A Monte Carlo based three-dimensional dose reconstruction method derived from portal dose images


Department of Radiation Oncology (MAASTRO), GROW, U.H. Maastricht, Maastricht, The Netherlands

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The verification of intensity-modulated radiation therapy (IMRT) is necessary for adequate quality control of the treatment. Pretreatment verification may trace the possible differences between the planned dose and the actual dose delivered to the patient. To estimate the impact of differences between planned and delivered photon beams, a three-dimensional (3-D) dose verification method has been developed that reconstructs the dose inside a phantom. The pretreatment procedure is based on portal dose images measured with an electronic portal imaging device (EPID) of the separate beams, without the phantom in the beam and a 3-D dose calculation engine based on the Monte Carlo calculation. Measured gray scale portal images are converted into portal dose images. From these images the lateral scattered dose in the EPID is subtracted and the image is converted into energy fluence. Subsequently, a phase-space distribution is sampled from the energy fluence and a 3-D dose calculation in a phantom is started based on a Monte Carlo dose engine. The reconstruction model is compared to film and ionization chamber measurements for various field sizes. The reconstruction algorithm is also tested for an IMRT plan using 10 MV photons delivered to a phantom and measured using films at several depths in the phantom. Depth dose curves for both 6 and 10 MV photons are reconstructed with a maximum error generally smaller than 1% at depths larger than the buildup region, and smaller than 2% for the off-axis profiles, excluding the penumbra region. The absolute dose values are reconstructed to within 1.5% for square field sizes ranging from 5 to 20 cm width. For the IMRT plan, the dose was reconstructed and compared to the dose distribution with film using the gamma evaluation, with a 3% and 3 mm criterion. 99% of the pixels inside the irradiated field had a gamma value smaller than one. The absolute dose at the isocenter agreed to within 1% with the dose measured with an ionization chamber. It can be concluded that our new dose reconstruction algorithm is able to reconstruct the 3-D dose distribution in phantoms with a high accuracy. This result is obtained by combining portal dose images measured prior to treatment with an accurate dose calculation engine. © 2006 American Association of Physicists in Medicine. [DOI: 10.1118/1.2207315]

Key words: Portal dosimetry, EPID, IMRT verification, pre-treatment verification, Monte Carlo dose calculation

I. INTRODUCTION

Intensity-modulated radiotherapy (IMRT) allows the delivery of the radiation dose conformally to the tumor and/or sparing of normal tissues. IMRT can result in steep dose gradients between target volumes and organs at risk and therefore good quality control procedures are necessary for the total process of treatment planning, patient setup, and beam delivery. Quality control procedures must be implemented before the start of the treatment, i.e., pretreatment verification should be performed. Various verification techniques and a large number of homogeneous and inhomogeneous phantoms are available for IMRT verification. Also, a variety of detectors are used, including film, arrays, and matrices of diodes and ionization chambers. Verification procedures can be divided according to the number of points that are checked in the phantom, and can be performed at a single point, in two dimensions (2-D), in multiple 2-D planes, i.e., semi three-dimensional (3-D), and using a full 3-D reconstruction. All methods have their own advantages and disadvantages.

Point detectors, e.g., ionization chambers, in a phantom can be used for checking the dose delivered at a single point. This verification technique is fast and simple and can be used for an absolute dose verification of both the total plan and a beam-by-beam verification. The difficulty with point detectors is that differences are difficult to interpret in terms of tumor coverage or dose in normal tissue because only a single point is verified.

An extension of the single point dose measurement is a 2-D plane measurement. Films, energy fluence detectors, or electronic portal imaging devices (EPIDs) are suitable for measuring a 2-D dose distribution in front of, inside, or behind a phantom. Films are widely used for checking the 2-D dose distribution inside a phantom. This measurement technique is accurate but labor intensive. The total plan can be verified and beam-by-beam measurements are also possible. The 2-D energy fluence detectors that can be placed on top of the treatment couch or attached to the accelerator head can decrease the workload compared to film measurements. However, these measurement devices have the disadvantage...
that only a verification of the energy fluence of a single beam can be performed. With energy fluence detectors, or for beam-by-beam verification in general, it is not clear how possible differences add up in the total dose delivered.

EPIDs can be used for the geometric verification of the patient position, of the multileaf collimator (MLC) setting or of the leaf trajectory control, but also for dosimetry purposes. If the EPID, which is attached permanently to the linear accelerator, is used for dosimetry this would decrease the workload compared to placing conventional dosimeters, energy fluence detectors, or films.

There are several possibilities to use EPIDs for pretreatment verification. The forward approach\(^1\) predicts the dose or the measured grayscale value at the portal imager based on the planned beams. This predicted dose is then compared with the measured dose and differences are quantified at the plane of the portal imager. The difficulty with this method is that it is not clear how differences at the plane of the EPID are related to the dose in the target volume or in normal tissues. A backward approach\(^2\) is used to relate the measured portal dose to the dose at a point, plane, or 3-D volume inside the patient. These reconstructed dose data are then compared with the corresponding planned dose values.

Verifying the entrance energy fluence or the dose inside a phantom with an ionization chamber, film or specially developed equipment is labor intensive. For the beam-by-beam analysis techniques it is not clear what the impact is of possible differences on the total 3-D dose distribution. With simultaneous film measurements it is possible to measure in multiple planes of the phantom and to reconstruct a semi-3-D dose distribution. This procedure is cumbersome for routine pretreatment verification due to the high workload for developing, digitizing, and comparing the films with the planned dose. Full 3-D dose distribution verification will gain more insight into possible delivery problems. Several approaches are presented in the literature. Gels\(^3\) have the possibility for measuring a 3-D dose distribution inside a phantom, but such a technique is not yet available for routine patient specific quality control and can only be used to verify class solution treatments. In an approach to pretreatment 3-D dose verification, Steciw et al.\(^4\) implemented a dose verification procedure based on the measured energy fluence, and then applied the same treatment planning system as used for the original planning to calculate the dose inside the patient and to estimate the impact of possible differences. Errors in the dose calculation engine of the planning system are, however, not detected in these approaches. Renner et al.\(^5\) used film dosimetry and a dose calculation algorithm based on a convolution/superposition algorithm to reconstruct the 3-D delivered dose in the patient. An independent pretreatment verification is necessary to detect all possible types of errors, e.g., incorrect treatment parameter transfer, dose calculation errors due to limitations of the algorithm itself and its implementation in the treatment planning system, uncertainties in the leaf position, and possible delivery errors of the linear accelerator.

To overcome the limitations of the current pretreatment verification models, that can only verify the dose at a single point or in a plane, and the limitation of entrance energy fluence detectors that can only measure a single beam with no information on how differences may add up or cancel out in the 3-D dose distribution, a 3-D dose reconstruction model is necessary.

The aim of our work is to develop a method that can reconstruct the 3-D dose distribution actually delivered to a phantom or patient, based on a measurement of portal dose images of the separate beams before the start of treatment. Our 3-D pretreatment dose verification should allow an independent verification of IMRT delivery by using measured portal images in combination with an independent dose calculation algorithm based on a Monte Carlo engine. Such a model would replace the need for performing film or ionization chamber measurements in a phantom for pretreatment quality control procedures of (IMRT) treatments. The result of the independent 3-D dose calculation must then be compared with the planned dose distribution. In the present study, this method is presented and verified for homogeneous phantoms. At a later stage the model will be extended to situations having inhomogeneous (tissue) densities.

II. MATERIALS AND METHODS

A. 3-D dose reconstruction model

The pretreatment dose reconstruction model is based on measuring portal images without the phantom in the beam and an independent dose calculation algorithm. The reconstruction model consists of four steps. First, a portal image is converted to a portal dose image. Second, from this portal dose image the energy fluence exiting the linac is extracted. The energy fluence is defined as the total energy of the photons passing through an area. The third step is to sample a phase space distribution from the energy fluence. And finally, a dose calculation based on a Monte Carlo calculation in a phantom is performed. These four steps will be discussed in the next sections in more detail.

1. Portal dose measurement

A portal image is measured under the same conditions as the actual treatment, but without an object placed in the beam. This portal image is converted into a portal dose image (PDI) using an in-house developed calibration model, similar to the procedure described by Chen et al.\(^7\) The calibration model converts the EPID image to a PDI with an error smaller than 1% (SD) for images without an object in the beam. In our study this PDI represents the 2-D dose distribution measured at a depth of 5 cm under full scatter conditions in a water tank, with a source-to-surface distance of 145 cm, resulting in an effective source-to-detector distance of 150 cm.

2. Energy fluence extraction

The primary dose is extracted from the measured PDI by deconvolving the portal dose with a lateral scatter kernel. The kernel is calculated from dose measurements with a miniphantom and dose measurements under full scatter con-
The portal dose $D$ can be described by a primary and a lateral scatter component $D^p$ and $D^s$, respectively,
\[
D = D^p + D^s.
\]
(1)

The lateral scattered dose $D^s$ is described by a convolution of the primary dose with a lateral scatter kernel $K$,
\[
D^s = D^p \otimes K.
\]
(2)

This kernel can be fitted from on-axis phantom scatter measurements at the position of the EPID. The phantom scatter correction factor $S_p$ is determined by using the collimator and total scatter correction factor $S_c$ and $S_{cp}$, respectively, measured at a source-to-detector distance of 150 cm, normalized to a 10 $\times$ 10 cm$^2$ field. The lateral scatter kernel is determined from the following equation:
\[
\frac{S_c E}{S_p} = c_7 \left( 1 + \int f(r) \cdot K(r) d^2 r \right),
\]
with
\[
K(r) = c_1 \exp(-c_2 r) + c_3 \exp(-c_4 r) + c_5 \exp(-c_6 r),
\]
and $c_7$ is a parameter describing the relation between the primary dose and the dose measured under full scatter conditions of a 10 $\times$ 10 cm$^2$ field, $f(r)$ is the primary dose profile $D^p(r)$ normalized to the central axis dose ($r=0$) for the measured field,
\[
c_7 = \frac{D^p_{10\times10}(r=0)}{D^p_{10\times10}(r=0)}, \quad f(r) = \frac{D^p(r)}{D^p(r=0)}.
\]
(5)

For a derivation of Eq. (3) see the Appendix.

The energy fluence $\Psi$ is assumed to be proportional to the primary dose $D^p$,
\[
\Psi = \frac{D^p}{\rho \mu(t) \exp(-\mu t)},
\]
(6)

with $\rho$ the density of the medium and $t$ the depth of the measurement point. The linear attenuation coefficient $\mu(\theta)$ at an off-axis angle $\theta$ is assumed to be described by the on-axis attenuation coefficient $\mu(\theta=0)$ multiplied by an off-axis correction factor described by Tailor et al.,
\[
\frac{\mu(\theta)}{\mu(\theta=0)} = 1 + 0.00181 \theta + 0.000202 \theta^2 - 0.00000942 \theta^3.
\]
(7)

3. Phase space reconstruction

The starting point of our dose calculation, based on the Monte Carlo code XVMC, is a phase space distribution (PSD). This PSD is reconstructed using a simple point source model that is located at the target of the linac. From the energy fluence distribution and a pre-defined spectrum, a random position and energy are sampled, respectively. Such a reconstructed PSD has eight parameters: three parameters indicating the position, three parameters describing the direction, and two parameters describing the energy and the statistical weight of the photon.

The direction and position of the individual photons is calculated by assuming that the photons originate from a point source located at the target. This position can be sampled from a point in the reconstructed energy fluence distribution. Six parameters indicating the position and direction of the photon below the collimator jaws, MLC, and blocks, are calculated.

The spectrum parameters are taken from the commissioning procedure of the XVMC code. Briefly, the energy of the photon $E$ is sampled from an on-axis spectrum and multiplied with a correction for the off-axis position to incorporate beam softening. The off-axis spectrum is derived from measured and simulated percentage depth dose curves and fitted to an analytical function. This on-axis energy spectrum $p(E)$ is described by a function characterized by four parameters: $l$, $b$, $E_{min}$, and $E_{max}$,
\[
p(E) dE = N E^l \exp(-bE) dE,
\]
(8)

with $E_{min} < E < E_{max}$, and $N$ a normalization factor, chosen in such a way that the sum of all chances is unity. The energy of the off-axis photons is sampled from the on-axis spectrum and an off-axis correction $s(\theta)$ is applied,
\[
E(\theta) = s(\theta) E,
\]
(9)

with $s(\theta)$ the off-axis correction for angle $\theta$,
\[
s(\theta) = \left( \frac{\mu(\theta=0)}{\mu(\theta)} \right)^{2.22}.
\]
(10)

The statistical weight of the photon is sampled from the measured energy fluence distribution. To incorporate the absolute value of the energy fluence at a specific point in the reconstructed phase space distribution, the energy fluence is divided by the off-axis correction factor so that the total energy fluence at an off-axis position is kept constant and equal to the energy fluence extracted from the portal dose image at that position.

4. Monte Carlo calculation

The reconstructed phase space distribution is the starting point for an independent dose calculation based on Monte Carlo calculation. The Monte Carlo code XVMC is used for a fast calculation of the 3-D dose inside the phantom. Briefly, this code was derived from the Voxel Monte Carlo code for electron dose calculations and several improvements such as variance reduction techniques are used for a fast and accurate dose calculation. For a detailed explanation of the code see Fippel et al.

B. Linear accelerator and Electronic Portal Imaging Device

The Electronic Portal Imaging Device (EPID) used for the measurements is a Siemens OptiVue 1000 amorphous silicon type EPID mounted on a Siemens Oncor Avant Garde linear accelerator (Siemens Medical Solutions, Concord, USA) equipped with 6 and 10 MV photons. The EPID is calibrated using an in-house developed model to measure a 2-D dose distribution in a water tank at 5 cm depth placed at 145 cm
from the focus of the linac. For the work presented here the distance of the EPIDs is fixed at a source-to-imager distance of 150 cm, with an effective area of $41 \times 41$ cm², the image size is resampled to $512 \times 512$ pixels with a pixel spacing of 0.8 mm. An additional copper plate of 3 mm is mounted on top of the EPID to reduce the over-response due to low-energy scattered photons that are incident on the sensitive layer of the EPID.8

C. Phantom study

The 3-D dose reconstruction method has been tested using homogeneous phantoms. First, the model parameters of the reconstruction algorithm are derived. Second, the method is compared to measurements in a water phantom with an ionization chamber. An irregularly shaped MLC segment is also measured with film in a homogeneous phantom. Third, an actual IMRT plan is delivered to a homogeneous phantom with various film inserts and an insert for an ionization chamber.

1. Model parameter derivation

The parameters of the model, i.e., the coefficients of the lateral scatter kernel, the energy fluence and spectrum parameters, are derived from measurements. For the determination of the lateral scatter kernel square fields of widths 3, 6, 10, 15, 20, 25, 30, and 35 cm were used, measured both under full scatter conditions and with a miniphantom. The source-to-detector distance is 150 cm and 100 monitor units for both 6 and 10 MV photons were given. The miniphantom is a polystyrene cylindrical phantom with a diameter of 3 cm and the effective point of measurement is located at 5 cm water equivalent below the surface. The detector (CC13, Scanditronix Wellhofer, Schwarzenbruck, Germany) is read out by an electrometer (Model 35040, Keithley Instruments Inc., Cleveland, OH) and positioned at a source-to-detector distance of 150 cm, equal to the point of measurement of the dose under full scatter conditions. A diagonal profile of the largest field is also measured with the mini-phantom to yield an estimate of $f(r)$ in Eq. (3), while the boundary values are chosen according to the field width. The shape of the profile inside the field is assumed to be independent of the field size. The lateral scatter kernel $K$ is fitted to Eq. (4). The coefficients $c_i$ are determined by a fitting procedure using an unconstrained nonlinear minimization problem implemented in MATLAB (MATLAB 7.0.4, The Mathworks, Natick, MA). The primary dose is calculated from the measured dose under full scatter conditions [Eq. (1)] in an iterative way using convolution and deconvolution, programmed using Fast Fourier Transforms for speed considerations.

The spectrum parameters $I$, $b$, $E_{\text{min}}$ and $E_{\text{max}}$ are taken from the commissioning procedure of the XVMC code using the Virtual Energy Fluence model.26 The fitted energy parameters can be found in Table I.

The on-axis attenuation coefficient $\mu(\theta=0)$ is determined by transmission measurements of a small field (3 cm x 3 cm) through a 10 cm thick polystyrene phantom.

<p>| Table I. Overview of the parameter values of the reconstruction model for 6 and 10 MV photons. |</p>
<table>
<thead>
<tr>
<th>Fit parameter</th>
<th>6 MV</th>
<th>10 MV</th>
</tr>
</thead>
<tbody>
<tr>
<td>$c_1$ (10⁻³)</td>
<td>6.6350a</td>
<td>11.321a</td>
</tr>
<tr>
<td>$c_2$ (10⁻⁴ cm⁻¹)</td>
<td>8.8738a</td>
<td>9.3365b</td>
</tr>
<tr>
<td>$c_3$ (10⁻⁴)</td>
<td>13.723a</td>
<td>2.4679b</td>
</tr>
<tr>
<td>$c_4$ (10⁻³ cm⁻¹)</td>
<td>3.0911b</td>
<td>1.7922a</td>
</tr>
<tr>
<td>$c_5$ (10⁻⁴)</td>
<td>1.0679b</td>
<td>1.1905a</td>
</tr>
<tr>
<td>$c_6$ (10⁻⁴ cm⁻¹)</td>
<td>0.9468b</td>
<td>1.4860b</td>
</tr>
<tr>
<td>$c_7$ (-)</td>
<td>0.8825b</td>
<td>0.8952a</td>
</tr>
<tr>
<td>$\mu(\theta=0)$ (10⁻² cm⁻¹)</td>
<td>4.68±0.05</td>
<td>3.57±0.05</td>
</tr>
<tr>
<td>$\rho$ (g/cm³)</td>
<td>1.000</td>
<td>1.000</td>
</tr>
<tr>
<td>$l$ (-)</td>
<td>0.1320b</td>
<td>0.8552b</td>
</tr>
<tr>
<td>$b$ (MeV⁻¹)</td>
<td>0.6478b</td>
<td>0.5552b</td>
</tr>
<tr>
<td>$E_{\text{min}}$ (MeV)</td>
<td>0.25b</td>
<td>0.25b</td>
</tr>
<tr>
<td>$E_{\text{max}}$ (MeV)</td>
<td>7.00b</td>
<td>11.00b</td>
</tr>
</tbody>
</table>

The uncertainty in the parameters is indicated by the actual fitting procedure of the phantom scatter correction factor. The model described this factor with a maximum error of 0.2% between measured and fitted $\mu$ values.

The parameters are taken from the commissioning procedure of the XVMC model and are fitted from a measured percentage depth dose curve. The maximum difference between the reconstructed curve and the measured curve was 1%.

2. Model verification measurements

To test the accuracy of the reconstruction model various symmetric field sizes are used with widths of 5, 10, 15, and 20 cm for both 6 and 10 MV photons. The percentage depth dose curves and off-axis profiles at various depths (5, 10, 20, and 30 cm) are measured with the CC13 ionization chamber in a water phantom (Blue Phantom, Scanditronix Wellhofer, Schwarzenbruck, Germany) with a source-to-surface distance of 90 cm and are compared to the reconstructed dose. Phantom measurements are used for checking the dose reconstruction model of an irregularly shaped MLC field using a photon beam of 6 MV. A phantom with dimensions $30 \times 30 \times 21$ cm consisting of square slabs of about 2 cm polystyrene thickness is used to position EDR2 films (Eastman Kodak Company, Rochester, NY) at different depths. For this measurement two films are placed at 10.5 and 15.0 cm depth from the surface (SSD 90.0 cm) of the phantom. The films are processed with the Kodak X-Omat 1000 film processor and digitized using a VXR-16 scanner (Vidar Systems Corporation, Herndon, VA).

3. IMRT plan and measurements

An IMRT plan, generated using our treatment planning system (XiO 4.2.2, CMS, St. Louis, MO), intended for a prostate treatment using 10 MV photons consisting of 5 beams (gantry angles of 36°, 100°, 180°, 260°, and 324°)
and 53 segments in total, is delivered to the phantom described before. Films are placed at 4.3, 8.6, 13.0, and 17.3 cm depth in the phantom. For checking the absolute dose, one polystyrene slab is replaced by another slab containing a hole, located at the isocenter, for the placement of a Farmer-type ionization chamber (NE 2505/3, NE Technology Ltd, Reading, UK).

4. PDI measurements and reconstruction

EPID measurements for the square fields and the MLC field have been performed without the phantom in the beam. For the IMRT plan, the various segments for a single beam are integrated in the EPID software (Coherence-Portal Imaging Workspace, Siemens) resulting in a single image per beam (gantry angle). The gantry angle for acquiring the EPID images is forced to 0°, because for portal dose acquisition a copper plate of 3 mm has been positioned on top of the EPID. This plate was, at the time of measurement not permanently fixed at the EPID and had to be placed manually on top of the EPID in order to perform the portal dose measurements. For the 3-D dose reconstruction, however, the gantry angle was changed to the planned gantry angle that was used for the film measurements.

The voxel size and calculation grid for the water phantom verification measurements is 0.50 × 0.50 × 0.50 cm³, whereas for the analysis of the film measurements a smaller grid of 0.25 × 0.25 × 0.25 cm³ is used. The number of simulated photons was adjusted until a statistical error smaller than 1% was achieved for all presented data.

III. RESULTS

A. Model parameters

The results of the on-axis phantom scatter measurements are shown in Fig. 1. From these measurements and the profile along the diagonal of the largest field, the lateral scatter kernel was fitted. This kernel is derived for a depth of 5 cm and a SSD of 145 cm and is shown in Fig. 1, while the parameters of the fit are also presented in Table I. The error of the fit was for both the 6 and 10 MV beam below 0.2% (maximum difference). The attenuation coefficient was chosen according to transmission measurements of a small field (3 cm × 3 cm) using a 10 cm thick polystyrene phantom.

The on-axis attenuation coefficients $\mu(\theta=0)$ are $4.68 \times 10^{-2}$ and $3.57 \times 10^{-2}$ cm⁻¹ for the 6 and 10 MV beam, respectively.

B. Model verification measurements

Figures 2–4 show reconstructed and measured beam profiles and depth dose curves for 6 and 10 MV photons. The agreement between ionization chamber measurements and reconstructed data is within a few percent. The profiles, shown in Figs. 2 and 3, are for all field sizes reproduced within 2% of the dose at the center of the profile, both inside and outside the field (excluding the penumbra) compared to ionization chamber measurements. Depth dose curves for all measured field sizes are reconstructed with an error smaller than 1% (maximum difference) at depths larger than the dose maximum ($D_{max}$). At shallow depths around $D_{max}$ this difference is larger, which is shown in Fig. 4. Absolute values of the reconstructed dose have been compared with measured output factors and are shown in Fig. 5. For small fields, e.g., 5 cm × 5 cm, there is an overestimation of the reconstructed dose compared to the measured dose, while for the larger field sizes, e.g., 20 cm × 20 cm, the opposite effect occurs. This difference is, however, small, <1.5% at the maximum, for the 6 and 10 MV beams.

The measured and reconstructed 2-D dose distribution at 15.0 cm depth in the phantom of the 6 MV MLC-shaped field is shown in Fig. 6, normalized at the central axis. A gamma evaluation, with 3% maximum dose difference and 3 mm distance-to-agreement criterion, is performed to quantify the differences between the measured and reconstructed dose distribution. The percentage of gamma-values, over a rectangular area defined by the outer leaf positions, smaller than unity, is larger than 99% for both films at 10.5 and 15.0 cm depth in the phantom.

C. IMRT verification measurements

The IMRT phantom plan has been checked by verifying the total dose distribution using film measurements in various planes. An ionization chamber measurement is used to verify the absolute dose at a single point in the phantom. The film measurements and reconstruction data are normalized on the central axis value. The gamma evaluation for the 10 MV IMRT plan is shown in Fig. 7. A histogram of gamma values is calculated inside the irradiated field, which is de-

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FIG. 1. Measured and fitted phantom scatter correction factors (left) and the corresponding lateral scatter kernel (right), for the 6 and 10 MV photon beams.
fined from $x=-5$ to $+5$ cm and $y=-10$ to $+10$ cm. Gamma values are generally below unity. The percentage of gamma-values $>1$ are 1.2%, 0.2%, 2.3%, and 0.6% for the data at 4.3, 8.6, 13.0, and 17.3 cm depth in the phantom. The somewhat higher number of gamma values $>1$ for the film placed at 13.0 cm depth was due to a difference between reconstructed and measured dose in a low dose area with a steep dose gradient toward the edge of the phantom. The dose difference between the reconstructed and measured absolute dose was small, 1%; the reconstructed dose was 2.13 Gy compared to 2.15 Gy for the dose measured with the ionization chamber.

**IV. DISCUSSION**

Our new method reconstructs the 3-D dose distribution in a homogeneous phantom based on the actual delivered
beams measured with an EPID prior to treatment without the object placed in the beam. An accurate and easy to use reconstruction model has been developed that is able to reconstruct the 3-D dose distribution in a homogeneous phantom, independent of the treatment planning system, and based on the actual measured portal images.

The energy parameters are taken from the commissioning procedure of the XVMC code. The parameters can easily be derived from a measured depth dose curve. The reconstructed percentage depth dose values are in good agreement with the measured data. There is, however, a small difference in the buildup region. This could be due to electron contamination, which is not modeled in the reconstruction model, but accurate measurements with the ionization chamber are also difficult to perform due to the lack of electron equilibrium. An extra source of this electron contamination could be incorporated in the model to yield accurate surface doses, but for dose verification at depths larger than $D_{\text{max}}$ this is not necessary. Also, head scatter is not taken into account very accurately. The measured energy fluence at the position of the EPID also contains head scatter fluence that is, due to the point source model, assumed to originate from the target as well.

The lateral scatter kernel is derived from the measured on-axis dose under full scatter conditions and a primary dose. The primary dose is reconstructed accurately from the portal dose. The reconstructed off-axis primary dose profiles at the position of the portal imager are in good agreement (maximum difference <2%, not shown) with the measured primary dose profiles, excluding the penumbra. This method of verification with a miniphantom is, however, limited to regions with a low dose gradient and cannot be used in regions with steep dose gradients because of the volume averaging effect due to the finite size, 3 cm diameter, of the miniphantom. However, the penumbra after the reconstruction of the dose inside the phantom is in good agreement with measurements, indicating that the primary dose extraction from the portal dose image is also valid in regions with a high dose gradient.

The agreement between the measured and reconstructed absolute dose is good, but there is a trend to overestimate the dose for smaller field sizes and underestimate the dose for larger field sizes compared to the measurements. Fippel et al.26 also found this difference for the XVMC dose calculations using the commissioning procedure of the Virtual Energy Fluence model. The reason for this small difference is still unknown and needs further investigation. For IMRT treatments that consist of a large number of small fields, this difference can be minimized by determining the absolute conversion factor for the reconstruction model at a smaller field size than the 10 cm × 10 cm field that was used.

**Fig. 4.** Measured and reconstructed percentage depth dose curves for a field size of 10 cm × 10 cm, 6 and 10 MV photons, normalized at 10 cm depth.

**Fig. 5.** Measured and reconstructed absolute dose values at 10 cm depth in a water phantom (SSD 90 cm) for 6 and 10 MV photons. For each measurement, 100 monitor units were given.

**Fig. 6.** Gamma evaluation, with 3% (of maximum) dose and 3 mm distance-to-agreement criteria, of a 6 MV MLC-shaped segment and a histogram of the gamma values (right). The measured portal dose image is shown in the right figure. The film measurement and dose reconstruction have been performed at 15.0 cm depth in the phantom.
For the square fields, the reconstructed off-axis and depth dose data show excellent agreement with the measurements in the water phantom. Also, the MLC-shaped segment shows a good correlation between measured and reconstructed dose values. At some points in the off-axis region around \((x,y) = (8, -7)\) cm there is a slightly larger difference. These small differences between measurement and reconstruction might be the result of differences in the acquisition of the portal image or uncertainties in the film measurement, voxel size effects of the dose calculation, repositioning of the leaves between two different fractions, or variation in the output of the linac.

The dose calculation is performed using a Monte Carlo dose engine. The presented 3-D dose reconstruction method is in principle not restricted to Monte Carlo based methods. Renner et al.\(^4\) used a similar approach by using film dosimetry and a dose calculation algorithm based on a convolution/superposition algorithm. Steciw et al.\(^5\) used the measured energy fluence and the treatment planning system’s algorithm to estimate the impact of possible differences. Errors in the dose calculation engine of the planning system are, however, not detected in this approach.

The reconstruction algorithm is also not restricted to be used with homogeneous phantoms. Monte Carlo calculations are in principle superior in the dose calculation accuracy for inhomogeneous phantoms. Our reconstruction model must then also be validated with an inhomogeneous phantom study. The presented approach can then be used for dose reconstruction in the planning CT data of the patient. This opens possibilities for actual patient specific pretreatment quality assurance procedures that can quantify the effects of differences between planned and delivered beams.

V. CONCLUSION

Our newly developed 3-D dose verification model allows an accurate and independent method for the verification of complex (IMRT) treatments and the quantification of possible differences between the 3-D dose distribution calculated with the treatment planning system and actual dose delivered. A set of measured portal images is the only input needed for the reconstruction model, which is the input for an accurate dose calculation based on a Monte Carlo dose engine.

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APPENDIX: DERIVATION OF THE LATERAL SCATTER KERNEL

The dose \(D\) measured under full scatter conditions can be described as the summation of a primary dose \(D^P\) and a lateral scattered dose \(D^S\),

\[
D = D^P + D^S
\]

the lateral scattered dose is calculated using the primary dose and a lateral scatter kernel \(K\),

\[
D^S = D^P \otimes K.
\]

The evaluation of Eqs. (A1) and (A2) at the central axis \(r = 0\) and dividing this value by \(D^P(r=0)\) leads to

\[
\text{FIG. 7. Reconstructed and measured dose distribution of a 10 MV IMRT prostate plan at 17.3 cm depth in the phantom are presented in the upper left and right plot, respectively. Differences between the reconstructed dose and film, shown as a gamma evaluation, are given in the lower left plot. A 3% of maximum dose and 3 mm distance-to-agreement criteria were used. In the lower right plot the reconstructed and film profiles for a horizontal and a vertical profile are shown. The vertical profiles are multiplied by a factor 1.5 in the lower right plot for a clear visual representation.}
\]
The collimator scatter correction factor $S_c$ and the total scatter correction factor $S_{c,p}$ are defined as the measured primary dose and the measured dose under full scatter conditions, respectively, divided by the corresponding values for the reference field size, i.e., 10 cm $\times$ 10 cm,

$$S_{c,p} = \frac{D}{D_{10\times10}} , \quad S_c = \frac{D^p}{D^p_{10\times10}} .$$  \hspace{1cm} (A4)

Normalizing the measured primary and the dose measured under full scatter conditions and dividing the total scatter correction factor by the collimator scatter correction factor leads to the phantom scatter correction factor $S_p$,

$$\frac{D(r=0)}{D^p_{10\times10}(r=0)} = \frac{S_{c,p}}{S_c} = \frac{D^p_{10\times10}(r=0)}{D^p_{10\times10}(r=0)} \left( 1 + \int f'(r')K(r')d^2r' \right) .$$  \hspace{1cm} (A5)

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\*Corresponding author: Wouter J.C. van Elmpt, MSc, Department of Radiation Oncology (MAASTRO), University Hospital Maastricht, P.O. Box 5800, NL-6202 AZ Maastricht, The Netherlands. Telephone: +31 88 55 66666. Fax: +31 88 55 66667. Electronic mail: wouter.vanelmpt@maastro.nl


