Received:
30 September 2014

Revised:
6 May 2015

Accepted:
12 May 2015

doi: 10.1259/bjr.20140645

Cite this article as:

Tan YI, Metwaly M, Glegg M, Baggarley SP, Elliott A. A dual two dimensional electronic portal imaging device transit dosimetry model based on an empirical quadratic formalism. *Br J Radiol* 2015; **88**: 20140645.

FULL PAPER

A dual two dimensional electronic portal imaging device transit dosimetry model based on an empirical quadratic formalism

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Objective: This study describes a two dimensional electronic portal imaging device (EPID) transit dosimetry model that can predict either: (1) in-phantom exit dose, or (2) EPID transit dose, for treatment verification.

Methods: The model was based on a quadratic equation that relates the reduction in intensity to the equivalent path length (EPL) of the attenuator. In this study, two sets of quadratic equation coefficients were derived from calibration dose planes measured with EPID and ionization chamber in water under reference conditions. With two sets of coefficients, EPL can be calculated from either EPID or treatment planning system (TPS) dose planes. Consequently, either the in-phantom exit dose or the EPID transit dose can be predicted from the EPL. The model was tested with two open, five wedge and seven sliding window prostate and head and neck intensity-modulated radiation

therapy (IMRT) fields on phantoms. Results were analysed using absolute gamma analysis (3%/3 mm).

Results: The open fields gamma pass rates were >96.8% for all comparisons. For wedge and IMRT fields, comparisons between predicted and TPS-computed in-phantom exit dose resulted in mean gamma pass rate of 97.4% (range, 92.3–100%). As for the comparisons between predicted and measured EPID transit dose, the mean gamma pass rate was 97.5% (range, 92.6–100%).

Conclusion: An EPID transit dosimetry model that can predict in-phantom exit dose and EPID transit dose was described and proven to be valid.

Advances in knowledge: The described model is practical, generic and flexible to encourage widespread implementation of EPID dosimetry for the improvement of patients' safety in radiotherapy.

There is much interest in using an electronic portal imaging device (EPID) for dose measurements.^{1–3} One of the major challenges for amorphous silicon (a-Si) EPID dosimetry is the presence of high *Z* material in the detector components that results in different response and scatter characteristics compared with a water-equivalent dosimeter.^{4–6} Various EPID dosimetry models have been proposed in the literature to work around the non-water-equivalent properties of EPID. These models can be broadly categorized into non-transit and transit models. Non-transit models are based on measurements without any object in the beam and are, therefore, limited to only pre-treatment quality assurance (QA). Ideally, patient QA should also allow actual treatment verification to complete the dose verification process and to detect errors that would otherwise be missed by pre-treatment QA.⁷

Transit models are desirable because they allow both pre-treatment and actual treatment verifications. Currently, there are only two commercially available EPID transit dosimetry

solutions, EPIgray® (Dosisoft, Cachan, France) and Dosimetry Check (Math Resolutions, Columbia, MD). The major drawback of EPIgray is that it only allows point comparison,^{8,9} which is unreliable for modulated fields with steep dose gradient. Dosimetry Check has the advantage of offering two dimensional (2D) and three dimensional (3D) dose verification. The model deconvolves the EPID image with a scatter kernel to retrieve the incidence fluence and uses this fluence as an input into an independent dose calculation algorithm for dose computation.^{10,11} An EPID transit dosimetry model using the convolution approach was also widely published by researchers at the Netherlands Cancer Institute (NKI-AVL) where the model is now in clinical use.^{12–18}

Instead of using the convolution approach, an empirical method is more practical for the majority of clinical centres with limited resources for computationally intensive simulations or mathematical modelling. 2D transit dosimetry using an empirical method was first described by Evans

et al¹⁹ and Symonds-Taylor et al²⁰ for an in-house imaging panel and later by Evans et al²¹ for a commercial scanning liquid-filled ionization chamber (SLIC) EPID. The technique used a quadratic equation established by Swindell²² and was based on a calibration method described by Morton et al²³ to derive coefficients for the conversion of EPID pixel value to equivalent path length (EPL). Although the technique was used for older model EPIDs with the purpose of designing compensators for breast irradiation, Kairn et al²⁴ proved that this method was also valid for a-Si EPID and suggested that the derived EPL matrix can be used as a form of treatment verification. Since dose, and not EPL, is the preferred metric for treatment verification, Kavuma et al^{25,26} extended the model by converting the 2D EPL matrix to entrance and exit doses for comparisons with the treatment planning system (TPS). However, the model by Kavuma et al²⁵ was partially dependent on the TPS for the conversion of EPL to dose. Tissue phantom ratios and envelope and boundary profiles from the pencil beam convolution algorithm, Eclipse™ TPS (Varian® Medical Systems, Palo Alto, CA), were used to calculate on-axis and off-axis doses. Furthermore, the model was tested only for conventional and wedge fields, and discrepancies were seen at the penumbra region.

In this study, we present a 2D EPID transit dosimetry model based on the empirical quadratic calibration method,²³ but without relying on any specific TPS for the conversion of EPL to dose. Different from previous studies,^{19–21,23–25} in addition to deriving quadratic equation coefficients from EPID-measured dose planes, coefficients were also derived from ionization chamber (IC) dose planes measured in water. Therefore, in the current model, EPL can be calculated using input from both EPID as well as TPS, which is conventionally modelled based on water measurements. The EPL, which is a property of the attenuating object, provided a link to the two different dosimetry systems and allowed a two-way relationship for the:

- (1) prediction of in-phantom exit dose from EPID-measured dose planes, for comparison with TPS-planned dose. (The in-phantom exit dose in this study was defined as dose at 1.5 cm upstream from the beam exit surface of the phantom.) and
- (2) prediction of EPID transit dose from TPS-computed dose planes, for comparison with EPID measurement during treatment.

This model was systematically tested with open, wedge and intensity-modulated radiation therapy (IMRT) fields on homogeneous and heterogeneous slab phantoms. Comparisons were made using 2D absolute global gamma analysis, and results were further validated against measurements with a commercial 2D array device.

METHODS AND MATERIALS

Quadratic formalism and implementation in this model

As shown by Swindell,²² a quadratic equation [Equation (1)] can be used to relate the reduction in intensity, represented by the ratio of exit (*M*) to entrance (*M*₀) dose, to the water-equivalent thickness or EPL of the attenuating object.

$$-\ln \left[\frac{M(i,j)}{M_0(i,j)} \right] = A(i,j) \times \text{EPL}(i,j) + B(i,j) \times \text{EPL}^2(i,j) \quad (1)$$

where “*A*” and “*B*” are the coefficients and (*i, j*) is the coordinate of pixels in a 2D plane. The linear coefficient, *A*, reflects the linear

attenuation coefficients of the attenuating object while the quadratic coefficient, *B*, describes the non-linear relationship owing to spectral variation in a photon beam.²² Since the beam spectrum changes with off-axis position because of differential beam hardening by the cone-shaped flattening filter, the *A* and *B* coefficients are expected to vary with off-axis position²² and thus, must be derived for each point (*i, j*) in the 2D plane.

To derive the *A* and *B* coefficients, a set of *M* and *M*₀ dose planes measured for homogeneous phantoms with known water-equivalent thicknesses under reference conditions is needed. By fitting Equation (1), pixel by pixel, to the set of dose planes measured with phantoms of different thicknesses, *A* and *B* coefficients can be obtained.^{23–25} With the *A* and *B* coefficients, the EPL of any attenuating object can be calculated from Equation (2) [inversion of Equation (1)] if given the amount of reduction in intensity (*M*/*M*₀). Although mathematically, the quadratic equation should be solved with a plus-minus sign, the negative sign was excluded since the EPL must be a positive value. An iterative numerical method was used to calculate the EPL, where the value from Equation (2) served as an initial approximation and the calculation repeated itself for five times to obtain a converged solution.^{24,25}

$$\text{EPL}(i,j) = \frac{-A(i,j) \pm \sqrt{A^2(i,j) - 4B(i,j) \times \ln \{ \text{FSF}(i,j) \times [M(i,j)/M_0(i,j)] \}}}{2B(i,j)} \quad (2)$$

where FSF is the field size correction factor, defined as the ratio of dose for the reference field size to dose for other field sizes of interest. Since the *A* and *B* coefficients were derived for a reference field size, the correction was necessary to account for change in collimator and phantom scatter for other field sizes. The FSF was tabulated from 2D dose planes measured for a range of field sizes and attenuator thicknesses, and the correction was applied based on the individual pixel location (*i, j*). The minimum field size that could be analysed by this model was dependent on the smallest field size available in the FSF table.

In this study, two separate sets of coefficients, (*A* and *B*)_{EPID} and (*A* and *B*)_{IC}, were derived from dose planes measured with EPID and IC in water, respectively. Figure 1 illustrates the function of these *A* and *B* coefficients and EPL in relating the two different dosimetry systems, EPID and IC (or TPS). Detailed descriptions of the dose plane measurements to derive *A* and *B* coefficients and the steps involved in the prediction process are given in the next section (the Commissioning process: calibration dose planes for *A* and *B* derivation section, the Path 1: predicting in-phantom exit dose section and the Path 2: predicting EPID transit dose section). All image processing and calculations were performed using MATLAB® software (MathWorks®, Natick, MA).

Commissioning process: calibration dose planes for *A* and *B* derivation

The *A* and *B* coefficients in the quadratic equation [Equation (1)] were derived from a set of *M* and *M*₀ calibration dose planes measured for homogeneous phantoms with a range of thicknesses under reference conditions. Since the EPID responds differently from IC measurements in water,^{4–6} two sets of calibration dose planes, measured separately with EPID and IC, were required for the

Figure 1. Diagram shows the role of quadratic equation coefficients (A and B) and EPL in relating two different dosimetry systems, electronic portal imaging device (EPID) and ionization chamber (IC) (or treatment planning system, TPS). Equations (1) and (2) are given in the text. From left to right is the path (Path 1) to predict in-phantom dose ratio $(M/MO)_{IC}$ from EPID-measured dose planes. From right to left is the path (Path 2) to predict EPID dose ratio $(M/MO)_{EPID}$ from TPS-exported dose planes. The predicted dose ratios $(M/MO)_{IC}$ and $(M/MO)_{EPID}$ were used to derive in-phantom exit dose and EPID transit dose, respectively, as described in the Path 1: predicting in-phantom exit dose section and Path 2: predicting EPID transit dose section. $(A$ and $B)_{EPID}$, coefficients derived from EPID calibration dose planes; $(A$ and $B)_{IC}$, coefficients derived from IC in water calibration dose planes.

commissioning of this model. All measurement were carried out on a Varian Clinac 21EX linear accelerator (Varian®, Medical Systems), and both sets of calibration dose planes, from EPID and IC, were measured for the same reference beam (6 MV, 100 MU and field size 20×20 cm) and the same set of thicknesses (0–35 cm).

Measurements of electronic portal imaging device calibration dose planes to derive $(A$ and $B)_{EPID}$

EPID images were captured with an aS500 a-Si EPID (Varian® Medical Systems) that has 384×512 pixels using standard manufacturer's software, IAS3. All images were captured using default manufacturer's settings that automatically performs dark field, flood field and defect pixel corrections as part of the standard image processing procedure. The distance from source to EPID was fixed at 140 cm. No modification was carried out on the EPID panel for practical implementation. Experimental setup for the EPID measurements is shown in Figure 2a. The $M0$ dose planes were captured without a phantom in the beam (0 cm) while M dose planes were captured with water-equivalent solid phantom (density 1.04 g cm^{-3} , Gammex Inc., Middleton, WI) of different thicknesses (5–35 cm) placed isocentrically (source to axis distance, SAD) at 100 cm.

Since the EPID was not absolutely calibrated to dose, the more appropriate term to describe the EPID measurements should be "EPID intensity image" or "EPID signal". However, the term "EPID dose plane" is used in this article for consistency with the IC or TPS dose planes.

Measurements of ionization chamber calibration dose planes to derive $(A$ and $B)_{IC}$

Meanwhile, the IC calibration dose planes were derived to simulate the isocentric setup shown in Figure 2b, where the $M0$ and M were doses at depth of 1.5 cm from the beam entrance and exit surface, respectively. The 1.5 cm build up for the $M0$ was

necessary to provide reasonable electronic equilibrium condition for the 6 MV beam used in this study and the 1.5 cm backscatter was chosen for a symmetric geometry. As a result, the dimension of water phantom was effectively $(1.5 + \text{"Thickness"} + 1.5) \text{ cm}$, where "Thickness" ranged from 0 to 35 cm.

To derive the IC dose planes, first, cross-line and in-line profiles were measured with CC04 IC (IBA Dosimetry, Schwarzenbruck, Germany) in a scanning water tank at different depths from 1.5 to 36.5 cm. All measured profiles were normalized to the central axis. After that, each of the one dimensional-relative profiles was duplicated, assuming a symmetric distribution, to fill a 2D matrix with 512×512 dimension. For each scan depth, there were two matrices representing the in-line and cross-line directions. Then, the in-line and cross-line matrices for the same scan depth were multiplied to generate a 2D-relative dose plane. The 2D-relative dose plane was cropped to 384 (rows) $\times$ 512 (columns) to match the EPID dimension. Finally, all the 2D-relative dose planes at different depths were converted into 2D absolute dose planes by multiplication with the central axis dose calculated from percentage depth dose (PDD). Since the PDD curves were measured at a fixed source to surface distance (SSD) of 100 cm and with full backscatter, corrections were made to account for the varying SSD owing to the SAD setup and the difference in backscatter thickness at the entrance and exit levels. To account for the effect of varying SSD, Mayneord factor²⁷ and tissue maximum ratio were used to correct for the difference in inverse square distance and phantom scatter, respectively [Equation (3)].²⁸

$$PDD_2 = PDD_1 \times \left[\left(\frac{SSD_2 + d_{\max}}{SSD_1 + d_{\max}} \right)^2 \times \left(\frac{SSD_1 + d}{SSD_2 + d} \right)^2 \right] \times \left[\frac{TMR(d, f_{s2})}{TMR(d, f_{s1})} \right] \quad (3)$$

where subscript "1" and "2" indicates the SSD and SAD setup, respectively, " $d_{\max}$ " is the depth of dose maximum, " d " is the

Figure 2. (a) Experimental setup for the electronic portal imaging device (EPID) entrance (M_0) and exit (M) calibration dose planes measurements. (b) Setup dimension for the ionization chamber (IC) entrance (M_0) and exit (M) calibration dose planes. Additional 1.5 cm build up was required for the IC setup dimension to provide adequate electronic equilibrium for the 6 MV beam used in this study while an additional 1.5 cm backscatter was chosen for a symmetric geometry. The ratio of M to M_0 , for both EPID and IC setup, included the same set of attenuator thicknesses (0–35 cm). SAD, source to axis distance.

depth of interest, “ fs_1 ” and “ fs_2 ” are the field sizes at depth “ d ”. The measurements performed in a scanning water tank, up to this point, represent full scatter condition. To account for the effect of different backscatter thickness as required for the setup shown in Figure 2b, a backscatter correction factor (BCF), defined as the ratio of dose without full backscatter to dose with full backscatter, was separately quantified through experimental measurements. The BCF values were measured at the central axis with IC in water-equivalent solid phantoms. Details of the experimental methods and series of BCF values as a function of field size, backscatter thickness and depth for the 6 MV beam investigated can be found in a previous publication.²⁹

Finally, before the IC dose planes were used to derive the A and B coefficients, all the M_0 and M dose planes were scaled to the same reference level using inverse square distance correction. In this study, the isocentre (100 cm from source), which is also the mid-plane of the phantom, was chosen as the reference level. The inverse square distance scaling was carried out so that the ratio of M and M_0 will reflect the reduction of dose owing to object attenuation entirely without the effect of distance.

Path 1: predicting in-phantom exit dose

Open field

Figure 1, from left to right, shows the flow to predict in-phantom exit to entrance dose ratio $(M/M_0)_{IC}$ from EPID-measured dose planes. Besides the EPID transit dose plane captured during the actual treatment delivery ($M_{EPID_{tx}}$), an additional measurement of the same treatment field without object in the beam was required ($M_{0_{EPID_{tx}}}$). The $M_{0_{EPID_{tx}}}$ can either be measured once for repeated use in the subsequent dose prediction or a new dose plane can be captured each time on the day of verification. By using the latter, machine output variation for the day was taken into account and would not be reflected in the results. All EPID images were smoothed with a moving average filter³⁰ using a window size of 8 pixels. Then, the ratio of $M_{EPID_{tx}}$ to $M_{0_{EPID_{tx}}}$ together with the A_{EPID} and B_{EPID} coefficients was used to calculate

the EPL of the attenuating object using Equation (2). FSF_{EPID} was also required in Equation (2) to correct for treatment fields with field size other than the reference 20×20 cm. The FSF_{EPID} is the field size correction factor tabulated from EPID-measured 2D dose planes for a range of field sizes (5×5 to 20×20 cm) and attenuator thicknesses (0–35 cm). The calculated EPL, which is a property of the object and independent of the dosimetry system, served as a bridge to convert from EPID to IC environment. The EPL matrix was used in combination with A_{IC} and B_{IC} coefficients to calculate the equivalent IC ratio $(M/M_0)_{IC}$ using Equation (1). Consequently, in-phantom exit dose ($M_{IC_{pred}}$) can be predicted by multiplying $(M/M_0)_{IC}$ with a, field size 20×20 cm, IC entrance dose plane ($M_{0_{IC_{fs20}}}$) that was previously derived as part of the commissioning process. An inverse square correction was also

Figure 3. Comparisons of MF_{EPID} and MF_{IC} for enhanced dynamic wedge field EDW15FS20 (wedge angle 15° , field size 20×20 cm) and EDW60FS20 (wedge angle 60° , field size 20×20 cm). The MF_{EPID} is the modulation factor derived from electronic portal imaging device measurements while the MF_{IC} is the modulation factor derived from ionization chamber measurements, represented by MapCHECK2 measurements.

added to scale dose from the reference level (100 cm) to the exit level.

$$M_{ICpred} = \left(\frac{M}{M_0}\right)_{IC} \times M_{0ICfs20} \div FSF_{IC} \times \left(\frac{100}{SDD}\right)^2 \quad (4)$$

where “SDD” is the source to exit distance. “FSF_{IC}” is the field size correction factor tabulated from IC-derived 2D dose planes for a range of field sizes (5 × 5 to 20 × 20 cm) and attenuator thicknesses (0–35 cm).

Modulated field (intensity-modulated radiation therapy and wedge)

The effect of beam modulation will not be reflected in the M_{ICpred} from Equation (4). This is because the ratio of M_{EPID_tx} to M_{0EPID_tx} was used to derive $(M/M_0)_{IC}$ and the division of two identical modulated fields with and without an object in the beam would have cancelled out the modulation effect. Therefore, for verification of modulated treatment fields, an additional modulation factor (MF) defined as the ratio of modulated to non-modulated dose plane was used [Equation (5)]. The modulated dose plane (M_{0EPID_tx}) was the EPID dose plane measured for the treatment field without an object in the beam. The non-modulated dose plane ($M_{0EPID_openFSx}$) was the EPID dose plane, also measured without an object in the beam, for a reference open field (100 MU; 20 × 20 cm) that was subsequently corrected with FSF_{EPID} based on the field size of the treatment field. The effective field size of irregular treatment fields was calculated according to the methods described by Monti et al.³¹ with the irregular field area for an IMRT delivery defined as the region with dose >50% of the maximum dose. Since both the modulated and non-modulated fields were measured by the EPID without an object in the beam, the MF is a relative factor that can be used to quantify beam modulation in a field.

$$MF_{EPID} = \frac{M_{0EPID_tx}}{M_{0EPID_openFSx}} \quad (5)$$

Equation (4) was modified to include MF, as shown in Equation (6), for the prediction of in-phantom exit dose for modulated fields.

$$M_{ICpred} = \left(\frac{M}{M_0}\right)_{IC} \times M_{0ICfs20} \div FSF_{IC} \times \left(\frac{100}{SDD}\right)^2 \times MF_{EPID} \quad (6)$$

In the above description, MF from EPID measurements was used to quantify beam modulation for the prediction of TPS in-phantom exit doses. Although the MF is a relative factor, the MF from EPID dose planes (MF_{EPID}) was found to be slightly different from the MF from IC measurements (MF_{IC}). Figure 3 shows the comparisons of MF_{EPID} and MF_{IC} for enhanced dynamic wedge (EDW) 60° (EDW60) and 15° (EDW15) fields. MapCHECK[®]2 measurements were used to represent the IC dose planes to save measurement time. It can be observed from Figure 3 that the discrepancies between MF_{EPID} and MF_{IC} were more prominent for EDW60

compared with EDW15. The difference between MF_{EPID} and MF_{IC} and the effect of different wedge angles could be owing to variation of EPID response with the changing field size and dose rate during delivery. To account for this effect, a generic correction was derived to convert MF_{EPID} to MF_{IC} for all wedge angles. This generic correction was derived from MF_{EPID} and MF_{IC} measured for the EDW60 field. A graph of MF_{EPID} against MF_{IC} for the EDW60 field was plotted and a linear fit was derived (Figure 4). The fitting formula was used to correct the MF_{EPID} to the equivalent of MF_{IC} , ($MF_{EPID \rightarrow IC}$). Two fitting formulae are shown in Figure 4, representing the EDW60 fields for field size 10 × 10 and 20 × 20 cm. Depending on the field size of the test field, the appropriate fitting formula was used for the correction. For wedge fields of other angles (θ), a weighting factor (W) was calculated and used to determine the fraction of total MF to be corrected [Equations (7) and (8)]. The weighting calculation was adapted from the concept of achieving a desired effective wedge angle by varying the proportion of open field irradiation and nominal wedge field irradiation.^{32,33}

$$\text{Weighting, } W = \frac{\left(\frac{\tan\theta}{\tan60}\right) \div WF}{\left(\frac{\tan\theta}{\tan60} \div WF\right) + \left(1 - \frac{\tan\theta}{\tan60}\right)} \quad (7)$$

where “WF” is the conventional wedge factor, defined as ratio of central axis dose measured with wedge 60° and open field.

$$MF_{corrected}^{EPID} = MF_{EPID \rightarrow IC} \times (W) + MF_{EPID} \times (1 - W) \quad (8)$$

where “ MF_{EPID} ” is the uncorrected MF from EPID measured dose planes as defined in Equation (5), and the “ $MF_{EPID \rightarrow IC}$ ” is the MF_{EPID} corrected to the equivalent of MF_{IC} using the fitting

Figure 4. Plot of MF_{EPID} vs MF_{IC} for EDW60 field and a linear fit of data. Two plots are shown, representing EDW60 field with field size 10 × 10 and 20 × 20 cm. The linear equation was used to convert MF derived from electronic portal imaging device to the equivalent of ionization chamber (or treatment planning system) and vice versa, $MF_{EPID} \leftrightarrow IC$.

formula shown in Figure 4. Accordingly, the mathematical expression to predict in-phantom exit dose for EDW fields was modified from Equation (6) to become Equation (9).

$$M_{ICpred} = \left(\frac{M}{M0}\right)_{IC} \times M0_{ICfs20} \div FSF_{IC} \times \left(\frac{100}{SDD}\right)^2 \times MF_{corrected}^{EPID} \quad (9)$$

As for IMRT fields, the in-phantom exit dose was also predicted using Equation (9). The MF correction was applied but with the W in Equation (8) assumed to be one ($W = 1$).

Path 2: predicting electronic portal imaging device transit dose

Open field

Figure 1, from right to left, shows the flow to predict EPID transit dose from TPS-exported dose planes. All treatment plans in this study were computed using AcurosXB algorithm from the Eclipse TPS (v. 10.0.28; Varian® Medical Systems), with dose-to-medium mode and 0.1 cm grid size. All TPS-exported dose planes were scaled to the same reference level (100 cm) using inverse square distance correction. The entrance ($M0_{TPS_tx}$) and exit (M_{TPS_tx}) dose planes for the treatment plan to be verified along with the A_{IC} and B_{IC} coefficients were used to calculate EPL of the object from Equation (2). FSF_{IC} was also required in Equation (2) to correct for treatment fields with field size other than the reference 20×20 cm. The FSF_{IC} was the field size correction factor tabulated from IC-derived 2D dose planes for a range of field sizes (5×5 to 20×20 cm) and attenuator thicknesses (0–35 cm). Subsequently, this EPL together with the A_{EPID} and B_{EPID} was used to calculate the equivalent EPID ratio ($M/M0$)_{EPID} from Equation (1). Finally, EPID transit dose ($M_{EPIDpred}$) can be predicted by multiplying ($M/M0$)_{IC} with an EPID dose plane captured with 20×20 cm open field without a phantom in the beam ($M0_{EPIDfs20}$). The $M0_{EPIDfs20}$ can be either the original EPID commissioning dose plane or a newly captured dose plane on the day of verification. By using the

latter, machine output variation for the day was taken into account.

$$M_{EPIDpred} = \left(\frac{M}{M0}\right)_{EPID} \times M0_{EPIDfs20} \div FSF_{EPID} \quad (10)$$

where FSF_{EPID} is the field size correction factor tabulated from EPID dose planes measured for a range of field sizes (5×5 to 20×20 cm) and attenuator thicknesses (0–35 cm).

Modulated field (intensity-modulated radiation therapy and wedge)

In the prediction of EPID transit dose from TPS dose planes, the MF for beam modulation quantification was computed from modulated and non-modulated dose planes exported from the TPS at the entrance level:

$$MF_{IC} = \frac{M0_{TPS_tx}}{M0_{TPS_openFSx}} \quad (11)$$

following the methods described in the Modulated field (intensity-modulated radiation therapy and wedge) section, the MF derived from TPS was corrected to the equivalent MF of EPID using the same linear equations shown in Figure 4. This corrected MF was then multiplied to Equation (10) to predict EPID transit dose for modulated fields:

$$M_{EPIDpred} = \left(\frac{M}{M0}\right)_{EPID} \times M0_{EPIDfs20} \div FSF_{EPID} \times MF_{corrected}^{IC} \quad (12)$$

In this case, the $MF_{corrected}^{IC}$ is:

$$MF_{corrected}^{IC} = MF_{IC \rightarrow EPID} \times (W) + MF_{IC} \times (1 - W) \quad (13)$$

where “ MF_{IC} ” is the uncorrected MF from TPS-exported dose planes as defined in Equation (11), and the “ $MF_{IC \rightarrow EPID}$ ” is the

Figure 5. (a) Homogeneous slab phantom and (b) heterogeneous slab phantom comprised of RMI phantom with rods with electron density ranging from 0.001 (air) to 1.69 (cortical bone) placed between two slabs of water-equivalent solid phantoms. The cross-section of the RMI phantom is shown in the inset. For dose computations with the treatment planning system, an additional 1.5 cm of water-equivalent material was added to the top and bottom surfaces of both phantoms to provide adequate electronic equilibrium and a symmetric geometry, similar to the condition used in the commissioning process.

MF_{IC} corrected to the equivalent of MF_{EPID} using the fitting formula shown in Figure 4. “W” is the weighting calculated according to Equation (7) for EDW fields with different wedge angle. For the prediction of EPID transit dose for IMRT fields, W was taken to be one.

Phantom measurements

The accuracy of the model was tested on homogeneous and heterogeneous slab phantoms (Figure 5). The homogeneous phantom comprised a water-equivalent solid phantom with a total physical thickness (PT) of 15 cm. The heterogeneous phantom, also with a total PT of 15 cm, included an RMI 467 phantom (Gammex Inc.) placed between two slabs of water-equivalent solid phantoms. The RMI phantom had cylindrical rods with electron densities relative to water (ρ_e) that ranged from 0.001 to 1.69 (air to cortical bone) to simulate all possible densities in a human body.

Non-modulated and modulated fields were tested. Non-modulated fields were open fields 10×10 and 20×20 cm. Modulated fields comprised five EDW fields (wedge angle 15° , 45° and 60° with field size 10×10 cm; and wedge angle 45° and 60° with field size 20×20 cm) and seven IMRT fields (three prostate and four head and neck fields). The IMRT fields were delivered using the sliding window technique with the number of segments in each field ranging from 114 to 178 segments. The extent of multileaf collimator travel distance for the seven IMRT fields ranged from 10.8 to 18.5 cm while the Y-jaw opening ranged from 10.5 to 18 cm. All beams were delivered at a fixed gantry angle of zero degrees and with the phantoms positioned at the isocentre. Results were analysed with the Varian Portal Dosimetry software (v. 10.0) using 2D global gamma analysis in absolute mode with 20% threshold for irregular fields and 100% of the field area for regular fields.

Independent verifications with two dimensional array

The phantom measurements described in the Phantom measurements section were repeated on the same day with a MapCHECK2

device to independently verify results from this model. The same homogeneous and heterogeneous phantoms were placed on top of the MapCHECK2 device. For consistency, MapCHECK2 dose planes were interpolated using a bilinear method to match the EPID resolution, and comparisons between MapCHECK2-measured dose planes and TPS-computed dose planes were also analysed with the Varian Portal Dosimetry software using 2D global gamma analysis in absolute mode.

RESULTS

Comparisons of in-phantom exit dose

Open field

Figure 6a,b shows comparisons of TPS-computed and EPID-predicted in-phantom exit dose for a 20×20 cm open field for homogeneous and heterogeneous phantoms, respectively. At the central axis, the TPS-computed exit dose was found to be 1.0% higher than the predicted dose for the homogeneous phantom and 1.5% lower than the predicted dose for the heterogeneous phantom. Additional central axis measurements with IC placed at the exit level of the two phantoms showed the same trend and confirmed that the TPS over- and under-estimates the exit doses by 1.5% for the homogeneous and heterogeneous phantom, respectively. The agreement between doses predicted from this model and doses physically measured by IC in the phantom proved that this EPID dosimetry model could accurately reflect the actual discrepancies between delivered and TPS-planned dose. The same verification for a 10×10 cm open field showed a similar finding with discrepancies of $\pm 2.0\%$.

The profiles in Figure 6a were normalized to the central axis and shown in Figure 7a as relative profile comparisons. Minor mismatches between TPS-computed and EPID-predicted relative profiles were observed at the shoulders and penumbra regions (indicated by arrows in Figure 7a). The same mismatch was also observed when the TPS-computed relative profile was compared with the IC-relative profile measured in a scanning water tank at the same depth (shown in Figure 7b), again proving that the

Figure 6. Profile comparisons between treatment planning system (TPS)-computed (solid line) and TPS-predicted in-phantom exit dose (dashed line) for a 20×20 cm open field on: (a) homogeneous phantom and (b) heterogeneous phantom. TPS inaccuracy in computing exit doses contributed to the observed offsets of the profiles.

Figure 7. Relative profile comparisons between: (a) treatment planning system (TPS)-computed (solid line) vs TPS-predicted from electronic portal imaging device (EPID) images (dashed line) and (b) TPS computed (solid line) vs ionization chamber (IC) measured in a scanning water tank (dashed line). Section of profiles from (a, b) were magnified and compared together in the inset (c). The good agreement between EPID-predicted and IC-measured profiles as well as the consistent mismatches between these profiles and the TPS-computed profiles at the shoulders and penumbra (indicated by arrows) implies that the discrepancies were inherent to the TPS and not from the EPID prediction model.

discrepancies were inherent to the TPS and not from the EPID prediction model. For clearer visualization of the relative profile comparisons, a section of the graphs covering the shoulder and penumbra region was magnified and compared together (TPS-computed, IC-measured and EPID-predicted) in the inset indicated as Figure 7c.

The on- and off-axis IC measurements described in the above paragraphs showed that this dosimetry model was capable of reflecting discrepancies between delivered and TPS-planned dose. While any disagreement is revealing a true discrepancy, the use of common gamma criterion, 3%/3 mm, is a problem especially when comparison is carried out at the exit level of the phantom, where the TPS-computed dose could be inaccurate.²⁹ Without correcting for the inaccuracy of TPS in computing exit doses, the

gamma pass rate using 3%/3 mm criterion was as low as 29.4% (10×10 cm open field tested on heterogeneous phantom). However, when the TPS-computed exit doses were corrected based on IC-measured doses at the central axis, the pass rates were >96.8% for all open fields tested on both phantoms. The TPS-computed exit dose plane was corrected as an overall offset based on IC measurement at the central axis. The minor TPS inaccuracies at the shoulders and penumbra were not corrected.

Modulated field (wedge and intensity-modulated radiation therapy)

TPS-computed exit doses for modulated fields were also corrected based on IC point dose measurements at the central axis. However, instead of performing IC measurements for each wedge and IMRT field on the different phantoms, the inaccuracies of TPS-computed

exit doses for these modulated fields were approximated from IC measurements of open fields with the same field size as the modulated fields and on the same phantom. For example, the effective field size for test field IMRT8 was 10.6×10.6 cm, and from the IC measurement with open field 10×10 cm at the exit level of the homogeneous phantom, we know that the TPS overestimated the exit dose by 2%. Subsequently, the TPS-computed exit dose for field IMRT8 on the homogeneous phantom was corrected as an overall offset of 2% before comparison was performed with the EPID-predicted dose. Results for gamma comparisons between EPID-predicted and TPS-computed exit doses corrected for inaccuracies were given in the column “With dose correction” in Table 1. Results for gamma comparisons for the same fields but without correcting for TPS inaccuracies were also presented in Table 1 under the column “Without dose correction”. MapCHECK2 results were included in the table for reference.

Comparisons between EPID-predicted and TPS-corrected exit dose and predicted in-phantom exit dose for wedge fields resulted in 3%/3 mm gamma agreement that ranged from 96.0% to 100% with a mean of 99.9% for homogeneous phantom and 97.5% for heterogeneous phantom. As for IMRT fields, the gamma pass rates ranged from 92.3% to 99.0% with a mean of 96.6% for homogeneous phantom and 96.3% for heterogeneous phantom. The IMRT fields with pass rates <95% concur with results from MapCHECK2 measurements. The overall mean gamma pass rate

(all modulated fields on both phantoms) from this model was 97.4%, in good agreement with MapCHECK2 measurement results of 97.5%. Figure 8 shows comparisons between EPID-predicted and TPS-corrected exit dose and EPID-predicted in-phantom exit dose for an EDW15FS10 field, an EDW60FS10 field and an IMRT field (IMRT4) tested on a heterogeneous phantom.

Comparisons of electronic portal imaging device transit dose

Open field

Comparisons between EPID-measured transit dose and that predicted by this model from TPS-exported dose planes showed good agreement for open fields. The TPS-exported exit dose planes were corrected for inaccuracies quantified from IC central axis dose measurements before the dose planes were used as input for the prediction of EPID transit dose. After correcting for TPS inaccuracy, the percentage of points that passed gamma criteria of 3%/3 mm was >97.3% for all open fields tested on both homogeneous and heterogeneous phantoms. Without correcting for TPS inaccuracy at the exit level, the lowest pass rate was 93.7%.

Modulated field

Table 2 gives the wedge and IMRT gamma comparison results between the EPID-measured transit dose planes and that predicted by this model from TPS-exported dose planes with and without correcting for TPS inaccuracy at the exit level.

Table 1. Gamma analysis results for comparisons between electronic portal imaging device-predicted in-phantom exit dose and treatment planning system (TPS)-computed exit dose with and without correcting for TPS exit dose inaccuracy. MapCHECK2 results were included for reference

Comparisons of in-phantom exit dose	Homogeneous phantom			Heterogeneous phantom		
	Gamma (3%/3 mm)			Gamma (3%/3 mm)		
	Without dose correction	With dose correction	MapCHECK2	Without dose correction	With dose correction	MapCHECK2
EDW15FS10	100.0	100.0	99.2	32.0	96.8	99.6
EDW45FS10	100.0	100.0	100.0	95.6	97.4	100.0
EDW60FS10	99.9	100.0	100.0	96.9	98.2	99.9
EDW45FS20	100.0	99.6	100.0	74.0	96.0	98.6
EDW60FS20	99.4	100.0	100.0	96.3	99.2	99.6
Mean wedge	99.9	99.9	99.8	79.0	97.5	99.5
IMRT1	99.2	99.0	97.6	91.1	98.1	96.0
IMRT2	94.8	98.3	96.0	91.2	98.1	95.4
IMRT3	89.1	97.2	98.3	96.5	96.7	97.6
IMRT4	88.4	96.7	97.3	94.4	97.7	96.0
IMRT5	86.2	97.1	96.4	96.9	97.3	95.9
IMRT6	87.3	93.5	95.6	95.4	92.3	94.0
IMRT7	89.4	94.5	94.6	95.0	93.7	92.7
Mean IMRT	90.6	96.6	96.5	94.4	96.3	95.4
Mean overall	94.5	98.0	97.9	87.9	96.8	97.1

EDW, enhanced dynamic wedge; IMRT, intensity-modulated radiation therapy.

Figure 8. Comparisons between treatment planning system (TPS)-corrected exit dose and electronic portal imaging device (EPID)-predicted in-phantom exit dose for an EDW15FS10 field (top row), EDW60FS10 field (middle row) and IMRT4 field (bottom row) tested on the heterogeneous phantom. From left to right is the EPID-predicted in-phantom exit dose plane, 3%/3 mm gamma comparison result (red indicating areas that failed the gamma criteria) and profile comparisons between TPS computed (solid line) and EPID predicted (dashed line) across the axis indicated by the arrow.

After correcting for TPS exit dose inaccuracy based on IC-measured dose, the predicted EPID transit dose agreed well with EPID-measured dose for all modulated fields on both phantoms. The gamma pass rate with 3%/3 mm criterion for wedge fields ranged from 97.9% to 100% with a mean value of 99.9% and 98.4% for homogeneous and heterogeneous phantom respectively. For IMRT fields, the percentage of points that passed the gamma criteria ranged from 92.6% to 98.7% with a mean pass rate of 96.5% (homogeneous phantom) and 96.2% (heterogeneous). The overall mean gamma pass rate from this model was 97.5% (MapCHECK2, 97.5%).

DISCUSSION

The exit dose was defined as dose at 1.5 cm upstream from the beam exit surface. With a constant underlying thickness of only

1.5 cm, the dose at the exit level should theoretically be less than the dose at the same point with full backscatter owing to reduced contribution from backscattered photons, which is dependent on beam energy, field size and depth.³⁴ In a previous study, we found that different TPS algorithms had variable accuracies in accounting for the backscatter effect and as a result, discrepancies between TPS and IC measured exit doses were observed.²⁹ For AcurosXB, the discrepancies in exit doses for a 6 MV beam at different depths and field sizes were up to 3%.²⁹ In this study, additional measurements were carried out with an IC at the exit level of both phantoms and the inaccuracy of TPS-computed exit doses was quantified and corrected. The IC measurements were only carried out for open fields. The TPS exit dose inaccuracies for modulated fields were approximated from the open field measurements of the closest setting. To ensure a reasonable

Table 2. Gamma analysis results for comparisons of electronic portal imaging device-measured transit dose and predicted from treatment planning system (TPS) dose planes with and without correcting for TPS exit dose inaccuracy. MapCHECK2 results are included for reference and are identical to those presented in Table 1

Comparisons of EPID transit dose	Homogeneous phantom			Heterogeneous phantom		
	Gamma (3%/3 mm)			Gamma (3%/3 mm)		
	Without dose correction	With dose correction	MapCHECK2	Without dose correction	With dose correction	MapCHECK2
EDW15FS10	100.0	100.0	99.2	61.9	98.0	99.6
EDW45FS10	99.6	100.0	100.0	88.5	97.9	100.0
EDW60FS10	99.9	100.0	100.0	97.5	98.3	99.9
EDW45FS20	98.9	100.0	100.0	80.7	98.2	98.6
EDW60FS20	98.9	99.5	100.0	95.1	99.5	99.6
Mean wedge	99.5	99.9	99.8	84.7	98.4	99.5
IMRT1	99.7	98.4	97.6	84.0	98.4	96.0
IMRT2	98.0	98.2	96.0	89.0	97.9	95.4
IMRT3	93.8	98.7	98.3	94.7	98.0	97.6
IMRT4	90.7	96.1	97.3	92.0	95.6	96.0
IMRT5	87.4	95.8	96.4	94.6	96.0	95.9
IMRT6	89.4	94.2	95.6	96.0	92.6	94.0
IMRT7	88.8	94.0	94.6	95.7	94.6	92.7
Mean IMRT	92.5	96.5	96.5	92.3	96.2	95.4
Mean overall	95.4	97.9	97.9	89.1	97.1	97.1

EDW, enhanced dynamic wedge; IMRT, intensity-modulated radiation therapy.

approximation, the parameters that affect the amount of back-scattered photons such as beam energy, field size, phantom type and measurement depth were chosen from the open field measurements that would represent the conditions of the modulated fields. Two sets of comparisons, with and without correcting for TPS inaccuracies, were presented in the results section. The purpose of presenting the results with TPS dose correction was to evaluate the effect of TPS exit dose inaccuracy on the results of this model. It could be observed from Tables 1 and 2 that the 3%/3 mm gamma comparison results improved considerably after accounting for TPS-computed exit dose inaccuracies. In the case where IC measurements at the exit level were not possible a higher gamma criterion should be considered. Table 3 presents the results for comparisons of in-phantom exit dose and comparisons of EPID transit dose using 4%/3 mm criterion, without any correction to the TPS-computed exit dose. The main disadvantage of this model was the use of dose close to the surface of phantom (1.5 cm from the entrance and exit surface) for comparisons in-phantom (Path 1) and as an input for EPID transit dose prediction (Path 2). Owing to the limited build-up and backscatter at the entrance and exit level, respectively, the inaccuracy of TPS-computed dose at these levels could be higher than for doses nearer to the centre.

As with any other empirical model, the limited number of measurements that can be practically carried out required certain assumptions to be made that may affect the accuracy of results. In

this model, two assumptions were made in using MF, defined as the ratio of modulated field to open field at the same level, to quantify beam modulation field. First, the MF derived from entrance dose planes was assumed to be similar for both EPID and IC (or TPS). Second, the MF derived at the entrance level was assumed to be representative of the MF at the exit level.

For this first assumption, a minor correction was made, as described in the Modulated field (intensity-modulated radiation therapy and wedge) section, to convert MF derived from EPID to the equivalent of IC and vice versa. This correction was derived from EDW60 measurements and used as a generic correction for all modulated fields, including IMRT. Without accounting for the difference in dose rate between an EDW60 and an IMRT delivery, the same generic MF correction derived from EDW60 measurements was applied to the IMRT fields. Figure 9a shows the comparisons of MF_{IC}^{EPID} and $MF_{corrected}^{EPID}$ for an IMRT field, with the $MF_{corrected}^{EPID}$ derived according to Equation (8) and W in the equation taken to be one. It can be observed from Figure 9a that the $MF_{corrected}^{EPID}$ agreed well with the MF_{IC} despite using a general correction derived from EDW60 measurements. For the second assumption, the MF derived from entrance dose planes was assumed to be the same as the MF at the exit level. This assumption is illustrated in Figure 9b,c, which shows comparisons of MF derived from entrance dose planes vs MF derived from exit dose planes for an IMRT field for both

Table 3. Gamma analysis results (4%/3 mm) for comparisons of in-phantom exit dose and comparisons of electronic portal imaging device transit dose. No correction was applied to the treatment planning system-computed exit dose

Type of therapy	Comparisons of in-phantom exit dose		Comparisons of EPID transit dose	
	Gamma (4%/3 mm)		Gamma (4%/3 mm)	
	Homogeneous phantom	Heterogeneous phantom	Homogeneous phantom	Heterogeneous phantom
Open FS10	100.0	97.3	100.0	95.7
Open FS20	99.1	99.5	93.4	99.6
Mean open	99.6	98.4	96.7	97.7
EDW15FS10	100.0	91.3	100.0	96.8
EDW45FS10	100.0	95.7	100.0	95.2
EDW60FS10	100.0	97.0	100.0	98.6
EDW45FS20	99.4	93.5	96.0	98.9
EDW60FS20	97.7	99.3	92.4	99.4
Mean wedge	99.4	95.4	97.7	97.8
IMRT1	100.0	96.7	99.9	93.6
IMRT2	99.1	96.8	99.4	94.1
IMRT3	95.2	99.1	95.5	97.6
IMRT4	98.5	98.5	96.9	96.2
IMRT5	97.0	99.1	93.8	96.1
IMRT6	95.0	98.7	93.8	98.3
IMRT7	92.9	98.1	90.3	98.7
Mean IMRT	96.8	98.1	95.7	96.4
Mean overall	98.1	97.2	96.5	97.1

EDW, enhanced dynamic wedge; IMRT, intensity-modulated radiation therapy.

TPS and EPID. Within the field of interest, the MFs derived from entrance and exit dose planes were reasonably comparable.

The other disadvantage of the described method was the limitation in maximum field size that can be analysed. Because a reference field size 20×20 cm was used to derive the *A* and *B* coefficients in this study, the quadratic equation can only be used to calculate EPL for measurements within this field area. Beyond the 20×20 cm area, the EPL calculation would be inaccurate because no proper coefficients were available. The field size 20×20 cm was selected considering that the EPID has an active area of 30×40 cm and, at 140 cm source to detector distance, the maximum field size at 100 cm must be less than 21×28 cm to avoid irradiation of EPID electronics. Although the detector distance could be reduced to maximize the irradiation field size, the 140 cm source to detector distance was chosen to allow for more clearance, as the ultimate aim of this study was to extend the application to actual patient treatment verification with different gantry angles.

Despite the disadvantages described in the previous paragraphs, this model has the advantages of being flexible, generic and practical to encourage widespread clinical implementation. Firstly, this model is flexible as it allows user to choose either to

compare dose in-phantom or at the EPID level. Secondly, the model is generic and practical as the methods involve only general measurements and do not require input specific to any manufacturer. This is in contrast to the method proposed by Kavuma *et al*,²⁵ where envelope and boundary profiles specific to the Eclipse TPS were required and thus, limiting its use to centres with Eclipse TPS. Similarly, the EPID transit dosimetry model described by Berry *et al*³⁵ was adapted from the Varian through-air portal dose image prediction algorithm and partly involved Monte Carlo simulations that may not be practical in most clinical departments. Researchers from Universita Cattolica del Sacro Cuore, Rome, Italy, proposed a generic measurement-based model but the method was only for single point comparisons, which may not be suitable for IMRT verifications.^{36–40} Although the model was later extended by Peca and Brown⁴¹ to perform 2D verification, it was only tested for 3D conformal fields.

In addition, the EPID dosimetry model proposed in this study is practical because all the EPID images were acquired in the clinical configuration without requiring panel modification or non-standard flood field calibration technique^{4,42–44} devised to prevent the removal of characteristic horns of a photon beam from an EPID image (“washed-out” effect). The washed-out effect owing to standard flood field calibration was not a problem in this dosimetry

model because the $(A \text{ and } B)_{\text{EPID}}$ coefficients were derived from EPID dose planes measured with standard flood field calibration and, therefore, the washed-out effect was inherent in the coefficients. Consequently, for Path 1, the EPL of attenuating objects could still be calculated correctly from the EPID dose plane that contained the washed-out effect using the $(A \text{ and } B)_{\text{EPID}}$ coefficients. Once the EPL was correctly calculated, the subsequent in-phantom dose prediction would not be a problem because the washed-out effect only affects EPID measurements. As for Path 2, the calculation of EPL from TPS-exported dose planes did not involve any EPID images and is, therefore, unaffected by the washed-out effect. Although the subsequent prediction of EPID transit dose plane from EPL and $(A \text{ and } B)_{\text{EPID}}$ coefficients would reflect the washed-out effect, this was not an issue because the predicted dose plane would be directly compared with EPID-measured transit dose plane that had the same feature. The ability to use EPID images captured under standard flood field calibration and without additional correction simplifies the EPID dosimetry model and makes it practical for other clinical departments to adopt the model.

Very often, the amount of additional time required to implement an EPID dosimetry programme must also be considered for practical reasons. The commissioning of the EPID part of this dosimetry model took approximately 2 h to complete, requiring a total of 32 EPID images measured for water-equivalent solid phantom of eight different thicknesses (0–35 cm, in 5 cm increment) and beams with four different field sizes (5×5 to 20×20 cm, in 5 cm increment). These images were inclusive of those required for tabulation of FSFs. As for the IC calibration dose planes, the required PDD and profiles were taken from the original machine commissioning measurements performed in a scanning water tank. An additional 3 h were required for the measurements of BCF values with IC in solid water phantoms with different backscatter thickness as a function of field size and depth. However, the BCF values for 6 MV beams were found to be very similar with linear accelerators from different manufacturers and these values had been previously published in Tan et al, 2014.²⁹ Once the model had been properly commissioned, the subsequent dose prediction took less than five minutes to complete for a typical conformal or IMRT field.

Figure 9. (a) Comparisons of MF_{IC} (solid line) and $\text{MF}_{\text{corrected}}^{\text{EPID}}$ (dashed line) for an intensity-modulated radiation therapy field. The MF_{IC} was derived from treatment planning system (TPS)-exported dose planes and the $\text{MF}_{\text{corrected}}^{\text{EPID}}$ was the MF_{EPID} (derived from electronic portal imaging device (EPID)-measured dose planes) corrected using the generic MF fitting formula according to Equation (8). (b) Comparisons of MF derived at the entrance (solid line) and exit level (dashed line) for TPS. (c) Comparisons of MF derived at the entrance (solid line) and exit level (dashed line) for EPID.

CONCLUSIONS

An empirical 2D EPID transit dosimetry model that can predict both in-phantom exit dose and EPID transit dose has been described. After correcting for TPS exit dose inaccuracy based on IC-measured dose, 2D absolute global gamma analysis (3%/3 mm) resulted in pass rates above 96.8% for all open fields investigated. For modulated fields, the overall mean gamma pass rate for in-phantom exit dose comparisons was 97.4% (range, 92.3–100%). For EPID transit dose comparisons, the overall mean pass rate was 97.5% (range, 92.6–100%). These first results on slab phantoms

indicate that the model is valid and the authors intend to extend the model for patient treatment verification at different gantry angles to detect major errors in the actual treatment delivery.

ACKNOWLEDGMENTS

The authors would like to thank Dr Kurian Oomen for his advice and assistance in the development of MATLAB® codes for this work. The first author is also grateful for the Academic Medicine Development Award scholarship from the National University Health System and Ministry of Health, Singapore.

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