 1998, Mobit et al 2000, 2006, Nunn et al 2008). In the study by Nunn et al (2008), the variation in response was up to 40% for x-ray beams in the energy range from 20–250 kVp as compared to Cobalt-60. While some studies have used Monte Carlo calculations to determine energy response values for TLDs, this has been shown not to be sufficiently accurate in all cases (Mobit et al 2000, 2006, Nunn et al 2008).

TLDs have been used for the dosimetry of kilovoltage x-ray beams with applications including measurements of skin doses (Ferguson et al 1995); dosimetric verification within tissue equivalent phantoms for comparing against planning system calculations (Alaei et al 2000); and quality assurance testing of kilovoltage x-ray beams (Nelson et al 2008). TLDs have also been used to measure BSFs of kilovoltage x-ray beams as mentioned elsewhere in this review (Coudin and Marinello 1998, Harrison et al 1990, Nelson and Hill 2011). It has also been shown that TLDs are suitable for the dosimetry of low energy brachytherapy sources with measurements in different solid phantoms (Meigooni et al 1988, 1994, Reniers et al 2004).

4.2. 2D dosimetry—radiochromic film

Radiochromic films are widely used in radiation dosimetry to measure a 2D dose distribution, are self-developing, have high spatial resolution, are very thin and can be cut into the required shape (Cheung et al 2004, Devic et al 2006, Niroomand-Rad et al 1998, Soares 2006, Sellakumar et al 2009). The most commonly used radiochromic films are the various versions of Gafchromic films (Ashland Specialty Ingredients, New Jersey, USA). The Gafchromic EBT, EBT2 and EBT3 films are used predominantly for dosimetry of radiotherapy beams while the XR-QA film is used mainly for quality assurance tests. The original EBT film has been superseded by the EBT2 and EBT3 films each with different characteristics such as the geometry and composition of the different layers in the film (Reinhardt et al 2012).


The Gafchromic EBT film was reported to have an energy response variation of between 4.5% and 10% in the kilovoltage x-ray beam range (Butson et al 2006b, Chiu-Tsao et al 2005, Lindsay et al 2010, Rink et al 2007, Sutherland and Rogers 2010). Several studies have examined the energy response of EBT2 film including at kilovoltage x-ray beam energies (Arjomandy et al 2010, Butson et al 2010, Lindsay et al 2010, Sutherland and Rogers 2010). In one study, Arjomandy et al (2010) examined the energy dependence of EBT2 film for kilovoltage x-ray beams at energies of 75, 125 and 250 kVp and found that energy dependence was relatively small being within measurement uncertainties of $1\sigma = \pm 4.5\%$. In a similar
study, Butson et al (2010) found that EBT2 film had energy response variations of up to 6.5% for x-ray beams with energies from 50 kVp to 10 MV. These energy response variations were taken into account by calibrating the film with the same or similar x-ray beam energies as used for the measurements.

Gafchromic EBT and EBT2 films have been used to measure BSFs in the energy range from 50 to 280 kVp (Butson et al 2007, Kim et al 2010, Smith et al 2011). The measured BSFs were compared to Monte Carlo calculations and/or published BSFs and gave differences of up to 3%. These results indicate that Gafchromic films are suitable for the direct measurement of BSFs of kilovoltage x-ray beams.

The Gafchromic EBT3 film has been found to be suitable for the measurement of relative output factors for x-ray beams in the energy range from 50–125 kVp (Gill and Hill 2013). They found that the agreement between output factors measured with EBT3 film and a parallel plate ionization chamber was generally better than 2%, with larger differences of up to 3.3% for the smallest field size.

Fletcher and Mills (2008) measured depth doses for 50 and 100 kVp x-ray beams using various dosimeters including several ionization chambers, a Wellhoefer photon diode and Gafchromic EBT film. Measured depth doses were compared to BEAMnrc Monte Carlo calculations and used an analytical program to generate the primary x-ray beam spectra. They found the agreement between the EBT film and Monte Carlo was quite good for both the 50 and 100 kVp x-ray beams. The maximum difference was up to 8% at depth but with the doses normalized at a depth of 20 mm.

Radiochromic films have found an application in relative dosimetry measurements of intraoperative x-ray units (Croce et al 2012, Eaton and Duck 2010, Ebert et al 2009). These units produce very low energy x-ray beams typically 50 kVp and some radiochromic films used for measurements showed significant nonlinearity, particularly the XR-QA film (Ebert et al 2009).

4.3. 3D dosimetry—gel dosimeters

The majority of radiation dosimeters can only provide dose information at a point or within a 2D plane. This limitation has been overcome through the development of different chemical dosimeters that contain ferrous sulphate (Fricke gel) or polyacrylamide gels (PAGs) (Baldock et al 2010). These gel dosimeters have the advantage that they can be made into any shaped volume and so be used to measure 3D dose distributions.

Many of these gel dosimeters contain a large proportion of water in their composition and are radiologically water equivalent for high energy photon beams (Brown et al 2008b, Keall and Baldock 1999, Kron et al 1993). The water equivalence of different gel dosimeters has also been examined for low energy photon beams (Boudou et al 2004, Pantelis et al 2004, Trapp et al 2002, Venning et al 2005b).

4.3.1. Genipin gels. The genipin dosimeter uses genipin which is extracted from the fruit of Gardenia jasminoides Ellis (Yao et al 2004). Genipin gelatin or genipin gel is a blue radiochromic gel which bleaches quantitatively after exposure to ionizing radiation (Jordan 2009).

Gorjiara et al (2011b) characterized the radiological properties of genipin gel for clinical kilovoltage x-ray beams in the energy range from 50 to 300 kVp. For energies below 150 keV, the photoelectric absorption cross sections were 3% greater than water due to the dependence on the atomic number. However, the calculated depth dose curves for genipin gel agree to within 1% as compared to those in water. This indicates that genipin gel has excellent radiological water equivalence for low energy x-ray beams.

In one study, Venning et al (2005b) compared the radiological properties of several PAGs and water for photons over a wide energy range by calculating the photon cross section data. The photon cross section ratios for the PAGs as compared to water were generally better than 3% of unity except for photon energies less than 100 keV where there were differences of up to 6%.

4.3.3. PRESAGE dosimeter. The PRESAGE dosimeter is a 3D dosimetry material consisting of an optically clear polyurethane matrix, containing a leuco dye that exhibits a radiochromic response when exposed to ionizing radiation (Adamovics and Maryanski 2006, Guo et al 2006). Exposure to ionizing radiation causes changes in the optical absorbance of the PRESAGE dosimeter, which can then be used to determine the absorbed dose distribution. PRESAGE has a number of advantages over other 3D dosimeters which include being insensitive to oxygen, having a solid form which means it can be designed to the required shape and that it is light absorbing rather than scattering which allows readout by optical CT scanner (Adamovics and Maryanski 2006, Baldock et al 2010, Guo et al 2006).

Brown et al (2008b) studied the radiological water equivalence of PRESAGE by determining the photon and electron interaction cross section data. They found that the mass–energy absorption coefficient ratio of PRESAGE to water was up to 5% more than unity in the energy range 10 to 100 keV and attributed this to the presence of high atomic number materials within the PRESAGE. In a more recent study, Gorjiara et al (2011a) investigated the radiological water equivalence of different formulations of PRESAGE by Monte Carlo methods. While the radiological water equivalence for low energy x-ray beams was better for the new formulations, correction factors were still required to convert the dose measured in the PRESAGE dosimeter to dose in water.

PRESAGE dosimeters have also been used for the dosimetry of kilovoltage x-ray beams for animal irradiators (Newton et al 2011, Thomas et al 2011). The x-ray units were generally operated at higher kilovoltage beam energies and the PRESAGE dosimeter was found to provide accurate 3D dosimetry as compared to ionization chambers and radiochromic film.

4.4. Relative dosimetry measurements

Relative dosimetry data of kilovoltage x-ray beams is measured for the commissioning of the x-ray unit, beam data collection for quality assurance testing, reference dosimetry measurements and treatment planning calculations. This data typically includes depth doses, profiles, relative output factors both for applicators and lead cutouts as well as BSFs.


There are however a number of limitations in using Farmer chambers for radiation dosimetry measurements. Some Farmer chambers are not water-proof, particularly those used for reference dosimetry and calibrated by a primary or secondary standards laboratory (Andreo et al 2000). A water proof sheath can be used on the Farmer chamber for measurements in...
water but care must be taken in the selection of an appropriate sheath (Ma et al. 2001, Ma and Seuntjens 1997). The other limitation is that the Farmer chamber is quite large, typically with a radius of 3 mm. This means the minimum depth of measurement is equal to the radius of the chamber or that surface dose measurements have a part of the Farmer chamber positioned above the phantom surface (Ma et al. 2001). This will lead to increased uncertainty in the dose measurements close to and at the phantom surface. It should be noted that many of the studies listed immediately above do not provide details of the minimum depth used and/or the use of any corrections applied if a part of the Farmer chamber was positioned partially above the phantom surface.

For these reasons, there have been a number of studies into the use of different dosimeters for the relative dosimetry of kilovoltage x-ray beams. These have included parallel plate ionization chambers, small volume ionization chambers, solid state detectors such as diodes and diamonds and radiochromic film.


The IPEMB dosimetry protocol for kilovoltage x-ray beams contains no recommendations for relative dosimetry measurements (Klevenhagen et al. 1996). However, the addendum to this protocol recommends relative dosimetry measurements should only be made with an ionization chamber with an air-kerma calibration factor, $N_K$, that varies by no more than ±5% over the energy range corresponding to an HVL of 0.15–4.0 mm Cu (Aukett et al. 2005). The addendum also states that this requirement would usually be met by cylindrical ionization chambers but not necessarily by parallel plate chambers designed for electron beams. These recommendations are based on two studies which compared different ionization chambers by calculating dose ratios over a range of depths for a range of x-ray beam energies (Aukett et al. 1999, Rosser 1996). However, dose ratios are not typically used in radiation dosimetry and it should be noted that taking ratios of small dose values can lead to large dose ratio differences while the actual doses difference can be quite small.

The AAPM TG-61 protocol states that a well designed cylindrical ionization chamber usually has a flat energy response for kilovoltage beams in the energy range 40–300 kVp (Ma et al. 2001). However, this is limited to the minimum depth of measurement being the outer radius of the chamber. The AAPM protocol also states that some parallel plate chambers, as designed for electron beams, are suitable for depth dose measurements with uncertainties of less than 3% (Li et al. 1997a). Depth-dependent correction factors may be required for some ionization chambers when used for depth dose measurements. However, published work by some of the authors of the protocol found that the correction factors were quite small for some detectors such as the NACP plane parallel plate ionization chamber (Ma et al. 1998, Seuntjens et al. 1999).

The IAEA TRS-398 protocol has two sets of recommendations for kilovoltage x-ray beams depending on the beam energy (Andreo et al. 2000). Low-energy kilovoltage x-ray beams are defined as x-ray beams with a potential of up to 100 kVp or a measured HVL of up to 3 mm Al while medium-energy kilovoltage x-ray beams have a beam potential greater than 80 kVp or a measured HVL greater than 2 mm Al. In the low-energy section, the protocol recommends that 'depth-dose distribution can be measured by using the same chamber as that used for reference dosimetry and a water-equivalent phantom' (Andreo et al. 2000). The IAEA TRS-398 protocol also states that the published depth dose data are not likely to match the clinical x-ray beam for both the kVp and the HVL and recommend that the data be measured for each clinical beam. They suggest that for the medium-energy x-ray beams, measurement
of the depth doses using either a scanning ionization chamber or parallel plate chambers would introduce uncertainties of no more than several per cent (Andreo et al 2000).

4.4.2. Detector comparisons. There has been a number of studies examining the suitability of different detectors for relative dosimetry measurements of kilovoltage x-ray beams (Aukett et al 1996, Gerig et al 1994, Hill et al 2009, Knöös et al 2007, Li et al 1997a, 1998, Munck Af Rosenschold et al 2008, Snow et al 2013). These detectors include Farmer type chambers, parallel plate ionization chambers as designed primarily for electron beams, scanning thimble chambers and solid state detectors such as diode and diamond detectors. Many of these detectors have the potential of better spatial resolution for either depth dose or lateral profile measurements.

Gerig et al (1994) measured the relative dosimetry data for x-ray beams in the energy range 100–300 kVp on a Pantak DXT300 x-ray unit and compared with the published data (BJR Supplement 17). They used an NACP parallel plate chamber for depth doses and a Scanditronix RK scanning chamber for the profiles. They found the agreement with the published depth dose data were very good except for the 300 kVp beam which was attributed to differences in beam quality and design of the x-ray unit.

Aukett et al (1996) reported on commissioning two DXT300 Therapax kilovoltage x-ray units in two different centres. Depth dose and relative output factors were measured with a variety of Farmer, thimble and parallel plate ionization chambers. Comparisons between different detectors showed differences in dose of up to 4% at a depth of 2 cm and attributed this to issues with measuring the dose accurately at the surface. Much larger differences in dose of up to 44% were observed as compared to the previously published dosimetry data. However, the published data were for x-ray units with different beam characteristics to those used in the study (BJR Report 25 1996, Niroomand-Rad et al 1987).

In two comprehensive studies, Li et al (1997a), 1998) investigated nine different detectors for the relative dosimetry of x-ray beams in the energy range 50–300 kVp. They found that the PTW N22342, PTW Markus and Scanditronix NACP parallel plate chambers could be used to measure depth doses with an uncertainty of less than 3% without applying any correction factors. While the Farmer chamber was considered the gold-standard, it could only be used with a minimum depth of 1 cm. They also found a good agreement in profile measurements between the Farmer, NACP and RK scanning ionization chambers.

A later study by the same group involved measurements of depth doses and profiles for highly filtered beams in the energy range 40–60 kVp and comparison with Monte Carlo calculations (Li et al 1998). The result of this work was that the PTW Markus and PTW N23342 parallel plate chambers gave doses within 2% for a 50 kVp x-ray beam while the NACP chamber gave results that were up to 10% greater for depths less than 0.3 cm.

In two studies, Knöös et al (2007), Munck Af Rosenschold et al (2008) compared the following dosimeters: the FC65-G Farmer chamber, NACP-02 parallel plate chamber (Wellhofer-Scanditronix, Germany) and PTW Roos parallel plate chambers and PTW diamond detector (PTW-Freiburg, Freiburg). In the first study, they undertook measurements of depth doses and profiles of 120 and 200 kVp x-ray beams. The Farmer chamber was used for the depth dose measurements except for depths less than 0.7 cm where the NACP chamber was used in order to avoid collisions with the applicator. The BEAMnrc Monte Carlo system was used to calculate both depth doses and profiles in a water phantom for both x-ray beams. The agreement between measured and calculated depth doses was better than 1% except near the water surface where there were differences of up to 2%. The second study involved measuring doses at the surface and at 2 cm depth and included the PTW Roos chamber (Munck Af Rosenschold et al 2008). They found differences of less than 2% between all dosimeters for
the relative doses measured at 2 cm depth compared to the surface doses for both 120 and 200 kV x-ray beams.

Hill et al (2009) examined ten different ionization chambers for the relative dosimetry of kilovoltage x-ray beams with energies ranging from 50 to 280 kVp. The various chambers studied were Farmer type chambers, scanning thimble chambers, parallel plate chambers and a PTW PinPoint chamber. The measured data were compared to BEAMnrc Monte Carlo calculations for depth doses for all x-ray beams. The results indicate that the Advanced Markus, Markus, NACP and Roos parallel plate ionization chambers were suitable for the measurement of depth doses for all x-ray beam energies with a total uncertainty of less than 3%. These parallel plate chambers were also suitable for measuring in the regions close to the water surface. The Farmer and scanning chambers had an excellent energy response but were not suitable for measurements at depths close to the surface due to perturbation effects which caused a flattening in the depth doses of up to several per cent right at the surface. The results of this comparison for 50 and 280 kVp x-ray beams and in the region close to the water phantom surface are shown in figures 2 and 3, respectively. Therefore using a larger detector, such as a Farmer chamber, for dose measurements near the water phantom surface will result in incorrect dose values.

Based on the results listed above, a number of parallel plate ionization chambers are suitable for the relative dosimetry of kilovoltage x-ray beams and have superior spatial resolution in the depth direction. In particular, the NACP, Markus, Advanced Markus, Roos and PTW N23342 parallel plate chambers were found to be acceptable for most depth dose measurements in the energy range from 40–300 kVp (Geric et al 1994, Hill et al 2009, Knöös et al 2007, Li et al 1997a, 1998, Munck Af Rosenschold et al 2008). In general, these parallel plate chambers give doses that are within a few percentage of either Monte Carlo calculations, published depth doses using BJR Report 25 or with doses measured at depth using Farmer type ionization chambers.

Therefore, these parallel plate chambers can be used to directly measure depth doses without needing to apply any depth correction factors with uncertainties of less than 3%. While Farmer type and scanning ionization chambers are generally suitable for both depth dose and profile measurements, they are limited in the minimum depth for which they can accurately measure. New ionization chambers or ones that have not been previously studied should only be used after dosimetric comparisons with a well characterized ionization chamber.

4.4.3. Relative output factors. Relative output factors are required to specify the dose output for a particular applicator and/or lead cutout as compared to the dose output for the reference applicator (Mayles et al 2007, Williams and Thwaites 2000). Generally, the output factor is defined at the surface of a water phantom but can also be defined at depth (Hill et al 2005, Williams and Thwaites 2000).


(ii) Measurements at depth in a water or a Solid Water equivalent phantom and application of PDD data (Evans et al 2001, Jurado et al 2005).

Figure 2. Measured percentage depth dose curves for a variety of ionization chambers and BEAMnrc calculations for a 50 kVp x-ray beam (Hill et al 2009).

The use of published BSF and depth dose data for the calculation of output factors leads to an increase in dose uncertainties (Chica et al 2008, Ma et al 2001, Munck Af Rosenschold et al 2008). There are significant differences between measured output factors and factors calculated using the ratio of published BSFs. Healy et al (2005) found differences of more than 10% in measured and calculated output factors for x-ray beams with HVL of 1 and 2 mm of Al and for field sizes of 2 mm diameter. It has also been shown that output factors should be measured in water or a solid phantom which is radiologically water equivalent particularly for lower energy photon beams (Healy et al 2005, Hill et al 2005). Output factors measured in Plastic Water and perspex have given differences of more than 5% as compared to those in water, which can potentially lead to a patient overdose or underdose.

For these reasons, it is recommended that output factors are measured in water (or a water equivalent solid phantom) for all the combinations of beam energy, applicators and/or field...
Figure 3. Measured percentage depth dose curves for a variety of ionization chambers and BEAMnrc calculations for a 280 kVp x-ray beam (Hill et al 2009).

sizes. The use of published BSF or output factors can be used as a verification of measured data and one should be aware of the uncertainties in the published data (BJR Report 25 1996, Chica et al 2008).

4.4.4. Relative dosimetry of very small field sizes. Several studies have determined the relative dosimetry data for kilovoltage x-rays used for intraoperative radiation therapy (IORT) and animal irradiators often with very small field sizes (Eaton and Duck 2010, Eaton 2012, Ebert et al 2009, Newton et al 2011, Pidikiti et al 2011). The IORT x-ray units typically deliver x-rays with peak potentials of 50 kVp. By comparison, animal irradiators use kilovoltage x-ray tubes generally at higher voltages of around 225 kVp but deliver x-ray beams with small field sizes ranging from 40 mm diameter down to 1 mm diameter. Depth dose and profile data have been determined in some of these studies using Gafchromic EBT2 film, PinPoint ionization
chambers, PRESAGE radiochromic dosimeters and stereotactic diodes. Newton et al. (2011) found there was consistency in the agreement in depth doses and profiles measured using the ionization chambers, Gafchromic EBT2 film and the PRESAGE dosimeter.

5. Backscatter factors

The BSF is defined as the ratio between a dose quantity measured on the central axis at the surface of a phantom facing the radiation source and the same dose quantity at the same position free in air (Benmakhlouf et al. 2011, Carlsson 1993, Mayles 2007, Petoussi-Henss et al. 1998). The reference phantom material for radiation therapy is water and there has been a lot of investigation into the BSF for water $B_w$.

The most commonly used definition of $B_w$ is the ratio of the water collision kerma at a point on the beam axis at the surface of a full scatter water phantom to the water collision kerma at the same point in the primary beam with no phantom present (Grosswendt 1984, Ma and Seuntjens 1999). This definition of $B_w$ has been used by the various kilovoltage x-ray beam reference dosimetry protocols (Andreo et al. 2000, Grimbergen et al. 1997, IPSM 1991, Klevenhagen et al. 1996, Ma et al. 2001). An alternative definition for $B_w$ is the ratio of the air-kerma at the surface of a phantom such as water to the air-kerma at the same point in space in the absence of the phantom or free-in-air; this definition is used more in diagnostic radiology (Benmakhlouf et al. 2011, Grosswendt 1984, Ma and Seuntjens 1999, Petoussi-Henss et al. 1998).

Grosswendt (1984) determined that $B_w$ could be calculated from the primary and scattered photon spectra at the surface of a water phantom by the following relationship:

$$ B_w = \frac{\int_0^E \left[ \left( \frac{d\phi}{dE} \right)_0 + \left( \frac{d\phi}{dE} \right)_w \right] \rho \mu_{en}(E) \, dE}{\int_0^E \left[ \left( \frac{d\phi}{dE} \right)_0 \right] \rho \mu_{en}(E) \, dE} $$

where $(\frac{d\phi}{dE})_0$ is the spectrum of primary photons, $(\frac{d\phi}{dE})_w$ is the spectrum of photons scattered within the water phantom at the point of interest on the surface, $\rho \mu_{en}(E)$ is the mass–energy absorption coefficient of the photons of energy $E$ in the water and the energy spectrum of the photon beam is from 0 to energy $E$.

It has been shown that the BSF varies as a complex function of the x-ray beam energy, field size, SSD and also with different materials in the phantom (Butson et al. 2008b, Grosswendt 1984, 1990, 1993, Johns and Cunningham 1983, Kim et al. 2006, 2010, Klevenhagen et al. 2000). The BSF has a maximum value for x-ray beams with a HVL of about 1.0 mm Cu or a peak potential of about 150 kVp (Johns and Cunningham 1983).

5.1. Monte Carlo calculations

5.2. Published BSFs and uncertainties

The AAPM and IPEMB reference dosimetry protocols for kilovoltage x-ray beams make use of $B_w$ based on Monte Carlo calculations (Grosswendt 1984, 1990, 1993, Klevenhagen et al 1996, Ma et al 2001). As stated in the AAPM protocol, these calculations have been independently verified by the measurements and independent Monte Carlo calculations (Klevenhagen 1989, Klevenhagen et al 1991a, Knight 1992, Ma et al 2001). The AAPM protocol presents $B_w$ for x-ray beams with a HVL ranging from 0.04 mm Al up to 5 mm Cu and for a wide range of field diameters and SSDs.

A number of investigations have shown that the use of published BSFs can lead to large uncertainties in dose calculations such as those used in the reference dosimetry. The study by Chica et al (2008) examined the uncertainty in absorbed dose calculations due to uncertainties in both BSFs and mass–energy absorption coefficients as calculated using the HVL of the radiation beam. They found that the uncertainty can be larger than 5% for some of the x-ray beams and recommended that $B_w$ be calculated using additional parameters such as kVp, in addition to HVL, so as to reduce the uncertainty to be less than 3%. This approach has been adopted in at least one kilovoltage dosimetry protocol (Grimbergen et al 1997).

Munck Af Rosenschold et al (2008) designed a full Monte Carlo model of an x-ray unit in order to determine $B_w$ for their 120 and 200 kV x-ray beams. This was in the context of comparing different reference dosimetry protocols and comparisons with the published beam data. They found the differences between published and calculated $B_w$ were within 3% but the use of calculated $B_w$ reduced the uncertainty in the absorbed dose calculation. However, they stated that using just the HVL as the x-ray beam quality specifier may be the main cause of such differences.

5.3. Measured BSFs

Measurement of the BSF requires the measurement of dose at the surface of a phantom and in-air at the same point. For low-energy x-ray beams, there may be problems with the accuracy of absorbed dose measurements with ionization chambers or other dosimeters (Klevenhagen et al 1991a). The size and shape of the dosimeter may cause perturbations of the photon fluence and the energy response of the detector may change if the x-ray spectrum changes between the reference and measuring points (Patrocinio et al 1996).

Thin window parallel plate ionization chambers have been used to measured BSFs for x-ray beams in the energy range from 16–40 kVp (Klevenhagen 1989, Kim et al 2006). They found measured BSFs agreed with previously published BSFs generally to within 3–5%. However, the larger differences occurred for x-ray beams with an HVL less than 1 mm and were attributed to differences in beam filtration.

TLDs have been used in several studies for comparison with BSF factors determined using ionization chamber measurements and Monte Carlo calculations. In two studies, lithium borate (TLD800) was used to measure BSFs for comparison with ionization chamber measurements and Monte Carlo calculations (Coudin and Marinello 1998, Harrison et al 1990). The study by Harrison et al (1990) was instrumental in identifying inaccuracies in published BSFs. The results from both studies found a good agreement between BSFs that were measured with TLDs and Monte Carlo calculations. However, there were differences of up to 9% as compared to ionization chamber measurements.

More recently, Nelson and Hill (2011) measured BSFs using TLD100 and TLD100H chips for x-ray beams in the energy range from 75 to 300 kVp and compared these with EGSnrc calculations and IPEMB published data. The results of this comparison are shown in...
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figure 4. measured bsfs using tld100 and tld100h chips for x-ray beams with hvl's of 2.2–7.2 mm al and published i pemb bsfs (nelson and hill 2011).

gafchromic ebt and ebt2 film have been used to measure bsfs in the energy range from 50 to 280 kvp. in two studies, measured bsfs using ebt film were compared to published bsfs and monte carlo calculations, giving differences of up to 2.5% (butson et al. 2007, kim et al. 2010). a later study by smith et al. (2011) showed a good agreement between bsfs measured using gafchromic ebt2 film and the published data with a maximum difference of 3%. this level of agreement was similar to the estimated uncertainties in the film measurements.

figures 4 and 5, respectively. the agreement between the values of these bsfs was generally within 5%, but with differences in beam spectra between the measurements and the published data.

<table>
<thead>
<tr>
<th>Beam Energy</th>
<th>Film Type</th>
<th>EBT</th>
<th>EBT2</th>
<th>BEAMnrc</th>
<th>AAPM TG-61</th>
</tr>
</thead>
<tbody>
<tr>
<td>50 kVp x-ray beam</td>
<td>2 cm diameter</td>
<td>1.091</td>
<td>1.064</td>
<td>1.098</td>
<td>1.096</td>
</tr>
<tr>
<td></td>
<td>4 cm diameter</td>
<td>1.132</td>
<td>1.113</td>
<td>1.146</td>
<td>1.141</td>
</tr>
<tr>
<td></td>
<td>6 cm diameter</td>
<td>1.188</td>
<td>1.188</td>
<td>1.172</td>
<td>1.169</td>
</tr>
<tr>
<td>100 kVp x-ray beam</td>
<td>2 cm diameter</td>
<td>1.124</td>
<td>1.098</td>
<td>1.120</td>
<td>1.122</td>
</tr>
<tr>
<td></td>
<td>4 cm diameter</td>
<td>1.218</td>
<td>1.197</td>
<td>1.186</td>
<td>1.211</td>
</tr>
<tr>
<td></td>
<td>6 cm diameter</td>
<td>1.269</td>
<td>1.234</td>
<td>1.246</td>
<td>1.258</td>
</tr>
<tr>
<td>280 kVp x-ray beam</td>
<td>2 cm diameter</td>
<td>1.064</td>
<td>1.051</td>
<td>1.059</td>
<td>1.052</td>
</tr>
<tr>
<td></td>
<td>4 cm diameter</td>
<td>1.096</td>
<td>1.080</td>
<td>1.095</td>
<td>1.097</td>
</tr>
<tr>
<td></td>
<td>6 cm diameter</td>
<td>1.139</td>
<td>1.100</td>
<td>1.134</td>
<td>1.137</td>
</tr>
</tbody>
</table>

A summary of BSFs measured with Gafchromic EBT and EBT2 films, Monte Carlo calculations and published data is shown in table 2.

Based on these studies, determination of BSFs for kilovoltage x-ray beams can be achieved in the clinic by measurements using TLD chips, Gafchromic film or Monte Carlo calculations.

6. Phantom materials

The reference material for radiation dosimetry measurements is water because it is the major component of human tissues and considered a surrogate for soft tissue (Andreo et al 2000, Metcalfe et al 1997, Thwaites 2007). Water is very easy to use, readily available and very cost effective. For these reasons, water is the reference medium for absorbed dose measurements in the reference dosimetry protocols for both x-ray and electron beams (Andreo et al 2000, Klevenhagen et al 1996, Ma et al 2001).

A phantom material is any material that mimics the radiological properties of water, including absorption and scattering of radiation (Metcalfe et al 1997, Thwaites 2007, Williams and Thwaites 2000). For a solid phantom to be considered water-equivalent, a given thickness of the phantom should have the same radiation absorption and scattering properties as the same thickness of water. The ICRU report 44 recommends if any solid phantom is to be considered water equivalent, it should not introduce uncertainties to the absorbed dose of greater than 1% otherwise appropriate correction factors are required (ICRU 1989). Radiological water equivalence may be assessed by considering a number of physical characteristics such as the mass density $\rho$, electron density $(\rho_e)$, the elemental composition and an effective atomic number $(Z_{eff})$ of the phantom and comparing to values in water (Metcalfe et al 1997).

There are many situations where a solid phantom may be more appropriate to use and may have a number of important advantages over water. These solid phantoms allow the use of non-waterproof radiation dosimeters, being easier and more reproducible to setup as well as giving less positional uncertainties in areas of a high dose gradient. Most solid phantoms are also easier to use for routine quality assurance testing and regular dosimetry measurements. For all of these reasons, some dosimetry protocols allow the use of solid phantoms in reference dosimetry particularly for lower energy x-ray and electron beams with appropriate correction factors (Andreo et al 2000, Klevenhagen et al 1996).
Table 3. Composition of phantom materials A150, PAGAT, perspex (PMMA), polystyrene, Plastic Water (PW), Plastic Water DT (PWDT), RMI-457 Solid Water (RMI-457), white polystyrene (RW3), Virtual Water (VW) and water.

<table>
<thead>
<tr>
<th>Phantom</th>
<th>Density (gm cm(^{-1}))</th>
<th>Fractional weights of the elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>A150</td>
<td>1.127</td>
<td>H(0.1014), C(0.7755), N(0.0351), O(0.0523), F(0.0174), Ca(0.0184)</td>
</tr>
<tr>
<td>PAGAT</td>
<td>1.026</td>
<td>H(0.1059), C(0.0681), N(0.0242), O(0.8088), Cl(0.0002), P(0.0002)</td>
</tr>
<tr>
<td>PMMA</td>
<td>1.190</td>
<td>H(0.0805), C(0.5998), O(0.3196)</td>
</tr>
<tr>
<td>Polystyrene</td>
<td>1.060</td>
<td>H(0.0774), C(0.9226)</td>
</tr>
<tr>
<td>PRESAGE</td>
<td>1.101</td>
<td>H(0.0892), C(0.6074), N(0.0446), O(0.2172), Cl(0.0334), Br(0.0084)</td>
</tr>
<tr>
<td>PW</td>
<td>1.013</td>
<td>H(0.0925), C(0.6282), N(0.0100), O(0.1794), Cl(0.0096), Ca(0.0795), Br(0.0003)</td>
</tr>
<tr>
<td>PWDT</td>
<td>1.039</td>
<td>H(0.0740), B(0.0226), C(0.4670), N(0.0156), O(0.3352), Mg(0.0688), Al(0.0140)</td>
</tr>
<tr>
<td>RMI-457</td>
<td>1.030</td>
<td>H(0.0809), C(0.6722), N(0.0242), O(0.1984), Cl(0.0013), Ca(0.0232)</td>
</tr>
<tr>
<td>RW3</td>
<td>1.045</td>
<td>H(0.0759), C(0.9041), O(0.0008), Ti(0.0120)</td>
</tr>
<tr>
<td>VW</td>
<td>1.030</td>
<td>H(0.0770), C(0.6874), N(0.0227), O(0.1886), Cl(0.0013), Ca(0.0231)</td>
</tr>
<tr>
<td>Water</td>
<td>1.000</td>
<td>H(0.1100), O(0.8900)</td>
</tr>
</tbody>
</table>

There are many solid phantom materials available for radiation dosimetry measurements including perspex, polystyrene and a range of epoxy resin based analogue materials. The fractional weight of many common phantom materials are listed in table 3 (Andreou et al (2000), Brown et al (2008b), Hill et al (2008), Ramaseshan et al (2008), Seco and Evans (2006)).

6.1. Testing for water equivalence

Testing for the radiological water equivalence of a solid phantom material has been performed using a number of experimental and analytical methodologies as follows:


(vi) Monte Carlo calculations of relative doses or phantom material correction factors (Gorjiara et al 2011a, 2011b, Hill et al 2010, Seuntjens et al 2005).

A number of studies have evaluated the use of various epoxy resin solid phantoms for both reference dose calibrations and relative dosimetry of megavoltage x-ray and electron beams (Allahverdi et al 1999, Tello et al 1995, Thomadsen et al 1995). It was found that radiation doses measured in these particular solid phantoms can differ by more than 1%, as compared to water, leading to the need for correction factors.
However, as emphasized by White (1978), radiological water equivalence is often assumed over a large energy range when in fact testing is often performed for a limited energy range. The method recommended by White for comparing the radiation properties of the solid phantom is to calculate the total mass-attenuation coefficient ($\mu/\rho$), total mass–energy absorption coefficient ($\mu/\rho_{en}$), electron mass stopping powers ($S/\rho$) and electron angular scattering powers ($\theta^{2}/\rho l$) over the energy range of interest.

6.2. Solid phantom dosimetry for kilovoltage x-ray beams

For therapeutic kilovoltage x-ray beams, a number of studies have investigated the relative dosimetry of solid phantoms in comparison to water. In early studies, the investigations involved varying comparisons including calculating mass attenuation coefficients, mass–energy absorption coefficients, BSFs, attenuation curves and/or depth doses in solid phantoms such as RW1, an older version of RMI Solid Water, polystyrene and in water (Hermann et al 1985, Shrimpton et al 1981, Stern and Kubo 1995). Both the RMI Solid Water and RW1 materials were found to be water equivalent even for low energies, while polystyrene gave depth dose differences of up to 10%.

Li et al (1999) compared the MS11 epoxy resin solid phantom, polystyrene and polymethyl methacrylate (PMMA) by measuring relative detector readings as a function of depth for 100 and 300 kVp x-ray beams. Their results indicated that doses measured in the MS11 phantom material were within 1% of those in water for both x-ray beam energies. In comparison, the relative doses measured in both the polystyrene and PMMA phantoms were higher than those in water.

Hill et al (2005) studied the radiological water equivalence of RMI-457 Solid Water (Gammex Inc., Wisconsin, USA) and Plastic Water (PW) (Computerized Imaging Reference Systems, Virginia, USA) by comparing relative and reference dosimetry measurements for x-ray beams in the energy range 75–300 kVp. They found that RMI-457 Solid Water was suitable for depth doses and output factors for all beam energies with a maximum deviation of 2.4% for the 75 kVp x-ray beam depth dose. In comparison, PW gave large differences in the depth doses of more than 20% and also in the relative output factors of up to 6%.

One study evaluated measured depth doses in PW, Solid Water, perspex (PMMA) and water for x-ray beam qualities with HVLs of 1–13 mm Al (Healy et al 2005). They found that Perspex over-predicts the dose as compared to water while doses measured in a Solid Water phantom gave a good agreement to the doses in water for all energies. However, the doses measured with PW were all consistently lower than the doses measured in water.

Gamma ray transmission measurements from a technetium-99m source were used to evaluate radiological water equivalence of RMI-457 Solid Water, PW and RW3 white polystyrene (PTW-Freiburg, Freiburg, Germany) (Hill et al 2008). Linear attenuation coefficients were determined from the transmission values and compared with calculated values determined from Monte Carlo calculated transmission curves using the EGSnrc Monte Carlo code and NIST values. The results showed that RMI-457 Solid Water, PW and solid phantoms had transmission values that agreed to those in water to within 1.5% for this gamma ray energy.

One study evaluated the radiological water equivalence of the Plastic Water diagnostic therapy (PWDT) solid phantom (Computerized Imaging Reference Systems, Virginia, USA) (Ramaseshan et al 2008). Depth doses were measured for a wide range of x-ray energies including 80, 100 and 250 kVp beams. It was found that doses measured with PWDT agreed to within 1.5% of those determined in water and Solid Water, confirming that the PWDT solid phantom is suitable for dosimetry of low energy x-rays.
Hill et al (2010) performed a study of the radiological water equivalence of ten commonly used solid phantoms and water for low energy photon beams using the PENELOPE Monte Carlo code. The calculations performed in all phantoms were depth doses, lateral beam profiles as well as dose to a small water volume at the surface of the solid phantom for x-ray beams with energies from 50–300 kVp. The results showed that the A150, PAGAT gel, PWDT, RMI-457 Solid Water and Virtual Water (VW) (CMNC, Tennessee, USA) could be considered radiologically water equivalent at all photon energies with dose differences of less than 2%. In comparison, the following phantoms gave significant dose differences as compared to water: PW, PMMA, polystyrene, PRESAGE and RW3. The results of the comparison of depth doses in the different solid phantoms for a 50 kVp x-ray beam (2 cm diameter field) and for a 180 kVp x-ray beam (8 cm diameter field) are shown in figures 6 and 7, respectively.
Based on these studies, it is recommended to use one of the solid phantoms which has been shown to have a good agreement in doses with water such as A150, MS11, RMI-457 Solid Water, PWDT and VW. A summary of the suitable and unsuitable solid phantom materials, along with the particular x-ray energy range, is presented in table 4. Any solid phantom should be tested in the clinic to ensure that it is suitably water equivalent and this should include the measured data for the x-ray beam energies used.

### 6.3. Effective atomic numbers of solid phantoms

It has been suggested that solid phantoms may be characterized for radiological water equivalence by calculation of their effective atomic number $Z_{\text{eff}}$ (Manohara et al 2008, Rao et al 1985, Taylor et al 2008, Yang et al 1987). The original definition of $Z_{\text{eff}}$ for a mixture was proposed by Mayneord and given by the following equation:

$$Z_{\text{eff}} = \left( \sum \alpha_i Z_i \right)^{1/n}$$

where $\alpha_i$ is the fractional electronic composition, $Z_i$ is the atomic number of the $i$th element and $n = 2.94$ (White 1977). The fractional electronic composition is given by the following equation:

$$\alpha_i = \frac{w_i (Z_i/A_i)}{\sum_i w_i (Z_i/A_i)}$$

where $A_i$ is the mass number and $w_i$ is the fractional chemical composition of the $i$th element.

This method has been used extensively in the literature with other authors using varying values of $n$ ≃ 2.94–3.5 (see e.g. Bazalova et al 2008, Johns and Cunningham 1983, White 1977). However, this methodology may only be valid in the region where the photoelectric effect is the dominant interaction process (Johns and Cunningham 1983). Several authors have extended this definition of $Z_{\text{eff}}$ to use different values of $n$ according to changes in the dominant

<table>
<thead>
<tr>
<th>Reference</th>
<th>X-ray energy range</th>
<th>Suitable phantom material</th>
<th>Unsuitable phantom material</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hermann et al (1985)</td>
<td>10–100 kVp</td>
<td>MS11</td>
<td>Plastic Water</td>
</tr>
<tr>
<td>Meigooni et al (1994)</td>
<td>50–300 kVp</td>
<td>RMI Solid Water</td>
<td>Polystyrene</td>
</tr>
<tr>
<td>Stern and Kubo (1995)</td>
<td>40–150 kVp</td>
<td>RMI Solid Water</td>
<td>PMMA</td>
</tr>
<tr>
<td>Li et al (1999)</td>
<td>40–150 kVp</td>
<td>MS11</td>
<td>Polystyrene</td>
</tr>
<tr>
<td>Reniers et al (2004)</td>
<td>1125 kVp</td>
<td>WT1 Solid Water</td>
<td>PMMA</td>
</tr>
<tr>
<td>Hill et al (2005)</td>
<td>75–300 kVp</td>
<td>RMI-457 Solid Water</td>
<td>Virtual Water</td>
</tr>
<tr>
<td>Healy et al (2005)</td>
<td>80–150 kVp</td>
<td>Solid Water</td>
<td>PMMA</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Virtual Water DT RMI-457</td>
<td>PRESAGE</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>RW3</td>
</tr>
</tbody>
</table>

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A number of studies have shown that $Z_{\text{eff}}$ varies as a function of photon energy if one compares the radiological properties of the phantom material over the energy range of interest (Manohara et al 2008, Naydenov et al 2004, Rao et al 1985, Taylor et al 2008, Yang et al 1987). The other limitation of many calculations of $Z_{\text{eff}}$ is that they are often limited to single photon energies and do not consider the spectrum of photons in an x-ray beam (Rao et al 1985).

An alternative definition is that of mean atomic number $\bar{Z}$ which is defined in the following equation (Andreo et al 2000):

$$\bar{Z} = \frac{\sum w_i Z_i^2}{\sum w_i Z_i}.
\tag{8}$$

This methodology is used by the IAEA TRS-381 and TRS-398 reference dosimetry protocols to characterize different solid phantom materials for radiation dosimetry measurements. However, this is a single parameter and is not necessarily valid over all photon energies. In addition, it has been shown that $\bar{Z}$ is not a good indicator of radiological water equivalence of a phantom for low energy x-rays (Hill et al 2010).

Based on these results, it is not recommended to use calculations of $Z_{\text{eff}}$ or $\bar{Z}$ for determining the radiological water equivalence of a solid phantom for low energy x-ray beams.

6.4. Effects of different phantom materials on BSF

It has been shown that the BSF varies according to the thickness of underlying phantom material and varies slightly for different SSDs (Klevenhagen 1982). In addition, there are changes in backscatter dose due to underlying high atomic number materials such as bone or lead as compared to a water phantom only (Butson et al 2008b, Healy et al 2005, Hill et al 2007, Huq et al 1992, Lanzon and Sorell 1993). This is due to the increased dose absorption in the high atomic number material leading to a reduction in the backscatter radiation at the phantom surface.

It has also been shown that the scattering properties of different solid phantoms can vary significantly as compared to water particularly for lower energy x-ray beams. Any differences in backscatter can have implications for radiation dosimetry measurements as these large water equivalent phantoms are used to provide full backscatter conditions (Klevenhagen et al 1996). Some materials such as PMMA and the ICRU soft tissue material are used routinely for radiation dosimetry measurements (Schauer et al 2000, Schwahn and Gesell 2008, Traub et al 1997). One study used Monte Carlo methods to calculate BSF ratios of various ICRU tissues as compared to water (Ma and Seuntjens 1999). They found negligible differences of less than 1% for the ICRU soft tissue but differences of up to 15% for the ICRU bone.

In the study by Hill et al (2010), it was found that different solid phantoms had significant variations in backscatter contribution to the dose to a small volume of water located at the surface of the solid phantom. They found that there were differences of up to 9.1% for polystyrene as compared to water only for a 50 kVp x-ray beam. There were significant differences in surface dose when the PW, PMMA and RW3 solid phantoms were used for backscatter. However, the differences in dose decreased as the x-ray beam energy became higher. Another study compared BSFs determined from measurements in ISO water phantoms and Perspex phantoms along with calculated BSFs in ICRU tissues and in PMMA (Schauer
et al 2000). They found that the BSFs measured in PMMA were up to 6.9% greater than those measured in the ISO water phantom.

Based on these results, it is recommended that one uses a radiologically equivalent phantom for any dosimetry measurements including an appropriate choice for backscatter material.

7. Treatment planning

7.1. Manual calculations

Treatment planning for kilovoltage x-ray beams is generally performed by manual calculations of either monitor units or treatment times. The dose is usually prescribed at the surface where the relative depth dose value is 100%. However in some cases, the radiation oncologist may wish to prescribe a dose at depth for which a depth dose correction is required. The energy of the particular beam selected is usually based on the depth dose characteristics with the surface dose of 100% and the required doses at depth to ensure sufficient dose coverage e.g. depth of the 95% or 90% dose value. Isodose curves can be used by the radiation oncologist for planning purposes in order to view a 2D map of the radiation dose. However, it should be noted that there will be a slight asymmetry in the isodoses due to the heel effect (Johns and Cunningham 1983, Mayles et al 2007).

The dose delivered to a patient for a particular x-ray beam energy will depend on a number of factors (Mayles et al 2007, Williams and Thwaite 2000).

(i) Depth dose value at the dose prescription point.
(ii) Treatment applicator.
(iii) Area of the radiation beam including any shielding.
(iv) Any offset between the end of the applicator and skin surface.

The x-ray unit is usually calibrated using a reference applicator and so the relative dose output for any other applicator needs to be considered by use of an applicator factor. The reference applicator is either a large diameter circular applicator or a 10 × 10 cm² square applicator.

Cutout factors are required if a lead cutout is used to define the beam shape. These cutout factors are usually measured in water or a water equivalent solid phantom. The cutout factor can be estimated by taking the ratio of the BSF for the cutout and the BSF for the open applicator but this method does not consider differences of scatter in the applicators used. In addition, the BSFs are most likely interpolated from a dataset of generic BSFs (BJR Report 25 1996, Klevenhagen et al 1996). For a very irregular cutout shape, a Clarkson integration method can be used to improve accuracy of the predicted cutout factor (Mayles et al 2007).

An inverse square law correction factor should be applied if there is a gap between the end of the applicator and the skin surface. However, additional correction factors may be required for very large gap distances of greater than 15 cm (Li et al 1997b).

Most current kilovoltage x-ray units provide dose delivery based on the selection of monitor units (Van Dyk 1999). However, many units can also provide dose delivery based on a timer mode and this will be the only option available for older x-ray units that do not have a monitoring ionization chamber.

7.2. Computer calculations

Modern radiotherapy treatment planning systems have been designed for dose calculations of megavoltage x-ray and electron beams (Mayles et al 2007, Van Dyk 1999). These dose
calculation algorithms such as pencil beam and convolution/superposition rely on electron density information from the CT data and dose kernels. However, these are applicable primarily for megavoltage x-ray beams where the Compton effect is the dominant interaction process and is dependent on the electron density of the particular medium.

There have been a limited number of investigation of using treatment planning systems for kilovoltage x-ray beams. Alaei et al (2000) investigated the accuracy of a convolution/superposition treatment planning system for predicting kilovoltage x-ray beam dosimetry. Customized dose calculation kernels for beam energies down to 20 keV were used in the planning system. They found that the agreement between TLD measured and computed doses was within 2% for materials similar to water. However, in the region around bone inhomogeneities, there were dose differences of up to 145%.

More recently, calculation algorithms for kilovoltage x-ray beams have been developed primarily for on-board imaging systems on linear accelerators (Ding et al 2008, Pawlowski and Ding 2011). These algorithms include corrections for different tissues types such as soft tissue and bone and have been shown to give dose errors of less than 3% (Pawlowski and Ding 2011). While these algorithms were designed for cone-beam CT dose calculations, there is the potential for them to be also used for therapeutic kilovoltage x-ray beams.

Several groups have investigated the use of Monte Carlo methods to calculate isodoses for patient CT data (Gao and Raeside 1997, Knöös et al 2007). The latter work presents patient dose calculations that were performed for both head/neck and thoracic treatments and proposed that such calculations would be of benefit to the radiation oncologist (Knöös et al 2007).

7.3. Dose to other tissues

The dose at the surface of a different medium, $D_{med,z=0}$ can be determined from the dose at the water phantom surface, $D_{w,z=0}$ by a correction factor, $C_{med}^{med}$, which is defined as follows (Ma et al 2001):

$$C_{med}^{med} = \frac{B_{med}}{B_{w}} \left[ \left( \frac{\mu_{en}}{\rho} \right)_{med,w} \right]_{air}$$

(9)

where $[\left( \frac{\mu_{en}}{\rho} \right)_{med,w}]_{air}$ is the ratio of the mass–energy-absorption coefficients of the medium and water averaged over the primary photon spectrum free in air and $B_{med}$ and $B_{w}$ are the BSFs for the medium and the water respectively (Ma et al 2001). The AAPM TG-61 report contains published correction factors for a number of different tissue types and are based on the previously calculated data (Ma and Seuntjens 1999, Seuntjens et al 1987b).

7.4. Shielding and interface effects

High density materials such as lead are often used for shielding of healthy tissues during kilovoltage x-ray beam treatments. Lead is an excellent shielding material due to its high atomic number and density as well as being malleable and so can be modified to the shape of the patient. However, it is important to note that the lead will also reduce backscatter and lead to a dose reduction in the dose at the surface (Das and Chopra 1995, Eaton and Doolan 2013, Healy et al 2008, Hill et al 2007, Huq et al 1992, Lanzon and Sorell 1993). This dose reduction is a function of field size, lead shielding thickness, beam energy and the depth of the lead shielding with a maximum reduction occurring for beam energies at approximately 100 kVp (Das and Chopra 1995, Healy et al 2008, Hill et al 2007). Figure 8 shows the surface dose ratio, which is defined as the ratio of measured surface doses with an underlying lead

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Figure 8. Measured surface dose ratios for lead shielding at a range of depths with a 100 kVp x-ray beam and for field sizes of 2, 5 and 8 cm diameter (Hill et al 2007).

sheet and in a full water phantom, for a 100 kVp x-ray beam and three field sizes with the lead sheet positioned at a range of depths.

It has also been shown that underlying bone or air cavities also lead to a reduction in backscatter and therefore a reduction in the surface dose (Butson et al 2008b, Hill et al 2007). Therefore, it is recommended that an assessment be made of any possible reduction in dose.

Related to this are conditions where there is a lack of full backscatter. For example, if you are treating the hand, the dose at the surface will not receive full backscatter thus leading to a reduction in the surface dose (Healy et al 2008, Klevenhagen 1982). It is recommended to treat such patients with additional backscatter material under the hand. Alternatively, one could increase the treatment time to counter the loss of backscatter with a factor calculated from the published data (Klevenhagen 1982).

There are also significant dose perturbations close to the interface between water/soft-tissue and a high density material (Das and Chopra 1995, Das 1997, Mitchell et al 1998, Regulla et al 2000). The comprehensive study by Das and Chopra (1995) investigated the variations in the backscatter dose perturbation factor (BSDF) as a function of the atomic number of the shielding material, beam energy, field size and distance from the interface. They showed that the BSDF can be more than a factor of 15 for a soft-tissue/lead interface. It is recommended that when lead is used behind the treatment volume, such as the lip or the ear, the lead is covered with a layer of low atomic number materials such as wax (Mayles et al 2007).

Lead shielding is commonly used for treatments using kilovoltage x-ray beams to define the shape of the radiation beam. However, lead can generate contaminant electrons which can significantly increase the dose right at the surface (Lee and Chan 2000, Lye et al 2010a). In the study by Lye et al (2010a), Monte Carlo calculations and Gafchromic film measurements were used to determine the amount of electron contamination. They found that the dose enhancement, due to electron contamination, was double the dose at the edge of the lead as compared to the central axis for the 150 kVp beam and three times for the 300 kVp beam. It was recommended that these contaminant electrons could be removed by wrapping the lead...
cutout with thin plastic wrap. It should be noted that increased electron contamination can also be generated from treatment cones made of high atomic number materials (Klevenhagen et al 1991b).

For kilovoltage treatments close to the eye, the use of internal eye shields can be used to reduce dose to critical structures such as the lens which have much lower dose tolerances (Amdur et al 1992, Baker et al 2002, Butson et al 2008a, Gordon et al 1995, Wang et al 2012). While the eye shield may provide a large reduction in the primary dose, scattered radiation can contribute up to 25% of the prescribed dose (Baker et al 2002). These studies showed that the actual dose to the lens, even with shielding, varied significantly according to the field size, shielded area and beam energy.

8. Monte Carlo methods for radiation transport

Monte Carlo calculation methods are widely used for solving radiation transport problems and a number of excellent review articles are recommended for more details (Bielajew 2007, Rogers 2006, Salvat and Fernandez-Varea 2009, Verhaegen and Seuntjens 2003). The Monte Carlo codes applicable for medical radiation physics are EGSnrc (Kawrakow 2000a, 2000b), FLUKA (Kling et al 2001), GEANT4 (Agostinelli et al 2003, Allison et al 2006), MCNPX (Kling et al 2001, Verhaegen and Seuntjens 2003, Waters et al 2007) and PENELOPE (Salvat et al 2006a, Sempau et al 2003). These codes are capable of simulating photon–electron transport through matter in the relevant energy range in medical physics applications covering from keV up to many MeV.

However, it should be noted that the different codes can give significant differences in their results, such as dose calculations, particularly in the low energy photon range (Almansa et al 2007, Reniers et al 2004, Rivard et al 2010, Verhaegen and Seuntjens 2003, Ye et al 2004). One component of these differences is the use of different photon cross section data (Hubbell 2006, Zaidi 2000). Photon cross section databases have developed with time using both theoretical calculations and the experimental data (Hubbell 1999, 2006). The most up-to-date databases are the Evaluated Photon Data Library (EPDL97) which is managed by the Lawrence Livermore National Laboratory and the XCOM database which is managed by the National Institute of Standards and Technology (Berger et al 1998, Cullen et al 1997). A number of studies have evaluated the differences between EPDL97 and XCOM along with older databases that are still available (Almansa et al 2007, Ye et al 2004, Zaidi 2000). These studies have shown that there are significant differences in cross section values particularly for the lower photon energies and for high atomic number materials.

Monte Carlo methods have been widely used to solve radiation transport problems for the characterization and dosimetry of low energy x-ray beams. These will now be discussed in relation to the most commonly used Monte Carlo codes for radiation transport calculations: EGSnrc, GEANT4, MCNP and PENELOPE.

8.1. EGSnrc

The EGSnrc Monte Carlo code has been verified by comparison with the experimental data in a number of studies involving low energy photon beams (Ali and Rogers 2008a, 2008b, Chibani and Li 2002, Kawrakow 2000a, Mainegra-Hing et al 2008, Reniers et al 2004, Rogers 2006, Verhaegen 2002). In one study, Mainegra-Hing et al (2008) compared correction factors for FACs and found a good agreement between Monte Carlo calculations and measured correction factors. Ali and Rogers (2008a) performed a range of benchmark calculations of the EGSnrc code in the kilovoltage energy range for charged particle backscatter coefficients.
with comparison against experimental measurements. They found the agreement between both sets of data generally within 4%.

Numerous studies have used EGSnrc/BEAMnrc to model therapeutic clinical kilovoltage x-ray units with beams in the energy range from 50 to 300 kVp (Fletcher and Mills 2008, Hill et al 2009, Keali et al 1999, Kim et al 2010, Knight and Nahum 1994, Knöös et al 2007, Lye et al 2010a, Ma et al 1998, Munck Af Rosenschold et al 2008, Natto 2002, Ubirch et al 2008, Verhaegen et al 1999). These studies used Monte Carlo simulations to generate the relative dosimetry data such as depth doses, relative profiles and BSFs for comparison with the measured data. In two recent studies, Knöös et al (2007), Munck Af Rosenschold et al (2008) used BEAMnrc to model a Gulmay kilovoltage x-ray unit. They used BEAMnrc to calculate depth doses and profiles in a water phantom as well as isodoses in patient CT data. Other studies have used BEAMnrc to model CT scanners or kilovoltage treatment units as used for animal studies (Bazalova et al 2009, Chow 2010, Omrane et al 2003).

EGSnrc/EGS4 has been used to determine the energy response of different radiation detectors (Agyingi et al 2006, Ma and Nahum 1995a, 1995b, Mobit et al 1998, 2000, Ubirch et al 2008). One study investigated the response of ionization chambers in medium-energy kilovoltage x-ray beams for the energy range 70–280 kVp (Ubirch et al 2008). They studied the influence of the central electrode dimensions on the beam quality correction factor \( k_Q \). The agreement between calculated and measured \( k_Q \) was better than 0.6% if the correct geometry was used for the central electrode. A number of these studies have used BEAMnrc to model FACs as used in primary dosimetry standards laboratories as well as the temperature–pressure correction factor \( P_{TP} \) for ionization chambers (La Russa and Rogers 2006, La Russa et al 2007, Lye et al 2010b).

Several studies have calculated doses using EGSnrc/EGS4 in water and solid phantom materials for kilovoltage x-ray beams (Gorjiara et al 2011a, 2011b, Hill et al 2008, 2009, Ma and Seuntjens 1999, Reniers et al 2004, Ye et al 2004). Reniers et al (2004) calculated the radial dose distribution in several phantom materials from low energy brachytherapy sources and compared with measured doses using TLDs. They found a good agreement between calculated and measured doses when the new cross section data were used, particularly for photon energies less than 100 keV.

8.2. Geant4

The software toolkit Geant4\(^6\) was originally developed by CERN for high-energy particle physics applications. Its radiation transport simulation capability has been substantially extended to the relatively low energies applicable to medical physics applications. For energies in the kilovoltage range in particular, Geant4 provides user-specified low energy and very low-energy (now called Geant4-DNA) electromagnetic radiation physics models that continue to be comprehensively tested for dosimetry calculations (Allison et al 2006, Amako et al 2005, Faddegon et al 2009, Francis et al 2011, Incerti et al 2010, Poon and Verhaegen 2005).

To date, Geant4 has been used in a number of Monte Carlo studies for low energy x-ray beams such as transmission values, dose calculations for different tissues, modelling of kilovoltage x-ray units and dosimetry of low energy brachytherapy sources (Clausen et al 2012, Landry et al 2010, Li et al 2012, Lian et al 2011, Rivard et al 2010).

The low-energy models in Geant4 use the Livermore and PENELOPE cross-sections for photon and electron transport, which have been validated down to energies of 250 eV. These models have been applied to brachytherapy (Afsharpour et al 2012) as well as radiobiology

\(^6\) geant4.web.cern.ch
and microdosimetry simulations (Byrne et al 2013, Chauvie et al 2007, Kuncic et al 2012). Geant4-DNA provides cross-section models for electrons in liquid water down to energies of just a few eV and is thus contributing to emerging radiobiology and nanodosimetry simulation studies (Bernal et al 2011, McNamara et al 2012).

8.3. MCNP

The most recently released version of the MCNP Monte Carlo code is MCNP6 (Goorley et al 2013). MCNP has been used to model low energy x-ray beams in a number of studies (Boudou et al 2004, 2005, Ding et al 2010, Edouard et al 2010, Hadid et al 2009, Nuttens et al 2008). Some of these studies used MCNP to calculate doses in water and various solid phantom materials as well as interface dosimetry including comparisons with other Monte Carlo codes (Chibani and Li 2002, Hadid et al 2010, Reniers et al 2004, Ye et al 2004). In one study, Traub et al (1997) calculated BSF for different solid calibration phantoms and found differences of up to 8% between the different phantom materials.

The MCNP code has also been used for calculating the dosimetry data from kilovoltage x-ray beams including depth doses and profiles by a full geometric model of the therapeutic kilovoltage x-ray unit (Natto 2002, Verhaegen et al 1999). In the study by Kim et al (2006), the MCNP code was used to calculate BSFs for kilovoltage x-ray beams and compared to measurements using a solid state detector. More recently, Rivard et al (2010) compared the dosimetry of different low energy brachytherapy sources from MCNPX and other Monte Carlo codes. They found that there was no statistically significant difference in calculated dose or kerma value between the Monte Carlo codes for the same beam spectrum.

8.4. PENELOPE


Several studies have evaluated the accuracy of the PENELOPE code for predicting x-ray spectra, by comparison with both the experimental data and theoretical calculations (Bote et al 2008, Llovet et al 2003, Salvat et al 2006b, Tian et al 2009). In one study, Llovet et al (2003) found that the PENELOPE code was suitable for calculating x-ray spectra emitted from targets bombarded with kilovoltage electron beams and gave a good agreement with the experimental spectral data. Subsequently, the PENELOPE code has been used for calculating x-ray spectra from tungsten and other targets (Garnica-Garza 2011).

A number of studies have used PENELOPE to evaluate the radiological water equivalence of different solid phantoms for low energy photons (Hill et al 2010, Marques et al 2010). The PENELOPE code has also been used for calculating the dosimetry data from kilovoltage x-ray beams including depth doses, profiles and BSFs as well as radial dose distributions from low energy brachytherapy sources (Assiamah et al 2007, Chica et al 2008, 2009, 2010, Croce et al 2012, Hill et al 2010, Ye et al 2004).

9. Conclusions and future perspectives

Kilovoltage x-ray beams will continue to provide a useful treatment modality in many radiotherapy departments. While being used predominantly for skin cancers, a number of
recent developments have seen their application in both intraoperative x-ray units and animal irradiators. Reported issues in the past relate to achieving accurate relative dosimetry and also with the recommendations for the use of generic published data. However, it has been shown that it is now possible to measure relative doses accurately using a wide range of dosimetry tools. In addition, in silico Monte Carlo techniques are being used to solve many problems for both relative dosimetry and for determining required factors for reference dosimetry.

There are still significant areas requiring investigation and development, in order to further improve the accuracy of the dosimetry and clinical application of kilovoltage therapy x-ray beams. Clear current examples are: the development and validation of kilovoltage treatment planning systems; the investigation of new dosimeters including 3D dosimetry systems; increasing use of Monte Carlo approaches and the increased accuracy and availability of dose-to-water based calibrations for kilovoltage x-ray beams. In addition, developments in kilovoltage x-ray beam use in related areas are likely to inform and strengthen the physics and dosimetry knowledge base for therapy beam use. For example, there is increased use of kilovoltage x-ray imaging on linear accelerators and other treatment units. There is also significant work in microbeam and irradiation systems for radiobiology work which require improved dosimetry and delivery methods. Thus it is expected that this area of clinical radiation physics will continue to provide significant research challenges.

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