A simple backprojection algorithm for 3D in vivo EPID dosimetry of IMRT treatments

Markus Wendling, a) Leah N. McDermott, b) Anton Mans, Jan-Jakob Sonke, Marcel van Herk, and Ben J. Mijnheer

Department of Radiation Oncology, The Netherlands Cancer Institute–Antoni van Leeuwenhoek Hospital, Plesmanlaan 121, 1066 CX Amsterdam, The Netherlands

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Treatment plans are usually designed, optimized, and evaluated based on the total 3D dose distribution, motivating a total 3D dose verification. The purpose of this study was to develop a 2D transmission-dosimetry method using an electronic portal imaging device (EPID) into a simple 3D method that provides 3D dose information. In the new method, the dose is reconstructed within the patient volume in multiple planes parallel to the EPID for each gantry angle. By summing the 3D dose grids of all beams, the 3D dose distribution for the total treatment fraction is obtained. The algorithm uses patient contours from the planning CT scan but does not include tissue inhomogeneity corrections. The 3D EPID dosimetry method was tested for IMRT fractions of a prostate, a rectum, and a head-and-neck cancer patient. Planned and in vivo-measured dose distributions were within 2% at the dose prescription point. Within the 50% isodose surface of the prescribed dose, at least 97% of points were in agreement, evaluated with a 3D γ method with criteria of 3% of the prescribed dose and 0.3 cm. Full 3D dose reconstruction on a 0.1×0.1×0.1 cm³ grid and 3D γ evaluation took less than 15 min for one fraction on a standard PC. The method allows in vivo determination of 3D dose-volume parameters that are common in clinical practice. The authors conclude that their EPID dosimetry method is an accurate and fast tool for in vivo dose verification of IMRT plans in 3D. Their approach is independent of the treatment planning system and provides a practical safety net for radiotherapy. © 2009 American Association of Physicists in Medicine. [DOI: 10.1118/1.3148482]

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I. INTRODUCTION

Due to the increasing complexity of nearly all steps in radiotherapy today, the demand for thorough and efficient verification of the dose delivered to the patient has also increased, either pretreatment or in vivo.1 Electronic portal imaging devices (EPIDs) provide an elegant solution for dosimetry when they are readily available on treatment machines for the purpose of position verification of the patient.

EPID dosimetry can be performed as transmission or non-transmission dosimetry, i.e., with or without an attenuating object (patient or phantom) in the beam. For both methods, we can distinguish between the forward approach, i.e., the EPID image is used to determine the fluence or dose at the position of the imager, or the backprojection/dose-reconstruction approach, i.e., the EPID image is used to estimate the dose in the patient or phantom. 7–18 For treatment verification, the EPID image, (energy) fluence, or dose is compared to the dose calculation of a treatment planning system (TPS) or another independent algorithm. This can be done in a point, plane [two dimensional (2D)], or volume (3D); for a recent review on EPID dosimetry see Ref. 19.

We described earlier a backprojection algorithm for transmission EPID dosimetry using (de)convolution methods, which can be used to estimate the dose in a phantom or patient in a plane parallel to the EPID, usually intersecting the isocenter. 16 Compared to pretreatment checks, in vivo dosimetry has the advantage that the dose actually delivered to the patient is verified. Currently, 2D in vivo EPID dosimetry is implemented in our hospital as the standard verification method for all intensity-modulated radiotherapy (IMRT) cancer treatments. 20–23 The exceptions are treatments with beam sizes that exceed a 25×25 cm² x-y square (because then the EPID electronics would be irradiated, leading to a decrease in the lifetime of the detector 24 ) and single-fraction treatments of the brain (because no intervention would be possible following an in vivo check); in these cases pretreatment EPID dosimetry is performed. For large beam sizes, the collimator jaws are moved inward for each segment (if necessary) so that the total adapted beam just fits into a 23×23 cm² x-y square. It has been shown that with the backprojection EPID dosimetry approach, dose reconstruction in 3D is possible for conformal treatments of breast cancer with tangential fields.11 Clinically, our original backprojection algorithm is used for 2D dose reconstruction; this verification is thus performed separately for each beam in 2D. However, treatment plans are usually designed, optimized, and evaluated based on the total 3D dose distribution using dose-volume parameters. When the delivered total dose distribution could accurately be reconstructed in 3D, then the delivered plan could
be assessed as a whole using, for example, the same dose-volume parameters as during planning. The purpose of this study was to investigate 3D dose reconstruction based on our simple backprojection algorithm using EPID transmission measurements and its use for 3D dose verification of IMRT treatments of several cancer sites. The model, which is independent of the TPS, was adapted on several points for higher accuracy based on physics principles without the need for additional calibration measurements. Clinical examples of a prostate, a rectum, and a head-and-neck cancer case were used to test our algorithm, both in a phantom and in vivo.

II. MATERIALS AND METHODS

II.A. Accelerator, EPID, and image acquisition

Measurements were performed on SL20i linear accelerators (Elekta, Crawley, UK) using 6, 10, and 18 MV photon beams. All linacs were equipped with a multileaf collimator (MLC) and a PerkinElmer RID 1680 AL5/Elekta iViewGT amorphous silicon EPID. On the EPIDs, an extra 2.5 mm thick copper plate was used as additional build-up for all measurements. Images were acquired using in-house developed software16,25 (a similar image acquisition is possible with the commercially available ELEKTA software), allowing the frame-averaged image and the respective number of frames to be retrieved. The EPID images were recorded at a resolution of 512 x 512 pixels and then resampled at 256 x 256 pixels for practical purposes, yielding an effective pixel size of 0.1 cm at the isocenter plane. Further details about the EPID, image acquisition, and processing can be found in Ref. 16. However, in contrast to the procedure outlined in Ref. 16 and the commercially available ELEKTA software, images were not acquired per segment but per field (i.e., frames were continuously acquired for each field from the start of the first segment until the end of the last segment—also between segments) using the iCom network protocol of the ELEKTA linac for triggering. These changes were introduced to adapt the image acquisition to high-speed segment delivery of linacs available today.

II.B. Calibration of the EPID and the backprojection algorithm

For dose reconstruction, we used a backprojection algorithm, which was described in detail earlier and which employs a measurement-based correction approach.16 The necessary parameters were determined by acquiring EPID images of square fields of several sizes, with and without a homogeneous polystyrene slab phantom of several thicknesses in the beam; the phantoms were centered at the isocenter. Ionization chamber measurements were used as reference data to fit the model parameters. As the set of parameters was determined for homogeneous situations, it is a “water-equivalent” model. The algorithm uses the external phantom or patient contours, which were taken from the planning CT scan in this study but does not include tissue inhomogeneity corrections.

For every image behind the phantom or patient, one additional image is required without the phantom or patient in the beam to estimate the transmission of the phantom or patient. The transmission is needed in our backprojection algorithm [see Sec. II C, Eq. (3)]. These images are acquired once per patient or phantom outside clinical hours or during an extra time slot, for instance, just before the first treatment fraction. The time necessary to perform these measurements is approximately equal to the treatment time of one IMRT fraction, which is approximately 10 min. As we did not know the precise EPID position (within the plane of the EPID, including possible detector sagging with gantry rotation), we used the outline of the beam from the TPS to match the EPID image before the dose was reconstructed. The contour of each EPID image was determined with a radiation field edge detection algorithm using global threshold segmentation and maximum gradient search;26 the accuracy of this algorithm is limited by the pixel size, i.e., in this study 0.1 cm at the isocenter. For each contour, the center of weight was then shifted to the center of weight of the beam outline from the TPS. The matching was done separately for each acquired image.

II.C. Extension from 2D to 3D

With our previously developed backprojection algorithm, 2D dose reconstruction in a plane through the isocenter parallel to the EPID is possible.16 By reconstructing the dose within the patient or phantom volume in multiple planes parallel to the EPID, the 3D dose distribution can be obtained for each beam. In this section, we describe this 3D dose reconstruction per beam by presenting the relevant equations and explaining the refinements of the model. Finally, we describe how the total 3D dose distribution per fraction is calculated. We start from the primary portal dose at the level of the EPID; image calibration, sensitivity matrix correction, EPID response, corrections for scatter within the EPID and scatter from the patient to the EPID were not adapted and can be found in Ref. 16 and references therein. All model parameters were determined for each nominal beam energy of the linac. When we mention in this section patient, this also refers to phantom.

For accurate dose reconstruction in 3D, the model was extended by adding three ingredients: (i) Explicit modeling of beam hardening with depth, (ii) introduction of depth dependence of the scatter component of the dose in the patient, and (iii) explicit (empirical) modeling of the dose build-up effect at the beam entrance.

In our model, the dose is reconstructed within the patient volume in multiple planes parallel to the EPID for each gantry angle. The gantry angle as well as the position and the external contours of the patient are needed for the dose reconstruction but will not explicitly be mentioned as variables in the following paragraphs. For a quantity $X$, we use the notation $X_{ij}(d_{	ext{reconst}})$. Each index pair $ij$ corresponds to a pixel of the EPID and also describes a line from this pixel $ij$ to the accelerator target (geometrical backprojection line $ij$); $d_{	ext{reconst}}$ is the distance of the reconstruction plane, which is parallel to the EPID, from the accelerator target. So the index
pair \( ij \), together with the argument \( d_{\text{reconst}} \), describes the point within the reconstruction volume, where the geometrical backprojection line \( ij \) intersects the reconstruction plane at distance \( d_{\text{reconst}} \). This point has a radiological and geometrical depth, for which we use the shorthand notation \( d_{\text{recons}}^\text{radiol} = d_{ij}^\text{radiol}(d_{\text{reconst}}) \) and \( d_{ij}^\text{geom} = d_{ij}^\text{geom}(d_{\text{reconst}}) \), respectively. Note that this coordinate system is “fixed to the gantry.”

The (reconstructed) dose \( D \) consists of two parts: The primary dose \( P \) and the scattered dose \( S \) within the patient,

\[
D_{ij}(d_{\text{reconst}}) = P_{ij}(d_{\text{reconst}}) + S_{ij}(d_{\text{reconst}}).
\]

(1)

To calculate the primary dose \( P_{ij}(d_{\text{reconst}}) \) in the patient in the reconstruction plane at the distance \( d_{\text{reconst}} \) from the primary portal dose \( P_{\text{EPID}} \) at the EPID, the inverse square law and an attenuation correction \( AC \) are used,

\[
P_{ij}(d_{\text{reconst}}) = P_{ij}(d_{\text{EPID}}) \left( \frac{d_{\text{reconst}}}{d_{\text{EPID}}} \right)^{-2} \cdot AC_{ij}(d_{\text{reconst}}),
\]

(2)

where \( d_{\text{EPID}} \) is the distance of the EPID (“imaging” layer) from the accelerator target and equals 160 cm for our EPID. The term \( (d_{\text{reconst}}/d_{\text{EPID}})^{-2} \) describes the inverse square law; note that this scaling factor does not depend on indices \( ij \). The attenuation correction \( AC \) for a reconstruction plane at distance \( d_{\text{reconst}} \) is described in Sec. II.C.1.

II.C.1. Attenuation correction: Beam hardening with depth

In the backprojection algorithm, the attenuation of a ray between the reconstruction plane and the EPID has to be taken into account. The attenuation is governed by the radiological path length of a ray through the patient from the reconstruction plane to the exit surface. The total attenuation of a ray between the entrance and the exit planes can be determined from measurements. Therefore, the primary transmission \( T_{ij}^{\text{primary}} \) behind a patient is calculated as the ratio of the primary portal dose image with the patient, \( P_{\text{EPID}} \), and the portal dose image without the patient, \( PD_{\text{EPID}} \), in the beam,

\[
T_{ij}^{\text{primary}} = P_{ij}(d_{\text{EOID}}) / PD_{ij}^{\text{EPID, without patient}}.
\]

(3)

In the original 2D algorithm, an exponential function \( \exp(-\mu_{AC} \cdot d_{\text{radiol}}) \) for the primary transmission is assumed, where \( \mu_{AC} \) represents the linear attenuation coefficient of water for a specific beam energy and \( d_{\text{radiol}} \) is the radiological path length of a ray through the patient. This model for the primary transmission as a function of the radiological path length \( d_{ij}^\text{radiol} \) was now refined by introducing a beam hardening coefficient \( \sigma_{BH} \),

\[
\tilde{T}_{ij}^{\text{primary}}(d_{ij}^\text{radiol}) = \exp[-\mu_{AC} d_{ij}^\text{radiol} \cdot (1 - \eta_{BH} d_{ij}^\text{radiol})] = \exp[-\mu_{AC} d_{ij}^\text{radiol} + \sigma_{BH} \cdot (d_{ij}^\text{radiol})^2],
\]

(4)

with \( \sigma_{BH} = \mu_{AC} \eta_{BH} \); to distinguish the depth-dependent model function for the primary transmission from the experimentally determined primary transmission \( T_{ij}^{\text{primary}} \) in Eq. (3), we use the symbol \( \tilde{T}_{ij}^{\text{primary}} \). The parameter \( \eta_{BH} \) in Eq. (4) accounts for a decrease in the linear attenuation coefficient with depth: Thus \( \eta_{BH} \) and \( \sigma_{BH} \) describe beam hardening with depth.\(^{27} \)

We obtained the beam hardening coefficient \( \sigma_{BH} \) from the dosimetric EPID calibration measurements at the reference field size (10 x 10 cm\(^2\)) behind isocentrically aligned polystyrene slab phantoms of several thicknesses by fitting the primary transmission \( T_{ij}^{\text{primary}} \) as a function of the radiological phantom thickness \( t_{ij}^\text{radiol} \),

\[
\left< T_{ij}^{\text{primary}} \right>_{\text{cROI}} = \exp[-\mu_{AC} (\eta_{BH} t_{ij} + \sigma_{BH} \cdot (t_{ij}^\text{radiol}^2))].
\]

(5)

The brackets \( \left< \cdot \right>_{\text{cROI}} \) represent the average over a central region of interest (cROI). The cROI is defined as a small region around the central axis of approximately 0.5 x 0.5 cm\(^2\) (projected into the isocenter plane).

With the fitted beam hardening coefficient \( \sigma_{BH} \) and the experimentally determined primary transmission \( T_{ij}^{\text{primary}} \) behind a patient with the radiological thickness \( t_{ij}^\text{radiol} \), the primary transmission \( T_{ij}^{\text{primary}}(d_{ij}^\text{radiol}) \) at an arbitrary radiological depth \( d_{ij}^\text{radiol} \) is calculated as

\[
T_{ij}^{\text{primary}}(d_{ij}^\text{radiol}) = \left[ T_{ij}^{\text{primary}}(d_{ij}^\text{radiol}) \right. \exp[\sigma_{BH} \cdot (t_{ij}^\text{radiol})^2]
\]

\[
\left. \times[-\mu_{AC} \cdot (t_{ij}^\text{radiol})^2 + (d_{ij}^\text{geom} / t_{ij}^\text{radiol})^2] \right].
\]

(6)

In this equation, the first term describes the situation without beam hardening and the second term, i.e., the exponential function, corrects the transmission for the beam-hardening effect. As in the original model, the radiological path lengths [see Eq. (6)] are replaced by the corresponding geometrical path lengths taken from the (planning) CT scan, i.e., no tissue inhomogeneities are taken into account. Therefore, the attenuation correction \( AC \) is now given by

\[
AC_{ij}(d_{\text{reconst}}) = \tilde{T}_{ij}^{\text{primary}}(d_{ij}^\text{radiol}) / T_{ij}^{\text{primary}}
\]

\[
= \left[ T_{ij}^{\text{primary}}(d_{ij}^\text{geom}) / T_{ij}^{\text{primary}}(d_{ij}^\text{radiol}) \right. \exp[\sigma_{BH} \cdot (t_{ij}^\text{geom})^2]
\]

\[
\left. \times[-\mu_{AC} \cdot (t_{ij}^\text{geom})^2 + (d_{ij}^\text{geom} / t_{ij}^\text{geom})^2] \right].
\]

(7)

In contrast to the radiological and geometrical depths, \( d_{ij}^\text{geom} = d_{ij}(d_{\text{reconst}}) \) and \( d_{ij}^\text{geom} = d_{ij}^\text{geom}(d_{\text{reconst}}) \), respectively, do not depend on \( d_{\text{reconst}} \). Note that the linear attenuation coefficient \( \mu_{AC} \) is not required to calculate the attenuation correction with Eq. (7) because the experimentally determined primary transmission \( T_{ij}^{\text{primary}} \) is used. If no beam hardening were present, i.e., \( \sigma_{BH} = 0 \), and the dose were reconstructed in the radiological midsurface, i.e., \( d_{ij}^\text{radiol} = t_{ij}^\text{radiol} / 2 \), then the attenuation correction would become \( 1 / \sqrt{T_{ij}^{\text{primary}}} \), as described previously.\(^{11,16} \)

Summarizing this part of the dose reconstruction, the input data for our algorithm are the primary portal dose image behind the patient \( P_{ij}(d_{\text{EPID}}) \) (“the transmitted dose”), the portal dose image without the patient \( PD_{ij}^{\text{EPID, without patient}} \) and the external contours of the patient taken from the planning CT scan. In order to backproject the transmitted dose to a reconstruction plane in the patient, the attenuation between this
reconstruction plane and the exit plane is required. First, from the two EPID images, the primary transmission $T_{ij}$ of the patient at the time of treatment is determined [Eq. (3)]; note that this is done independent of the CT scan and enables us to detect changes in patient geometry. Then this (total) primary transmission $T_{ij}^{\text{primary}}$ is used to estimate how the primary transmission changes in the patient with depth $[\bar{T}_{ij}^{\text{primary}}(d_{ij}^{\text{radio}})]$, Eq. (6). Here the external contours from the patient are used in order to “distribute the (total) 2D primary transmission over three dimensions,” giving a reconstruction-plane dependent attenuation correction $AC_{ij}(d_{\text{recons}})$ [Eq. (7)].

II.C.2. Scatter within the patient

The second adaptation of the reconstruction algorithm was done for the scatter contribution $S_{c}$ to the dose within the reconstruction plane [see Eq. (1)], which is separated into a thickness and a field-size dependence in our original model,

$$S_{ij}(d_{\text{recons}}) = \{Pr_{ij}(d_{\text{recons}}) \cdot SPR_{ij}^{\text{ref}}[\bar{T}_{ij}^{\text{primary}}]\} \otimes K_{ij}^{\text{mid}}.$$  \hspace{1cm} (8)

The primary patient dose is weighted with $SPR_{ij}^{\text{ref}}$, the scatter-to-primary ratio determined under reference conditions (see below). $SPR_{ij}^{\text{ref}}$ 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. 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.

In the original model, the scatter-to-primary ratio $SPR_{ij}^{\text{ref}}$ is only dependent on the primary transmission $T_{ij}^{\text{primary}}$ and therefore independent of depth. For the estimation of the parameters for the scatter contribution, EPID measurements were performed behind isocentrically aligned slab phantoms of several thicknesses for the reference field size (10 × 10 cm$^2$), i.e., the same amount of scatter material was present above and below the isocentric dose-reconstruction plane at depth $d_{ij}^{\text{geom}} = r_{ij}^{\text{geom}}/2$. Therefore, for dose reconstruction at an arbitrary plane at depth $d_{ij}^{\text{geom}}$ in the new model, the scatter contribution is calculated as if the same amount of material below the reconstruction plane were present as above it. Therefore $SPR_{ij}^{\text{ref}}$ becomes a function of twice the geometrical depth and Eq. (8) changes into

$$S_{ij}(d_{\text{recons}}) = \{Pr_{ij}(d_{\text{recons}}) \cdot SPR_{ij}^{\text{ref}}[\bar{T}_{ij}^{\text{primary}}(2d_{ij}^{\text{geom}})]\} \otimes K_{ij}^{\text{mid}}.$$  \hspace{1cm} (9)

For the calculation of the primary transmission $\bar{T}_{ij}^{\text{primary}}(2d_{ij}^{\text{geom}})$ at $2d_{ij}^{\text{geom}}$ with Eq. (6), the radiological path lengths are replaced by the corresponding geometrical path lengths. The concept of the depth dependence of the scatter component of the dose is illustrated in Fig. 1.

II.C.3. Build-up effect

The third adaptation was the empirical build-up correction. This correction was based on the depth-dose curve along the central beam axis through the isocenter for the reference field size (10 × 10 cm$^2$). We determined the EPID-based depth-dose curve from the EPID-based 3D dose distribution for an isocentrically aligned slab phantom of reference thickness (20 cm). From an EPID measurement behind this phantom, we reconstructed the EPID-based 3D dose distribution using Eqs. (1), (2), (7), and (9) and changed the distance $d_{\text{recons}}$ of the reconstruction plane between 90 and 110 cm. The reference depth-dose curve $D_{\text{reference}}^{\text{reference}}$ was determined with a small ionization chamber (Semiflex of 0.125 cm$^3$, PTW-Freiburg, Freiburg, Germany) in a water tank (PTW-Freiburg, Freiburg, Germany). Both depth-dose curves were normalized to the nominal linac output. By fitting the deviations between the EPID-based and reference depth-dose curves along the central axis with an exponential function, we determined the dose–build-up correction parameter $\alpha_{\text{BCP}}$

$$\langle D_{ij}(d_{\text{recons}}) \rangle_{\text{ROI}} \cdot [1 - \exp(- \alpha_{\text{BCP}} d_{ij}^{\text{geom}})] = \langle D_{\text{reference}}(d_{\text{recons}}) \rangle_{\text{ROI}}.$$  \hspace{1cm} (10)

Finally, the 3D dose distribution $D_{ij}^{3D}$, is calculated in our new model by

$$D_{ij}(d_{\text{recons}}) = D_{ij}(d_{\text{recons}}) \cdot [1 - \exp(- \alpha_{\text{BCP}} d_{ij}^{\text{geom}})].$$  \hspace{1cm} (11)

By changing the distance $d_{\text{recons}}$ of the reconstruction plane between a minimum and a maximum value, the 3D dose distribution is calculated within these boundaries for each gantry angle.

II.C.4. Dose grids

The reconstruction grid for the total 3D dose distribution, the “total-dose grid,” was parallel to the axes of the CT grid.
and had a resolution of $0.1 \times 0.1 \times 0.1$ cm$^3$. It encompassed the patient volume, excluding the CT scanner couch.

The dose was reconstructed per beam on a 3D grid based on multiple planes parallel to the EPID. This “beam-dose grid” encompassed the total-dose grid and had also a resolution of $0.1 \times 0.1 \times 0.1$ cm$^3$. The 3D dose distributions of all beams were resampled on the total-dose grid using Catmull–Rom cubic-spline interpolation$^{28}$ and summed to obtain the 3D dose distribution for the total treatment fraction.

### II.D. EPID versus TPS: Depth-dose curves, in-phantom and in vivo dose verification

To investigate the accuracy of our model for 3D dose reconstruction, depth-dose curves along the central beam axis through the isocenter were determined from EPID images behind a slab phantom. These curves were compared to depth-dose curves from our clinical TPS (Pinnacle V8.0h, Philips Medical Systems, Eindhoven, The Netherlands). As an example, we used square fields of 10 MV photons and a 20 cm thick, isocentrically aligned, homogeneous phantom of $30 \times 30$ cm$^2$ polystyrene slabs. The comparison was made for the original model and its refinements, as well as for different field sizes ($3 \times 3, 5 \times 5, 10 \times 10$, and $20 \times 20$ cm$^2$). All depth-dose curves were normalized to the nominal linac output.

Three clinical step-and-shoot IMRT plans were used to evaluate the accuracy of our 3D EPID dosimetry method for IMRT treatments: a prostate, a rectum, and a head-and-neck cancer case. Details of the patient plans are listed in Table I. In all cases, a dose of 2 Gy/fraction was prescribed to a point in the planning target volume (PTV). All plans were calculated with inhomogeneity correction using Pinnacle’s adaptive convolution superposition approach, which is a fast implementation of the collapsed cone convolution superposition algorithm. All plans were recalculated on the 20 cm thick, isocentrically aligned, homogeneous phantom of $30 \times 30$ cm$^2$ polystyrene slabs using a dose grid of $0.2 \times 0.2 \times 0.2$ cm$^3$ voxels and verified by transmission EPID dosimetry. Note that the same machine settings, such as energy, monitor units, MLC parameters, gantry and collimator angles, as in the patient plans were used. A correction was made for the actual output of the linear accelerator for the in-phantom verifications.

**In vivo** EPID dosimetry was performed at the first three treatment fractions, following our clinical protocol for 2 Gy/fraction treatments. For both in-phantom and **in vivo** dosimetry, the long-term change in response of the EPID (Ref. 24) and the gantry angle-dependent attenuation by the actual treatment couch were taken into account; these issues are important for both 2D and 3D EPID dosimetry.

### II.E. $\gamma$ evaluation, dose-volume histograms

The 3D dose distributions from EPID and TPS were compared by using a 3D $\gamma$-evaluation method$^{29-31}$ using 3% of the prescribed dose as dose-difference and 0.3 cm as distance-to-agreement criteria. Note that for the in-phantom verifications, the prescribed dose was adapted to the “dose at the prescription point.” Because of the different dose-grid resolutions of the planned dose distributions (see Table I), all 3D dose distributions were first resampled on a $0.2 \times 0.2 \times 0.2$ cm$^3$ grid; the dose distributions from the TPS were further interpolated “on the fly” during the 3D $\gamma$ evaluation.
to a $0.1 \times 0.1 \times 0.1 \text{ cm}^3$ grid. The 3D $\gamma$ distributions were statistically evaluated within the 50% isodose surface (relative to the prescribed dose). Within the 50% isodose surface, the mean $\gamma$ index ($\gamma_{\text{mean}}$), the 99th percentile ($\gamma_{1\%}$) $^{31,32}$ (symbol chosen in analogy with $D_{1\%}$ in dose-volume histograms (DVHs), as 1% of points have an equal or higher $\gamma$ value), and the percentage of points within $\gamma$ agreement ($P_{\gamma<1}$) were determined.

For one treatment fraction of the prostate case, DVHs were determined for the PTV, the prostate with seminal vesicles, and the organ at risk (OAR), the rectal wall. These volumes were defined on the CT scan during the planning process. The DVHs were determined for the 3D dose distribution reconstructed from EPID measurements and the 3D dose distribution from the TPS. Several DVH points were determined: The median dose value $D_{50\%}$, the “minimum” dose value $D_{99\%}$, and the “maximum” dose value $D_{1\%}$.

III. RESULTS

III.A. Depth-dose curves

In Fig. 2(a) depth-dose curves for a $10 \times 10 \text{ cm}^2$ field of 10 MV photons are shown. The original EPID model (“EPID original model”) shows the exponential decay with inverse-square-law dependence. The introduction of beam hardening with depth and depth dependence of the scatter component in the model (“EPID 3D, no build-up”), improved the agreement; however, the build-up in the first few centimeters was not predicted correctly. When that part was explicitly modeled (“EPID 3D”), the agreement improved further. Approximately beyond the first 1 cm, the deviation was within 3% of the local planned dose [see Fig. 2(b)].

In Fig. 3, the depth-dose curves of a 10 MV photon beam are shown for various field sizes. Note that the curves for all field sizes agreed with the calculations performed by the TPS at the isocenter. The deviations were larger than 3% at depths less than approximately 1 cm and at depths larger than 17 cm for some of the fields.

III.B. In-phantom test

All plans were checked using the slab phantom to make sure that our method worked when the anatomy of the “patient” was perfectly known and no tissue inhomogeneities were present. Statistical data of the 3D $\gamma$ evaluation are summarized in Table I. For the prostate, the rectum, and the head-and-neck case, the percentages of points within $\gamma$ agreement, $P_{\gamma<1}$, were 99.9%, 98.1%, and 96.6%, respectively. At the dose prescription point, the dose values of the 3D dose distributions based on EPID measurements and the TPS agreed within 1%.

III.C. In vivo dosimetry

III.C.1. Prostate cancer treatment example

In Fig. 4, the results of 3D in vivo EPID dosimetry for an IMRT treatment fraction of prostate cancer are shown. The total 3D dose distribution reconstructed from EPID treatment images, the planned 3D dose distribution from the TPS, and the resulting 3D $\gamma$ evaluation are presented in three orthogonal planes through the isocenter; the dose distributions are overlayed on the patient CT data. The reconstructed and planned dose distributions agreed, which was quantified using a 3D $\gamma$ evaluation. The 3D $\gamma$ distributions were statisti-
The mean had a deviation of 17%, but because of the small absolute PTV were within 3%. For the OAR, the maximum dose values agreed within 1% for both the PTV and the OAR.

In order to illustrate the additional value of total 3D dose reconstruction, these data were compared to single-beam 3D dose-reconstruction results. As an in vivo example, we performed a single-beam dose verification of the beam at a gantry angle of 40°; a plane parallel to the EPID through the isocenter from the resulting 3D γ distribution is depicted in Fig. 6(a). In Fig. 6(b), the same plane, but now taken from the total 3D γ distribution (see Fig. 4), is shown. The (small) deviations for the single beam (green/yellow/red regions) vanished in the γ evaluation of the total 3D dose distribution. Note that the dose-difference criterion of 3% of the prescribed dose corresponds to 6.0 cGy for the total dose but to approximately 1.3 cGy for this individual beam.

### III.C.2. Rectum and head-and-neck cancer treatment examples

For a treatment fraction of the rectum case, the results in three orthogonal planes through the isocenter are shown in Fig. 7 for the reconstructed and planned dose distribution and the 3D γ evaluation. The results for the three fractions were similar (see Table I). Within the 50% isodose surface (relative to the prescribed dose), the mean γ index was 0.35, γ1% was 1.15, and the percentage of points within γ agreement, Pγ<1%, was 98.2%; all values were averaged over the three fractions. For the head-and-neck example (see Fig. 8, Table I), on average, for the three fractions, 97.6% of points were in γ agreement, with an average γmean of 0.39 and an average γ1% of 1.12, within the 50% isodose surface.

The calculation time for the 3D dose reconstruction was less than 15 min for one fraction on a standard PC with an Intel dual-core processor (1.9 GHz). The 3D γ evaluation took, on average, approximately 2 s/fraction.
coronal

FIG. 8. 3D in vivo dose verification results for an IMRT treatment fraction of head-and-neck cancer: Plots as for Fig. 4. The white cross marks the isocenter. The white and yellow contours represent the planning target volume (patient- and tumor-specific clinical target volume, 0.5 cm margin) and an organ at risk (spinal cord), respectively. The orange line shows the 50% isodose surface (relative to the prescribed dose).

IV. DISCUSSION

A simple backprojection algorithm for 3D in vivo transmission EPID dosimetry was developed, which is accurate within $\gamma$ criteria of 3% of the prescribed dose and 0.3 cm and is independent of a TPS. The new method was based on our previously developed backprojection algorithm, where 2D dose reconstruction in a plane through the isocenter parallel to the EPID was possible. Various refinements of our EPID dose-reconstruction model were introduced to improve the accuracy of the existing model for depths other than the plane through the isocenter in order to enable 3D dose reconstruction. These improvements were based on physical phenomena (beam hardening with depth and depth dependence of the scatter component of the dose) as well as empirical fitting (dose build-up). Note that this model improvement did not need any additional measurements (assuming the depth-dose curve with an ionization chamber is available, as it is usually used to commission the TPS).

The depth-dose curves illustrate the improvements made (see Fig. 2). With the original model, there was only agreement around the isocenter, where the fit of the EPID to the reference data was performed. However, with the new model the EPID-based depth-dose curve and the depth-dose curve from the TPS agreed along almost the whole curve. In our approach, the depth-dose curve for the $10 \times 10 \text{cm}^2$ field agreed best with the plan as this field size was chosen to fit the reference data. But also depth-dose curves for other field sizes ($3 \times 3$ to $20 \times 20 \text{cm}^2$) were well described by our new model (see Fig. 3). Because this is the range of effective field sizes for most of the IMRT segments in our clinical plans, these corrections can be applied to IMRT fields as well.

Using the linear beam-hardening model [see Eq. (4)], we derived a beam-hardening parameter $\eta_{BH}=0.004 \text{ cm}^{-1}$ for 10 MV. In the formal concept developed by Kleinschmidt, the beam-hardening coefficient is defined through the (differential) change in mean linear attenuation coefficient. Based on the published accelerator spectra, the initial beam-hardening coefficient, i.e., at zero depth, shows no energy dependence and its mean value is estimated to approximately 0.006 cm$^{-1}$, corresponding to 0.003 cm$^{-1}$ in our parametrization. This is in agreement with our result of 0.004 cm$^{-1}$, considering the simplicity of our beam-hardening model and that the response of our detector, i.e. the EPID, is, in principle, energy dependent. The latter effect has, however, been reduced by the additional 2.5 mm thick copper plate.

The build-up correction parameter describes an empirical correction of the depth-dose curve [see Eq. (11)]. It was only fitted for the reference field size ($10 \times 10 \text{cm}^2$) and the reference phantom thickness (20 cm) along the central axis, ignoring off-axis variations. Note that the depth-dose curve for 10 MV already showed some “build-up behavior” without the build-up correction (see Fig. 2, “EPID 3D, no build-up”) due to the (improved) parametrization of the scatter contribution to the total dose in the dose-reconstruction model. Our modeling involved fits to reference data measured with an ionization chamber in slab-geometry phantoms. In particular, to estimate the scatter-to-primary ratio SPR [used in Eq. (9)], phantoms of thicknesses ranging from 4 to 44 cm were used in an isocentric setup with the ionization chamber at the isocenter. For the smallest thickness, this corresponds to a depth of 2 cm, which is just a bit upstream of the position of dose maximum for a 10 MV photon beam, resulting in the “rudimentary” build-up effect. The additional improvement toward the reference depth-dose curve is done through the build-up correction parameter $\alpha_{BCP}$. The estimated value for 10 MV was $\alpha_{BCP}=2.3 \text{ cm}^{-1}$, indicating that the correction influences only the first 2 cm; after 2 cm the effect is less than 1%. This means that one has to be careful with the dose-reconstruction results at small depths.

For the in-phantom dose verification, the dose distributions resulting from the EPID measurements and the TPS agreed within $\gamma$ criteria of 3% and 0.3 cm for at least 96.6% of points within the 50% isodose surface. We conclude that...
the linac settings, such as the number of monitor units and MLC parameters, were correctly transferred from the TPS to the treatment machine. Moreover, it proves that for a homogeneous object, the 3D dose-reconstruction algorithm based on EPID transmission images is accurate within y criteria of 3% and 0.3 cm.

The EPID images were matched to the beam outline from the TPS before the dose was reconstructed to compensate for EPID shifts including possible detector sagging. The EPID is shifted to make off-axis fields fit on the EPID. As these shifts are done manually, a gantry-angle dependent EPID sagging correction would not be sufficient. Since the matching of the EPID images to the beam outlines from the TPS was done separately for each image, the EPID positions did not have to be the same for the image acquisition with and without the attenuating object in the beam. However, if for a certain field the fluence delivery would uniformly be shifted by some distance, i.e., irradiating the wrong spot with the correct fluence, our current matching procedure would attribute the fluence shift to an EPID shift and we would therefore not detect this error. This is a limitation of the current procedure, but at present, it is difficult to determine the EPID shift independently. In the future, automatic EPID position readout may help to solve this problem. However, if the transmission of the attenuating object would change due to the fluence shift, our transmission-measurement based verification method would create a warning (see below).

On one hand, our dose-reconstruction model itself does not account for tissue inhomogeneities but uses water-based correction parameters and a “homogeneous representation” of the patient by the external contours. On the other hand, the EPID measurements are taken behind the actual patient (and thus “contain” the patient’s inhomogeneities) and the result of our dose-reconstruction algorithm is compared to the dose calculation of the TPS including inhomogeneity corrections. We think that in a treatment verification, it is preferable to verify the original treatment plan instead of a derived plan, e.g., a plan with the inhomogeneity corrections switched off or a plan that has separately been transferred to the treatment machine. By using the original plan, we can check an integral part of the radiotherapy-process chain from planning to patient irradiation. The examples presented here and elsewhere21,23 show that our simple and fast approach is still accurate within 3% and 0.3 cm for the verification of prostate, rectum, and head-and-neck cancer treatments despite the presence of tissue inhomogeneities (gas pockets for the prostate and rectum case, and air cavities for the head-and-neck case), which are typically relatively small. Furthermore, when the geometrical path lengths are used instead of the radiological ones to calculate the primary transmission at an arbitrary geometrical depth [see Eq. (6) used in Eqs. (7) and (9)], many of the terms contain the ratio of these path lengths, thus reducing the effect of inhomogeneities on the reconstructed dose distribution. For large inhomogeneous regions such as lung, the current algorithm cannot be used directly. An adaptation has been developed and is used both in 2D and 3D.37

Performing in vivo dosimetry is not much different from performing in-phantom dosimetry in our approach. The patient or the phantom has to be positioned on the treatment table and the treatment fields have to be given, while the EPID images are acquired. As outlined above, images without the patient or phantom are needed in our dose-reconstruction algorithm. In principle, the same “without images” can be used in both cases. Also the model for the dose reconstruction is the same. The attenuation of the treatment table is taken into account for both in-phantom and in vivo dosimetry.

However, for the in-phantom verification, we know exactly the “anatomy,” i.e., the external contours and the position. In such a “pretreatment” approach, patient anatomy changes are excluded and a correction is made for the actual output of the linear accelerator. This enables an “undisturbed” comparison between measured and calculated doses in the phantom. During clinical in vivo dosimetry, this is not possible but also not desired since changes in patient anatomy and/or accelerator output should not be ignored but detected. If the in vivo dose verification fails for an unknown reason, we resort to an in-phantom check; this is our current clinical procedure for 2D EPID dosimetry. The in-phantom dose verification also works for treatment sites with large tissue inhomogeneities. However, in-phantom dose verification costs approximately 1 h additional time on the treatment machine, whereas for in vivo dosimetry, only 10 min extra are needed for the measurement of the images without the patient.21

In vivo measurements are usually performed for more than one fraction. This gives the possibility to distinguish between systematic and random deviations in the reconstructed in vivo data; an example of causes for random deviations are gas pockets in prostate cancer treatments or obstructing table arms.21 An obstructing table arm (behind the patient) only influences the reconstructed dose but not the real dose to the patient and can easily be identified in the EPID images. If possible, these table arms are moved out of the beam. This is easier to do during a measurement on a phantom, where the measurement can also be repeated if necessary.

If different plans are transferred from the TPS to the treatment machine for the phantom test and the patient treatment, a transfer error in the clinical plan will be missed when the in-phantom plan is properly transferred. It is safer to use the same plan for the in-phantom test as for the patient treatment.

The in vivo dose is based on a portal image with the patient in the beam in our algorithm. The transmitted dose is backprojected into a volume with the planning anatomy at the planning position. However, as the transmitted dose changes with the entrance fluence (e.g., caused by wrongly transferred monitor units or missing wedges) and with anatomy changes, the reconstructed dose depends on both the entrance fluence and the anatomy at the treatment time. Our transmission-based dose-reconstruction approach is different from non-transmission-based dose-reconstruction methods, which are based on measurements without or before the patient (see Ref. 19). The latter images are not in-
fluenced by patient anatomy changes, whereas in our approach they are. Note that the dose comparison is done versus the TPS dose calculation based on the planning anatomy. Even though the dose might not accurately be reconstructed in the case of changed anatomy, any changes that make the reconstructed dose different from the planned dose (above 3% and 0.3 cm) will give a warning in our verification process, demanding further action to find the reason for the difference and to investigate a possible clinical impact on the patient.

Situations can be envisaged where our method would fail. In the event that an entrance fluence change would be compensated by an anatomy change of the patient in such a way that the reconstructed dose would not change, our method would not detect the changes. But it is highly unlikely that this would happen for all beams at once. Or imagine a homogeneous slab-geometry phantom, “treated” with a 4-field box technique at gantry angles of 0°, 90°, 180°, 270°. When we change, for instance, the source-surface distance for the 0° beam in such a way that the other beams still fully travel through the phantom, the transmitted fluence is invariant to the phantom position shift (except for a very small change in phantom-to-EPID scatter) and we will not detect this with our dosimetry method, where we assume that the phantom is at the correct position. Although this is unlikely for a real patient with an irregular body contour, we like to stress that our EPID dosimetry method is not a replacement of a proper position verification.

In order to estimate how a change in anatomy influences the total dose in the patient, we consider the dose in the radiological midsurface defined by \( d_{ij}^{\text{radiol}} = T_{ij}^{\text{radiol}} / 2 \). When we neglect that the (position of the) radiological midsurface changes with a changing anatomy, neglect beam hardening, and assume that the dose is proportional to the primary dose in the patient, then the dose in the radiological midsurface is proportional to \( P_{ij}^{\text{EPID}} / \sqrt{T_{ij}^{\text{primary}}} \) according to Eqs. (2) and (7), with \( P_{ij}^{\text{EPID}} \) as the primary portal dose behind the patient and \( T_{ij}^{\text{primary}} \) as the primary transmission [see Eq. (3)]. A change in primary portal dose by \( \times \% \) will change the primary transmission also by \( \times \% \), and therefore, the dose in the radiological midsurface will change by \( \times \% / 2 \). So as a rule of thumb, half of the change in the transmission is seen as a change in the midsurface dose. As an example, a change in patient thickness of 1 cm will change the transmission for a 6 MV photon beam by approximately 5% (see Refs. 27 and 33 for the linear attenuation coefficient), and therefore, the dose at the radiological midsurface by 2.5%. This (crude) estimate only depends on measurements but not on a CT scan. Again, the comparison in our method is done against the planned dose distribution based on the original anatomy. When the induced change is larger than 3% or 0.3 cm, we will detect it.

The external contours of the patient are needed in our dose-reconstruction algorithm. As outlined above, in the case of changed anatomy, our method will still give a warning even when the external contours from the planning CT scan are used. Therefore, the “warning approach” does not depend on the availability of a cone-beam CT scanner and can therefore also be used in a hospital without such a device. However, as long as the patient contours within the dose-reconstruction grid are available, a cone-beam CT scan or any contour scan could be used in our algorithm without compromising the accuracy. When the anatomy changes, the treatment-time anatomy is a prerequisite for an accurately reconstructed dose distribution.

A reconstructed 3D dose distribution provides the possibility to determine DVHs in vivo. It is important to verify the location and shape of the delineations on which the DVHs are based and to redelineate these structures if (large) changes have occurred. Important clinical parameters can be estimated from these DVHs, such as reported for the normal tissue complication probability after breast irradiation. This makes valuable information available for the radiation oncologist since in vivo dose data are presented in a way familiar to a physician. In contrast, 2D or 3D \( \gamma \) distributions usually require some experience to be interpreted. In our hospital, we intend to use 3D dose reconstruction based on EPID transmission images clinically for all IMRT patients, starting by using it when a dose error is found by means of 2D in vivo EPID dosimetry and more in-depth analysis is required. For example, a serious plan transfer error was analyzed and we estimated an underdosage of 20% for a part of the PTV in the first fraction—an estimate based on an analysis of the actual 3D dose distribution. By resending the plan, the correct settings were transferred to the treatment machine and the error was “diluted” by delivering the correct plan for the remaining fractions.

van Zijtveld et al. described the potential impact of a malfunctioning leaf detected during pretreatment verification. Differences between measured and predicted portal dose images of about 10% were observed. Due to the combination of the different treatment fields, the dose differences reduced to 5% in the reconstructed 3D dose distribution. The local underdosage in part of the PTV could not be detected due to the large tumor size and they concluded that for clinical evaluation one cannot rely on the comparison of DVH and DVH parameters only. Indeed, in 3D dose reconstruction, the cumulative effect of all beams is evaluated. Small deviations seen in single-beam dose reconstruction may vanish in the total plan comparison if they are relatively unimportant (see Fig. 6), e.g., when these deviations are not (strongly) correlated. We would like to emphasize that it is essential to realize that such deviations might be hidden in 3D. However, after certain pass-fail criteria (e.g., based on the prescribed dose, \( \gamma \) and/or DVH parameters) have carefully been chosen by taking into account what is clinically acceptable, one actually wants to hide such small deviations; the pass-fail criteria determine what is “within specs.” In this respect, 3D EPID dosimetry allows a better discrimination of relevant from irrelevant errors. In vivo EPID dosimetry, as any other types of patient-specific quality assurance, is not a replacement of regular machine quality assurance but should be regarded as a safety measure for the individual patient treatment.

3D dose reconstruction based on EPID transmission im-
ages has been demonstrated on phantoms using 3D convolution/superposition methods. For non-transmission 3D EPID dosimetry, advanced dose calculation algorithms (from TPSs, Monte Carlo) have been used. Our 3D dose-reconstruction algorithm is a measurement-based convolution/correction approach. Compared to the above mentioned advanced dose calculation algorithms, our method is relatively simple but fast and still sufficiently accurate to detect “noteworthy” errors in vivo. In this regard, it meets the current need for a fast and accurate, TPS-independent dose verification system as a safety net for patient treatment.

V. CONCLUSIONS

A model for 3D in vivo dose reconstruction from EPID images was developed. For IMRT treatments of a prostate, a rectum, and a head-and-neck cancer patient, it was demonstrated that the model is accurate within criteria of 3% and 0.3 cm inside the patient when the external patient contour is known. The algorithm is fast and is independent of the TPS and provides a safety net for patient treatment.

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1. Present address: Department of Radiation Oncology, Academic Medical Center, Meibergdreef 9, 1105 AZ Amsterdam, The Netherlands. Electronic mail: m.wendling@amc.uva.nl
2. Present address: Department of Radiotherapy, University Medical Center, Postbus 85500, 3584 CX Utrecht, The Netherlands
3. Author to whom correspondence should be addressed. Electronic mail: b.mijnheer@nki.nl


