Radiotherapy treatment verification using radiological thickness measured with an amorphous silicon electronic portal imaging device: Monte Carlo simulation and experiment

T Kairn¹, D Cassidy², P M Sandford¹‡, A L Fielding¹

¹ School of Physical and Chemical Sciences,
Queensland University of Technology, GPO Box 2434, Brisbane, Qld 4001, Australia
² Cancer Care Services, Royal Brisbane and Women’s Hospital,
Herston, Qld 4029, Australia

E-mail: a.fielding@qut.edu.au

Abstract.
This work validates the use of an amorphous-silicon, flat-panel electronic portal imaging device (α-Si EPID) for use as a gauge of patient or phantom radiological thickness, as an alternative to dosimetry. The response of the α-Si EPID is calibrated by adapting a technique previously applied to scanning liquid ion chamber EPIDs, and the stability, accuracy and reliability of this calibration are explored in detail. We find that the stability of this calibration, between different linacs at the same centre, is sufficient to justify calibrating only one of the EPIDs every month, and using the calibration data thus obtained to perform measurements on all of the other linacs. Radiological thickness is shown to provide a reliable means of relating experimental measurements to the results of BEAMnrc Monte Carlo simulations of the linac-phantom-EPID system. For these reasons we suggest that radiological thickness can be used to verify radiotherapy treatment delivery and identify changes in the treatment field, patient position and target location, as well as patient physical thickness.

PACS numbers: 87.53.Bn, 87.53.Oq, 87.53.Wz, 87.64.Aa

Keywords: Portal dosimetry, EPID, Monte Carlo dose calculation, radiological thickness


‡ Present address: Radiation Oncology Services, Princess Alexandria Hospital, Woolloongabba Qld 4102, Australia
1. Introduction

Amorphous silicon electronic portal imaging devices (a-Si EPIDs) are beginning to be used for radiotherapy treatment dose verification in a clinical setting (McDermott et al. 2005, Nijsten, Mijnheer, Dekker, Lambin & Minken 2007). The application of a technology primarily designed for beam-target-alignment to the more-complex problem of radiation dosimetry raises some challenging physical issues (Fielding et al. 2002, Swindell & Evans 1996, Evans et al. 1999, Mohammadi & Bezak 2005, Symonds-Taylor et al. 1997). One means of monitoring the delivery of a radiotherapy treatment, using an EPID, involves generating a predicted image of each field at the detector and comparing this with the actual EPID signal. Monte Carlo (MC) simulations provide a means of producing reliable models of a-Si EPID panels and generating the required image predictions (Parent, Seco, Evans, Fielding & Dance 2006, Parent, Seco, Evans, Dance & Fielding 2006). However, reliably comparing the resulting simulation data with experimental results (other than in relative terms) remains a challenge. The purpose of this paper is to suggest a clinically viable method of generating qualitatively and quantitatively comparable data from experiment and MC simulation. This method is suggested as a means of monitoring radiotherapy treatment delivery from fraction to fraction.

Specifically, this work aims to examine the process of determining the parameters required for the a-Si, flat-panel EPID to be used as a radiological thickness gauge. Following previous authors (Morton et al. 1991, Fielding et al. 2002) we use a variable-thickness PMMA phantom to derive a relationship between phantom thickness and EPID signal, and refer to this process as a ‘calibration’ for ‘PMMA-equivalent radiological thickness’ (abbreviated to ‘radiological thickness’). Here radiological thickness is used to denote the thickness of the imaged phantom as measured via the transmission of a megavoltage photon beam. The use of the radiological thickness calibration method has been described previously for a scanning liquid ion chamber EPID (Evans et al. 1999), a scanning solid-state EPID (Morton et al. 1991, Evans et al. 1995) and for a fluoroscopic camera-based EPID (Fielding et al. 2002). The current study applies this method to the Elekta iView GT a-Si flat panel EPID (Elekta Ltd, Crawley, UK).

In order to establish the clinical utility of using EPID-based radiological thickness measurements for dose verification, it is necessary to confirm the stability over time of the required calibration coefficients and correction factors as well as the radiological thickness measurements themselves. Similarly, the degree of transferability of these data between linacs (using calibration data obtained from one machine to measure the radiological thickness of an object imaged on another) is an important but previously unexplored question, of great relevance to potential clinical use.

To run the requisite simulations, to produce data for comparison with these experiments, the EGSnrc/BEAMnrc MC codes are used. These codes have been robustly established as capable of reliable simulation of medical photon-beam linear
accelerators (Fix et al. 2004, Fix et al. 2005, Kawrakow et al. 1996, Keall et al. 2003, Walters et al. 2002, Tzedakis et al. 2004, Aljarrah et al. 2006, Lovelock et al. 1995) and a-Si EPIDs (Kirkby & Sloboda 2005b, Kirkby & Sloboda 2005a, Siebers et al. 2004, Parent, Seco, Evans, Fielding & Dance 2006). Various authors have used MC simulations to evaluate both the deposition (Kirkby & Sloboda 2005b, Kirkby & Sloboda 2005a, Siebers et al. 2004, Parent, Seco, Evans, Fielding & Dance 2006) and attenuation (Kirkby & Sloboda 2005b, du Plessis & Willemse 2003) of energy that occurs as a radiotherapy photon beam traverses a phantom-EPID system. Kirkby and Sloboda (Kirkby & Sloboda 2005a, Kirkby & Sloboda 2005b) have studied the effects on beam energy and the consequent variation of EPID response when various physical thicknesses of phantom and EPID buildup plate are placed in the path of the beam. Siebers et al (Siebers et al. 2004) and Parent et al (Parent, Seco, Evans, Fielding & Dance 2006) have examined the EPID dose deposition arising from differently shaped and modulated fields. The current study builds on this work by suggesting a means of evaluating an absolute (rather than relative) property, radiological thickness, which can be used to make quantitative comparisons between simulation and experimental results.

Radiological thicknesses measured at the EPID (in cm) can be used to reveal the same information (eg. changes in field size, patient position and organ location) that is usually sought through calculations of EPID dose (in Gy) (Chen et al. 2006, Nijsten, van Elmpt, Jacobs, Mijnheer, Dekker, Lambin & Minken 2007). If a predicted radiological thickness map is obtained from MC simulation and compared with the actual radiological thickness maps obtained throughout the treatment, then any difference between these can be attributed to a change in linac output, dose delivery, patient position or patient physical thickness.

This work has two main aims: firstly, to show that a-Si EPIDs can be calibrated to accurately and reproducibly derive patient or phantom radiological thickness; and secondly, to show that a full Monte Carlo simulation of the linac-patient-EPID system can be performed to produce a predicted radiological thickness image for comparison with measurement.

2. Experimental method

2.1. Linac, EPID and phantoms

Experimental measurements used in this study were made using three Elekta Precise linear accelerators located at the same centre, each producing a nominal 6 MV photon beam governed by a flattening filter, a multi-leaf collimator and two pair of orthogonal jaws. Each exposure used in this study uses 100 MU delivered at a nominal dose rate of 600 MU/min. The associated EPID is an Elekta iView GT a-Si EPID (with iView software version 3.02), mounted on the linac gantry such that its detecting surface is $\approx$ 160 cm from the photon source.

Like most contemporary commercial EPIDs (Antonuk 2002), this detector consists
of a metallic buildup layer over a phosphor (in this case, a Kodak Lanex Fast screen) which delivers optical photons to an $\alpha$-Si-mounted photodiode-transistor array.

The automatic uniform normalisation of the EPID’s signal, which is usually made by Elekta’s iView image acquisition software, is deactivated for this study. However, the images generated by this device are nevertheless corrected by iView, by normalising out any unflatness in the beam and pixel sensitivity variation in the EPID (by dividing by an open flood field image, $I_{\text{open}}(x,y)$) and subtracting any electronic offset signal from the image (by subtracting a dark field image recorded either at the time of EPID quality assurance testing, $I_{\text{dark}}(x,y,t_0)$, or at a time close to the time that the raw image is recorded, $I_{\text{dark}}(x,y,t)$). The resulting image, described by (Parent, Seco, Evans, Fielding & Dance 2006)

$$I(x, y) = \frac{I_{\text{raw}}(x, y) - I_{\text{dark}}(x, y, t)}{I_{\text{open}}(x, y) - I_{\text{dark}}(x, y, t_0)},$$

is recorded by iView in the hospital information system (HIS) format and analysed using code written in the Interactive Display Language, IDL (ITT Visual Information Solutions, Boulder CO) (Perez 1998). This permits detailed off-line analyses of the EPID images as well as qualitative and quantitative comparison with the results of our MC simulations. It is through calculations using our IDL code that the EPID is calibrated and then used to measure the radiological thickness of various types of radiotherapy phantoms.

The phantoms imaged in this experiment are placed between linac and detector on a ‘tennis racket’ couch support with varying source-to-surface distances (SSDs). For our radiological thickness calibration, eight uniform blocks of a poly(methyl methacrylate)-like material (henceforward referred to as ‘PMMA’) are used. Each of these blocks has a mass density of $\rho = 1.17 \pm 0.05$ g/cm$^3$, an electron density of $\rho_e = 3.65 \times 10^{23}$ electrons/cm$^3$, a physical thickness of $3.8 \pm 0.1$ cm and an area normal to the beam direction of $40.0 \times 40.0$ cm$^2$. These are combined to produce phantoms with total physical thicknesses ranging from 3.8 to 30.4 cm. The maximum physical thickness of 30.4 cm is chosen for its comparability with typical patient physical thicknesses. The accuracy of this calibration, over a range of field sizes, is studied by imaging the PMMA blocks using $5 \times 5$, $10 \times 10$, $15 \times 15$ and $20 \times 20$ cm$^2$ (on axis) fields. The reliability of the method for measuring the radiological thickness of a tissue-like material, other than PMMA, is tested by imaging combinations of four $20.0 \times 20.0 \times 5.0$ cm$^3$ blocks, one $20.0 \times 20.0 \times 0.5$ cm$^3$ sheet and one $20.0 \times 20.0 \times 0.2$ cm$^3$ sheet of Solid Water (GAMMEX rmi, Middleton, USA), using a $10 \times 10$ cm$^2$ field. These blocks are set up with a fixed air gap to the EPID, with total thicknesses and SSDs as listed in Table 2.

Additionally, the method is tested for a more realistic clinical geometry by imaging an anthropomorphic phantom. Here, we use a Supertech head phantom (Supertech, Elkhart, USA), comprised of a human skull (with bone density $\rho \approx 1.4 – 1.8$ g/cm$^3$ and tooth-density $\rho \approx 2.7$ g/cm$^3$), without cervical spine, suspended in a head-shaped block of opaque tissue-equivalent plastic (with density $\rho \approx 1.0$ g/cm$^3$), with lower-density materials filling the cranium and the oral and nasal cavities. This phantom is set up
isocentrically and imaged using \((10 \times 10 \text{ and } 25 \times 25 \text{ cm}^2)\) anterior fields, with a \(\simeq 90\) cm SSD.

Images of all of these phantoms are converted to radiological thickness, for comparison with MC results, as described in Section 2.2.

EPID exposures taken in the absence of any phantoms also contribute to the radiological thickness calibration through \(I_0\), the open-field image, and through the field size correction (see Section 2.2). Although the tennis racket has a negligible effect on our results, this structure is kept in the path of the beam during the ‘open field’ exposures, for the sake of consistency with the other calibration exposures.

In describing and analysing this linac-phantom-detector setup, \(x\)- and \(y\)-axes are defined as orthogonal to the beam axis and parallel to the surface of the phantom/detector. All positions in the \(x-y\) plane are measured at the level of the EPID’s active layer (approximately 160 cm from the source), and all field sizes quoted are measured at the isocentre (100 cm from the source). Throughout this paper, \(I(x, y)\) is used to denote the signal intensity as measured by the EPID.

### 2.2. Calibration and radiological thickness measurement

We expose stacks of zero to eight blocks of 3.8-cm-thick PMMA to a field with dimensions \(25 \times 25 \text{ cm}^2\) at the machine isocentre, which is 100 cm from the photon source. (This is the ‘reference field size’ referred to below.) This calibration is conducted with the phantom set up isocentrically.

Images of these stacks are obtained from the EPID and used to calibrate the detector to make radiological thickness measurements, following the quadratic calibration method described in detail by Morton et al. (Morton et al. 1991). The decrease in the EPID signal, \(I(x, y)\), from its open field value, \(I_0(x, y)\), as phantom physical thickness is increased can be described by an exponential, so that the negative logarithm of the transmission, \(T(x, y)\), becomes

\[
-\ln (T(x, y)) = -\ln \left( \frac{I(x, y)}{I_0(x, y)} \right) = \alpha(x, y)t_p + \beta(x, y)t_p(x, y)^2
\]

where \(t_p(x, y)\) is the physical thickness of the phantom, and \(\alpha\) and \(\beta\) are treated as free fitting parameters. The physical thickness \(t_p(x, y)\) is evaluated at each point \((x, y)\), to account for the slightly varying physical thickness of material in the path of the beam, arising from its divergence. This physical thickness is calculated at each point as

\[
t_p(x, y) = \frac{t_p(0, 0)\sqrt{x^2 + y^2}}{SDD(0, 0)\sin\theta},
\]

where \(t_p(0, 0)\) is the physical thickness on the central axis, SDD is the source-detector distance and \(\theta\) is the angle between the pixel at any point \((x, y)\), the source and the central pixel \((0, 0)\), or

\[
\theta = \tan^{-1} \left( \frac{\sqrt{x^2 + y^2}}{SDD(0, 0)} \right).
\]
Equation (2) is fitted to the EPID data for each pixel at the various physical thicknesses, allowing two-dimensional maps of $\alpha(x,y)$ and $\beta(x,y)$ to be generated. Inversion of equation (2) gives the relationship,

$$t_p(x,y) = \frac{-\alpha(x,y) \pm \sqrt{\alpha(x,y)^2 - 4\beta(x,y) \ln[I(x,y)/I_0(x,y)]}}{2\beta(x,y)},$$

which can be used to produce a first approximation to the radiological thickness of an irradiated object, by replacing $I(x,y)$ with the intensity detected with the object in the path of the beam, $I_{\text{object}}(x,y)$.

Importantly, the calibration used here is carried out using a reference field size of $25 \times 25 \text{ cm}^2$. Consequently, equation (5) provides an adequate description of an object’s radiological thickness only when the object is measured using a $25 \times 25 \text{ cm}^2$ field. To calibrate the EPID as a gauge that will reliably measure the radiological thicknesses of materials subjected to fields of various sizes, corrections for field area and resulting phantom scatter must be calculated. The EPID image when corrected for field size and scatter is described by (Evans et al. 1995)

$$I_{\text{corrected}}(x,y) = F \cdot \frac{1 + SPR_{\text{ref}}}{1 + SPR} \cdot I_{\text{object}}(x,y)$$

where $F$ denotes an open field output factor, which is equal to the ratio of the open-field EPID signal at the field size for which $I_{\text{object}}(x,y)$ is measured to the open-field EPID signal, at the reference field size. $SPR$ is the scatter-to-primary ratio, calculated in the presence of the calibration phantom, and can be approximated by (Swindell & Evans 1996)

$$SPR = k_0 At$$

where $A$ is the area of the field (measured at the centre of the phantom) at which $I_{\text{object}}(x,y)$ is measured (or the area of the reference field, for $SPR_{\text{ref}}$) and $t$ is the thickness of the object being imaged. The parameter $k_0$ combines the effects on the $SPR$ of the system’s geometry and the calibration material’s electron density, being equal to

$$k_0 = \rho_e \left( \frac{d\sigma}{d\Omega} \right)_{\theta=0} \left( \frac{(L_1 + L_2)^2}{(L_1 L_2)^2} \right).$$

$L_1$ and $L_2$ are the distances from the source to the isocentre and from the isocentre to the detector, respectively (shown in Figure 1). Examination of the Klein-Nishina equation indicates that the zero-deflection-angle differential cross-section, $\left( \frac{d\sigma}{d\Omega} \right)_{\theta=0}$, is equal to $r_0^2$, the square of the classical electron radius, which is $79.408 \times 10^{-27} \text{ cm}^2$. Since the electron density of our PMMA is $\rho_e = 3.65 \times 10^{23}$ electrons/cm$^3$, for this material,

$$\rho_e \left( \frac{d\sigma}{d\Omega} \right)_{\theta=0} = 0.029 \text{ cm}^{-3}.$$  

Therefore, for our experimental setup, as described above, $k_0 = 2.06 \times 10^{-5} \text{ cm}^{-3}$. 


Using this calibration to calculate the radiological thickness image for an object from its EPID image involves a straightforward process of iteration. At each \((x, y)\)-position: \(t\) is estimated as the physical thickness of the phantom, measured on the central axis; \(SPR\) and \(SPR_{ref}\) are calculated from equation (7); these are substituted into equation (6) to calculate \(I_{\text{corrected}}\); this value, \(I_{\text{corrected}}\), is used as \(I_{\text{object}}\) and is substituted along with \(\alpha\) and \(\beta\) into equation (5) to give a new value of \(t\); this corrected value of \(t\) is substituted back into equation (7) and the process is repeated. Following previous authors (Fielding et al. 2002) who saw convergence in \(t(x, y)\) after three iterations, and having established that in our system \(t(x, y)\) has converged after five iterations, we choose to repeat this calculation five times to obtain our radiological thickness measurements. Central axis values of the radiological thickness and other parameters of interest are calculated as mean values, over a 2 cm\(^2\) region of interest, at the centre of the EPID.

Predictions of the radiological thicknesses of the planar phantoms imaged in this work are made by correcting the physical thicknesses of these phantoms \((t_p)\), with reference to the relationship between the electron density of each phantom \((\rho_e)\) and the electron density of the calibration material \((\rho_{e,\text{PMMA}})\),

\[
t_{p,\text{corr}} = t_p \left( \frac{\rho_e}{\rho_{e,\text{PMMA}}} \right). \tag{10}
\]

In this study, radiological thickness calibrations and measurements are repeated after 14 days, 28 days and 42 days, to investigate the reliability of the procedure. (Dates of measurements are determined by linac availability, and calibration data sets are not acquired with all measurements.) The conversion to radiological thickness is performed, for each measurement image, using calibration data recorded on the same machine on the same day (in instances where such data was recorded). Additionally, this analysis is also performed with each measurement image converted to radiological thickness using every other calibration data set available. (All of these calculations are made with reference to a value for \(I_0(x, y)\) which is obtained using a 25 × 25 cm\(^2\) open field exposure made on the same day as the measurement image.) This allows us to verify that this technique remains valid even when calibration data are recorded infrequently and on different linacs.

3. Simulation method

3.1. Simulation model and run parameters

To investigate the underlying physical effects which have an influence on the radiological thickness calibration and measurement, we employ simulations using the EGSnrc MC code (Nelson et al. 1985, Kawrakow & Rogers 2003). User codes applied in this work are: BEAMnrc (Rogers et al. 1995, Ma et al. 1997, Kawrakow & Walters 2006, Rogers et al. 2004), with which the transmission of the treatment beam through the linear accelerator is simulated; DOSXYZnrc (Kawrakow & Walters 2006, Walters et al. 2005),
which uses the phase-space data produced by the accelerator simulation to calculate doses within our phantom-detector system; and CTCREATE, with which a voxelised virtual phantom suitable for DOSXYZnrc calculations can be generated from CT data.

In our BEAMnrc and DOSXYZnrc simulations, we use the PRESTA-II electron step algorithm (Rogers et al. 2004). The boundary crossings in the EPID, where the precise transport of electrons through the thin layers (and small voxels) that make up the model detector is very important, are accounted for using EGSnrc’s EXACT algorithm.

To avoid the consumption of computational time by trajectory calculations for particles which are unlikely to reach the phantom, we choose to terminate the histories of electrons whose energies fall below 700 keV and photons whose energies fall below 10 keV, throughout the accelerator. Similarly, where secondary particles arise from electron interactions, only the production of those particles which have energies exceeding these limits is modeled explicitly. For our 6 MV beam simulation, the use of these electron and photon energy cutoffs is expected to have a negligible effect on the results (Rogers et al. 1995, Kawrakow & Walters 2006).

Maximum statistical accuracy in our results is obtained with a minimum computation-time cost by employing the ‘range rejection’ and ‘photon splitting’ variance reduction techniques in the BEAMnrc and DOSXYZnrc simulations (Kawrakow & Walters 2006, Kawrakow et al. 2004, Rogers et al. 2004).

In our BEAMnrc simulations of the treatment head, $1 \times 10^7 - 2 \times 10^8$ initial electrons are used to produce between $1 \times 10^8$ and $3 \times 10^8$ particles at the linac’s exit plane (depending on field size). The DOSXYZnrc component of the simulation, incorporating the transmission of the beam through the various phantoms and the deposition of dose in the model EPID, is run for $5 \times 10^8$ to $50 \times 10^8$ particle histories. As serial jobs, these simulations can take up to 60 hours to complete. When run in parallel, however, using a 128-processor SGI Origin 3000 supercomputer, these simulations can be completed within 4-5 hours.

3.2. Virtual linac, EPID and phantoms

The linear accelerator itself is modeled in BEAMnrc according to the manufacturer’s specifications for an Elekta Precise linear accelerator, incorporating the various materials and geometries that are specific to this apparatus. This virtual accelerator was commissioned to accurately model the output of the real accelerator operating at a nominal 6 MV, via a procedure described in detail elsewhere (Tzedakis et al. 2004, Aljarrah et al. 2006).

Briefly, depth-dose and lateral profiles in a simple virtual water tank are calculated by the simulation, for on-axis square fields ranging in size from $5 \times 5$ to $40 \times 40$ cm$^2$. These data are compared to profiles measured experimentally and the model parameters defining the initial electron beam characteristics are iteratively varied until the dose calculated in the simulation matches the dose measured by the experiment. (This was achieved with a monoenergetic, parallel, circular electron beam with a gaussian radial
DOSXYZnrc is used to calculate dose in a model EPID which is located beneath an appropriately sized volume of air, under each phantom examined (as shown in Figure 1). This virtual detector is modeled according to the manufacturer’s specifications, so that it matches as closely as possible the Elekta iView GT a-Si EPID used for the experimental measurements. Like previous authors, (Parent, Seco, Evans, Fielding & Dance 2006), we include all component layers of the EPID explicitly (from its front cover to its rear support), rather than using the more-efficient method of modeling the components beyond the phosphor as a uniform source of scatter (such as a volume of water (Siebers et al. 2004)). The model does not, however, explicitly include the plastic housing and electronics which lie around the sides of the detector. To account for these, the model extends 0.5 cm beyond the dimensions of the clinical EPID, in each direction, to permit the generation of additional scatter into the phosphor. The EPID thus modeled has a total area of $42 \times 42 \text{cm}^2$. The model EPID has a pixel size of $0.25 \times 0.25 \text{cm}^2$, which is relatively large compared to the pixel size of the clinical device, but which was chosen to balance computational efficiency with resulting image detail.

Calibration phantoms used here are modeled in DOSXYZnrc to simulate the PMMA blocks described in Section 2.1, with SSDs matching those listed in Table 2. Measurements of radiological thickness are made using a series of virtual phantoms, also based on the phantoms described in Section 2.1.

Water volumes measuring $20 \times 20 \times N \text{cm}^3$, where $N = 5, 10$ and $20$ cm, are simulated at SSDs listed in Table 2 (which produce a constant air gap between phantom and EPID), to provide comparisons with the experimental results.

The Supertech head phantom described in Section 2.1 is incorporated into a DOSXYZnrc simulation by CT scanning it and converting this CT data into a voxelised
virtual phantom. The requisite CT data is obtained using a Siemens Sensation 4 scanner (Siemens AG, Munich, Germany). The CT slices are converted into .egsphant format, readable by DOSXYZnrc by using CTCREATE to assign a material and a density to each voxel in the phantom, based on its CT number (Kawrakow et al. 1996). This .egsphant data is merged with an .egsphant file describing the EPID generated using DOSXYZnrc, producing a hybrid phantom incorporating the head phantom and the EPID, aligned and separated appropriately to replicate the experimental system.

3.3. Calibration and radiological thickness measurement, by simulation

The model EPID is calibrated for radiological thickness using a method similar to that used for the measured data (see Section 2.2). Using the phase space file resulting from running the BEAMnrc linear accelerator simulation for a field size (at 100 cm) of $25 \times 25$ cm$^2$, DOSXYZnrc is run for each of the nine (including zero) physical thicknesses of the PMMA phantom, and nine corresponding dose maps at the EPID’s phosphor screen are generated. These files are imported to IDL and analysed with reference to equation (2), to produce maps of $\alpha(x, y)$ and $\beta(x, y)$. In combination with the output factors ($F$) and SPRs calculated for this simulated system, these coefficients can be used to derive the radiological thickness of each of the objects (PMMA, water and head phantom) imaged by the virtual EPID.

4. Results

4.1. Radiological thickness calibration

Figures 2(a), (b) and (c) illustrate comparisons between measurement and simulation results for profiles in the open field intensity, transmission (for phantom thicknesses of 15.2 and 30.4 cm only) and the linear calibration coefficient, respectively.

In Figure 2(a), the simulated and experimental open field profiles are normalised to their respective values on the central axis of the field. The experimental profile for the $25 \times 25$ cm$^2$ field detected by the EPID in the absence of the phantom is very flat, due to the shape of the beam and any variations in pixel sensitivity having been removed from the signal by the flood field correction (see equation (1)). In contrast, the nonuniform profile of the beam which arises from its passage through the flattening filter, as well as the effect that this varying energy spectrum has on EPID response, remain present in the simulation results, which have not been flood-field corrected.

More usefully, this beam-profile effect can be negated by evaluating the transmission, $T(x, y)$, for both sets of data. This effectively removes the flood field correction from the experimental data and produces a set of profiles which describe the effects that the phantom is having on the beam, permitting a more direct comparison of simulated and experimental results. Figure 2(b) shows the results of such a calculation for the system in the presence of 4 and 8 blocks of PMMA ($t_p(0, 0) = 15.2$ and 30.4 cm). This illustration indicates that the intensity of the beam after transmission through the
Figure 2. (a) Open-field intensity, $I_0(x,0)$. (b) Transmission, $T(x,0)$ when field is passing through 30.4 cm of PMMA (lower data points) and 15.2 cm of PMMA (upper data points). (c) Linear calibration coefficient, $\alpha(x,0)$. (d) Quadratic calibration coefficient, $\beta(x,0)$. Open data points are from simulation, solid data points are from experiment. Error bars are omitted for clarity.

Figure 3. $-\ln(T(0,0))$ vs physical thickness on the central axis (CAX) for a 25 × 25 cm² field. Open (small) data points are from simulation, solid (large) data points are from experiment. Solid line shows the result of fitting equation (2) to the experimental data. Dotted line is an arbitrary straight line included for comparison.
PMMA is, in both cases, reduced below its open field intensity, with the thicker phantom producing the greater reduction. (Figure 3 illustrates the relationship between phantom physical thickness and beam intensity more explicitly.)

The change in the energy spectrum of the beam towards the edges of the field is reflected in the profiles for $\alpha(x,0)$ and $\beta(x,0)$ shown in Figures 2(c) and (d). These results for $\alpha$ are derived from the experimental and simulated calibration images as described in Section 2.2 and are indicative of the effect that the real and simulated flattening filters have on the energy of the beams (Symonds-Taylor et al. 1997).

Figures 2(b), (c) and (d) show very good qualitative and quantitative agreement between the experimental and simulation results, which is achieved without normalising, applying conversion factors or any additional manipulation of the data.

Figure 3 illustrates an intermediate result in this calculation. Here, equation (2) is fitted to the negative logarithm of the transmission through the various physical thicknesses of PMMA, measured on the central axis. The agreement between simulation and experiment makes these two sets of data points indistinguishable. Also included in Figure 3 is a straight line, which indicates the deviation of the data from linearity. This confirms that, although the second-order calibration coefficient $\beta$ has a small magnitude (as shown by Figure 3(d)), it is nonetheless necessary for accurately describing the photon beam transmission.

4.2. Radiological thickness measurement: Stability and reproducibility

| Table 1. | Measured radiological thickness of phantoms determined via experiment, using different linacs on different days. (All thicknesses are in cm. All fields are 10 $\times$ 10 cm$^2$. Number in brackets gives uncertainty (standard deviation) in last digit.) |
|---|---|---|
|    | linac 1 | linac 2 | linac 3 |
| day 0 | 4.52(6) | 4.55(2) | 4.55(2) |
| day 9 | 4.53(2) |       |       |
| day 14 | 4.57(6) | 4.58(2) |       |
| day 28 | 4.60(6) | 4.56(2) |       |
| day 42 | 4.57(6) | 4.56(2) |       |

To test the stability and reproducibility of the radiological thickness calibration, radiological thicknesses are determined for a uniform block of solid water with a physical thickness of 5 cm, using measurement and calibration images taken during a six week period. The expected result of this radiological thickness measurement is $t_{p,corr} = 4.57$ cm (from equation (10)).

Table 1 shows a typical set of results, obtained by using a calibration data set from each linear accelerator to derive radiological thickness values for measurement images obtained using the same linear accelerator at different times. These data are in agreement with each other and with the prediction $t_{p,corr}$. 


When the radiological thickness for each exposure is derived using the corresponding calibration data from the same machine on the same day, the radiological thickness which results from averaging these 9 measurements is $4.57 \pm 0.02$ cm. When a single calibration data set is used to derive a radiological thickness from each of the different measurement exposures, the radiological thickness which results from averaging these 13 measurements is $4.57 \pm 0.06$ cm.

To investigate the effect of the chosen calibration data set on the measured thickness, each of the different calibration data sets is also used to derive a radiological thickness from a single exposure. The radiological thicknesses which results from averaging these 9 measurements is $4.57 \pm 0.03$ cm.

These radiological thickness measurements remain consistent when evaluated using data collected over a period of six weeks and across three different linear accelerators.

4.3. Radiological thickness of planar phantoms: Effect of field size and physical thickness

Figures 4(a) and (b) show the results of performing the radiological thickness analysis on planar blocks of PMMA, exposed to square fields of varying size. Data points shown in these figures describe the radiological thickness as detected in a region of interest in the centre of the EPID. Figure 4(a) indicates the effect of changing the measurement field size without including the commensurate corrections for field size and resulting scatter in the thickness derivation. Here, radiological thickness strongly increases as the measurement field size decreases. By contrast, Figure 4(b) shows the results, for both simulation and experiment, when the appropriate field size and scatter corrections are included in the derivation. Here, strong agreement between simulation, experiment and physical thickness is shown, at all field sizes studied (see Table 2 for more detail), indicating that the corrections described by equations (6) to (9) are appropriately correcting the radiological thickness result.

Radiological thickness results determined for the water phantoms are listed in Table 2 and suggest that when a water-equivalent phantom is measured using this method, its electron density (and the fact that it differs from the electron density of PMMA) affects its measured radiological thickness. Data listed in Table 2 indicate the precision with which radiological thickness can be measured, using the $\alpha$-Si EPID, with all results for physical thicknesses of water between 5.0 and 20.0 cm falling within 1% of their expected values ($t_{p,corr}$).

Additionally, Table 2 lists radiological thicknesses of experimentally imaged water-equivalent blocks with thicknesses of 5.0, 5.2 and 5.5 cm as well as 20.0, 20.2 and 20.5 cm. These results indicate that it is possible to detect changes in physical thickness of at least 0.2 cm via experimental radiological thickness data.

Table 2 also shows the results of using simulations to determine radiological thicknesses of a planar, water phantom, at selected experimental geometries. These results are subject to statistical noise, and are therefore less precise than the
Figure 4. Measured radiological thickness (cm) vs physical thickness (cm), for 40 × 40 cm² PMMA blocks of various physical thicknesses, at various field sizes, (a) without field size and scatter correction and (b) with field size and scatter correction. Open data points are from simulation, solid data points are from experiment. Plot symbols: diamonds are for 5 × 5 cm² field; squares are for 10 × 10 cm² field; triangles are for 15 × 15 cm² field; circles are for 20 × 20 cm² field; and crosses are for 25 × 25 cm² field. (Simulation data are omitted from (a), for clarity.) Solid lines indicate parity between physical and radiological thicknesses. Dotted lines interpolate between data points as a guide only.

Experimental measurements (especially as physical thickness increases). Nevertheless, these data are in general agreement with both the predictions \( t_{p,corr} \) and the experimentally evaluated radiological thicknesses.

4.4. Radiological thickness of head phantom: Effect of inhomogeneity

The efficacy of the radiological thickness measurement as a means of relating simulation and experimental results is illustrated by Figures 5(a), (b) and (c), which indicate the information contained in radiological thickness profiles, derived via simulation and experiment, for the head phantom.

For reference, Figures 5(a) and (b) show the details of the system used to generate the results shown in Figure 5(c). Figure 5(a) illustrates an EPID image of the head
Table 2. Measured radiological thickness of phantoms determined via experiment ($t_{expt}$) and simulation ($t_{MC}$), compared to their physical thickness ($t_p$) and their physical thickness corrected for the difference in electron density between PMMA and water ($t_{p,corr}$). (All distances are in cm. All fields are $10 \times 10$ cm$^2$. Number in brackets gives uncertainty (standard deviation) in last digit.)

<table>
<thead>
<tr>
<th>Material</th>
<th>SSD</th>
<th>$t_p$</th>
<th>$t_{p,corr}$</th>
<th>$t_{expt}$</th>
<th>$t_{MC}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>PMMA</td>
<td>84.8</td>
<td>30.4</td>
<td>30.4</td>
<td>30.22(3)</td>
<td>30.9(5)</td>
</tr>
<tr>
<td>PMMA</td>
<td>86.7</td>
<td>26.6</td>
<td>26.6</td>
<td>26.35(3)</td>
<td>26.9(3)</td>
</tr>
<tr>
<td>PMMA</td>
<td>88.6</td>
<td>22.8</td>
<td>22.8</td>
<td>22.70(3)</td>
<td>22.9(3)</td>
</tr>
<tr>
<td>PMMA</td>
<td>90.5</td>
<td>19.0</td>
<td>19.0</td>
<td>18.92(3)</td>
<td>19.1(3)</td>
</tr>
<tr>
<td>PMMA</td>
<td>92.4</td>
<td>15.2</td>
<td>15.2</td>
<td>15.25(2)</td>
<td>15.3(3)</td>
</tr>
<tr>
<td>PMMA</td>
<td>94.3</td>
<td>11.4</td>
<td>11.4</td>
<td>11.42(2)</td>
<td>11.5(3)</td>
</tr>
<tr>
<td>PMMA</td>
<td>96.2</td>
<td>7.6</td>
<td>7.6</td>
<td>7.54(2)</td>
<td>7.7(2)</td>
</tr>
<tr>
<td>PMMA</td>
<td>98.1</td>
<td>3.8</td>
<td>3.8</td>
<td>3.86(2)</td>
<td>3.8(2)</td>
</tr>
<tr>
<td>Water</td>
<td>100.0</td>
<td>5.00</td>
<td>4.57</td>
<td>4.56(2)</td>
<td>4.6(3)</td>
</tr>
<tr>
<td>Water</td>
<td>99.8</td>
<td>5.20</td>
<td>4.76</td>
<td>4.73(2)</td>
<td></td>
</tr>
<tr>
<td>Water</td>
<td>99.5</td>
<td>5.50</td>
<td>5.03</td>
<td>5.01(2)</td>
<td></td>
</tr>
<tr>
<td>Water</td>
<td>95.0</td>
<td>10.00</td>
<td>9.15</td>
<td>9.15(2)</td>
<td>9.2(3)</td>
</tr>
<tr>
<td>Water</td>
<td>90.0</td>
<td>15.00</td>
<td>13.72</td>
<td>13.71(2)</td>
<td></td>
</tr>
<tr>
<td>Water</td>
<td>85.0</td>
<td>20.00</td>
<td>18.30</td>
<td>18.28(3)</td>
<td>18.8(4)</td>
</tr>
<tr>
<td>Water</td>
<td>84.8</td>
<td>20.20</td>
<td>18.48</td>
<td>18.48(3)</td>
<td></td>
</tr>
<tr>
<td>Water</td>
<td>84.5</td>
<td>20.50</td>
<td>18.75</td>
<td>18.74(3)</td>
<td></td>
</tr>
<tr>
<td>Supertech head</td>
<td>$\approx$ 90</td>
<td>$\approx$ 25</td>
<td>$\approx$ 12</td>
<td>$\approx$ 12</td>
<td></td>
</tr>
</tbody>
</table>

phantom, produced via MC simulation (from a $25 \times 25$ cm$^2$ exposure), marked with the position of the $10 \times 10$ cm$^2$ field used in the thickness evaluation. Figure 5(b) shows a CT slice through the head phantom, made at $y = -6.125$ (measured at the EPID plane, where the $10 \times 10$ cm$^2$ field projects to approximately $16 \times 16$ cm$^2$). This position is selected for analysis because it is very heterogeneous, with densities ranging from approximately $0.3 \text{ g/cm}^3$ to approximately $1.8 \text{ g/cm}^3$.

Figure 5(c) shows the profiles in radiological thickness which result from performing this analysis, using simulation and experimental data, for the $10 \times 10$ cm$^2$ field. These data show good qualitative and quantitative agreement, with the simulation result effectively replicating all of the major features of the experimental profile.

5. Discussion

The accuracy of the calibration for radiological thickness examined in this work is indicated by the results determined for the PMMA phantom, shown in Figure 4(b) and Table 2. At all the field sizes studied here, when calibrated using PMMA, the EPID measures a radiological thickness of PMMA which is consistently within 2 $\%$ of its physical thickness. As expected, when the measured material has the same composition and density as the calibration material, the measured radiological thickness functions as an indicator of the physical thickness of each phantom.
The radiological thickness measurement is also expected to be affected by the properties of the field used. Because the calibration is completed under reference conditions (using a square field with dimensions $25 \times 25$ cm$^2$ applied for 100 MU), any change in the beam used to acquire the image of the measured object must be accounted for. For the measurements described herein, the relevant change is to the size of the incident field. By omitting the requisite corrections for field size and scatter, the results shown in Figure 4(a) show how changes to the measurement field can affect the resulting radiological thickness.

The current study examines only on-axis, square fields, where the relationship between field size and EPID response can easily be predicted and corrected for (as is shown by the data in Figure 4(b)), but it suggests that radiological thickness could profitably be used to monitor the properties of the beam delivering a simple radiotherapy treatment, while also indicating the necessity of developing a means to account for field size changes for irregular, conformal fields.

These experimental results are also shown to be amenable to quantitative comparison with simulation results. Data illustrated in Figures 2, 3, 4(b) and 5(c) and
in Table 2 all show good agreement between the results of simulation and experimental measurements of the phantoms’ radiological thicknesses and the parameters used to determine them. In particular, Figure 5(c) shows how MC simulations incorporating a heterogeneous anthropomorphic phantom can predict the radiological thickness of the same phantom measured using the clinical radiotherapy system. Here, the variation in radiological thickness seen in the experimental measurement, arising from the presence of high and low density materials in the phantom (see Figure 5(b)), is replicated by the radiological thickness profile derived from simulation data.

This agreement results from applying the calibration procedure described in Section 2.2 to both systems, producing maps of radiological thickness (measured in cm) which can be compared without recourse to correction, normalisation or any other manipulation of the data. Since the original raw EPID images acquired using simulation and experiment differ qualitatively from each other, these results and the procedure they validate have strong practical value. These results constitute a novel application of MC simulation results to the successful replication and confirmation of experimental data.

5.1. Clinical utility

Any development of this method as a clinical technique for monitoring radiotherapeutic treatments demands that the reliability and reproducibility of the results be established. This is discussed in Section 4.2, where results of repeatedly determining the radiological thickness of a single, planar, water-equivalent phantom are compared. Different calibration data sets and different measurement images are obtained experimentally on different days with different linear accelerators (at the same centre) are shown to produce consistent results, over a six week period. The consistency of these radiological thickness measurements arises from the consistency (over time and across linacs) of the calibration coefficients (α and β) and the field-size factor (F). From these results it is apparent that where all machines at a given centre are of the same type and are maintained according to the same protocols, calibration data from one machine can be used to determine radiological thicknesses from images obtained on all machines.

We suggest the following protocol for the clinical application of this technique, for centres with one or more Elekta Precise linacs and Elekta iView EPID. A calibration can be completed according to the method described in Section 2, every month. This calibration may be completed using one or more of the linacs (depending on the conformity between machines). It is advisable to make a weekly check of the radiological thickness of a uniform object with each linac, to monitor the performance of each individual EPID panel and to confirm the stability of the system. On each machine, on each day, where radiological thickness images are going to be determined, an additional open 25 × 25 cm² field image needs also to be acquired, to provide an accurate \( I_0(x, y) \) for that machine on that day. The daily acquisition of this \( I_0(x, y) \) image accounts for the individual linac output, but in doing so makes the resulting radiological thickness results less sensitive to variations in this property. (Linac output needs to be monitored using...
existing QA techniques.) The MC simulation technique described in Section 3 should be used to generate a radiological thickness image based on each patient’s planning CT. This becomes the reference data set to which all subsequent radiological thickness images, generated using the treatment beam, can be compared. Radiological thickness images of each patient derived using treatment-time EPID images can also be compared to each other, to monitor inter-fraction changes in the dose delivery or the patient. The dependence of the results upon the internal anatomy of the patient suggests that this method may be applicable to the identification of internal organ motion or tumor shrinkage, in circumstances where the external physical thickness profile of the patient may appear unchanged. Apart from the EPID images already taken routinely as part of the radiotherapy treatment, use of this method requires minimal additional imaging and no increase in patient dose.

6. Conclusions

Radiological thickness can be accurately derived from both MC simulation data and from experimental measurements and thereafter used to relate and compare images from these two otherwise disparate sources. Indeed, very good agreement can be found between the radiological thicknesses, as well as the coefficients used in their calibration, evaluated for the simulation and experimental systems.

In using an EPID to derive radiological thickness, the important factors affecting the results are the output of the linac and the physical thickness and electron density of the imaged object. Measurements of the radiological thickness of various homogeneous, planar phantoms have been used here, to confirm the accuracy and reliability of the technique, to relate experimental results to simulation results, to exemplify the potential effect of field size variation on the measurement and to indicate the accuracy with which field-size dependence can be removed from the results, when simple, square fields are used. Additionally, images of a head phantom are used to generate radiological thickness profiles which confirm that this method remains viable for more clinically realistic geometries.

This study therefore suggests that radiological thickness determined using a calibrated a-Si EPID may be used to verify the delivery of radiotherapy treatments and detect changes to the physical thickness and anatomy of imaged patients as well as changes to the treatment beam. This study has also shown the feasibility of predicting radiological thickness using a MC simulation of the linac, phantom and EPID.

Acknowledgments

This work was funded by the National Health and Medical Research Council (NHMRC), through a project grant, and completed with the support of the Queensland University of Technology (QUT) and the Royal Brisbane and Women’s Hospital (RBWH). The authors would especially like to thank the staff of the RBWH for technical assistance in
obtaining the experimental measurements described in this paper. The authors are also grateful to Elekta for the provision of manufacturing specifications which permitted the detailed simulation of their linear accelerators and a-Si EPIDs. Computational resources and services used in this work were provided by the High Performance Computing (HPC) and Research Support Group, Queensland University of Technology, Brisbane, Australia.

References

Kawrakow I & Walters B R B 2006 Medical Physics 33(8), 3046–3056.
Kirkby C & Sloboda R 2005b Medical Physics 32(8), 2649–2658.