Accurate two-dimensional IMRT verification using a back-projection EPID dosimetry method

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The use of electronic portal imaging devices (EPIDs) is a promising method for the dosimetric verification of external beam, megavoltage radiation therapy—both pretreatment and in vivo. In this study, a previously developed EPID back-projection algorithm was modified for IMRT techniques and applied to an amorphous silicon EPID. By using this back-projection algorithm, two-dimensional dose distributions inside a phantom or patient are reconstructed from portal images. The model requires the primary dose component at the position of the EPID. A parametrized description of the lateral scatter within the imager was obtained from measurements with an ionization chamber in a miniphantom. In addition to point dose measurements on the central axis of square fields of different size, we also used dose profiles of those fields as reference data for our model. This yielded a better description of the lateral scatter within the EPID, which resulted in a higher accuracy in the back-projected, two-dimensional dose distributions. The accuracy of our approach was tested for pretreatment verification of a five-field IMRT plan for the treatment of prostate cancer. Each field had between six and eight segments and was evaluated by comparing the back-projected, two-dimensional EPID dose distribution with a film measurement inside a homogeneous slab phantom. For this purpose, the comparison between EPID and film measurements for each field, both in the central part of the beam and in the penumbra and low-dose regions. It can be concluded that our modified algorithm is able to accurately predict the dose in the midplane of a homogeneous slab phantom. For pretreatment IMRT plan verification, EPID dosimetry is a reliable and potentially fast tool to check the absolute dose in two dimensions inside a phantom for individual IMRT fields. Film measurements inside a phantom can therefore be replaced by EPID measurements.

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I. INTRODUCTION

The challenge of external beam radiotherapy for cancer treatment is to irradiate the tumor with a high dose, while the surrounding healthy tissue suffers as little as possible radiation damage. Due to the increasing complexity of nearly all steps in modern radiotherapy, the demand for a thorough verification of the dose delivered to the patient has also increased, either pretreatment or in vivo.

For pretreatment verification various treatment parameters, such as beam energy, number of monitor units, and (multileaf) collimator settings, have to be verified to ensure the correct dose delivery to a patient. These parameters can be used to calculate a dose distribution within a phantom. A variety of methods is available to check the absolute dose at specific points, which is usually done with ionization chamber measurements, and to verify relative dose distributions, e.g., by film measurements in specific planes or by gel dosimetry in three dimensions.1 In vivo dose verification is often done by placing dosimeters, such as diodes, thermoluminescence dosimeters, or metal oxide semiconductor field effect transistors (MOSFETs), on the skin of patients or inside patients to derive the dose at specific points within the patient (for a review see Ref. 2).

These measurements are very labor intensive and yield merely a limited amount of information; often, the dose is only determined at a single point. One would like to have an alternative method to verify dose delivery in two or, preferably, three dimensions. EPID dosimetry is a very promising approach for this purpose. EPIDs—electronic portal imaging devices—are widely used for setup verification during radiotherapy. These devices are easy to use and data acquisition is fast. Many types of EPIDs combine good reproducibility of the response,2–7 the possibility to measure dose distributions in two dimensions with high spatial resolution, and a digital format of the images.

Basically, there are two approaches to EPID dosimetry and in principle both are suitable for pretreatment verification and in vivo dosimetry. In the “forward approach,” the measured (and sometimes processed) portal image is compared to a predicted dose or photon fluence at the plane of the EPID, which is calculated with the treatment planning system (TPS) or an independent algorithm.8–17 In the “backward approach,” portal images are used to reconstruct the dose within the patient or phantom.18–21 This back-projection method makes it possible to directly compare the calculated with the delivered dose distribution in the patient or phan-
tom. While the dose is verified in only one plane with the forward approach, three-dimensional dose reconstruction is potentially possible with the back-projection method.\textsuperscript{19,20,21}

Variations on these approaches have also been described in the literature. In a “hybrid” method, McNutt \textit{et al.}\textsuperscript{22,23} treat the EPID as part of an extended volume and use a convolution/superposition algorithm to predict a portal dose image. The primary energy fluence is then iteratively adapted until predicted and measured portal dose distributions agree. Finally, the converged primary energy fluence is back-projected and convolved with the dose deposition kernel yielding the dose distribution within the patient or phantom in three dimensions. In the pretreatment verification method of Warkentin \textit{et al.},\textsuperscript{24} EPID images are acquired without a phantom in the beam and these EPID images are deconvolved with a (two-component) “dose-glare” kernel for the EPID to yield the two-dimensional primary fluence. This fluence is then convolved with a phantom dose-deposition kernel to yield the dose distribution in a solid-water phantom; for absolute dosimetry, cross calibration with an ionization chamber is performed. However, this method cannot be applied for \textit{in vivo} dosimetry purposes.

Our back-projection method has been described for liquid-filled matrix ionization chamber EPIDs,\textsuperscript{19,20,26} but is also applicable to other types of EPIDs. The calibration of the EPID for back-projection dosimetry consists of two parts. First, a dosimetric calibration is needed to establish the dose-response relationship by relating EPID pixel values to dose values at the position of the imager. Second, the parameters for the back-projection algorithm have to be determined to enable the conversion from the dose at the EPID position to the dose inside the patient or phantom. This is done by applying correction procedures for the scatter component of the dose within the EPID and the scatter from the patient or phantom to the EPID. Furthermore, the scatter component within the patient or phantom, in combination with the attenuation of the beam, is accounted for to obtain the total dose at specific points in the patient or phantom.

In our back-projection approach,\textsuperscript{19,20} the relationship between the EPID signal and the “real dose” (defined as the dose measured with a calibrated ionization chamber) has, until now, been derived from data only on the central axis. As a consequence, the errors in dose reconstruction at off-axis positions are usually larger than those on the central axis. This is relevant since the treatment outcome does not only critically depend on the dose delivered to the tumor (usually close to the isocenter), but also on the dose delivered to normal tissue. Moreover, sensitive anatomical structures are often close to the location of steep dose gradients within the patient. Therefore, it is essential to improve the accuracy of the method in the penumbra region and outside the field. This especially applies to IMRT fields, which may have many steep dose gradient/penumbra regions, spread throughout the target volume and organs at risk.

The purpose of this study was to improve and evaluate the accuracy of the back-projection method for verifying two-dimensional dose distributions in phantoms irradiated with intensity-modulated beams. The back-projection method was originally developed for a liquid-filled matrix ionization chamber EPID;\textsuperscript{19,20} we adapted the algorithm and applied it to an amorphous silicon (aSi) EPID. The aim of this study was also to reveal how well the EPID back-projection method compares with the \textit{de facto} “gold standard” film dosimetry procedure for pretreatment IMRT verification.

There are currently three aSi EPID systems commercially available for portal imaging: the PortalVision aS500 by Varian Medical Systems (Palo Alto, CA), the BEAMVIEW TI by Siemens Medical Solutions (Erlangen, Germany), and the iViewGT by Elekta (Crawley, UK). The panels for both Siemens and Elekta EPIDs are manufactured by PerkinElmer (Fremont, CA). Depending on the hard- and software used, the dosimetric properties of the various aSi EPID systems may differ. The basic dosimetric properties of aSi EPIDs have been reported for both the Varian Portal Vision aS500 (Refs. 5 and 25) and the Elekta iViewGT (Refs. 7 and 26) systems. An Elekta aSi EPID was used for our study; however, while parameter details may vary, the principles underlying the (improved) back-projection algorithm outlined here are also valid for other types of EPIDs.

### II. MATERIALS AND METHODS

#### A. Accelerator and EPID

The measurements were done using an 18 MV photon beam of an SL20i linear accelerator (Elekta, Crawley, UK). The accelerator was equipped with a multileaf collimator (MLC), consisting of 40 leaf pairs with a projected leaf width at the isocenter of 1 cm.\textsuperscript{27} For all measurements described in this paper the gantry angle was set to 0°.

A PerkinElmer RID 1680 AL5/Elekta iViewGT aSi EPID was used for all measurements. This imager has a 1 mm thick copper plate on top of the scintillation layer. An extra 2.5 mm thick copper plate was mounted directly on top of the standard plate (replacing the aluminum cover plate) both as additional buildup material and to absorb scattered low-energy photons from the phantom or patient; this modification has negligible impact on image quality.\textsuperscript{26} The EPID has a sensitive area of 41 × 41 cm². In this paper, the axis of an EPID image parallel to the plane of gantry rotation is called \(x\), the axis perpendicular to it, \(y\).

Images were acquired using in-house developed software\textsuperscript{7,26} (a similar image acquisition is possible with the commercially available Elekta software). In the acquisition mode used, EPID frames are acquired every 285 ms and stored in a buffer. When the signal of a frame increases above a set threshold (“beam-on” trigger), the image acquisition is started. When the signal of a frame drops below the threshold (“beam-off” trigger), the image acquisition is stopped. The signal of all frames between beam-on and beam-off is averaged; two pre-beam-on and two post-beam-off frames are included in the average to make sure that no signal before the beam-on trigger and after the beam-off trigger is missed, which is especially important for segments having a small number of monitor units. This raw EPID image is processed with flood-field and (dynamic) dark-field images to optimize the image quality for patient position.
verifying. (Note that dynamic dark-field images are continuously acquired every 30 s; the EPID is not irradiated.) Therefore, all segments of one field will usually have the same dynamic dark field, whereas segments of different fields will have different ones.) The processed image is then stored, with the number of frames in the image header. Consequently, in a multiple-segment field, one image is obtained per segment.

The EPID images were recorded at a resolution of 512 × 512 pixels, yielding an effective pixel size of 0.05 cm in the isocenter plane. When we mention in this study the “central pixel value” or “central pixel dose” of the EPID, we refer to the “average pixel value” or the “average pixel dose” of a region of 9 × 9 pixels at the center of the EPID.

The distance between the accelerator target and the touch guard of the EPID was approximately 157 cm. To accurately estimate the effective source-detector distance (SDD), i.e., the distance from the target to the “imaging” layer of the EPID, we placed a thin brass plate with well-known dimensions in the isocenter plane perpendicular to the beam axis. The plate was irradiated with a field larger than the plate, and the plate’s dimensions in the portal image were measured. From this experiment we concluded that for this accelerator the effective SDD of the aSi EPID is 160.0 cm ± 0.5 cm.

B. Calibration of the EPID and the back-projection algorithm

Several steps are necessary to reconstruct the dose in the phantom or patient from the pixel values of the EPID. By multiplying the acquired (frame-averaged) EPID image by the number of frames, the time-integrated EPID signal is obtained (knowledge of the delivered number of monitor units is not necessary). The resulting response of the aSi EPID has been shown to be linear with dose, although a small ghosting effect remains, which is mainly a function of the number of exposed frames. In this study, no correction for ghosting was made. Note that the flood-field calibration corrects both for variations in pixel sensitivity and non-flatness of the flood field. For dosimetry purposes, the latter is an “overcorrection” and needs to be “factored out.” This is achieved by using the so-called sensitivity matrix, which is determined experimentally.

Our original implementation of the back-projection algorithm has been described in detail elsewhere. Briefly, for dose reconstruction, i.e., relating pixel values in the EPID images with absolute dose values in the phantom or patient, we have to account for

(i) the dose-response of the EPID;
(ii) the lateral scatter within the EPID;
(iii) the scatter from the phantom or patient to the EPID;
(iv) the attenuation of the beam by the phantom or patient;
(v) the distance from the radiation source to the EPID plane and to the dose-reconstruction plane; and
(vi) the scatter within the phantom or patient.

In order to determine the necessary parameters, EPID images of square fields of several sizes are recorded, with and without a polystyrene slab phantom of several thicknesses in the beam. In this study, we used field sizes of 3, 4, 5, 6, 8, 10, 15, and 20 cm. The phantoms were between 11 and 49 cm thick; this thickness range was chosen to encompass more than sufficiently “typical patient thicknesses” for 18 MV photon beams. For the phantom measurements an isocentric setup (SAD [source-axis distance] setup) was used.

Ionization chamber measurements are used as reference data to fit the model parameters for steps (i), (ii), and (vi). In the original algorithm, the reference dose is only determined at a single point on the central axis. In this study, the method was extended by using dose profiles (instead of point dose measurements alone) as a reference in step (ii) to account more accurately for the lateral scatter within the EPID. This was done as follows.

After the conversion of the EPID pixel values into dose values according to the dose-response relation, the resulting image is called dose image \( D^\text{EPID} \). The portal dose image \( PD^\text{EPID} \), which is the dose image after correction for lateral scatter in the EPID, is obtained by

\[
PD_{ij}^\text{EPID} = D_{ij}^\text{EPID} \odot K_{ij}^\text{EPID,1} \odot K_{ij}^\text{EPID,2},
\]

where \( K_{ij}^\text{EPID,1} \) and \( K_{ij}^\text{EPID,2} \) are EPID scatter correction kernels, \( \odot \) and \( \odot^{-1} \) denote the convolution and deconvolution operator, respectively. Every pixel in the EPID image is referred to by its indices \( i \) and \( j \); \( r_{ij} \) is the distance of a pixel \( ij \) from the central axis; \( c_1, \mu_1, c_{\text{DR}}, c_2, \) and \( \sigma \) are the kernel parameters. In the original method only the first kernel was applied. This was sufficient when only point measurements were used as reference input data in the model. We introduced the second kernel to improve the agreement for dose profiles at the plane of the EPID, in order to arrive at a more accurate dose reconstruction—especially in the penumbra region, which is important for IMRT fields.

Convolution and deconvolution operations were performed in the frequency domain using the fast Fourier transform in two dimensions for computational speed. In order to prevent wrap-around effects, the EPID images were first padded with zeros at all edges. With this procedure it is assumed that the dose at the edges of the imager is negligible. For this reason, profiles for the largest field size (20 × 20 cm²) were not taken into account.

The dose reconstruction is done per acquired EPID image, i.e., per segment (or per field for nonsegmented fields). Note that for every image behind the phantom or patient, one additional image is required without the phantom or patient in the beam to estimate its transmission. In order to summarize
the back-projection method and to elucidate the extensions of the method, the details are given in the Appendix.

For all profile measurements a small ionization chamber (Semiflex 0.125 cm³, PTW-Freiburg, Freiburg, Germany) was used in combination with an electrometer (Keithley Instruments Inc., Ohio). This ionization chamber has an inner diameter of approximately 5 mm. For profile measurements at the level of the EPID, the ionization chamber was located in a cylindrical PMMA miniphantom (diameter 4 cm, buildup 3 cm for an 18 MV photon beam), which was placed in an empty water phantom (PTW-Freiburg, Freiburg, Germany) for accurate positioning in two dimensions. However, with the water phantom the ionization chamber in the miniphantom could not be moved to SDDs much larger than 140 cm; the profiles were therefore measured at an SSD of 140 cm and their coordinates were scaled to the actual EPID level of 160 cm (for absolute dose determination, the ionization chamber in the miniphantom was set up without the water phantom at an SSD of 160 cm at the central axis). Dose profiles in the full-scatter water phantom [source-surface distance (SSD)=90 cm] were measured with the ionization chamber at 10 cm depth. For absolute dosimetry, the Semiflex ionization chamber was calibrated under reference conditions (at 10 cm depth in a 20 cm thick polystyrene slab phantom with SSD=90 cm, 200 monitor units, 10×10 cm² field) against a calibrated Farmer-type ionization chamber (NE 2571 0.6 cm³, NE Technology Ltd, Reading, UK).

An approach using film dosimetry (see Sec. II C) instead of ionization chamber measurements as a reference method to estimate the parameters for the second EPID kernel, $K_{\text{EPID},2}$, was also studied. Measurements with film should be done under full-scatter conditions and therefore the method presented so far had to be adapted. First, all parameters for the back-projection algorithm (see the Appendix) were estimated using only reference values on the central axis, i.e., only kernel $K_{\text{EPID},1}$ was used at the EPID level. Then, the dose profile in the $x$ direction of a 10×10 cm² field was taken from a film measurement at 10 cm depth in a 20 cm thick polystyrene slab phantom at an SSD of 90 cm. This profile was used as a reference for the reconstructed EPID midplane dose profile for that field in order to adjust the parameters for the second kernel $K_{\text{EPID},2}$; we will call the “film-adjusted” second kernel $\tilde{K}_{\text{EPID},2}$. In principle, this approach should be iterated, i.e., after $K_{\text{EPID},2}$ is fit for the first time, all earlier steps in the parameter estimation for the back-projection procedure should be repeated. Then, $K_{\text{EPID},2}$ should be fit for the second time, etc., until all parameters for the back-projection algorithm have converged. However, because introducing $K_{\text{EPID},2}$ is only a small modification, affecting mainly the penumbra region, we omitted the iteration process and used the first fit result for $K_{\text{EPID},2}$.

In this paper, all presented results for EPID dose reconstruction inside a phantom were obtained for an isocentrically positioned polystyrene slab geometry phantom (20 cm thick, SSD=90 cm). In principle, the dose is reconstructed in the radiological midsurface of the phantom. Due to the simple geometry and setup of the phantom, this plane coincides both with the geometrical midplane of the phantom and a plane through the isocenter.

C. Film dosimetry

For the film measurements EDR2 films (Eastman Kodak Company, Rochester, NY) were used. The film was placed perpendicular to the beam axis at 10 cm depth in the polystyrene slab geometry phantom (20 cm thick, SSD=90 cm), i.e., the film was located at the plane to which the EPID dose was back-projected.

The films were processed using a Kodak X-OMAT 3000 RA film processor and digitized with a Lumiscan 75 film scanner (Lumiys Inc., now part of Eastman Kodak Company). The digitized images were corrected for background and by linear scaling for geometric distortion in both the film-feeding direction of the scanner and the direction perpendicular to it. The resolution of the scanned images was 0.1 mm in both directions. The images were smoothed using a running average filter with a size of 0.5×0.5 mm² for noise reduction and for making the effective film image resolution approximately equal to the EPID pixel size at the isocenter.

The sensitometric curve of the film (pixel values of the film versus absolute dose) was determined by irradiating films with a different number of monitor units (0, 25, 50, 75, 100, 150, 200, 250, 300, 350) for a field size of 10×10 cm². The absolute dose at the film position was determined with the calibrated Farmer-type ionization chamber. The sensitometric curve was fit using a fourth-order polynomial (rms=1.08), which was then used to convert film pixel values into absolute dose. Good dosimetric results can be obtained with this method.

D. Pretreatment verification: EPID versus film

In order to assess the accuracy of our method for pretreatment verification, we used a clinical step-and-shoot IMRT plan for a prostate cancer treatment generated with our TPS (Pinnacle V7.4, Philips Medical Systems, Eindhoven, The Netherlands). This plan consisted of five fields, and each field had between six and eight segments; the total number of segments was 37. In order to avoid underexposure of the EDR2 films, the original number of monitor units for each field was multiplied by a factor of 5 to reach optimal dose values.

With the EPID, one image was acquired for each segment with the phantom in the beam and one image for each segment without the phantom in the beam (per-segment image acquisition and storage also allows verification of MLC leaf positions). The back-projection to the midplane was also done separately for each segment. The two-dimensional midplane dose distributions of all segments of each field were then summed to obtain the total midplane dose of that specific field. The film was irradiated simultaneously with the corresponding EPID image acquisition for each field, to prevent (small) differences in MLC leaf (re-)positioning and accelerator output variations. For comparison of the two dose
distributions, the field edges of the (summed) EPID midplane dose image and of the film dose image were first matched for each field—since both measurements were performed simultaneously, the field edges had to agree. Then, profiles of EPID and film dose distributions were compared and a $\gamma$ evaluation was performed in two dimensions.

**E. $\gamma$ evaluation**

The $\gamma$ index is a useful tool to compare dose distributions that have both low- and high-dose gradient regions. It combines a dose-difference criterion with a distance-to-agreement criterion. A $\gamma$ index smaller/larger than unity means that both distributions agree/disagree for that point with respect to the chosen criteria. In this study, the two-dimensional dose distributions measured with EPID and with film were compared. The $\gamma$ index was calculated for every pixel of the EPID image. We used 2% of the maximum dose as dose-difference criterion and 2 mm as distance-to-agreement criterion.

During the determination of the parameters for the back-projection algorithm (see Sec. II B), the same method was used for the comparison of EPID dose profiles with dose profiles measured with an ionization chamber. The optimization of the parameters of EPID scatter kernel $K_{EPID,2}$, $c_2$ and $\sigma$ [see Eq. (2)], was done as follows. For each EPID-ionization chamber profile pair (i.e., for each field size) a $\gamma$ profile was calculated over the length of the EPID profile and from this $\gamma$ profile a mean $\gamma$ index was computed. The average (over all field sizes) of the mean $\gamma$ indices was minimized in the fit procedure. An overall shift for all ionization chamber dose profiles was allowed in the fit to be able to correct for (small) positioning errors of the ionization chamber/miniphantom and of the EPID. This shift was found to be smaller than 0.2 cm.

**III. RESULTS**

**A. Scatter in the EPID**

In Fig. 1, central axis dose values of square fields of different size are shown. These were determined at the position of the EPID without a phantom in the beam, both with an ionization chamber in a miniphantom and with the EPID (dose values before and after EPID scatter correction are shown). The ionization chamber data were used as reference values for the (primary) portal dose. If no scatter correction was performed for the EPID, the EPID curve was steeper than the one for the ionization chamber. When the reference curve was used to fit the scatter kernel $K_{EPID,1}$, the maximum difference between EPID and ionization chamber measurements became smaller than 0.2% (see Fig. 1).

As in previous studies, ionization chamber and EPID data were only fit and compared on the central axis so far. In Fig. 2(a) we compare dose profiles measured with an ionization chamber in a miniphantom with those determined with the EPID after the first scatter correction with the deconvolution kernel $K_{EPID,1}$. As could be expected from the results shown in Fig. 1, the agreement was very good for the central axis region, but became worse in the penumbra and in the tails of the profiles. Generally, the EPID dose profiles showed a steeper dose falloff in the penumbra than the ionization chamber data due to blurring of the measurement by the ionization chamber.

In Fig. 2(b), EPID profiles are again compared with ionization chamber profiles, but now both EPID scatter kernels, $K_{EPID,1}$ and $K_{EPID,2}$, were applied. Compared to Fig. 2(a), the agreement improved considerably and was very good for both the absolute dose and for the shape of the profiles, al-
through some discrepancy remained in the penumbra regions. The profiles for all field sizes were fit simultaneously and therefore the result is a compromise. The average of the mean γ indices of all profiles decreased from 0.4 to 0.2 by using an additional kernel $K^{EPID,2}$. The kernel parameters were $c_1=4.0 \times 10^{-3}$, $\mu_1=0.011$ cm$^{-1}$, $c_{DR}=1.2$ for $K^{EPID,1}$ and $c_2=1.0$, $\sigma=0.41$ cm for $K^{EPID,2}$. Both EPID scatter kernels are displayed in Fig. 3. The parameters for the film-adjusted second kernel $K^{EPID,2}$ are presented as part of the next section.

### B. Midplane dose

In Fig. 4, midplane dose profiles of square fields of different size measured with an ionization chamber in a full-scatter water phantom are shown together with midplane dose profiles reconstructed from EPID images. For Fig. 4(a), the EPID reconstruction was done using only one kernel, $K^{EPID,1}$, to correct for the lateral EPID scatter [see Fig. 2(a)]. For the EPID profiles shown in Fig. 4(b), both EPID scatter kernels, $K^{EPID,1}$ and $K^{EPID,2}$, were applied [see Fig. 2(b)]. Note that no extra fitting of parameters was performed for the midplane dose profiles. After the EPID scatter kernels $K^{EPID,1}$ and $K^{EPID,2}$ were determined, as described in Sec. II B, the remaining parameters for the back-projection method were estimated in each case using values on the central axis only (see the Appendix). The agreement of the midplane dose profiles, especially in the penumbra region, clearly improved by using a better fit at the level of the EPID. The discrepancy on the central axis was smaller than 1%. Overall, the profiles from reconstructed EPID images of square fields agreed with the ionization chamber measurements within the 2% / 2 mm γ criteria both in the center and in the penumbra of the fields; in small parts of the profiles tails, the maximum dose difference was approximately 3% of dose maximum.

In Fig. 5 we compare the midplane dose profile of a 10 × 10 cm$^2$ field determined with film and other methods: (1) the ionization chamber measurement shows a shallower penumbra than the film; (2) when only EPID kernel $K^{EPID,1}$ is used, the reconstructed EPID dose profile has a steeper penumbra than the film, and deviations exist especially in the shoulders and tails of the profiles; (3) when kernel $K^{EPID,2}$ is determined to fit the EPID profile measurement to the film (with parameters $c_2=1.0$, $\sigma=0.38$ cm, displayed in Fig. 3) the best agreement is obtained: shoulder and tail regions agree very well; however, there is still some discrepancy in the upper part of the penumbra region. Note that due to $K^{EPID,2}$ the penumbra for the EPID has broadened.
In Fig. 6 we compare the reconstructed EPID midplane dose image of a 10 × 10 cm² field with a film measurement in the homogeneous slab phantom using the evaluation method with the 2%/2 mm criteria. The histogram shows the distribution of indices. Over an area of 14 × 14 cm² (which approximately represents the area within the 3% isodose line), 99.8% of the pixels satisfied the chosen criteria with a maximum γ index of 1.12. When the pixels with dose values smaller than 10% of the maximum dose were disregarded, all points satisfied the chosen γ criteria and the maximum γ index was 0.96.

**C. Pretreatment IMRT verification: EPID versus film**

A clinical step-and-shoot IMRT plan was delivered to a 20 cm thick polystyrene slab phantom. Figure 7 shows the results for one field consisting of eight segments as an example. The midplane dose distribution for this field, which was reconstructed from EPID images, is shown in Fig. 7(a) and illustrates the typical intensity modulation of the fields. In Fig. 7(b) γ profiles from the reconstructed EPID midplane dose image and from the film dose image are shown, demonstrating very good agreement. The γ index distribution for the comparison of EPID and film measurements is presented in Fig. 7(c), and shows agreement within the 2%/2 mm γ criteria. In the area of 14 × 14 cm² only a few pixels had a γ index larger than unity with a maximum of 1.09. In Fig. 8, γ histograms for all five IMRT fields are shown. The histograms were calculated for a 14 × 14 cm² square, which amply encompasses each field. All γ distributions had a mean value below 0.4. Nearly all γ indices were below unity. The percentages of pixels with γ indices above unity were 0.04%, 0.03%, 0.05%, 0.01%, and 0.14% with maximum γ indices of 1.09, 1.15, 1.19, 1.15, and 1.47 for fields A, B, C, D, and E, respectively.

**IV. DISCUSSION**

**A. Scatter in the EPID**

In the back-projection algorithm, we need the primary dose component at the level of the EPID. We used an ionization chamber in a miniphantom at the EPID position as a reference detector, because in this way only the primary dose component is measured. Within the EPID, mainly lateral x-ray scatter takes place, but also optical photon scatter occurs. For these reasons, the uncorrected EPID has a steeper field size-dependent response than the ionization chamber in a miniphantom (see Fig. 1). Note that the EPID was normalized to the ionization chamber measurement of the 10 × 10 cm² field. This point was chosen arbitrarily and the normalization factor is compensated by the fit parameter $c_{DR}$ of kernel $K_{EPID,1}$ [see Eq. (2)]. The EPID images are...
corrected with kernel $K^{\text{EPID},1}$, which describes the overall scatter effect, i.e., both the “dosimetric scatter” (x rays) and the “glare” (optical photons). After correction of the EPID images with kernel $K^{\text{EPID},1}$, the average pixel dose agreed with the ionization chamber measurement within 0.2%. However, the EPID dose profiles deviated in the region of the “horns” from the profiles measured with the ionization chamber in a miniphantom. Moreover, the EPID profiles had a steeper penumbra [see Fig. 2(a)].

These differences may have (at least) the following causes: (i) the (dimensions of the) ionization chamber, (ii) the size of the miniphantom, and (iii) the increased low-energy response of the EPID.

(i) Compared to the EPID, the ionization chamber has a lower spatial resolution due to volume averaging effects.\textsuperscript{35} Moreover, in principle a gas-filled ionization chamber itself is not a good instrument to determine the dose in the penumbra, because of the lack of electron equilibrium in this region.\textsuperscript{35,36}

(ii) The miniphantom is used to measure the primary dose component. On one hand, the diameter of the miniphantom has to be large enough that lateral electron equilibrium is achieved; on the other hand, it has to be small (“mini”) with respect to the field size, so that negligible side scatter takes place.\textsuperscript{32,33}

(iii) Amorphous silicon EPIDs are known to have an over-response to low-energy photons.\textsuperscript{16,17,37–40} The distribution of low-energy photons reaching the EPID is in principle field size and position dependent (when there is a patient or phantom in the beam, it also depends on the exact patient or phantom geometry). Due to the shape of the flattening filter, low-energy photons contribute relatively more to the energy spectrum off-axis. However, this effect was reduced by adding a 2.5 mm extra copper plate to the detector (see Sec. II A). This copper plate also reduces the effect of low-energy photons scattered from the phantom or patient on the EPID under our measurement conditions, i.e., at an SDD of 160 cm for the EPID and hence with a large air gap between EPID and phantom or patient.

All effects were combined in one additional empirical kernel, $K^{\text{EPID},2}$, at the level of the EPID, and we forced the EPID dose profiles to agree with the data from the ionization chamber in a miniphantom. From Fig. 2(a) it is obvious that this kernel had to exhibit a blurring effect. For this purpose a Gaussian convolution kernel was chosen, as it is a commonly used and well-understood blur function, though other functions might work equally well. At the level of the EPID the agreement is excellent [see Fig. 2(b)]. The validity of the assumptions of our dose-reconstruction algorithm (as detailed in the Appendix) was tested after the back-projection into the midplane of the phantom (discussed below).

B. Midplane dose

After accounting for scatter from the phantom to the EPID, for attenuation of the beam by the phantom, and for scatter within the phantom, the midplane dose was reconstructed. Profiles of square fields back-projected from EPID images were compared with those measured with an ionization chamber that was located in a full-scatter water phantom to measure the total dose (see Fig. 4). When only kernel $K^{\text{EPID},1}$ was used to correct the EPID scatter, the penumbra of the EPID midplane dose profiles were steeper than those of the ionization chamber [see Fig. 4(a)]. This is reasonable considering the pixel size of the EPID (0.5 mm in the isocenter plane) and the inner diameter of the Semiflex ionization chamber (5 mm), limiting the resolution for profile measurements. However, the agreement improved by using a combination of two kernels to correct (mainly for the lateral scatter within) the EPID [see Fig. 4(b)]. The profiles of square fields down to $3 \times 3$ cm$^2$ were well reproduced with the EPID considering both absolute dose and shape.

We investigated the effect of using film instead of ionization chamber data on the reconstruction of a midplane dose profile (see Fig. 5). Also with film a steeper penumbra was measured than with the ionization chamber. Nevertheless, the overall disagreement between ionization chamber and film is quite small. If one reconstructs the dose from EPID images using only kernel $K^{\text{EPID},1}$, even a steeper penumbra is obtained with the EPID than with film—however, with obvious differences in shoulders and tails. Adjusting the EPID dose

Fig. 8. Histograms of the $\gamma$ indices of all IMRT fields for an area of 14 $\times$ 14 cm$^2$ encompassing each field. The histogram of field A corresponds to the $\gamma$ distribution shown in Fig. 7(c).
reconstruction to the film data with kernel $K_{\text{EPID},2}$ improved the agreement at the cost of the penumbra region. This is the result of the fitting procedure: fewer points contribute in the penumbra region relative to shoulders and tails. The parameters for $K_{\text{EPID},2}$ were very similar to those of kernel $K_{\text{EPID},2}$ determined with the ionization chamber reference (see Fig. 3). Therefore, the respective midplane profiles are virtually indistinguishable.

Although the pixel size of the EPID is effectively 0.5 mm in the isocenter plane, the reference data for the EPID profiles were measured with a small ionization chamber (Semi-flex) with an inner diameter of the measuring volume of approximately 5 mm. Kernel $K_{\text{EPID},2}$ was based on these reference data, so in practice the resolution of the reconstructed EPID images is approximately 5 mm. We are aware of the limitations of this ionization chamber to measure dose in steep dose gradient regions such as in the penumbra. Nevertheless, this ionization chamber was our reference detector of choice for the measurement of the reference profiles, because this ionization chamber is also used in our hospital to collect the data for the commissioning of our TPS. Therefore, all planned dose distributions will have this “ionization chamber effect.” In a clinical pretreatment situation, one wants to verify that the delivered dose agrees with the planned dose, i.e., we aim for consistency between measurements and calculations. Therefore, in our opinion, fitting kernel $K_{\text{EPID},2}$ in the described way is a reasonable approach. If one chooses film as a reference detector, the procedure can be adopted as described in Sec. II B.

The de facto gold standard for two-dimensional dosimetry in many institutions is radiographic film. With film the dose inside a phantom can be measured independently. The film was scanned with a resolution of 0.1 mm. The scanned film images were smoothed using a running average filter with a size of $0.5 \times 0.5$ mm$^2$. This procedure removes noise from the film images, which is important since the $\gamma$ distribution would be underestimated with noise in the dose distribution. In Fig. 6, by comparing the dose distributions from EPID and film for a $10 \times 10$ cm$^2$ field, we demonstrated that our back-projection method agreed very well with film dosimetry.

C. IMRT verification

As a test for a clinical pretreatment verification situation, an IMRT plan was delivered to the homogenous slab phantom and the EPID reconstruction was compared to film dosimetry. Again, excellent agreement was obtained for all fields (see Figs. 7 and 8). In our hospital, 3% and 3 mm are currently used as criteria for the $\gamma$ evaluation for pretreatment verification of IMRT prostate plans. We like to emphasize that the $10 \times 10$ cm$^2$ field and the five IMRT fields completely satisfied those criteria. In this paper, however, we have chosen to use 2% and 2 mm to assess the limitations of our back-projection method (one can easily “translate” from 2%/2 mm to 3%/3 mm by scaling the $\gamma$ index with a factor of 1.5). Even for 2%/2 mm criteria, the agreement is excellent, with only a very small percentage of points not satisfying those criteria.

The $\gamma$ histogram for the $10 \times 10$ cm$^2$ field shows a slightly worse distribution of $\gamma$ values than the histograms for the IMRT fields [Fig. 6(b) versus Fig. 8]. The main reason is the nature of the $\gamma$-evaluation method: combining a dose-difference with a distance-to-agreement criterion will usually result in smaller $\gamma$ indices for a modulated field compared to a more flat field. Note that this effect is intended in the $\gamma$-evaluation method, because points of two dose distributions are said to agree when at least one of the criteria is satisfied.

It can be argued that the film dose image should be smoothed with an “ionization chamber-like” response kernel. In that approach, one would exclude potential differences between EPID and film dose distributions that are due to resolution/averaging effects. We would like to note that when a broader averaging kernel is used for the film image (e.g., $5 \times 5$ mm$^2$, making the effective film resolution equal to the diameter of the ionization chamber), the dose distributions of EPID and film for the $10 \times 10$ cm$^2$ field and the five IMRT fields agree within the 2%/2 mm $\gamma$ criteria, but the $\gamma$ distributions improve only slightly due to the somewhat better agreement in the penumbra region. For the $14 \times 14$ cm$^2$ square region of interest, more than sufficiently encompassing each field, the percentages of pixels with $\gamma$ indices above unity and the maximum $\gamma$ indices (given in brackets) decrease to $0.05\%$ (1.02) for the $10 \times 10$ cm$^2$ field and $0.02\%$ (1.06), $0.0\%$ (0.92), $0.0\%$ (0.88), $0.09\%$ (1.23) for the five IMRT fields A, B, C, D, and E, respectively (compare to Sec. III B and Sec. III C). An opposite effect, however, is the noise reduction by the large kernel, which increases the $\gamma$ values. This is reflected by the increase of the mean $\gamma$ index from 0.34 to 0.39 for the $10 \times 10$ cm$^2$ field and from 0.31 to 0.38 on average for all IMRT fields.

The time required to perform a field-by-field dose verification of a five-field prostate plan with 37 segments as described in this study—either by EPID or by film—is estimated. Note that the original number of monitor units for each field was multiplied by a factor of 5 to reach optimal dose values for measurements with the EDR2 films. The delivery of those five fields takes approximately 15 min. Handling, developing, and scanning of the films take approximately 15 min, yielding 30 min for the whole film measurement and processing procedure. In case of a more sensitive film, for which the monitor unit scaling would not be necessary, we would need 3 min + 15 min = 18 min.

In a clinical pretreatment verification, the dose determined with the EPID is compared with the TPS dose calculation, and therefore the scaling of the monitor units can be omitted. However, for EPID dosimetry, the whole plan has to be delivered twice, because the images without the phantom in the beam also have to be acquired. This yields in total 2 $\times$ 3 min = 6 min for the EPID measurements. Moreover, with EPID dosimetry, the result is immediately available. This is particularly advantageous for in vivo dosimetry of a series of fractions, where the images without a patient in the beam
have to be acquired only once, which implies that if this is done prior to treatment, the result is immediately available after each fraction. Overall, EPID and film dosimetry take approximately 10 min and (at least) 20 min, respectively, for a field-by-field dose verification of the five-field prostate plan.

The back-projection algorithm enables accurate, simple, and potentially fast field-by-field IMRT pretreatment verification inside a phantom. Therefore, the EPID can replace film for this purpose. In the future it might become more important to have an alternative method to film dosimetry, as the processing facilities for radiographic films in many hospitals may disappear because of the digitization of radiography.

Due to the excellent agreement between reconstructed and actual dose values in the phantom, our EPID dosimetry method could potentially be extended to in vivo verification. For this application, however, the position and geometry of the patient should also be known. The accuracy of two-dimensional dose reconstruction considerably improves when contour information is used for the attenuation correction. This information can usually be obtained from the planning CT scan.20

So far no inhomogeneity corrections are implemented in the model. In general, this will lead to lower accuracy in the dose reconstruction for situations where inhomogeneities, such as air cavities, are present, for example in an anthropomorphic phantom or patient. However, under certain circumstances, the dose can be reconstructed accurately (in three dimensions) despite the inhomogeneity, for instance in the case of two (almost) opposing beams for a breast cancer treatment.20

As for any quality assurance protocol, it has to be decided in each institute whether EPID dosimetry is necessary and—if yes—for what purpose it should be used. This will determine the required accuracy, and hence complexity, of the dose-reconstruction method. In our opinion, EPID dosimetry as described in this work is valuable for finding the dose-reconstruction method. In our opinion, EPID dosimetry can replace film dosimetry in three-dimensional verification of IMRT fields inside phantoms.

V. CONCLUSIONS

We have shown that the improved back-projection algorithm can be applied to an aSi EPID and provides an accurate method to verify the dose of IMRT fields in two dimensions inside a homogeneous slab phantom. The algorithm performs well both in the center of a field (target volume), in the penumbra(s), and in the tails of the dose distributions. This is important for IMRT treatments with potentially many steep dose gradients and for situations where organs at risk are located close to the target volume. The EPID is an accurate and potentially fast alternative to film for field-by-field pretreatment verification of IMRT inside a phantom.

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APPENDIX: DESCRIPTION OF THE BACK-PROJECTION ALGORITHM

This appendix summarizes the details of the back-projection algorithm. Most of the elucidations and equations can also be found in Refs. 19, 20, and 42 (and references therein). They are given here for a complete description with our extensions for the benefit of the reader. We will start with the description of the back-projection algorithm after the image calibration,7,26 and the sensitivity matrix correction;29 these issues will not be discussed. When we mention in this appendix patient, this also refers to phantom.
1. Dose-response function

The dose $D_{ij}^{\text{EPID}}$ at a certain pixel $ij$ of the EPID consists of two parts: the portal dose $PD_{ij}^{\text{EPID}}$ of radiation reaching the EPID directly and the scatter part $Sc_{ij}^{\text{EPID}}$ of (lateral) scatter within the EPID. Therefore,

$$D_{ij}^{\text{EPID}} = PD_{ij}^{\text{EPID}} + Sc_{ij}^{\text{EPID}}. \quad (A1)$$

In order to measure dose with the EPID, we first have to relate dose values with pixel values, i.e., we have to determine the dose-response function, $f_{\text{DR}}$, of the EPID

$$PV_{ij}^{\text{EPID}} = f_{\text{DR}}(D_{ij}^{\text{EPID}}). \quad (A2)$$

where $PV_{ij}^{\text{EPID}}$ is the time-integrated pixel value at a certain pixel $ij$ of the EPID after field-foam, (dynamic) dark-field, and sensitivity corrections. We assume that the dose-response function $f_{\text{DR}}$ is equal for all pixels of the EPID. (In order to keep the notation general, we use $f_{\text{DR}}$, instead of an explicit linear function for sA EPIDs.)

Note that $D_{ij}^{\text{EPID}}$ [see Eq. (A1)] cannot directly be obtained nor can a reference value be measured to determine $f_{\text{DR}}$ absolutely. Instead, we use an ionization chamber in a miniphantom to measure the dose in air on the central axis (CAX), providing a reference value $PD_{\text{CAX}}^{\text{EPID}}$ for the portal dose $PD_{ij}^{\text{EPID}}$ [see Eq. (A1)]. This is done for the reference field size (typically $10 \times 10 \text{ cm}^2$) at the SDD of the EPID for several numbers of monitor units without a phantom in the beam. Under the same conditions, EPID images are acquired. Therefore, we can determine the relative dose-response function, $f_{\text{DR}}^{\text{rel},10\times10\text{ cm}^2}$, by fitting a function to relate EPID pixel and dose values

$$\langle PV_{ij}^{\text{EPID}} \rangle_{\text{CAX}} = f_{\text{DR}}^{\text{rel},10\times10\text{ cm}^2}(PD_{ij}^{\text{CAX}}). \quad (A3)$$

The data are determined as a function of the number of monitor units. The brackets $\langle \cdot \rangle_{\text{CAX}}$ represent the average over a small central region of interest (cROI) of the EPID at the central axis. Typically, the cROI is defined as a region around the central axis of approximately 0.5 × 0.5 cm² (projected into the isocenter plane).

2. Scatter within the EPID

In our algorithm, the scatter component is modeled as the convolution of the portal dose with a scatter kernel $K_{ij}^{\text{EPID}}$

$$Sc_{ij}^{\text{EPID}} = PD_{ij}^{\text{EPID}} \otimes K_{ij}^{\text{EPID}}. \quad (A4)$$

Therefore, with Eq. (A1)

$$D_{ij}^{\text{EPID}} = PD_{ij}^{\text{EPID}} + PD_{ij}^{\text{EPID}} \otimes K_{ij}^{\text{EPID}}$$

$$= PD_{ij}^{\text{EPID}} \otimes [\delta(r_{ij}) + K_{ij}^{\text{EPID}}]$$

$$= PD_{ij}^{\text{EPID}} \otimes K_{ij}^{\text{EPID},1}, \quad (A5)$$

with

$$K_{ij}^{\text{EPID},1} = \delta(r_{ij}) + K_{ij}^{\text{EPID}}, \quad (A6)$$

where $\delta(r_{ij})$ represents the $\delta$ function and $r_{ij}$ is the distance of a pixel $ij$ from the central axis. As the (primary) portal dose at the position of the EPID is needed for our back-projection algorithm, we have to deconvolve the dose image with the scatter kernel,

$$PD_{ij}^{\text{EPID}} = D_{ij}^{\text{EPID}} \otimes K_{ij}^{\text{EPID},1}. \quad (A7)$$

We introduce the scatter kernel $K_{ij}^{\text{EPID},1}$ with the following model:

$$K_{ij}^{\text{EPID},1} = c_{\text{DR}} \begin{cases} c_1 \cdot e^{-\mu_1 r_{ij}} & \text{for } r_{ij} \neq 0 \\ 1 & \text{for } r_{ij} = 0, \end{cases} \quad (A8)$$

where $c_1$, $\mu_1$, and $c_{\text{DR}}$ are the kernel parameters; $c_{\text{DR}}$ is a dose-response correction factor for the EPID, which has to be introduced because in Eq. (A7) it is assumed that the dose $D_{ij}^{\text{EPID}}$ has been determined absolutely, which is not the case (see Sec. I of this appendix). The $\delta$ function from Eq. (A6) is implemented here as the term for $r_{ij} = 0$.

To experimentally derive the parameters $c_1$, $\mu_1$, and $c_{\text{DR}}$ for the EPID scatter kernel $K_{ij}^{\text{EPID},1}$, square fields of different size are measured with the EPID without a phantom in the beam and averaged over the central region of interest. Then, the dose is measured in air on the central axis with an ionization chamber in a miniphantom to the same field sizes at the same SDD. These ionization chamber measurements are then used as reference data for the portal dose $PD$:

$$\langle PD_{ij}^{\text{EPID}}(fs) \rangle_{\text{CAX}} = PD_{\text{CAX}}^{\text{EPID}}(fs). \quad (A9)$$

The data are determined as a function of field size $fs$. By adjusting the kernel parameters $c_1$, $\mu_1$, and $c_{\text{DR}}$ [see Eq. (A8)], the portal dose $PD_{ij}^{\text{EPID}}$ [see Eq. (A7)] is fit to the reference data, and the kernel $K_{ij}^{\text{EPID},1}$ is obtained for a specific EPID-photon beam energy combination.

The portal dose image $PD_{ij}^{\text{EPID}}$ can now be calculated for any field size and shape. However, for an accurate two-dimensional dose reconstruction within a patient with our back-projection model, we do need an accurate description of the portal dose image over the whole field, including the penumbra region. To assess and improve the accuracy in two dimensions, profiles are also measured with the ionization chamber in the miniphantom under the same conditions using the scanning mechanism of a water phantom. As the profiles from EPID portal dose images are steeper in the penumbra region than the ionization chamber profiles, the portal dose image is convolved with a second kernel $K_{ij}^{\text{EPID},2}$.

For this purpose, a Gaussian is chosen as a commonly used and well-understood blur function, though other functions may work equally as well

$$PD_{ij}^{\text{EPID}} = D_{ij}^{\text{EPID}} \otimes K_{ij}^{\text{EPID},1} \otimes K_{ij}^{\text{EPID},2}, \quad (A10)$$

with

$$K_{ij}^{\text{EPID},2} = \frac{c_2}{2\pi\sigma^2} \cdot e^{-r_{ij}^2/2\sigma^2}, \quad (A11)$$

with the kernel parameters $c_2$ and $\sigma$. For fitting the parameters of $K_{ij}^{\text{EPID},2}$ a suitable method for comparing EPID with ionization chamber dose profiles has to be used (e.g., the $\gamma$-evaluation method).
Instead of profile measurements with the ionization chamber in a miniphantom at the EPID level, film measurements under full-scatter conditions (at 10 cm depth in a 20 cm thick polystyrene slab phantom with SSD = 90 cm) can be used as an alternative reference to estimate the parameters for the second EPID kernel $K^{\text{EPID,2}}$. The parameters of this “film-adjusted” second EPID kernel, $K^{\text{EPID,2}}$, are determined by fitting the reconstructed EPID midplane dose profile(s) to the corresponding film measurement(s).

3. Scatter from patient to EPID

The portal dose image $P_{\text{D,EPID}}$ behind a patient includes a component $S_{\text{patient}} \rightarrow \text{EPID}$ due to scatter from the patient to the EPID,

$$ P_{\text{ij,EPID}} = P_{\text{ij}} + S_{\text{patient}} \rightarrow \text{EPID}. \quad (A12) $$

$P_{\text{ij,EPID}}$ is the primary portal dose, which results from radiation coming directly from the radiation head of the accelerator, $\text{Sc}^{\text{EPID}}$, and is required for our back-projection method. The primary portal dose is obtained by subtracting the scatter contribution from the portal dose,

$$ P_{\text{ij,EPID}} = P_{\text{ij}} = P_{\text{ij}} \rightarrow \text{EPID} - S_{\text{patient}} \rightarrow \text{EPID}. \quad (A13) $$

In order to estimate the scatter contribution, we first calculate the total transmission image $\gamma_{\text{total}}$ by dividing the portal dose image with a patient in the beam by the portal dose image without a patient in the beam and use Eq. (A12) for the separation into a primary and a scatter contribution,

$$ \gamma_{\text{total}} = \frac{P_{\text{ij,EPID}} \rightarrow \text{EPID,with patient}}{P_{\text{ij,EPID}} \rightarrow \text{EPID,without patient}} = \frac{P_{\text{ij}} + S_{\text{patient}} \rightarrow \text{EPID}}{P_{\text{ij}} \rightarrow \text{EPID,without patient}} = \gamma_{\text{primary}} + \frac{S_{\text{patient}} \rightarrow \text{EPID}}{P_{\text{ij}} \rightarrow \text{EPID,without patient}} \quad (A14) $$

where $\gamma_{\text{primary}}$ is the “primary transmission,” i.e., the transmission, when no scatter from the patient would reach the EPID, and is given by

$$ \gamma_{\text{primary}} = \frac{P_{\text{ij}}}{P_{\text{ij}} \rightarrow \text{EPID,without patient}}. \quad (A15) $$

Note that $\gamma_{\text{total}}$ depends on field size, but $\gamma_{\text{primary}}$ has to be field size independent by definition. This fact is used to assess the scatter contribution $S_{\text{patient}} \rightarrow \text{EPID}$ as follows. The total transmission $\gamma_{\text{total}}$ is experimentally determined as a function of field size $f$s, by irradiating a phantom of “reference thickness” (typically 20 cm) with square fields of different size. For very small field sizes, there is negligible scatter from the phantom to the EPID. Thus, in the limit of zero field size, the total transmission $\gamma_{\text{total}}$ equals the primary transmission $\gamma_{\text{primary}}$. The values of the total transmission $\gamma_{\text{total}}$ averaged over the cROI, $\langle \gamma_{\text{total}} \rangle_{\text{cROI}}$, are plotted as a function of field area, $f^2$, and extrapolated to zero field size. By adjusting $S_{\text{patient}} \rightarrow \text{EPID}$ in Eq. (A15) using a model explained below, $\gamma_{\text{primary}}$ at the cROI is fit for all field sizes to the zero-field-size limit of $\gamma_{\text{total}}$ at the cROI,

$$ \langle \gamma_{\text{total}} \rangle_{\text{cROI}} \rightarrow 0 = \lim_{f^2 \to 0} \langle \gamma_{\text{total}} \rangle_{\text{cROI}} \rightarrow 0, \quad (A16) $$

i.e., the right side of this equation is a constant and the left side has to be made equal to that constant for all field sizes to achieve the field size independence of $\gamma_{\text{primary}}$. The field area is used as the independent variable instead of the field size, because the scatter-to-primary ratio for the scatter from the phantom to the EPID, and therefore also $\gamma_{\text{total}}$ see Eq. (A14), increases for small field sizes and for large air gaps approximately linearly with the field area. For our data, a second-order polynomial appeared sufficient for a reasonable fit of $\langle \gamma_{\text{total}} \rangle_{\text{cROI}}$ and the extrapolation to zero field size.

The scatter from the patient to the EPID is modeled as a convolution of the portal dose image with a scatter kernel $K_{\text{patient}} \rightarrow \text{EPID}$,

$$ S_{\text{ij}} \rightarrow \text{EPID} = P_{\text{ij}, \text{EPID,with patient}} \otimes K_{\text{patient}} \rightarrow \text{EPID}. \quad (A17) $$

In principle, the primary portal dose $P_{\text{ij,EPID}}$ should be used here instead of the portal dose $P_{\text{D,EPID,with patient}}$, suggesting an iterative approach. However, for relatively large air gaps and a copper plate filtering (scattered) low-energy photons (as in our setup), the scatter contribution is small and the portal dose itself can be used in good approximation. The parts of the image without the patient in the beam are excluded from the scatter computation. As kernel we choose a constant,

$$ K_{\text{ij}} \rightarrow \text{EPID} = \epsilon_{\text{patient}} \rightarrow \text{EPID}. \quad (A18) $$

Therefore, with Eq. (A17),

$$ S_{\text{ij}} \rightarrow \text{EPID} = \left( \sum_{i,j} P_{\text{D,ij,EPID,with patient}} \right) \cdot \epsilon_{\text{patient}} \rightarrow \text{EPID}. \quad (A19) $$

The summation runs over all pixels of the imager (except for the parts without the patient in the beam). Thus, in our model the scatter component $S_{\text{patient}} \rightarrow \text{EPID}$ is a constant offset over the whole EPID image. This homogeneous scatter component is proportional to the portal dose $P_{\text{D,EPID,with patient}}$ integrated over the whole EPID image, and therefore approximately proportional to both the portal dose on the central axis and the field area. There is no dependence of the scatter on the thickness of the patient in our model.

In the analytical scatter model of Swindell and Evans, the scatter-to-primary ratio is for vanishingly small scattering volumes proportional to the irradiated volume, i.e., to the field area and the thickness of the patient or phantom. In our model, the dependence on field area is described correctly, but this is not the case for the dependence on thickness. In order to obtain the parameter $\epsilon_{\text{patient}} \rightarrow \text{EPID}$ for the scatter kernel $K_{\text{patient}} \rightarrow \text{EPID}$, the required data ($\gamma_{\text{total}}$ for several field sizes) are measured at the so-called reference thickness, for which a representative thickness of a clinical situation for the considered beam energy is chosen. This is certainly an approximation. However, under our conditions with small scatter-
ter contribution [due to relatively large air gaps and a copper plate filtering (scattered) low-energy photons], we found that the errors made here do not hamper an accurate back-projection into the midplane (see also Sec. 5 of this appendix).

Other models for the kernel, such as a Gaussian or a Lorentzian, have also been tested and can improve the fit slightly. More rigorous treatments of scatter from a patient or a phantom to the EPID, especially for smaller air gaps, are described in the literature.43–49

4. Scatter within the patient

The total (reconstructed) dose \( D_{ij}^{\text{mid}} \) in the patient consists of two parts: the primary dose \( P_{ij}^{\text{mid}} \) and the scattered dose \( S_{ij}^{\text{mid}} \) within the patient,

\[
D_{ij}^{\text{mid}} = P_{ij}^{\text{mid}} + S_{ij}^{\text{mid}}. \quad (A20)
\]

We choose to reconstruct the dose in the radiological mid-surface, which is defined as the surface connecting those points in the patient, where the transmission of the patient above and below that surface is equal for rays emerging from the radiation source in the direction of the portal imager. To calculate the primary dose in the radiological mid-surface of the patient, \( P_{ij}^{\text{mid,radial}} \), the inverse square law ISQL and an attenuation correction \( AC \) are used. For the attenuation we assume an exponential function \( \exp(-\mu_{AC} \cdot r_{\text{radial}}) \) for the primary transmission \( T_{ij}^{\text{primary}} \) [see Eqs. (A14) and (A15)], where \( \mu_{AC} \) represents the linear attenuation coefficient of water for a specific beam energy and \( r_{\text{radial}} \) the radiological path length of a ray through the patient. In principle, a model of the beam’s energy spectrum should be used for \( \mu_{AC} \). However, with the above assumption, an effective attenuation correction can be calculated directly from the experimental primary transmission \( T_{ij}^{\text{primary}} \). The following equation is valid for the reconstruction of the primary dose in the radiological mid-surface:

\[
P_{ij}^{\text{mid,radial}} = P_{ij}^{\text{EPID}} \cdot \text{ISQL} \cdot AC
= P_{ij}^{\text{EPID}} \left( \frac{d_{\text{reconst}}}{d_{\text{EPID}}} \right)^{-2} \frac{1}{e^{-\mu_{AC} r_{\text{radial}}}}
= P_{ij}^{\text{EPID}} \left( \frac{d_{\text{reconst}}}{d_{\text{EPID}}} \right)^{-2} \frac{1}{\sqrt{T_{ij}^{\text{primary}}}} \quad (A21)
\]

where \( d_{\text{EPID}} \) is the distance of the EPID (“imaging” layer) from the accelerator target and equals 160 cm for our EPID; \( d_{\text{reconst}} \) is the distance of the reconstruction surface from the accelerator target. In our study, with a homogeneous slab geometry phantom aligned symmetrically around the isocenter plane, the surface of reconstruction is always the isocenter plane, hence the term “midplane,” at 100 cm from the target, i.e., \( d_{\text{reconst}} = 100 \text{ cm} \); \( r_{\text{radial}} \) is the radiological thickness of the patient determined at pixel \( ij \).

For dose reconstruction in an arbitrary geometrical plane parallel to the EPID, the attenuation has to be corrected for the radiological path length \( f_{ij}^{\text{const}–\text{exit,radial}} \) along a ray through the patient from the reconstruction plane down to the exit surface, determined at pixel \( ij \). The primary transmission for this path is then given by \( (T_{ij}^{\text{primary}} \cdot f_{ij}^{\text{const}–\text{exit,radial}})^{-1} \). The ratio of the radiological pathlengths in the exponent is approximated by the ratio of the geometrical pathlengths \( f_{ij}^{\text{const}–\text{exit,geom}} \) by using three-dimensional contour information from the patient, e.g., from the CT scan;20 note that for a homogeneous phantom, those ratios are equal. Therefore, for dose reconstruction in an arbitrary geometrical plane,

\[
P_{ij}^{\text{mid,geom}} = P_{ij}^{\text{EPID}} \cdot \left( \frac{d_{\text{reconst}}}{d_{\text{EPID}}} \right)^{-2} \frac{1}{(T_{ij}^{\text{primary}} \cdot f_{ij}^{\text{const}–\text{exit,geom}})^{-1}}. \quad (A22)
\]

The calculation of the scatter component \( S_{ij}^{\text{mid}} \) within the reconstruction plane is in our model separated into a thickness and a field size dependence. First, the midplane dose is weighted with the \( SPR^{\text{ref}} \), the scatter-to-primary ratio determined under reference conditions (see below). [Note that in Ref. 20, the term normalized scatter-to-primary ratio, NSPR, was used.] The \( SPR^{\text{ref}} \) is a function of the primary transmission \( T_{ij}^{\text{primary}} \), and accounts for the amount of scatter which depends on the (radiological) thickness of the patient. Then, the result is convolved with the scatter kernel \( K_{ij}^{\text{mid}} \), accounting for the field-size dependence of the scattered dose distribution in the reconstruction plane,

\[
S_{ij}^{\text{mid}} = [P_{ij}^{\text{mid}} \cdot SPR^{\text{ref}}(T_{ij}^{\text{primary}})] \otimes K_{ij}^{\text{mid}}. \quad (A23)
\]

In order to parametrize \( SPR^{\text{ref}} \), it is first obtained experimentally and then fit as a function of \( T_{ij}^{\text{primary}} \). The dose is measured with an ionization chamber at the isocenter in iso-centrically aligned slab phantoms of several thicknesses at the reference field size (typically 10 × 10 cm²); the ionization chamber measurements are used as the reference values for the total dose \( D \) on the central axis. EPID measurements are performed with and without the phantom and are used to calculate the primary transmission, \( T_{ij}^{\text{primary}} \) [see Eq. (A15)], and the primary mid-surface/midplane dose, \( P_{ij}^{\text{mid}} \) [see Eqs. (A21) and (A22)], in the central region of interest. The experimental scatter-to-primary ratio \( SPR^{\text{exp}} \) is then calculated for every phantom thickness as

\[
SPR^{\text{exp}} = \frac{D_{\text{exp}}^{\text{mid}} - (P_{ij}^{\text{mid}} \cdot CAX)}{(P_{ij}^{\text{mid}} \cdot CAX)} , \quad (A24)
\]

and then fit to a polynomial function, \( SPR^{\text{ref}} \), of the primary transmission \( T_{ij}^{\text{primary}} \)

\[
SPR^{\text{ref}}(T_{ij}^{\text{primary}}) = SPR^{\text{exp}}(T_{ij}^{\text{primary}}). \quad (A25)
\]

For our data a third-order polynomial was used. Thus, from the primary transmission at every pixel \( ij \), the scatter-to-primary ratio can be calculated for every pixel \( ij \) by using the “\( SPR^{\text{ref}} \) polynomial.” In principle, the scatter-to-primary ratio depends on depth. In our current model, however, \( SPR^{\text{ref}} \) only depends on the primary transmission \( T_{ij}^{\text{primary}} \) and is
therefore independent of depth (see Sec. 5 of this appendix).

As a model for the scatter kernel \( K_{\text{mid}} \) [see Eq. (A23)], we use

\[
K_{ij}^\text{mid} = c_{\text{mid}} e^{-\mu_{\text{mid}} r_{ij}} \frac{r_{ij}}{r_{ij}^2 + \Delta^2},
\]

(A26)

where \( c_{\text{mid}}, \mu_{\text{mid}}, \) and \( \Delta \) are the parameters of kernel \( K_{\text{mid}} \). The parameter \( \Delta \) was introduced to prevent division by zero for \( r_{ij} = 0 \); this a slight modification from Ref. 20. Here, \( c_{\text{mid}} \) is a scaling factor, \( \mu_{\text{mid}} \) is the linear attenuation coefficient for water, which depends on energy (and therefore indirectly on patient thickness and off-axis distance). Because there is a whole distribution of photon energies within the patient, both from the primary and the scattered radiation, \( \mu_{\text{mid}} \) will be the effective attenuation coefficient for the photon energy spectrum in the midsurface or midplane of the patient. Instead of using a model for the beam's energy spectrum, we decided to treat \( \mu_{\text{mid}} \) as a free parameter in our fit procedure. Beam hardening is not directly taken into account.

The midplane scatter kernel \( K_{\text{mid}} \) is obtained by measuring the dose with an ionization chamber in a phantom and changing the field size \( f_{\text{s}} \) for the reference phantom thickness (typically 20 cm). EPID measurements are performed for the same field sizes with and without the phantom in the beam. By adjusting the kernel parameters \( c_{\text{mid}}, \mu_{\text{mid}}, \) and \( \Delta \) for kernel \( K_{\text{mid}} \), the EPID midplane dose \( D_{ij}^\text{mid}(r_{ij}) \) is fit to the ionization chamber measurements \( D_{\text{CAM}} \) on the central axis in the phantom,

\[
(D_{ij}(f_{\text{s}}))_{\text{ROIs}} = D_{\text{CAM}}(f_{\text{s}}).
\]

(A27)

5. Summary

The cross calibration of the EPID is performed (i) against an ionization chamber in a miniphantom at the level of the EPID for several field sizes and (ii) against an ionization chamber in a slab phantom for several field sizes at the reference thickness and for several thicknesses at the reference field size. The (reference) field sizes and phantom thicknesses were chosen to be representative of clinical situations. The effects of field size and patient/phantom transmission are separated in the model. Errors/shortcomings in the determination of the parameters in an earlier step can be compensated later in the procedure. For example, errors in the field-size dependence of the scatter from the phantom to the EPID might be compensated during the cross calibration of the field-size dependence of the total midplane dose.

The model is “as physical as possible.” However, it is primarily designed to describe the measured data for the clinically relevant range of field sizes and thicknesses and to accurately predict them. In the end, the accuracy in the phantom after the dose reconstruction is most relevant. For this study, the agreement on the central axis in the phantom between EPID dose reconstruction and reference measurements was better than 1% for all data used to estimate the parameters.

The dose reconstruction for one segment of a treatment field can be summarized as follows:

(i) take an EPID image with and without a patient and convert the pixel values to relative dose values with the inverse relative dose-response function;
(ii) calculate the portal dose \( PD_{ij}^\text{EPID} \) [Eq. (A10)] for both images;
(iii) calculate the primary portal dose \( P_{ij}^\text{EPID} \) [Eqs. (A13) and (A19)];
(iv) calculate the primary transmission \( T_{ij}^\text{primary} \) [Eq. (A15)];
(v) calculate the primary dose in the radiological mid-surface, \( P_{ij}^\text{primary,rad} \) [Eq. (A21)] or in the geometrical midplane, \( P_{ij}^\text{primary,geom} \) [Eq. (A22)];
(vi) calculate the scatter-to-primary ratio for the scatter within the patient from the primary transmission \( T_{ij} \) and the \( \text{SPR}_{\text{ref}} \) polynomial;
(vii) calculate the scattered midplane dose [Eq. (A23)]; and
(viii) calculate the total midplane dose [Eq. (A20)].

By reconstructing the dose in different planes from the target, i.e., by changing \( d_{\text{con}} \) and \( p_{\text{con}} \) [see Eq. (A22)], for each gantry angle and finally adding them up, a three-dimensional dose distribution can be obtained for simple fields.20 It should further be noted that currently no inhomogeneity correction is implemented in our back-projection algorithm. These issues are part of future investigations.

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