IMRT verification by three-dimensional dose reconstruction from portal beam measurements

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A method of reconstructing three-dimensional, in vivo dose distributions delivered by intensity-modulated radiotherapy (IMRT) is presented. A proof-of-principle experiment is described where an inverse-planned IMRT treatment is delivered to an anthropomorphic phantom. The exact position of the phantom at the time of treatment is measured by acquiring megavoltage CT data with the treatment beam and a research prototype, flat-panel, electronic portal imaging device. Immediately following CT imaging, the planned IMRT beams are delivered using the multiple-static field technique. The delivered fluence is sampled using the same detector as for the CT data. The signal measured by the portal imaging device is converted to primary fluence using an iterative phantom-scatter estimation technique. This primary fluence is back-projected through the previously acquired megavoltage CT model of the phantom, with inverse attenuation correction, to yield an input fluence map. The input fluence maps are used to calculate a “reconstructed” dose distribution using the same convolution/superposition algorithm as for the original planning dose calculation. Both relative and absolute dose reconstructions are shown. For the relative measurements, individual beam weights are taken from measurements but the total dose is normalized at the reference point. The absolute dose reconstructions do not use any dosimetric information from the original plan. Planned and reconstructed dose distributions are compared, with the reconstructed relative dose distribution also being compared to film measurements. © 2002 American Association of Physicists in Medicine. [DOI: 10.1118/1.1494988]

Key words: IMRT verification, electronic portal imaging, megavoltage CT

I. INTRODUCTION

The use of intensity modulation in radiation therapy (IMRT) has been shown to enable the delivery of highly conformal dose distributions.1 Escalation of the dose delivered to the target, while maintaining an acceptable dose to surrounding organs at risk, is only possible using IMRT in a significant number of cases. The technology to plan and deliver such treatments has received significant research interest in recent years, and routine IMRT is now performed by a growing number of groups around the world. Methods of delivery include the use of multiple-static fields shaped with a multi-leaf collimator (MLC); dynamically collimated fields using a number of fixed gantry angles; dynamic, intensity-modulated arc therapy, and tomotherapy.

With the routine introduction of such techniques, there is significant current interest in verification. There are two reasons why verification is of particular interest in IMRT.

(i) Verification that the planned fluence distributions have been successfully delivered. A particular concern is that these new procedures may not be adequately covered by standard verification protocols. Point dose measurements frequently used as part of standard verification procedures may be difficult to apply in some IMRT cases, where the measurement point falls on a steep dose gradient. Commissioning and quality assurance of new types of planning and delivery techniques requires careful attention, including checking of small-field output factors in dose calculations and the effects of MLC leaf position error on dose delivery.

(ii) Verification of patient positioning. Where high dose gradients are used to spare organs at risk close to a target volume, special attention should be given to ensuring that the patient is correctly positioned. This may include organ motion both between fractions and during any given treatment fraction.

In a recent review, Essers et al. recommend that “portal dose measurements [should be made] during treatments with a large dose inhomogeneity ... for instance in the thorax region or when using intensity-modulated beams.” 2

The first approach to verifying IMRT treatments was to deliver the patient-optimized intensity-modulated beams to an anthropomorphic or simple geometric phantom and measure the resultant dose distribution using film.3,4 The measured dose distribution is then compared to distributions for the patient-optimized beams and recalculated using the particular phantom’s geometry. Burman et al. and Tsai et al. both describe the integration of these pretreatment, film-based measurements into comprehensive IMRT quality assurance procedures by combining the results in vivo point dose measurements using TLDs.5,6

The use of electronic portal imaging devices (EPIDs) is also of interest for IMRT verification. Studies of the basic dosimetric performance of EPIDs have been presented for
camera-based, liquid ionization chamber-based and amorphous silicon flat-panel systems. Pretreatment verification of 2-D intensity-modulated beam portals has also been demonstrated for both camera-based and liquid ionization chamber-based systems. In vivo dosimetric measurements were demonstrated using a camera-based EPID by Kirby et al., who reported multiple-static field verifications showing the agreement between prescribed and measured exit doses to within 3%, and McNutt et al., who used a convolution/superposition model, treating the EPID as part of an “extended phantom” to create portal dose images, showing agreement to within 4% (2 SD). An iterative method was then used to reconstructed dose distributions in phantom, showing agreement to within 3% (2 SD). Similar in vivo measurements using the liquid ionization chamber have been presented Boellaard et al., showing a reconstructed midplane dose in clinical practice that agreed with planned dose distributions to within 2% (larynx) and 2.5% (breast). Other in vivo dosimetric measurements have been demonstrated by Chang et al., who presented a phantom study of 70 intensity-modulated prostate fields showing that profiles matched to within 3.3% 1 SD and central-axis doses to 1.8% 1 SD. Kroonwijk et al., who showed in vivo dosimetry for prostate treatments and Partridge et al., who showed an in vivo breast dose measurement.

A number of the in vivo measurements describe dose to a point outside of the patient (e.g., the exit surface, or a portal plane). To describe the dose distribution in three dimensions inside the patient, some sort of transit dose calculation, or back-projection of the measured beam portals is required. Wong et al., described a comparison of a forward ray tracing calculation through the planning CT cube with exit dose measurements made with film and TLD, but found matching the geometry of the film and CT cubes a problem. Agreement of 3% was seen after matching, but local differences of 5%–10% were still apparent. A slightly different approach by Aoki et al., was then extended by Hansen et al., who aligned EPID images with “beam’s eye view” DRRs and back-projected the primary fluence from the EPID images through the planning CT and calculated dose. Phantom measurements showed agreement of TLD, film, and transit dose measurements to within 2%.

A problem with transit dose calculations using the planning CT volume (as pointed out by the above authors) is that patient set-up errors or organ motion on the day of treatment must be taken into account. Portal images and DRRs can only be used to correct for rigid body motion, for nonrigid transformations further 3-D anatomical information is required to avoid inaccurate and potentially misleading results. A solution to this problem is to have a “treatment time” CT, taken either using the megavoltage treatment beam (MVCT), a CT scanner close to the treatment machine (i.e., inside the treatment room) or additional kilovoltage CT hardware added to the treatment machine.

Nakagawa et al., presented verification images of a rotational arc therapy treatment, where measured fluence was back-projected through a single-slice MVCT image, although full dosimetric calculations were not presented. The single-slice approach has been further developed by the tomotherapy research group, showing first input fluence measurements, made by back-projecting measured exit fluence through a treatment-time MVCT image, and then full dosimetric reconstructions. Results are shown that agree to within 3 mm (high dose gradients) or 3% (low dose gradients).

Our aim in the work presented here is to demonstrate the feasibility of three-dimensional dosimetric reconstruction of IMRT distributions using a conventional LINAC. Treatment-time volume images are obtained using megavoltage conebeam CT. An inverse-planned IMRT treatment is delivered to an anthropomorphic phantom using the multiple-static field delivery technique and the transmitted beams recorded using an amorphous silicon flat-panel imager. The measured signal is converted to primary fluence using a method similar to that described by Hansen et al., and back-projected through the MVCT model of the phantom to yield the input fluence. The dose is then calculated from the measured input fluence maps and the MVCT model using the same planning system dose calculation as for the original plan. Planned and reconstructed dose distributions (both relative and absolute) are presented and the applications and limitations of the technique are discussed.

II. METHOD

The concept investigated by the work described here is summarized by Fig. 1. A phantom was CT scanned using a stereotactic localization system and an inverse IMRT plan generated. The phantom was set up for treatment on a linear accelerator—using the same stereotactic system—and translated to the defined treatment position. A megavoltage CT
scan was then acquired using an amorphous silicon flat-panel imager before delivering the IMRT beams using the multiple-static field technique. Beam delivery was recorded using the flat-panel imager. The delivery was then repeated with radiographic film inserted in the phantom. The IMRT beams were also delivered to the flat-panel imager with the phantom removed, to provide a measurement related to the input fluence. The following sections describe each stage of this process in detail and present the results of the dosimetric measurements.

A. kVCT data acquisition treatment planning

Three slices from the neck section of an Alderson-Rando phantom were placed in a stereotactic head localization frame and CT scanned with a Siemens Somatom scanner (slice thickness 3 mm, slice resolution 0.6 mm × 0.6 mm). The slices were mounted on a 15° shallow wedge such that the chin of the phantom was raised. This makes the axis of the phantom neck parallel to the rotation axis of the scanner, thus reducing the field of view required to acquire complete CT data (this was important due to the small field of view of the flat-panel imager, as discussed below). An IMRT plan was then generated using the Virtuos/KonRad inverse planning system (DKFZ Heidelberg/MRC Systems, Heidelberg, Germany). For the purposes of this study, a single cylindrical “organ at risk” (OAR) running vertically through the phantom was defined. A “horseshoe”-shaped target was then defined that partially wraps around the OAR. The plan was generated with five equispaced beams (36°, 108°, 180°, 252°, and 324°), each divided in to a maximum of five intensity levels. A transverse slice through the treatment isocenter of the planning CT cube, showing the target and organ-at-risk outlines, is shown in Fig. 2. All dose calculations described here were performed using an in-house convolution/superposition algorithm in preference to the planning system’s standard pencil-beam model. The algorithm employs kernels constructed by the weighted summation of monoenergetic dose spread arrays, following the method of Mackie et al.

B. Detector optimization and calibration

The detector used for all imaging measurements described here was an amorphous silicon (a-Si) flat-panel device comprised of a matrix of 256 × 256 pixels with a pitch of 800 μm (RID256L, Perkin Elmer Optoelectronics, Germany). Each pixel consists of an a-Si:H photodiode switched by a thin-film transistor. This array is placed in contact with 134 mg/cm² Gd₂S₂O:Tb phosphor screen (Lanex fast, Kodak, USA). The detector as-supplied was fitted with a 0.6 mm aluminum cover plate.

Photoelectric interaction of lower-energy scattered radiation has been shown to produce an over-response in the as-supplied detector when compared to a water-equivalent detector. Monte Carlo studies using the GEANT detector simulation tool³⁶ have shown that replacing the 0.6 mm aluminum plate with 3.0 mm copper provides an optimum improvement in spectral response—by preferential absorption of the low-energy scattered radiation—with negligible degradation in spatial resolution.³⁷ The total energy deposited in the amorphous silicon layer of the detector was scored for a series of monoenergetic pencil beams with energies from 200 keV to 6 MeV in steps of 200 keV and the optimum copper build-up plate thickness determined. This Monte Carlo derived energy response function is particularly important when correcting for scattered radiation; see Sec. II C. This method implicitly assumes that the signal given by the amorphous silicon detector is directly proportional to the dose deposited in the detection layer, and is independent of energy. This assumption is not strictly true, since both the phosphor screen and—to a lesser extent—the amorphous silicon layer can be expected to exhibit an energy-dependent response. In this case, however, the effect of energy dependence is kept small due to the beam filtration provided by the 3.0 mm copper plate.³⁷,³⁸ A full Monte Carlo simulation of the detector including optical effects in the phosphor screen would be interesting, but is beyond the scope of this work.

Available integration times on the imager ranged from 80–6400 ms per frame. The detector reads out using an ana-
log to digital converter in a half-row interleaved raster scan, with two charge amplifier ASICs working simultaneously. For example, the incoming data stream for a frame arrives at the acquisition computer in the following order: \((1,1) (129,1) (2,1) (130,1) \cdots (128,1) (256,1) (1,2) (129,2)\), and so on, where \((u,v)\) is the column and row number of each pixel. The signal charge transfer time (effective dead time) for each pixel is 2.4 \(\mu\)s, with the pixel integrating continually for the rest of the frame.

The small size of the detector (20 cm \(\times\) 20 cm) limits this study to small phantoms and a small target volume, although the method described is readily applicable to larger detectors. Since the entire object must be visible in every projection for good quality CT reconstructions, the detector was mounted at a fixed distance of 130 cm from the source using an in-house constructed support frame, giving a field of view at the isocenter of just 15.4 cm \(\times\) 15.4 cm. The resultant small phantom-to-detector distances (less than 25 cm) means that the amount of scattered radiation incident on the detector was high.

All images were filtered to remove the effect of defective pixels, using a locally applied median window filter, and dark-offset corrected during acquisition. The linearity of the detector was measured by taking the signal recorded at the dark-offset corrected during acquisition. The linearity of the detector was high.

The matrices of coefficients \(B\) and \(C\) were fitted using a series of images of known thicknesses \(t\) of water-equivalent plastic acquired under reference conditions. The scatter correction \(\Phi^\text{scatter}\) is described in Sec. II.C. This calibration scheme is known to be useful for removing artefacts caused by pixel-to-pixel variations in sensitivity in the detector and off-axis changes in beam energy, giving high quality and uniform images suitable for CT reconstruction. An absolute calibration was also performed using the following method: the imager was placed at 130 cm, symmetrically at the center of a 10 cm \(\times\) 10 cm, 6 MV photon beam. The integration time of the detector was set to 100 ms, averaging over 100 frames for one image. In the middle of the averaged frame sequence, a 10 MU beam was delivered using a dose rate of 200 MU/min. The mean signal over a 16 \(\times\) 16 pixel region of interest (12.8 mm \(\times\) 12.8 mm at the plane of the imager) at the center of the image was then calculated. This method relates the signal recorded by the imager to the absolute dose delivered under reference conditions. This calibration does not take into account the variations in responsivity of the detector outside the averaged region of interest or off-axis variation in beam energy. Given that the detector is not water-equivalent, systematic errors in absolute dose are also expected with field size. Dose reconstruction using both the relative and absolute calibration methods are presented.

### C. Scattered radiation correction

With the detector placed close to the exit surface of the phantom, a large amount of scattered radiation is incident on the detector surface. The increased buildup plate thickness of the modified detector helps to preferentially absorb lower-energy multiply scattered radiation, but single Compton scatter must still be corrected for. The method used was based on the iterative Monte Carlo kernel-based technique described by Hansen et al.\(^{34}\) and later implemented by Spies et al.\(^{40}\) The raw image is first converted to effective radiological water thickness, using the relative thickness calibration described in Eq. (1). The scattered radiation present in the image is then estimated by convolution of the thickness map with a thickness-dependent pencil beam scatter kernel. For the work described here, the GEANT-calculated spectral response of the detector was incorporated into the precalculated pencil beam kernels. The procedure can be summarized in the following form, where the scatter in a given iteration \(n\) is given by

\[
\Phi^\text{scatter}(n) = W \sum_{i,j} \sum_{k,l} \Phi^\text{primary}(i,j) \cdot \xi(t_{ij}, r_{ijkl}),
\]

where \(r_{ijkl}\) is the distance from the scoring pixel \(i, j\) to the presumed center of the scatter kernel \(\xi\) at pixel \(k, l\) and \(w\) is a scaling factor. The thickness map \(t^{(n)}\) is calculated using the scatter estimate from the previous iteration \(\Phi^\text{scatter}(n-1)\) using the following equation:

\[
0 = C(t^{(n)})^2 + Bt^{(n)} - \ln \left( \frac{\Phi^\text{primary}}{\Phi^\text{primary} - \Phi^\text{scatter}(n-1)} \right)
\]

(the scatter contribution is set to zero for the first iteration). In practice, the iterative process is found to converge on an acceptably accurate solution within three or four iterations.

### D. The LINAC

All beams were delivered using the 6 MV photon beam of a Siemens Primus LINAC (Siemens Medical Systems, Concord, CA, USA) equipped with the standard 40 leaf-pair MLC. Each leaf projects to 1 cm wide at the isocenter. The IMRT beams were delivered automatically as a series of multiple static fields using the Siemens Lantis control system. The control system of the LINAC limits the smallest dose deliverable by each single beam to an integer number of monitor units. For the IMRT beams, each beam segment is simply rounded to the nearest integer number of MUs. For the megavoltage CT data acquisition, the minimum dose deliverable for each projection was limited to 1 MU. The alignment of all phantoms was carried out using a stereotactic localization system and the treatment room lasers. All beams were delivered at 200 MU/min.

### E. Geometric alignment, helix and stereotaxy

Due to the nonideal trajectory of the detector during gantry rotation (because of sagging of the detector support frame under gravity) a geometric correction was performed. Projections of a cylindrical phantom\(^{31}\) with a helix of high-contrast...
ball bearings set into its surface were acquired for every gantry angle to be used during MVCT reconstruction. The motion of the support frame was known to be quite reproducible. The helix phantom was aligned using the treatment room lasers and imaged immediately prior to the Alderson Rando phantom. For a particular gantry angle, the projected positions of the ball bearings on the detector give sufficient information to calculate the three-dimensional rigid body transformation from the treatment room (stereotactic) coordinates to the detector coordinates \((u, v)\). A full discussion of this method is beyond the scope of this paper.

### F. Megavoltage CT Imaging

The phantom was set up on the LINAC using the same stereotactic localization system as for kV CT and imaged and translated so that the center of the target point, as defined by the inverse plan, coincided with the rotation isocenter of the LINAC. A megavoltage CT dataset was then acquired taking 120 projection at 3° intervals around the phantom. Beam delivery was automated using the Lantis system, delivering 1 MU per projection at 200 MU/min with the gantry stationary for each beam. The limit of 1 MU per projection was limited by the LINAC control system, as discussed above. The detector was not automatically synchronized with the LINAC, so images were acquired by setting the maximum integration time available of 6400 ms and triggering the detector manually. This long integration time ensures that the entire 1 MU pulse is captured, given that there can be a random delay of up to 2–3 s at the start of each beam.

The CT dataset was scatter corrected using the iterative technique described in Sec. II C, and reconstructed using a cone-beam implementation of the Feldkamp algorithm. The correction for the nonideal scanning geometry using the method discussed in Sec. II E was performed during back-projection. This is preferable to rebinning the projection data, since it removes one interpolation step, which would potentially degrade spatial resolution. Because Compton interactions dominate in the megavoltage region, the reconstructed megavoltage images are almost linear with electron density. However, the dose calculation algorithm used expects CT numbers (Hounsfield units, HU) so the absolute values in the reconstructed images have to be converted. For this work, the reconstructed megavoltage CT values were assumed to be equal to electron density and were converted to HU using the electron density to CT number lookup tables provided by the CT scanner calibration. Since the same look-up table is used to convert from CT numbers back to electron density during the dose calculation, this step should not introduce significant systematic error.

### G. IMRT beam delivery and verification

The IMRT beams were delivered and recorded with the flat-panel imager by acquiring sequences of images. Each image was an average of ten frames, each frame acquired using an integration time of 100 ms. These images could then be summed to give images of each individual beam segment, or the complete intensity-modulated beam for each gantry angle. An independent measurement of the delivered dose in the phantom was acquired by placing radiographic film (Kodak X-Omat V) between two slices of the Rando phantom. (Note: this necessitated the removal of the phantom from the stereotactic positioning system, dismantling to insert the film and repositioning. The phantom could not then be MVCT scanned because of the presence of the film. The geometric error arising from this process is estimated to be 1–2 mm.) As a final independent check, the phantom was removed and the IMRT beams delivered, in air, directly to the flat-panel imager. In this way, the directly measured input fluences could be compared to the in vivo measured, scatter-corrected back-projected input fluences.

### H. Back-projection and dose calculation

The IMRT beam image sequences were iteratively scatter corrected using the method described above and then summed to give the measured primary fluence at the plane of the detector. These primary fluence maps were back-projected through the megavoltage CT dataset to a plane 30 cm above the isocenter. Inverse attenuation corrections were performed using the reconstructed electron density values. The input fluence \(\Phi_{u,v}^{\text{IN}}\) for detector pixel \(u, v\) is given by

\[
\Phi_{u,v}^{\text{IN}} = \Phi_{u,v}^{\text{Primary}} \exp \left( \int \frac{\mu_{x,y,z}}{t_{x,y,z}} dr \right),
\]

where \(\Phi_{u,v}^{\text{Primary}}\) is the measured primary fluence. The integral is performed along the ray line \(r\) (from the point \(u, v\) to the focus) over the attenuation coefficient \(\mu_{x,y,z}\) and pathlength \(t_{x,y,z}\) values for each CT voxel \(x,y,z\) traversed. The nonideal geometry was corrected for using the same helix transformation method as for the CT reconstruction. The resultant reconstructed input fluence maps were then used, together with the MVCT data set, as input for the convolution/superposition dose calculation.

The two different calibration schemes described in Sec. II B were used to derive \(\Phi_{u,v}^{\text{Primary}}\), providing two different dose reconstructions. For the relative method each measured beam was scatter corrected to produce primary fluence at the detector plane, the relative calibration was applied and the input fluence estimated by back-projecting through the MVCT cube and correcting for attenuation. The input fluence maps were converted to monitor units such that total number of monitor units was the same for the reconstruction as for the original plan. This was done by applying a single weighting factor to all of the reconstructed input fluence maps, thus maintaining the “measured” relative beam weights. The relative beam weights are therefore independent of the original plan. The dose was calculated using the same convolution/superposition algorithm as for the original plan and the dose distribution normalized at a reference point in the center of the target volume. For the absolute method, the primary fluence at the detector plane was scatter corrected and converted to absolute dose using the single calibration factor derived from a small region of interest at the center of the detector. These absolute numbers (Gy) were back-projected and attenuation corrected as described above, and used as
III. RESULTS

A. MVCT images and geometric correction

A sample reconstructed slice from the MVCT dataset is presented in Fig. 2. For the purposes of comparison with the neighboring kV CT image, several MVCT slices have been averaged to give comparable slice thicknesses. The bone contrast in the image is noticeably lower, since contrast in the MV image results almost solely from electron density variations, rather than changes in atomic number. Figure 3 shows the gantry sag effect. The shift in the projected position of the isocenter is shown as a function of gantry angle. Distances are quoted at the plane of the isocenter.

B. Beam segments

Figure 4 parts (i)–(vii) show in vivo measured beam segments for a sample beam from one direction (108°). Due to the small size of these beams, no image contrast is visible within each subfield (one beam element, or Bixel, is 1 cm × 1 cm at the isocenter). The collimator positions (field edges), however, are very easily visible and could be used for on-line monitoring of the delivery hardware performance. The intensity-modulated distribution resulting from the sum of the beam segments is then displayed in panel (viii) (scatter corrected and back-projected through the MVCT cube to yield input fluence). For comparison, the “ideal” model of the intensity-modulated produced by the inverse planning system is displayed in panel (ix). This is provided simply to give an idea of the shape of the ideal beam. For a numerical comparison, this “ideal” fluence should be convolved with a model accounting for the radiation source size and the point spread function of the detector.

C. Input fluence matrices

As a test of the scatter correction, back-projection, and inverse attenuation methods, line sections through both the relative reconstructed and the directly measured (in air) input fluences matrices are presented in Fig. 5. The line sections are the average over a 2 mm wide strip through the center of the 108° beam both (a) parallel and (b) perpendicular to the MLC leaf travel direction. The positions of the line sections are indicated in Fig. 4, panel (ix). All distances are quoted at the plane of the isocenter. Generally very good agreement is seen between the directly measured and reconstructed sections, with the greatest discrepancy occurring at the rightmost peak of Fig. 4(a). This good level of agreement is a direct indication that the iterative scatter correction and back-projection methods work correctly, with the scatter corrected and back-projected fluence map matching the directly measured one. It should be noted, however, that, because the direct input to the dose calculation with no further correction. The output of the planning system is then given directly in Gray and, again, no further scaling was applied.
relative normalization scales each beam according to this peak intensity, the normalization point in this beam is probably shifted systematically down because of the small effective field size of the peak.

Because the two sections are taken from two separate deliveries, it is unclear whether the small differences seen are the result of physical differences in the delivered fluence, or arise as an artifact of uncertainty in the measurements and calculations. The accuracy of the these measurements is estimated to be in the region of ± 1 mm, ± 2% dose (relative). An assessment of the repeatability of the delivery system is beyond the scope of this work. Panel (c) shows the effective water thickness of the phantom as seen by the 108° beam. The values were calculated by back-projecting through the MVCT cube along the central axis of the detector in the \( \mu \) (lateral) direction. Water thickness was obtained by ray tracing the raw “electron density” values in the MVCT cube (i.e. before conversion to HU) and dividing the sum by the effective attenuation coefficient of water for a 6 MV beam.

The relative back-projected input fluence matrices were normalized using to the maximum recorded fluence in each beam—the sum of the maximum fluences of all beams is defined to be 100%. Table II shows the results of this process, and compares the resulting values to the normalization values originally produced by the planning system. Some clear discrepancies are seen between the two sets of figures, with the high planned weightings of the 36° and 324° beams reduced and the values of the remaining beams correspondingly increased. One reason for this may be that peak values for very small peaks (1 cm \( \times1 \) cm) in the measured, reconstructed fluence matrices are not well reproduced, as shown earlier in figure 5a. It should also be noted that the prescribed beam weightings, given in terms of monitor units, do not account for the difference in output factor that will be seen in the different-sized field measurements. For a beam

<table>
<thead>
<tr>
<th>Beam angle (degrees)</th>
<th>Relative beam weighting (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>36°</td>
<td>29</td>
</tr>
<tr>
<td>108°</td>
<td>14</td>
</tr>
<tr>
<td>180°</td>
<td>16</td>
</tr>
<tr>
<td>252°</td>
<td>15</td>
</tr>
<tr>
<td>324°</td>
<td>27</td>
</tr>
</tbody>
</table>

Table II. Relative beam weightings for the five IMRT beams. The planned beam weighting were generated by the inverse planning system. The “measured” weightings are the relative maximum values of the measured, back-projected input fluence matrices for each beam.
comprised of many small subfields, the measured fluence would be expected to be lower than a corresponding simple field with the same number of monitor units.

A possible source of systematic error in the back-projection is the assumption in Eq. (4) that attenuation is a simple exponential function of thickness $t$. For the small, centrally placed fields used in this study, significant off-axis changes is beam spectrum and beam-hardening in the phantom are not expected, but for application to more realistic-sized fields the effect of this approximation should be studied further. Off-axis beam quality and beam hardening are taken into account in the detector calibration and scatter correction—using a linear-quadratic model described in Eq. (1)—but the coefficients $B$ and $C$ from Eq. (1) cannot be used directly to model the beam alone, since they contain information about both the beam and spatial changes in gain and nonlinearity of the detector itself.

D. Relative isodose comparisons

Figure 6 shows surface renderings of the planned and reconstructed 90% isodose surfaces, viewed from (approximately) the anterior direction. The two surfaces are drawn to exactly the same scale. By inspection, it can be seen that the broad shape and size of the two surfaces are very similar, indicating that the measured dose distribution is similar to that planned. Two obvious differences, however, are visible: first the measured distribution is generally smoother than the plan calculation and, second, the measured distribution is truncated in the superior–inferior direction. Any blurring of the measured data caused by either x-ray or optical scattering within the flat-panel detector, or by the finite spatial resolution of the detector, is, of course, not modeled by the planning calculation. These smoothing effects do not seem to seriously degrade the reconstructed dose distribution. The truncation of the measured, reconstructed dose distribution is caused by the large beam penumbrae parallel to the MLC leaf-travel direction, as seen in Fig. 5(b). The output factors used in the planning calculation are not well tested for these very small fields. Again, this problem is highlighted in this case by the use of a very small target volume.

A more quantitative examination of the differences between the planned and relative reconstructed dose distributions is gained from comparing the cumulative dose-volume histograms (DVHs) of the target and organ at risk, as presented in Fig. 7. The DVHs show the effect of smoothing in the reconstructed dose distribution with a uniformly slightly higher dose to the OAR. The slight target underdose in the reconstruction is a result of the superior–inferior truncation discussed above. Despite these obvious differences, generally good agreement is seen between the two sets of DVHs, indicating good agreement between the plan and the reconstruction.

Figure 8 shows a contour plot of a transverse slice through the center of the target volume. Isodose lines for the planned and reconstructed dose distributions are overlaid,
showing the 60%, 75%, 90%, and 100% contours, together with the outlines of the organ at risk, target, and phantom surface. Because of the careful geometric calibration procedure used, the absolute positions of the two dose distributions are accurately known, making a direct comparison of the spatial distributions possible. A very good reproduction of the 90% isodose contour is seen, with agreement to within ±2 mm in most places. The mean difference between the planned and reconstructed dose cubes, calculated for a centrally placed volume of interest 14.3 cm × 14.3 cm (transverse) × 6.6 cm (axial) was 0.6% with a standard deviation of ±2.3% (calculated with a voxel size of 2.24 × 2.24 × 3.0 mm³), this is in good agreement with the results of Kapatoes.32,33 The position of the 60% isodose contour is less well reproduced, with the reconstruction showing the dose distribution to be shifted out by 3–4 mm in some places, although this could be an effect of the different dose gradients present at 60% and 90%.

An alternative measure of the difference between the planned and reconstructed dose distributions was calculated by applying a tolerance rule. The two distributions were said to be within tolerance if, for a given point in one distribution, a sufficiently similar dose was found within a short distance from the other distribution. In x,y,z dose space, this criterion is equivalent to specifying points within a four-dimensional hyperellipse with semiaxes equal to the tolerances in x,y,z and dose. Using levels of ±2 mm, ±2% and a similar volume of interest to that used above (bounding the high-dose region), 32.3% of the volume of interest is out of tolerance. Increasing the tolerance to ±3 mm, ±3% the percentage out of tolerance drops to 7.3%, with values of 1.9% and 0.56% for ±4 mm, ±4% and ±5 mm, ±5%, respectively. This method is effectively the same as the γ distribution method described by Low et al.43 with the hyperellipse reduced to a two-dimensional figure with semiaxes equal to the dose difference ΔDM and distance to agreement ΔdM in the notation of Low et al.

E. Comparison of relative reconstruction with film

The dose measured directly by film, relative dose reconstructed from the EPID measurement and planned doses are shown in Fig. 9. The section is inclined at an angle of 15° to the transverse plane, intersecting close to the center of the target volume but moving gradually superior in the anterior direction. Again, agreement of the 90% isodose contours are good, being within ±3 mm most of the time. The 60% isodose (high gradient) is again less accurate, deviating by up to 5 mm in some places. There are several additional sources of error in this comparison, including the accuracy of the film developing and scanning process, errors associated with removing and repositioning the phantom between the two measurements, and the uncertainty in locating the exact oblique section through the reconstructed dose cube. There was also no method available of accurately registering the dose distribution measured by the film with the reconstruction, so the two distributions shown in Fig. 9 were aligned by eye.

F. Absolute dose reconstruction

The results of the absolute dose reconstruction are shown in Fig. 10. A clear systematic error is present, with the absolute dose at the reference point calculated to be 1.055 Gy, compared to a value of 1.022 Gy in the original plan. A line section through the treatment isocenter in the RL direction is shown in panel (b). The ±1.7% 1 SD error bars indicated are estimated just from the uncertainty in the absolute calibration (±0.9%) and errors in measurement. No attempt has been made to estimate the accuracy of the electron density conversion and dose calculation. An estimation of the robustness and repeatability of absolute dose measurements using α-Si flat-panel detectors is beyond the scope of this work.

IV. DISCUSSION

The accurate reconstruction of in vivo dose delivery relies on high-quality treatment-time, anatomical information. For the phantom study presented here, megavoltage CT was used to acquire this information. Because of limitations of the LINAC control system, the dose delivered to the phantom was unfeasibly high for clinical use. To avoid severe undersampling artefacts in the Feldkamp CT reconstruction, 120 projections were required, which with a minimum of 1 MU per projection gives 120 MU. The time taken to automatically acquire a complete CT dataset was also about 17 min. In the future, is it hoped that data will be acquired using continuous gantry rotation, making it possible to reduce the dose per projection and acquire the complete dataset in 1–2 min.

The image quality required for verification must be sufficient to resolve low-contrast soft tissues (∼2% contrast) with sufficient spatial resolution for position verification (1–2 mm). There is a tradeoff between noise and spatial resolution in CT reconstruction, giving less noisy reconstructed images if spatial resolution restrictions can be relaxed. For example,
The effect of the CT set on the dose calculation was studied by running the same dose calculation, with the same (planned) input fluence matrices and substituting the kV and MV CT cubes. For the superposition calculation, no significant differences were observed, indicating that the conversion from electron density to CT number does not introduce significant inaccuracy. The choice of planning algorithms, however, was found to be important. Dose calculations were performed using the kV CT dataset and the planned input fluence matrices for both the standard pencil-beam algorithm and the convolution/superposition algorithm mentioned in Sec. II A. Significant differences in the calculated dose were seen, particularly in the region close to the entry of the $108^\circ$ and $252^\circ$ beams, which show the greatest deviation from normal incidence to the phantom. Because of this, the convolution/superposition algorithm was used for all results presented in this paper. By using the same planning calculation for both the plan and the verification measurement, the comparison should be independent of the dose calculation algorithm. This is in contrast to other transit dose calculation methods, where a separate dose deposition algorithm is used. The generally good agreement between the film measured doses and the results of the convolution/superposition algorithm indicates that this algorithm performs reasonable well, although it would be interesting to compare the measured, reconstructed dose to a full Monte Carlo dose calculation. A further discussion of dose calculation is given by Ahnesjö, but is outside of the scope of this work.

The scatter correction technique described requires an estimate of the water-equivalent thickness of the patient in the beam’s eye view. For the work described here this was calculated performed iteratively from an EPID image, although it could also have been calculated directly by ray tracing through the MVCT cube. An advantage of using the iterative technique is that the scatter correction is not dependent on the MVCT and verification using “portal dose images” could be carried out without the need for MVCT, as demonstrated by McNutt et al. Use of the iterative technique also makes the resulting thickness map independent of errors in the reconstructed MVCT number and does not require the assumption of a single effective attenuation coefficient to convert to thickness. (Note: the MVCT reconstruction is not independent of the scatter correction, since the CT projections are scatter corrected.) With a larger detector, an increase in the patient-to-detector distance should be possible, resulting in a smaller scatter fraction and less severe scatter correction problem.

A major problem with understanding the differences between planned 3-D dose distributions and the measured, reconstructed distributions presented in the last section, is the large number of possible sources of error. An attempt has been made in the method described here to remove or isolate as many of these as possible, but giving unambiguous reasons for the remaining discrepancies is still difficult. The method of localization using a stereotactic frame and the treatment room lasers is the standard method of localization for IMRT for the data acquisition system described in this paper, 0.27 MU per projection and 120 projections gives a signal-to-noise ratio (SNR) 4.2 in a reconstructed image with $0.6 \times 0.6 \times 0.6 \text{mm}^3$ voxels, whereas increasing the voxel size to $1.2 \times 1.2 \times 1.2 \text{mm}^3$ increases the SNR to 16.9. (Note: for this example the dose per projection delivered to the phantom was still 1 MU, although only 0.27 MU per projection was used for the reconstruction.)

A possible source of systematic error in the dose calculation arises from the conversion of reconstructed “MVCT numbers” to HU. The assumption made in this work that the MVCT reconstruction was linear in electron density may be inaccurate for a number of reasons: (i) any inaccuracy in the scattered radiation correction process will lead to “cupping artefacts” in the CT reconstruction, with radial changes in the effective CT number; (ii) uncorrected nonlinearities in the detector response could lead to reconstructed values that are not exactly linear in electron density; and (iii) uncertainties in the calibration of the detector could lead to inaccuracies in the reconstructed absolute electron density. A possible future method for improving both the reconstructed image quality and reconstructed CT number accuracy would be to use an elastic matching technique to fuse the MVCT and planning CT data.

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at DKFZ. For head and neck patients treated in face masks, it has been shown to be accurate to within 3 mm. For this work, the phantom was attached directly to the stereotactic frame and accuracy is estimated to be within 1–2 mm. The relative agreement between dose distributions was generally good, indicating that this method is promising for relative IMRT verification. Significant problems exist with using such a method for accurate, absolute dosimetry.

The major limitation of the work presented was the small size of the flat-panel detector available. This necessitated the use of a small phantom and correspondingly small target volumes and intensity-modulated fields. For a full assessment of the likely clinical accuracy of this technique, further investigations should be carried out with a larger detector (40 cm × 40 cm detectors are commercially available, replacing the 20 cm × 20 cm research prototype described here).

The work described here has been restricted to the use of megavoltage CT to acquire anatomical information immediately pretreatment. The dose reconstruction method, however, is equally applicable to other CT data. For example, kilovoltage cone-beam CT integrated into the treatment LINAC or “CT on rails” could also be used (where a conventional CT scanner is placed in—or very close to—the treatment room, and the patient scanned immediately before treatment, while immobilized). An obvious area of future work is the development of strategies for using such on-line anatomical information to modify treatment plans, in so-called “adaptive radiotherapy.” The kind of information provided by the in vivo dose reconstruction method described here could provide useful input for developing these adaptive radiotherapy strategies.

V. CONCLUSIONS

A proof-of-principle study has been presented describing a method of making three-dimensional, in vivo dosimetric measurements of IMRT beams. A sample IMRT treatment was inverse planned on the neck section of an Alderson Rando phantom and delivered. The position of the phantom at the time of treatment was recorded by taking a megavoltage CT scan. The IMRT beams were delivered using the multiple-static field technique and measured using a flat-panel imager. Measured fluence was converted to primary fluence at the plane of the detector using an iterative technique to correct for phantom scatter. This primary fluence was then back-projected through the MVCT reconstruction of the phantom—using an inverse attenuation correction—to get an estimate of the input fluence. The 3-D dose distribution was calculated using the same convolution/superposition algorithm as for the original plan. Both absolute and relative dose reconstructions were shown.

Verification measurements at various stages of the process were presented. First, the individual multiple-static field segments were shown. Next, the derived input fluence distributions were compared with directly measured input fluences, showing good agreement. The results of the relative dose reconstruction were analyzed using dose/volume histograms of the defined target and “organ at risk” and contour plots of planar sections though the 3-D distributions. Agreement between the planned and reconstructed relative dose distributions were found to be mostly within ±2 mm and areas of low dose gradient, rising to 3–4 mm in higher gradient regions. The overall deviation between the planned and reconstructed dose distributions had a mean of 0.6% with a standard deviation of ±2.3%. The discrepancy between the reconstructed and planned absolute dose at the reference point was 0.033 cGy (5.5%).

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