Monte Carlo-based adaptive EPID dose kernel accounting for different field size responses of imagers

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The aim of this study is to present an efficient method to generate imager-specific Monte Carlo (MC)-based dose kernels for amorphous silicon-based electronic portal image device dose prediction and determine the effective backscattering thicknesses for such imagers. EPID field size-dependent responses were measured for five matched Varian accelerators from three institutions with 6 MV beams at the source to detector distance (SDD) of 105 cm. For two imagers, measurements were made with and without the imager mounted on the robotic supporting arm. Monoenergetic energy deposition kernels with 0–2.5 cm of water backscattering thicknesses were simultaneously computed by MC to a high precision. For each imager, the backscattering thickness required to match measured field size responses was determined. The monoenergetic kernel method was validated by comparing measured and predicted field size responses at 150 cm SDD, 10×10 cm² multileaf collimator (MLC) sliding window fields created with 5, 10, 20, and 50 mm gaps, and a head-and-neck (H&N) intensity modulated radiation therapy (IMRT) patient field. Field size responses for the five different imagers deviated by up to 1.3%. When imagers were removed from the robotic arms, response deviations were reduced to 0.2%. All imager field size responses were captured by using between 1.0 and 1.6 cm backscatter. The predicted field size responses by the imager-specific kernels matched measurements for all involved imagers with the maximal deviation of 0.34%. The maximal deviation between the predicted and measured field size responses at 150 cm SDD is 0.39%. The maximal deviation between the predicted and measured MLC sliding window fields is 0.39%. For the patient field, gamma analysis yielded that 99.0% of the pixels have $\gamma < 1$ by the 2%, 2 mm criteria with a 3% dose threshold. Tunable imager-specific kernels can be generated rapidly and accurately in a single MC simulation. The resultant kernels are imager position independent and are able to predict fields with varied incident energy spectra and a H&N IMRT patient field. The proposed adaptive EPID dose kernel method provides the necessary infrastructure to build reliable and accurate portal dosimetry systems. © 2009 American Association of Physicists in Medicine. [DOI: [10.1118/1.3158732](https://doi.org/10.1118/1.3158732)]

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I. INTRODUCTION

Contemporary radiation therapy modalities such as intensity modulated radiation therapy (IMRT) tend to deliver highly conformal dose to targets while sparing critical organs and normal tissues. IMRT requires strict and dedicated plan verification due to the mechanical complexity and potentially steep dose gradients. To ensure acceptable delivery of a plan,

both geometrical and dosimetric verifications are required. Although electronic portal imaging devices (EPIDs) are routinely utilized to facilitate patient geometrical setup, considerable efforts have been made to exploit it as a full-fledged dosimeter, as shown in several review papers.^{1–3} The majority of current EPIDs are based on amorphous silicon (*a*-Si) flat panel devices. Their response characteristics with respect to various input parameters have been investigated by sev-

eral groups,^{4–11} which suggest the appropriateness of using this type of EPID for dose measurement. Throughout this paper, EPID refers to the amorphous silicon-based EPID unless otherwise stated.

There are different sorts of algorithms to predict dosimetric portal images (DPIs).^{4,12–21} Among them, Monte Carlo (MC) simulation is highly attractive given its ability to accurately reproduce measurements. It has been demonstrated that MC is able to achieve agreement between the predicted and measured DPIs.^{12,14,22} However, to calculate to a desirable precision, MC can be extremely time consuming. Using 16 simultaneous processes at our institution, our in-house MC code requires 20 min to compute a 5×5 cm² field DPI with 2.5% precision. When computing a 25×25 cm² field DPI with the same settings, simulation time increases to about 9 h. Therefore, for an average multifield plan, at least several hours are required to obtain statistically meaningful MC-generated DPIs. Due to this, in the foreseeable future at our institution, pure MC will not be the method of choice for clinical DPI calculations.

Most researchers alternatively resort to the kernel convolution method because the fast Fourier transform (FFT) is generally applicable to substantially speed up the computation, and, more importantly, the same level of precision can be obtained through convolution while only requiring a small fraction of the simulation time of MC. In this method, the predicted DPI is calculated by convolving EPID dose response kernels with the fluence transported to the sensitive plane of the imager. The fluence can be either from the treatment planning system (TPS) or from a MC simulation. Using MC-generated phase spaces identical to those of the previously mentioned tests to determine the incident fluence, an unoptimized in-house kernel convolution code calculates a 5×5 cm² field EPID image in about 5 min and a 25×25 cm² field image in about 10 min. The time difference between the two calculations is largely due to the disk I/O operation to read particles from the phase spaces.

Various types of EPID kernels are reported in the literature, including parametrized kernels^{15,21,23–26} and MC-based kernels.^{17,24,27–31} EPID kernels can be formulated as parametrized functions. These parameters can be obtained by fitting kernel convolution results to EPID measurements on a set of test patterns. It has been reported that parametrized kernels can fit MC-based kernels well.²⁴ However, its accuracy may be limited by the optimization algorithm, which searches for the most optimal kernel parameters. The objective function of this optimization problem would inevitably have local suboptimal solutions which would trap the searching algorithm. Moreover, the objective function is built upon the test patterns and the optimal solution would be sensitive to the selection of these patterns.

Kernels can also be generated by MC. In some cases, the resultant kernel is adjusted by a glare kernel, which is either explicitly computed by simulation²⁴ or empirically fitted from measurement.^{29,30} The glare kernel is intended to account for optical reflection and diffusion in the screen layer, which results in signal blurring. The optical glaring effect is most prominent for the camera-based metal-phosphor EPID,

while its influence on the modern amorphous silicon-based EPID is believed to be negligible for most cases.²⁴ Another noteworthy point on MC kernel computation is that the downstream backscattering of EPIDs makes a significant contribution to the resultant signal. This phenomenon has been analyzed^{32,33} and compensated for by adding an extra backscattering layer to the EPID model in the EPID MC prediction algorithm.¹⁴ When generating the EPID kernel, the backscattering should also be explicitly modeled.^{17,24}

Most MC-based kernels are incident energy spectra independent and spatially invariant once they have been generated. In other words, it is assumed that the EPID has an identical response regardless of any variation of the location and the energy spectrum of the incident fluence. Unfortunately, several studies have clearly invalidated this assumption. Studies^{4,34,35} show that the *a*-Si-EPID response is higher for low energy photons. It is also well established that the treatment beam from the linear accelerator (LINAC) gradually becomes harder toward the central axis.^{36,37} In addition, the beam hardening can be caused by beam modifiers such as the multileaf collimator (MLC) (Ref. 38) and even by the patient/phantom placed in the beam.³⁹ A further development for improving MC-based kernel generation is to take the variation of the incident energy spectra into consideration. Kernels are created in terms of the energy spectra of open and blocked MLC fields and depend on the radial distance from the central axis.¹⁷ However, these kernels are generated for a limited number of situations and do not fully account for the variation of the energy spectra incident on the EPID when a patient attenuates the beam.

This work will demonstrate a more flexible set of MC-based EPID dose kernels, along with a method to generate and tune them to an individual imager. For simplicity, we will only focus on 6 MV photon beams in this study. However, the method is naturally extendable to other beam energies. The fluence used for the kernel convolution is obtained from the previously developed in-house MC calculation system, which is commissioned to match our LINACs.⁴⁰ The system is capable of simulating LINAC head components such as the jaws,⁴¹ MLC,⁴² patient/phantom,^{43,44} and EPID.¹⁴ A phase space can be written at a user defined plane to provide fluence for the kernel convolution.

The specific goals of this paper include (1) to investigate the field size responses of various EPIDs and the possible causes which lead to the variation, (2) to introduce an efficient MC-based kernel calculation method which accounts for the field size response of an individual imager, (3) to introduce a weighted fluence scoring method to improve the approximation of the energy dependence of EPID response, and (4) to illustrate imager-specific kernel tuning for investigated EPIDs.

II. METHODS AND MATERIALS

II.A. The dosimetric investigation of Varian *a*-Si-EPIDs

II.A.1. Detectors

Two types of Varian (Varian Medical Systems, Palo Alto, CA) EPIDs, the aS500 and aS1000, are investigated in this

study. The two EPIDs have almost identical designs, including various protective layers, a copper buildup layer that filters out scattered radiation, a $\text{GdO}_2\text{S:Tb}$ (Kodak Lanex Fast B) screen layer that converts incident radiation into visible light, and an array of photodiodes and thin film transistors that convert light to electrical signal for readout. The geometry and material composition are described in more detail by several groups.^{12,14,24} The major difference between the two EPIDs is the resolution of the detector array. The aS500 has 512×384 pixels at a resolution of 0.784×0.784 mm² each, forming a 40×30 cm² active region. The resolution of aS1000 doubles that of aS500, covering the same region but with 1024×768 pixels. The detector arrays are identical. The increased resolution is achieved with software/hardware modifications.

The EPIDs are attached to the LINACs with robotic supporting arms, which move the imagers laterally, longitudinally, and vertically. The imagers can be positioned into a field for pretreatment measurement, adjusted beneath the patient for transmission dosimetry measurements, or removed from the beam path altogether. In this study two different types of supporting arms, the R arm and E arm, are examined. The supporting arm type should not alter the dosimetric characteristics of the EPID itself, but it might change downstream backscattering, resulting in varied responses.

Five imagers from three institutions are used in this study, with different combinations of imager type and arm type. The imagers are aS1000-A (E arm, Virginia Commonwealth University), aS1000-B (E arm, SUNY Upstate Medical University), aS500-A (E arm, Virginia Commonwealth University), aS500-B (E arm, Calvary Mater Newcastle Hospital), and aS500-C (R arm, Calvary Mater Newcastle Hospital).

II.A.2. EPID calibration method

When used for imaging purposes, an EPID image usually undergoes flood field and dark field calibrations. The flood field image is obtained by irradiating the EPID with the largest allowable field size in order to evaluate the relative sensitivity of the pixels. The dark field image, captured without any irradiation, measures the background electronic noise which needs to be removed from the resultant signal. The routine calibration method can be summarized in the following equation:

$$I(x,y) = \left[\frac{I_{\text{raw}}(x,y) - \text{DF}(x,y)}{\text{FF}(x,y) - \text{DF}(x,y)} \right] [\text{FF}(x,y) - \text{DF}(x,y)]_{\text{mean}}, \quad (1)$$

where x and y are the pixel coordinates on the imager, FF is the flood field integrated image, DF is the dark field integrated image, I is the corrected EPID image data, and I_{raw} is the raw EPID response caused by incident particles.

In this study, except where explicitly noted, EPID measurements are made with the imager vertical readout at -5 cm. At this position, the optical distance indicator (ODI) readout is 103.4 cm, the imaging cassette surface is 105 cm from the source, and the sensitive screen is 107 cm from the source. For clarity, we define this position to have a source to

detector distance (SDD) of 105 cm. This is the position used for the pretreatment IMRT verification at our institution. When calibrating the imager, the flood field is taken at this position with a 38×28 cm² open field, irradiated with 100 MU. The beam collimated by this field size fully covers the 40×30 cm² active region of the detector while still maintaining a margin to avoid damaging the surrounding electronics. The dark field is also taken at this position with the radiation beam off. The dark field image is typically taken for 30 frames, while the flood field image requires about 200 frames.

The open-field beam is not perfectly flat and has a horn-shape profile due to the design of the flattening filter. Since the flood field image is generated by an open-field irradiation, the routine calibration method washes out the intrinsic beam horns which are of interest to dosimetric applications. Therefore, the method proposed by Greer⁴⁵ is utilized to further process the FF and DF corrected EPID images. Basically, a relative one dimensional pixel variation profile is obtained by irradiating various lateral positions on the EPID with an identical field. The corresponding relative EPID beam response profile is obtained by dividing the FF profile by the pixel sensitivity profile. This one dimensional profile is then interpolated and rotated to get a two dimensional matrix, which is used to correct the EPID image and convert it into a DPI.

II.A.3. Field size response

Field size responses in water or output factors are a dosimetric property of each LINAC. It is an essential part of the machine QA and frequently checked to verify the stability of the machine output. Five different Varian LINACs from the participating institutions are used in this study. Each LINAC was matched to the Varian “gold standard” data when they were commissioned. The output factors of the LINACs, measured in water phantoms with an ionization chamber, have been cross compared and found to deviate by less than 0.3% for the investigated field sizes.

In this study, EPID field size responses are measured to evaluate the dosimetric agreement among various EPIDs, as well as to guide the imager-specific kernel tuning. The studied field sizes include 5×5 , 10×10 , 15×15 , and 25×25 cm². EPID images are acquired with the AM-MAINTENANCE software utility at one institution (Calvary Mater Newcastle Hospital) and with the utility software IAS-MONITOR version 3 at the remaining institutions (Virginia Commonwealth University and SUNY Upstate Medical University). All images are taken with 100 MU irradiation and with the integrated acquisition mode. The measurements are repeated three times and the mean images are calculated to reduce uncertainty. Also, to obtain a robust estimation of EPID field size response, the pixel values of the central 1×1 cm² regions of interest (ROIs) of the mean images are averaged to improve the statistics. Responses are normalized to the 10×10 cm² field response for each imager so that comparisons may be made with other imagers.

Moreover, for two imagers, EPID measurements with and without the robotic arm attached are performed to ascertain the effect of supporting arm backscattering on the imager response. Since the supporting arm is the closest object downstream from the EPID detector housing, it makes a considerable contribution to the backscattering. For each investigated EPID, a normal set of field size responses is taken with the EPID positioned to the 105 cm SDD with the robotic supporting arm. In the subsequent experiment, the supporting arm is removed from the EPID and the imager is again placed at the 105 cm SDD using the treatment couch with the “tennis racket” insert removed to support the imager in such a way that no materials are immediately behind the imager. An additional set of field size responses is acquired. The comparison between these two sets of responses quantifies the backscattering from the arm.

II.B. MC-based EPID monoenergetic dose kernel (monokernel)

An EGSnrc (Ref. 46) MC code based on a previous EGS4-based (Ref. 47) MC DPI prediction code¹⁴ is used to compute EPID monoenergetic dose response kernels, referred to as monokernels in the remaining text. A set of monokernels is required to compute EPID images by the convolution system described later.

The EPID model used in the MC computation consists of 26 layers. The EPID, which is created according to manufacturer-provided information, includes the first 25 layers. In addition, a water slab is added to the original model in an effort to model complex downstream backscattering. The thickness of the compensating water slab is determined for each imager by the ratio of measured EPID responses at two field sizes.¹⁴ To compute each monokernel, monoenergetic photons are directed toward the center of the EPID model. Whenever energy is deposited at the screen layer, the lost energy is accumulated in a two dimensional 1024×1024 scoring matrix in which each element has the same size as one aS500 pixel, 0.784×0.784 cm². After the simulation is finished, the accumulated energy deposition in the scoring matrix is normalized to the total number of incident particles to obtain the response per particle. There is, however, a fundamental drawback to the monokernel computation method. Since the observed EPID field size response varies from one imager to another, the thickness of the backscattering water slab also changes for different imagers. Therefore, the monokernels may need to be recomputed for each imager, which requires extra time.

Each monokernel is validated against the EPID MC results under three scenarios: (1) An artificially generated monoenergetic parallel 5×5 cm² photon beam, (2) artificially generated monoenergetic divergent 5×5 and 25×25 cm² photon beams, and (3) artificially generated monoenergetic divergent 5×5 and 25×25 cm² photon beams with 10% contaminant electrons. Recall that when a 105 cm SDD is used, the fluence is scored at 107 cm to match the location of the sensitive screen. The set of monokernel was validated

with respect to the EPID MC results for 6 MV simulations of 5×5 , 10×10 , 15×15 , and 25×25 cm² fields.

II.C. MC-based EPID monoenergetic all-in-one dose kernel (all-in-one kernel)

Due to the fact that the monokernel may require considerable computation resources to be regenerated for each imager, a so-called EPID all-in-one monoenergetic dose kernel is introduced, which is referred to as the all-in-one kernel hereafter. The all-in-one kernel allows a tunable backscattering thickness and can be converted to an imager-specific monokernel precisely and efficiently.

The single backscattering water slab beneath the original EPID model is replaced with a series of thin 1 mm thick subwater slabs. As with computing the monokernel, monoenergetic photons are directed onto the EPID model. For each source particle, the depth of the greatest penetration in the imager is tracked and scored using the corresponding EGSnrc LATCH bit. The trajectories of all secondary particles are tracked separately. In the EGSnrc system, these particles inherit the LATCH automatically from the primary particle. Therefore, the maximum depth of penetration of a secondary particle also depends on all of its primary particles up to the source particle. To implement LATCH tracking in the all-in-one kernel generating code, the IAUSFL flag of the EGSnrc system is set to capture all interactions in the AUSGAB subroutine that may directly or indirectly cause energy deposition. In AUSGAB, whenever an interaction is recorded, the LATCH variable is immediately compared to the current layer of interaction. If the layer is deeper than that stored in the LATCH, the variable is updated. Finally, energy deposition in the screen layer is scored separately for the main imager (not backscattered from the slabs) and for each backscattering slab in terms of the content of the LATCH. The resultant all-in-one kernel contains a set of subkernels as follows:

$$K^{\text{all-in-one}}(E, x, y) = \{K^{\text{EPID}}(E, x, y), K^{\text{bs1}}(E, x, y), K^{\text{bs2}}(E, x, y), \dots, K^{\text{bsN}}(E, x, y), \dots\}, \quad (2)$$

where x and y are the two dimensional indices for each kernel element, $K^{\text{EPID}}(E, x, y)$ is the contribution from the main imager, and $K^{\text{bsi}}(E, x, y)$ is the contribution from the i th backscattering slab, indexed from upstream to downstream.

The all-in-one kernel can be easily assembled into the monokernel with desired backscattering as

$$K^{\text{mono}}(E, x, y) = K^{\text{EPID}}(E, x, y) + K^{\text{bs1}}(E, x, y) + K^{\text{bs2}}(E, x, y) + \dots + K^{\text{bsN}}(E, x, y), \quad (3)$$

where N is the number of backscattering kernels to be included to compensate for the downstream backscattering of a specific imager. This way, only one set of all-in-one kernels is required to obtain monokernels for any imager. The all-in-one kernels therefore can be computed once to a high precision.

FIG. 1. EPID image prediction via convolution using imager-specific monoenergetic kernels. The kernels are tuned from the all-in-one kernels in terms of the measured EPID field size responses. The particle fluence is generated by the in-house Monte Carlo system and scored into energy-differential fluence matrices. For the pretreatment EPID prediction, the exit particle fluence from the LINAC head modeling, optionally attenuated by MLC, is scored; for the after-patient EPID prediction, either DOSXYZnrc (Ref. 43) or VMC++ (Ref. 44) can be utilized to obtain after-patient particle fluence. The scored energy-differential fluences are subsequently convolved with imager-specific monoenergetic kernels to compute the EPID image.

II.D. EPID image prediction algorithm by convolution

II.D.1. Overview of the prediction algorithm

The flow chart for the EPID image prediction algorithm using convolution is shown in Fig. 1. The whole system can be described by the following equation:

$$I^{\text{predicted}}(x, y) = \sum_{j=1}^N (K^{\text{mono}}(E_j, x, y) \otimes \Phi_j(x, y)), \quad (4)$$

where $I^{\text{predicted}}(x, y)$ is the predicted EPID image, $\Phi_j(x, y)$ is the energy-differential particle fluence at the imager plane for the j th energy bin, $K^{\text{mono}}(E_j, x, y)$ is the imager-specific monokernel at the middle energy E_j over the energy bin j , converted from the corresponding all-in-one kernel, and N is the total number of energy bins spanning the whole energy spectrum of the incident beam. The energy-differential fluence and the monokernels are convolved for each energy bin and the accumulated contribution from all bins gives the predicted DPI. The convolution part of the prediction algorithm utilizes FFTW, a C-based FFT library.⁴⁸

Objects upstream from the imager such as the LINAC head and MLC are simulated by a previously developed MC system,^{41,42} including simulations through the patient using either DOSXYZnrc (Ref. 43) or VMC++ (Ref. 44). The resultant exit particles are transported to the sensitive plane of the imager to extract the energy-differential fluence, which is scored with an incident energy-dependent weighting strategy to better resemble the characteristic of the EPID response, as detailed in Sec. II D 2. Only photons are analyzed since other types of particles would presumably be filtered out by the copper layer of the imager before they reach the screen layer or make negligible energy deposition in the screen layer. The hypothesis is tested when the monokernel convolutions are compared directly to the full EPID MC simulations. Namely, the kernel convolutions can reproduce the MC results with pure photon incident fluence and photon and electron mixed incident fluence.

The number of energy bins, N , should be determined in such a way that the variation of EPID response in each bin is acceptably low and the response can be approximated reasonably well by an invariant monokernel. The choice of N should also balance the speed of the convolution and the

prediction accuracy. The prediction would be improved if there are more energy bins, since the energy depositions within a narrower bin can be better represented by a single kernel than a wider bin. However, the increased number of bins would require more per-bin convolutions, as well as more source particles to achieve the desired statistics for the energy-differential fluence.

The EPID response changes with respect to incident energy.^{4,34,35} The response peaks at about 0.15 MeV and decreases gradually toward the high energy end of the spectrum. To keep minimal energy variation in each bin, the energy bins are intentionally selected to be dense at the lower end of the energy spectrum and sparse at the higher end in our current implementation. There are a total of 28 energy bins, including 10 bins from 0.0 to 1.0 MeV, each spanning 0.1 MeV, 5 bins from 1.0 to 2.0 MeV, each spanning 0.2 MeV, 8 bins from 2.0 to 10.0 MeV, each spanning 1.0 MeV, and 5 bins from 10.0 to 20.0 MeV, each spanning 2.0 MeV. Correspondingly, 28 monokernels are required at the middle energies of these bins.

II.D.2. Scoring energy-differential fluence

The purpose of the convolution algorithm in this study is to replace the MC DPI prediction algorithm to speed up the calculation. We will focus on an arbitrary j th energy bin over the energy spectra of the incident beam and work through the fluence scoring method so that the convolution result is sufficiently close to the MC result. The energy bin is bounded by E_j^{lower} and E_j^{upper} and contains M incident photons. These photons are assumed to impinge on the EPID at $(x_1, y_1), (x_2, y_2), \dots, (x_M, y_M)$ and have energies $e_1, e_2, \dots, e_M$ anywhere between E_j^{lower} and E_j^{upper} . In addition, the M photons have weights $w_1, w_2, \dots, w_M$ ranging from 0 to 1.

A straightforward formulation of the convolution for the j th energy bin is given by

$$I_j^{\text{convolution}}(x, y) = K^{\text{mono}}(E_j, x, y) \otimes \Phi_j^{\text{particle}}(x, y), \quad (5)$$

where

$$\Phi_j^{\text{particle}}(x, y) = \sum_{i=1}^M w_i * \delta(x - x_i, y - y_i). \quad (6)$$

In this formula, $\delta(x - x_i, y - y_i)$ is the impulse function centered at (x_i, y_i) . The particle fluence $\Phi_j^{\text{particle}}(x, y)$ is the two dimensional count of the statistical weights of M incident photons. The monokernel should be representative of the energy deposition within this energy bin. A good compromise is to use the monokernel $K^{\text{mono}}(E_j, x, y)$ at the middle of the energy bin, i.e., $E_j = (E_j^{\text{lower}} + E_j^{\text{upper}})/2$. The convolution calculated using this formula may still significantly deviate from the MC result, because the energy-dependent EPID response of an arbitrary incident photon is replaced by the response at the middle energy E_j .

To further increase the accuracy of the convolution, the weighted particle fluence is scored. The weighted particle fluence for the j th energy bin may be defined as

$$\Phi_j^{\text{weighted}}(x, y) = \sum_{i=1}^M \frac{IE_i}{IE_j} * w_i * \delta(x - x_i, y - y_i), \quad (7)$$

where IE_j is the integrated energy of the monokernel at E_j and IE_i is the integrated energy of the monokernel at e_i , the energy of the incident photon.

When $\Phi_j^{\text{particle}}(x, y)$ is used for convolution, the incident photon with energy e_i will have the incident energy invariant contribution $w_i * K^{\text{mono}}(E_j, x, y)$, which only relies on the middle energy E_j of the energy bin. However, when $\Phi_j^{\text{weighted}}(x, y)$ is used, the contribution $(IE_i/IE_j) * w_i * K^{\text{mono}}(E_j, x, y)$ would have the integrated energy identical to that caused by $w_i * K^{\text{mono}}(e_i, x, y)$, that is, the contribution changes with the incident energy. Therefore, if $\Phi_j^{\text{particle}}(x, y)$ is replaced by $\Phi_j^{\text{weighted}}(x, y)$ in the convolution, it would mimic the energy dependence of the EPID response. In the remaining text, these two methods will be referred as unweighted and weighted fluence scoring methods, respectively.

II.D.3. Imager-specific monokernel tuning

For each imager, the monokernel backscattering thickness is tuned to match the measured EPID field size responses. Starting with a set of all-in-one kernels, monokernels are assembled with increased backscattering thicknesses. Each set of monokernels is used by the convolution algorithm to calculate 5×5 , 10×10 , 15×15 , and 25×25 cm² field DPIs. Correspondingly, the field size responses for these four fields are measured for the imager under investigation. By an exhaustive search in the convolution results, the imager-specific set of monokernels is recognized as a set of kernels by which the convolved field size responses best match measured ones.

The agreement criterion for the measured and convolved field size responses is the least squares deviation with respect to the measurements, which is defined as

$$D = \sum_{i=5,10,15,25} (\text{FS}^{\text{simulated}}(i \times i) - \text{FS}^{\text{measured}}(i \times i))^2, \quad (8)$$

where $\text{FS}^{\text{simulated}}(i \times i)$ and $\text{FS}^{\text{measured}}(i \times i)$ are field size responses of the $i \times i$ cm² field from convolution and measurement, respectively.

II.D.4. Additional test fields

If the backscattering is constant with respect to the imager position, the imager-specific monokernel tuning should also be imager position independent. To validate this, 5×5 , 10×10 , 15×15 , and 18×18 cm² fields are measured at 150 cm SDD, which is the after-patient EPID imaging position in our institution. The maximal investigated field size is reduced from 25×25 to 18×18 cm² field to keep the maximal field projected to 150 cm within the active area of the imager. The corresponding four fields are simulated at 150 cm SDD using the imager-specific monokernels tuned to match the field size responses at 105 cm SDD.

Because the energy-differential fluence is scored prior to the convolution, *a priori* knowledge regarding the energy spectra incident on the EPID is not required to predict images by our convolution system. To validate that imager-specific monokernels, though tuned by the open-field responses, can predict EPID images that are formed by different incident energy spectra, a series of test fields is created in which the energy spectra are modified by blocking different time fractions of the incident beam. As in previous studies,^{42,49} 10×10 cm² fields are created using the sliding window technique by sweeping uniform 5, 10, 20, and 50 mm MLC gaps. Further validation is performed by measuring and simulating a head-and-neck (H&N) IMRT patient field and performing gamma analysis⁵⁰ with 2%, 2 mm criteria. To avoid the detrimental effects of stochastic noise in the high resolution images on the reported gamma values, values are reported for using both the measured and the computed images as reference images. This test only showcases the feasibility of using the kernels for the portal dosimetry application; intensive comparison with patient field measurements is beyond the scope of this work.

III. RESULTS

The MC simulations, with or without the kernel convolution, predict the absolute response of the imager. Converting this response to the measured response per monitor unit requires application of a calibration procedure similar to that used for patient dose modeling⁵¹ and is simply determined by matching measured and simulated 10×10 cm² fields. In the following results, all measurements are calibrated as such to make them comparable to the simulations. To quantitatively analyze the difference between simulations and measurements, the integrals of the central 1×1 cm² regions are compared. For the 0.784×0.784 mm² pixel size, a 1×1 cm² region contains about 13×13 pixels. The integral of this region leads to 13-fold improvement in statistical precision. Thus, for an EPID MC simulation run with 2.5% precision, the integral of the central 1×1 cm² region would have around 0.2% precision.

III.A. Field size response of various a-Si-EPIDs

Measured field size responses for five Varian EPIDs on accelerators with matched outputs, including two aS1000s with E arms, two aS500s with E arms, and one aS500 with an R arm, are plotted in Fig. 2(a) and the deviations with respect to the aS1000-A are plotted in Fig. 2(b). The responses of the two aS1000s are similar to each other with the maximal deviation less than 0.1%. They are significantly different from the aS500s with the maximal deviation of 1.3%. The aS500s are different from one another with the maximal deviation of 0.6% regardless of the type of the supporting arm.

Arm-on and arm-off field size responses for two aS500s with an E arm and an R arm, respectively, are shown in Fig. 3(a) and the deviations from the aS1000-A are in Fig. 3(b). The arm-on responses are different from the arm-off responses for both imagers with the maximal deviation of

FIG. 2. (a) Measured EPID field size responses for five Varian imagers normalized to the 10×10 cm² field. (b) Deviations of EPID field size responses from the imager aS1000-A. Two aS1000 imagers have similar responses with less than 0.1% maximal deviation. Three aS500 imagers have different responses from aS1000 imagers with the maximal deviation of 1.3%. Moreover, the aS500 imagers themselves do not have identical responses regardless of the type of the robotic arm. The maximal deviation is around 0.6%.

0.8%, suggesting that the supporting arm alters the field size responses due to backscattered particles. Both of the arm-on responses are closer to the responses of aS1000-A, which is supported by a mounting arm, and therefore suffers from backscattering. Moreover, the two aS500s have similar responses with the deviation of 0.2% when the supporting arms are removed. This suggests that the two imagers have identical intrinsic dosimetric characteristics, which is as expected since they have the exact same detector design.

These measurements demonstrate that the cause of different field size responses for different EPIDs is probably the downstream backscattering. This variation in EPID responses must be taken into consideration when generating the convolution kernel.

FIG. 3. (a) Measured EPID field size responses for two aS500 imagers normalized to the 10×10 cm² field with and without the robotic arms attached. (b) Deviations of EPID field size responses from the imager aS1000-A. The EPID field size responses with robotic arms attached show significant deviations from those without arms. The maximal deviations are 0.8% and 0.6% for aS500-B and aS500-C, respectively. In addition, the responses without arms are similar to each other, with the maximal deviation of 0.2%.

III.B. Monokernel

A set of monokernels from 0.05 to 19.0 MeV is calculated with 1 cm of backscattering water added to the default EPID model. The identical EPID model configuration is used in the previously developed EPID MC code to make the MC results comparable to the convolution results. For clarity, only three monokernels at 0.15, 3.5, and 19.0 MeV are shown in Fig. 4(a). The spreading of the energy deposition varies significantly with respect to the incident energy. Figure 4(b) gives the integrated energies of all kernels, which shows the energy dependence of the EPID response. The EPID is much more sensitive at low energy than at high energy.

These monokernels are convolved with artificially generated monoenergetic parallel photon beams, divergent photon beams, and divergent photon and electron beams. The inte-

FIG. 4. (a) Monoenergetic kernels calculated at 0.15, 3.5, and 19.0 MeV. (b) Integrated energies of kernels from 0.15 to 19.0 MeV. Monoenergetic kernels have different shapes with varied incident energies. In addition, EPID is more sensitive to low incident energy than high incident energy.

grals of the central 1×1 cm² regions of all convolution results have the deviations of less than 0.1% from those of corresponding MC results. This indicates that tilted kernels are unnecessary; although the kernels are generated using pencil beams normal to the EPID model at the imager surface, the convolution method reproduces the MC results with divergent incident beams.

Results comparing use of the set of monokernels with EPID MC simulations to compute 5×5 , 10×10 , 15×15 , and 25×25 cm² fields are shown in Fig. 5. The maximal deviation between convolved and MC field size responses in terms of the central 1×1 cm² regions is 0.23%. Note that the convolution code only scores photon contributions to the image formation while the EPID MC code simulates both photon and electron processes. Agreement between the convolution and MC code here and above in the monokernel tests indicates that the electron contribution to the EPID PDI is negligible. In the difference plot, there are impulses on the

FIG. 5. The central profiles of MC simulated and kernel convolved EPID images for 5×5 , 10×10 , 15×15 , and 25×25 cm² fields, respectively. It is an absolute response comparison, that is, two sets of EPID images are not normalized to the 10×10 cm² field as in comparing measured to convolved EPID images. The differences are calculated with respect to convolution results. Kernel convolution can reproduce MC results reasonably well except for large, symmetrical deviations at the steep-gradient field edges, which are attributed to the high frequency losses in kernel convolutions.

beam edges. Note that there are one positive and one negative impulse on each beam edge and they are symmetrical. It suggests that the convolution results have good alignments with the MC results but the convolution algorithm cannot perfectly reproduce the steep-gradient edges. These edges contribute high frequency components in the Fourier domain. Since FFT assumes the band-limited input and output, it is normal to lose some of these high frequency components.

III.C. All-in-one kernel

A set of all-in-one kernels is calculated from 0.05 to 19.0 MeV. Twenty-five layers of backscattering water slabs, 1 mm each, are added to the original EPID model to compute the kernels. The all-in-one kernel at 1.1 MeV is plotted in Fig. 6(a). This kernel consists of 26 subkernels, 1 subkernel for the original EPID model and the other 25 subkernels from each of the backscattering water slabs.

In Fig. 6(b), the monokernel assembled from all-in-one kernel is compared to the directly computed monokernel, both having 1 cm of backscattering water. These two kernels are statistically equivalent to each other but not exactly identical, even though both kernel calculations use the same initial random number seed. The reason for this is that the random number sequence in the two simulations gets out of step as soon as the first particle interacts in the backscattering material beyond 1 cm depth in the all-in-one kernel simulation. This results in the two simulations being essentially uncorrelated. Nonetheless, the absolute differences between these two kernels are bounded by one standard deviation (STD); thus they can be stated as being statistically equivalent to each other.

FIG. 6. (a) All-in-one kernel calculated at 1.1 MeV. The kernel includes contributions from the original EPID model (EPID subkernel) and from subbackscattering water slabs (water subkernels, only two are plotted for clarity). Water subkernels are low frequency components of the all-in-one kernel. (b) Difference between the assembled monokernel from all-in-one kernel and directly calculated monokernel at 1.1 MeV, both having 1 cm backscattering water. The differences are bounded within one STD, which were calculated simultaneously in the monokernel MC simulation.

According to Fig. 6(a), the backscattering kernels are relatively broad, meaning that they are mostly comprised by the low frequency component in the Fourier domain. When assembling the monokernel from the all-in-one kernel, the integrated energy of the monokernel would increase with more backscattering kernels, but the high frequency peak of the monokernel would remain unchanged. Therefore, the monokernel compensated with varied backscattering would scale the convolution result up or down. The amount of backscattering would have little influence on the high gradient region such as the beam edge. The backscattering effects are demonstrated in Fig. 7 for 5×5 and 25×25 cm² fields. Monokernels used in the convolution are assembled with 0, 1, and 2 cm backscatterings, respectively. The backscattering effect is more prominent for the large field than the small field, since a point in the large field receives more contribution from its surrounding points.

FIG. 7. Backscattering effect illustrated by kernel convolution with 0, 1, and 2 cm backscattering water for 5×5 and 25×25 cm² fields. Differences are calculated with respect to 1 cm results. The effect is more prominent for the 25×25 cm² field than the 5×5 cm² field since a pixel in the 25×25 cm² field receives more backscattering contributions from surrounding pixels.

III.D. Comparison between two fluence scoring methods

Comparisons of directly using the energy-differential fluence and weighting the energy-differential fluence by kernel integral at the incident photon energy interpolated from Fig. 4(b) with pure MC simulations are shown in Fig. 8. Results are shown for 5×5 , 10×10 , 15×15 , and 25×25 cm²

FIG. 8. Top: The central profiles of kernel convolutions and MC simulations for 5×5 , 10×10 , 15×15 , and 25×25 cm² fields, respectively, using the unweighted fluence scoring method. Bottom: The weighted method. Both plots are amplified to emphasize the differences. The weighted method improves the prediction accuracy of the kernel convolution with respect to MC. The kernel convolution with the unweighted method overestimates the MC results. Note that the convolved and MC results are not normalized to the 10×10 cm² field. See Table I for the deviations between the convolution and MC in terms of central 1×1 cm² regions.

TABLE I. The deviations of kernel convolved EPID responses from MC simulated for 5×5 , 10×10 , 15×15 , and 25×25 cm² fields in terms of the integrated responses in the central 1×1 cm² regions. The fluences were scored using weighted and unweighted methods, respectively. The weighted method brings down the maximal deviation from 0.98% to 0.23%. Note that MC simulated and kernel convolved images were not normalized to the 10×10 cm² field.

Fluence scoring method	Deviations of convolved EPID responses from MC (%)			
	5×5	10×10	15×15	25×25
Weighted	0.17	0.06	0.02	0.23
Unweighted	0.93	0.69	0.72	0.98

fields. The deviations of the field size responses by the weighted method and by the unweighted method from those by MC are evaluated in terms of the integrals of the central 1×1 cm² regions and are detailed in Table I. Use of the weighted method improves the profile agreement and reduces the maximal field-size response deviation from 0.98% to 0.23%.

III.E. Imager-specific monokernel tuning

To establish the envelope of field size responses possible by altering the effective backscatter in the range of 0–2.5 cm, two sets of monokernels are assembled from the all-in-one kernels with minimal and maximal amounts of backscattering, i.e., 0 and 25 backscattering subkernels, respectively. These two sets of monokernels are used to compute 5×5 , 10×10 , 15×15 , and 25×25 cm² field responses and are plotted with measured field size responses for the five investigated EPIDs in Fig. 9. It is clearly seen that measured responses are within the capture range of the simulated responses. In other words, the field size response of any

FIG. 9. The measured and calculated EPID field size responses. The measured responses are bounded by calculated responses with 0 and 2.5 cm kernels. It suggests that any imager in the current study can be simulated by kernel convolution with backscattering water between 0 and 2.5 cm.

FIG. 10. Top: The central profiles of measured and convolved 5×5 , 10×10 , 15×15 , and 25×25 cm² fields for aS500-A normalized to the 10×10 cm² field. The kernel for the imager aS500-A requires 1.2 cm backscattering water. Middle: The differences are calculated with respect to the simulated profiles. Kernel convolution can reproduce measurement reasonably well except for large, asymmetrical deviations at the steep-gradient field edges. Bottom: A DTA analysis with a 1% dose threshold suggests that all pixels have DTAs less than 1 mm, including the pixels on the field edges. Note that the pixels under threshold are not included in the DTA analysis and are given the DTA value of 0 mm.

investigated imager can be modeled using an imager-specific backscattering water of thickness between 0 and 2.5 cm.

Imager-specific monokernel tuning was performed for the aS500-A. By an exhaustive searching in the precomputed field size responses using monokernels of backscattering water between 0 and 2.5 cm, a monokernel of 1.2 cm backscattering was found to give the best agreement between measured and simulated field size responses. Lateral profiles computed with this kernel set are compared with measurements for 5×5 , 10×10 , 15×15 , and 25×25 cm² fields in Fig. 10. Reasonable agreement is found with asymmetrical impulse differences on the beam edges, which suggests minor misalignments due to our modeling or due to the jaw calibration. To evaluate the impact of these misalignments, a

distance to agreement (DTA) analysis was performed on simulated and measured images for pixels with dose greater than 1% of the in-field dose (Fig. 10). All DTA values are less than 1 mm, which suggests that these misalignments are small and inconsequential. Given the steep gradients at the beam edges, it is not surprising that even small misalignments would cause large dose differences.

The kernel tuning procedure was repeated for the remaining imagers and results are presented in Table II. The required backscattering thicknesses of the tuned monokernels are from 1.0 to 1.6 cm. The maximal deviation between convolved and measured field size responses in terms of the central 1×1 cm² regions is 0.34%.

III.F. Additional test fields

Additional fields were simulated and measured for the aS500-A to further validate the tuned imager-specific monokernels with 1.2 cm backscattering water. 5×5 , 10×10 , 15×15 , and 18×18 cm² fields were simulated at 150 cm SDD and compared to corresponding measurements. The simulations agreed with the measurements within 0.39%.

Profiles of the simulated and measured 10×10 cm² fields generated by sweeping different MLC gap widths, thereby altering the incident energy spectra, are shown in Fig. 11. Similar to the open-field comparisons shown in Fig. 10, the simulations and measurements agree within 0.39%, except the edges. The DTA analysis suggests that the modeling errors at the edges are small.

The measured and simulated H&N IMRT patient fields are shown in Fig. 12. The absolute difference shows that the modeling error is small and the only relative large deviations between the measurement and simulation are in the high gradient regions. Lateral and longitudinal profiles are therefore extracted along the high gradient regions in the measured and simulated images (Fig. 12). The simulated profiles stick tightly to the measured ones. Only a few points drift away significantly from the measured profiles. Gamma analysis for pixels with dose greater than a 3% dose threshold with 2%, 2 mm criteria finds 99.8% and 99.0% of the pixels with $\gamma < 1$ depending on which image is used as the reference.

TABLE II. The water backscatter thickness required for each imager to match the measured field size responses and the residual deviations. The deviations are evaluated in terms of the integrated responses in the central 1×1 cm² regions.

Imager	Water backscatter (cm)	Deviations between measured and convolved (%)			
		5×5	10×10	15×15	25×25
aS1000-A	1.6	0.27	—	0.03	0.24
aS1000-B	1.6	−0.02	—	0.06	0.13
aS500-A	1.2	0.24	—	0.03	0.16
aS500-B	1.3	−0.34	—	−0.16	0.02
aS500-C	1.0	0.20	—	0.24	−0.12

FIG. 11. Top: The measured and simulated central profiles of 10×10 cm² fields formed by 5, 10, 20, and 50 mm MLC sliding windows. Middle: The differences are calculated with respect to the simulated profiles. Kernel convolutions can reproduce measurements reasonably well except for large, asymmetrical deviations at the steep-gradient field edges. Bottom: A DTA analysis with a 1% dose threshold suggests that all pixels have DTAs less than 1 mm, including the pixels on the field edges. The convolution system is robust regardless of the variation of the incident energy spectra. Note that all measurements are normalized by one single normalization factor. The factor is determined by matching the measured and simulated responses of 10×10 cm² open fields.

IV. DISCUSSION

The study of field size responses on five imagers from three institutions suggests that the downstream backscattering plays an important role on their dosimetric characteristics. As clear evidence, for two imagers studied, the supporting robotic arms caused up to 0.8% deviation in the field size responses. When the arms are mounted back to the imagers, the maximal deviation of the field size responses of the five imagers is 1.3%. Among these imagers, two aS1000s agree within 0.1%; three aS500s have up to 0.6% deviation. It is interesting to note that two aS500s with the arms removed have up to 0.2% deviation and two aS500s with the same type of arms attached have up to 0.3% deviation. These deviations might be caused by other factors such as manufacturing differences in the screen layer thicknesses and in the cable arrangements.

An MC-based EPID kernel calculation method was proposed, which allows adjusting the backscatter specific for each individual imager. In generating the kernel, the backscattering contribution is divided into a series of subkernels. When creating an imager-specific EPID kernel, the number of included backscattering subkernels is tunable until the simulated field size responses match with the measured responses.

The method of adding a fixed thickness water slab was used previously to model the backscattering but the thickness had to be determined for each imager.^{14,17,24} In this study, we

FIG. 12. Top left: The measured EPID image of a H&N IMRT patient field. Top middle: The simulated EPID image. Top right: The absolute difference between the measured and simulated. The overall difference is small. The only relative large deviations occur in the high gradient regions. Bottom left: The simulated and measured lateral profiles. Bottom right: The simulated and measured longitudinal profiles. The profiles are intentionally selected to cross the high gradient regions, shown as the white lines in the measured and simulated images. The simulated profiles are in agreement with the measured profiles.

further quantify the relationship between field size responses and backscatter. This relationship can be directly used to estimate the backscattering thickness. Instead of creating the imager-specific kernels by separate MC simulations, the all-in-one kernel method reduces the backscattering modeling to a one time effort. For the imagers studied, the water thickness to compensate for the backscattering ranges from 1.0 to 1.6 cm. The difference of 0.6 cm in the thickness accounts for the maximal 1.3% deviation among the imagers.

Previously, it was reported that the supporting arm results in nonuniform localized backscattering.^{32,33} The backscattering can lead to about 5% signal increase along the longitudinal direction and make the longitudinal profile asymmetrical. Although the localized backscattering is not investigated in this study, the proposed adaptive kernel method could be extended to account for this problem. Basically, local kernels could be created with a different backscatter thickness as a function of location. The local kernels could then be used to model local backscattering sources to improve the accuracy of the prediction.

Besides the improved kernel calculation method, an energy-differential kernel convolution algorithm was developed to best reproduce MC results. The resolution of the energy bin was determined by the energy dependence of the EPID response. A weighted fluence scoring method was used, which shows better agreement to MC than the unweighted method. Using tuned imager-specific monokernels, the convolution algorithm predicted the measured field size responses within the maximal deviation of 0.34% for all involved imagers.

Kernels tuned with the open-field responses can be used to reliably predict EPID images with varied incident energy spectra generated by different MLC sliding window fields by

up to 0.39% deviation and for pretreatment IMRT QA. Indeed, for a H&N IMRT case, gamma analysis on the simulated and measured images with a 3% dose threshold suggests that at least 99.0% of the pixels have $\gamma < 1$ by 2%, 2 mm criteria. The separation of incident fluence into differential energy bins makes the kernels excellent candidates for the treatment time patient EPID image prediction, since the energy spectrum of the after-patient fluence would change from one patient to another and possibly change during the treatment course of the same patient.

V. CONCLUSIONS

The EPID field size responses are affected by the backscattering downstream the imagers. The responses deviate by up to 1.3% for five investigated imagers in this study for otherwise matched accelerators. The inconsistency of the field size responses among different imagers can be taken into account by using the imager-specific monokernels with different backscatter thickness. Such kernels for all imagers can be efficiently generated in a single MC simulation. Using the imager-specific monokernels, the field size responses of five imagers can be modeled to within 0.34% deviation. The simulations by the monokernels agree well with measurements for different incident energy spectra, different imager positions, and a H&N IMRT patient field. The modeling capability offered by the imager-specific monokernels lays a solid foundation for developing other portal dosimetry applications.

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¹P. Munro, "Portal imaging technology: Past, present, and future," *Semin. Radiat. Oncol.* **5**, 115–133 (1995).

²K. A. Langmack, "Portal imaging," *Br. J. Radiol.* **74**, 789–804 (2001).

³W. van Elmpt, L. McDermott, S. Nijsten, M. Wendling, P. Lambin, and B. Mijnheer, "A literature review of electronic portal imaging for radiotherapy dosimetry," *Radiother. Oncol.* **88**, 289–309 (2008).

⁴B. M. McCurdy, K. Luchka, and S. Pistorius, "Dosimetric investigation and portal dose image prediction using an amorphous silicon electronic portal imaging device," *Med. Phys.* **28**, 911–924 (2001).

⁵E. E. Grein, R. Lee, and K. Luchka, "An investigation of a new amorphous silicon electronic portal imaging device for transit dosimetry," *Med. Phys.* **29**, 2262–2268 (2002).

⁶P. B. Greer and C. C. Popescu, "Dosimetric properties of an amorphous silicon electronic portal imaging device for verification of dynamic intensity modulated radiation therapy," *Med. Phys.* **30**, 1618–1627 (2003).

⁷L. N. McDermott, R. J. Louwe, J. J. Sonke, M. B. van Herk, and B. J. Mijnheer, "Dose-response and ghosting effects of an amorphous silicon electronic portal imaging device," *Med. Phys.* **31**, 285–295 (2004).

⁸P. Winkler, A. Hefner, and D. Georg, "Dose-response characteristics of an amorphous silicon EPID," *Med. Phys.* **32**, 3095–3105 (2005).

⁹P. Winkler and D. Georg, "An intercomparison of 11 amorphous silicon EPIDs of the same type: Implications for portal dosimetry," *Phys. Med. Biol.* **51**, 4189–4200 (2006).

¹⁰L. N. McDermott, S. M. Nijsten, J. J. Sonke, M. Partridge, M. van Herk, and B. J. Mijnheer, "Comparison of ghosting effects for three commercial a-Si EPIDs," *Med. Phys.* **33**, 2448–2451 (2006).

¹¹P. B. Greer, "Off-axis dose response characteristics of an amorphous silicon electronic portal imaging device," *Med. Phys.* **34**, 3815–3824 (2007).

¹²E. Spezi and D. G. Lewis, "Full forward Monte Carlo calculation of portal dose from MLC collimated treatment beams," *Phys. Med. Biol.* **47**, 377–390 (2002).

¹³S. C. Vieira, M. L. Dirks, K. L. Pasma, and B. J. Heijmen, "Dosimetric verification of x-ray fields with steep dose gradients using an electronic portal imaging device," *Phys. Med. Biol.* **48**, 157–166 (2003).

¹⁴J. V. Siebers, J. O. Kim, L. Ko, P. J. Keall, and R. Mohan, "Monte Carlo computation of dosimetric amorphous silicon electronic portal images," *Med. Phys.* **31**, 2135–2146 (2004).

¹⁵A. Van Esch, T. Depuydt, and D. P. Huyskens, "The use of an aSi-based EPID for routine absolute dosimetric pre-treatment verification of dynamic IMRT fields," *Radiother. Oncol.* **71**, 223–234 (2004).

¹⁶W. J. van Elmpt, S. M. Nijsten, B. J. Mijnheer, and A. W. Minken, "Experimental verification of a portal dose prediction model," *Med. Phys.* **32**, 2805–2818 (2005).

¹⁷W. Li, J. V. Siebers, and J. A. Moore, "Using fluence separation to account for energy spectra dependence in computing dosimetric a-Si EPID images for IMRT fields," *Med. Phys.* **33**, 4468–4480 (2006).

¹⁸S. Cilla, P. Viola, L. Azario, L. Grimaldi, M. Craus, G. D'Onofrio, A. Fidanzio, A. G. Morganti, and A. Piermattei, "Comparison of measured and computed portal dose for IMRT treatment," *J. Appl. Clin. Med. Phys.* **7**, 65–79 (2006).

¹⁹C. Talamonti, M. Casati, and M. Bucciolini, "Pretreatment verification of IMRT absolute dose distributions using a commercial a-Si EPID," *Med. Phys.* **33**, 4367–4378 (2006).

²⁰L. N. McDermott, M. Wendling, B. van Asselen, J. Stroom, J. J. Sonke, M. van Herk, and B. J. Mijnheer, "Clinical experience with EPID dosimetry for prostate IMRT pre-treatment dose verification," *Med. Phys.* **33**, 3921–3930 (2006).

²¹P. Winkler, A. Hefner, and D. Georg, "Implementation and validation of portal dosimetry with an amorphous silicon EPID in the energy range from 6 to 25 MV," *Phys. Med. Biol.* **52**, N355–N365 (2007).

²²L. Parent, J. Seco, P. M. Evans, A. Fielding, and D. R. Dance, "Monte Carlo modelling of a-Si EPID response: The effect of spectral variations with field size and position," *Med. Phys.* **33**, 4527–4540 (2006).

²³J. Chang and C. C. Ling, "Using the frame averaging of aS500 EPID for IMRT verification," *J. Appl. Clin. Med. Phys.* **4**, 287–299 (2003).

²⁴C. Kirkby and R. Sloboda, "Comprehensive Monte Carlo calculation of the point spread function for a commercial a-Si EPID," *Med. Phys.* **32**, 1115–1127 (2005).

²⁵M. Wendling, R. J. Louwe, L. N. McDermott, J. J. Sonke, M. van Herk, and B. J. Mijnheer, "Accurate two-dimensional IMRT verification using a back-projection EPID dosimetry method," *Med. Phys.* **33**, 259–273 (2006).

²⁶W. J. van Elmpt, S. M. Nijsten, R. F. Schiffeleers, A. L. Dekker, B. J. Mijnheer, P. Lambin, and A. W. Minken, "A Monte Carlo based three-dimensional dose reconstruction method derived from portal dose images," *Med. Phys.* **33**, 2426–2434 (2006).

²⁷T. R. McNutt, T. R. Mackie, P. Reckwerdt, and B. R. Paliwal, "Modeling dose distributions from portal dose images using the convolution/superposition method," *Med. Phys.* **23**, 1381–1392 (1996).

²⁸V. N. Hansen, W. Swindell, and P. M. Evans, "Extraction of primary signal from EPIDs using only forward convolution," *Med. Phys.* **24**, 1477–1484 (1997).

²⁹B. Warkentin, S. Steciw, S. Rathee, and B. G. Fallone, "Dosimetric IMRT verification with a flat-panel EPID," *Med. Phys.* **30**, 3143–3155 (2003).

³⁰S. Steciw, B. Warkentin, S. Rathee, and B. G. Fallone, "Three-dimensional IMRT verification with a flat-panel EPID," *Med. Phys.* **32**, 600–612 (2005).

³¹K. Chytky and B. M. McCurdy, "Investigation of tilted dose kernels for portal dose prediction in a-Si electronic portal imagers," *Med. Phys.* **33**, 3333–3339 (2006).

³²L. Ko, J. O. Kim, and J. V. Siebers, "Investigation of the optimal backscatter for an aSi electronic portal imaging device," *Phys. Med. Biol.* **49**, 1723–1738 (2004).

³³J. A. Moore and J. V. Siebers, "Verification of the optimal backscatter for an aSi electronic portal imaging device," *Phys. Med. Biol.* **50**, 2341–2350 (2005).

³⁴A. E. Schach von Wittenau, C. M. Logan, M. B. Aufderheide 3rd, and D. M. Slone, "Blurring artifacts in megavoltage radiography with a flat-panel imaging system: Comparison of Monte Carlo simulations with measurements," *Med. Phys.* **29**, 2559–2570 (2002).

³⁵C. Kirkby and R. Sloboda, "Consequences of the spectral response of an

- a-Si EPID and implications for dosimetric calibration," *Med. Phys.* **32**, 2649–2658 (2005).
- ³⁶R. C. Tabor, V. M. Tello, C. B. Schroy, M. Vossler, and W. F. Hanson, "A generic off-axis energy correction for linac photon beam dosimetry," *Med. Phys.* **25**, 662–667 (1998).
- ³⁷D. Sheikh-Bagheri and D. W. Rogers, "Monte Carlo calculation of nine megavoltage photon beam spectra using the BEAM code," *Med. Phys.* **29**, 391–402 (2002).
- ³⁸J. O. Kim, J. V. Siebers, P. J. Keall, M. R. Arnfield, and R. Mohan, "A Monte Carlo study of radiation transport through multileaf collimators," *Med. Phys.* **28**, 2497–2506 (2001).
- ³⁹G. Jarry and F. Verhaegen, "Patient-specific dosimetry of conventional and intensity modulated radiation therapy using a novel full Monte Carlo phase space reconstruction method from electronic portal images," *Phys. Med. Biol.* **52**, 2277–2299 (2007).
- ⁴⁰P. J. Keall, J. V. Siebers, B. Libby, and R. Mohan, "Determining the incident electron fluence for Monte Carlo-based photon treatment planning using a standard measured data set," *Med. Phys.* **30**, 574–582 (2003).
- ⁴¹D. W. Rogers, B. A. Faddegon, G. X. Ding, C. M. Ma, J. We, and T. R. Mackie, "BEAM: A Monte Carlo code to simulate radiotherapy treatment units," *Med. Phys.* **22**, 503–524 (1995).
- ⁴²J. V. Siebers, P. J. Keall, J. O. Kim, and R. Mohan, "A method for photon beam Monte Carlo multileaf collimator particle transport," *Phys. Med. Biol.* **47**, 3225–3249 (2002).
- ⁴³I. Kawrakow and B. R. Walters, "Efficient photon beam dose calculations using DOSXYZnrc with BEAMnrc," *Med. Phys.* **33**, 3046–3056 (2006).
- ⁴⁴I. Kawrakow, "VMM++, electron and photon Monte Carlo calculations optimized for Radiation Treatment Planning," in *Advanced Monte Carlo for Radiation Physics, Particle Transport Simulation and Applications: Proceedings of the Monte Carlo 2000 Meeting*, Lisbon, Portugal, edited by A. Kling, F. Barao, M. Nakagawa, L. Távora, and P. Vaz (Springer, Berlin, 2001), pp. 229–238.
- ⁴⁵P. B. Greer, "Correction of pixel sensitivity variation and off-axis response for amorphous silicon EPID dosimetry," *Med. Phys.* **32**, 3558–3568 (2005).
- ⁴⁶I. Kawrakow, "Accurate condensed history Monte Carlo simulation of electron transport. I. EGSnrc, the new EGS4 version," *Med. Phys.* **27**, 485–498 (2000).
- ⁴⁷W. R. Nelson, H. Hirayama, and D. W. O. Rogers, "The EGS4 code system," Stanford Linear Accelerator Center, Technical Report No. SLAC-265, 1985.
- ⁴⁸M. Frigo and S. G. Johnson, "The design and implementation of FFTW3," *Proc. IEEE* **93**, 216–231 (2005).
- ⁴⁹M. R. Arnfield, J. V. Siebers, J. O. Kim, Q. Wu, P. J. Keall, and R. Mohan, "A method for determining multileaf collimator transmission and scatter for dynamic intensity modulated radiotherapy," *Med. Phys.* **27**, 2231–2241 (2000).
- ⁵⁰D. A. Low, W. B. Harms, S. Mutic, and J. A. Purdy, "A technique for the quantitative evaluation of dose distributions," *Med. Phys.* **25**, 656–661 (1998).
- ⁵¹B. Bednarz and X. G. Xu, "Monte Carlo modeling of a 6 and 18 MV Varian Clinac medical accelerator for in-field and out-of-field dose calculations: development and validation," *Phys. Med. Biol.* **54**, N43–N57 (2009).