Dosimetric verification of intensity modulated beams produced with dynamic multileaf collimation using an electronic portal imaging device

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Dose distributions can often be significantly improved by modulating the two-dimensional intensity profile of the individual x-ray beams. One technique for delivering intensity modulated beams is dynamic multileaf collimation (DMLC). However, DMLC is complex and requires extensive quality assurance. In this paper a new method is presented for a pretreatment dosimetric verification of these intensity modulated beams utilizing a charge-coupled device camera based fluoroscopic electronic portal imaging device (EPID). In the absence of the patient, EPID images are acquired for all beams produced with DMLC. These images are then converted into two-dimensional dose distributions and compared with the calculated dose distributions. The calculations are performed with a pencil beam algorithm as implemented in a commercially available treatment planning system using the same absolute beam fluence profiles as used for calculation of the patient dose distribution. The method allows an overall verification of (i) the leaf trajectory calculation (including the models to incorporate collimator scatter and leaf transmission), (ii) the correct transfer of the leaf sequencing file to the treatment machine, and (iii) the mechanical and dosimetric performance of the treatment unit. The method was tested for intensity modulated 10 and 25 MV photon beams; both model cases and real clinical cases were studied. Dose profiles measured with the EPID were also compared with ionization chamber measurements. In all cases both predictions and EPID measurements and EPID and ionization chamber measurements agreed within 2% (1σ). The study has demonstrated that the proposed method allows fast and accurate pretreatment verification of DMLC.

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I. INTRODUCTION

Dose distributions delivered by multiple field irradiation techniques can often be significantly improved by modulating the two-dimensional intensity profile of the individual x-ray beams.1–9 Intensity modulated (IM) beam profiles, calculated by means of inverse treatment planning, can be realized in several ways. One technique is the use of dynamic multileaf collimation (DMLC). So far, the “sliding window” technique in combination with DMLC has received the most attention in the literature. This technique is based on moving each leaf pair of a multileaf collimator (MLC) independently but unidirectionally across the treatment field while the beam is on, effectively sweeping apertures of variable widths across the treatment field. The width of the aperture varies between leaf pairs. Moreover, for each leaf pair the width is also a function of time. Convery and Rosenbloom have described the basic algorithm to calculate the required leaf trajectories; the algorithm has the form of an optimization problem in which the beam on time is minimized.10 Three groups have independently developed analytical equations to calculate the leaf trajectories,11–13 which were further developed by others.14–16 Compared to the algorithm of Convery and Rosenbloom, computation times of the analytical approaches are much shorter. The first group of patients were irradiated using DMLC in 1995 at Memorial Sloan–Kettering Cancer Center, NY.17 Recently, treatment of head and neck cancer patients with DMLC has started in our clinic.

A disadvantage of the dynamic technique is that it is difficult to verify due to its complexity. Bortfeld et al. have therefore proposed to produce intensity modulated beams by superpositioning a number of partially overlapping, static, irregularly shaped fields produced with the MLC.18 Controlling the delivery of a sequence of small doses and static leaf settings was considered to be a straightforward extension of existing linear accelerator control. However, in contrast to DMLC,14–16 a method to avoid tongue-and-groove underdosage has not yet been described for the static technique.

For verification of segmented beam delivery Curtin-Savard and Podgorsak proposed the use of the scanning liquid-filled electronic portal imaging device (EPID).19 Without a patient in the beam, portal images were acquired for each subfield of the leaf sequence and converted into a dose rate distribution. Subsequently, the images were converted into absolute dose distributions by multiplication with the corresponding monitor unit setting. Finally, the individual
dose distributions were summed to produce a dose distribution at the measurement depth. These distributions were then compared with dose distributions predicted by a treatment planning system. Because of the use of the monitor unit setting, the comparison with the dose distribution of the planning system is basically a verification of the relative dose distribution. The applied EPID is relatively slow in returning to the initial state. Therefore, they had to apply a 60 s rest interval between measurement of subsequent segments, yielding long overall measurement times of typically 1 h. Due to the scanned signal readout and the measurement of dose rate instead of dose, the applied EPID is not suitable for high precision dosimetric verification of DMLC.

To test the reproducibility and accuracy of DMLC, the use of film has been reported.20–22 At Memorial Sloan-Kettering Cancer Center film dosimetry is performed in a flat homogeneous phantom for each field prior to the first treatment. Measured dose distributions are compared with corresponding calculated dose distributions.22 A disadvantage of film dosimetry is that it is time consuming since it requires developing and scanning of the film, furthermore a sensitometric curve is needed to convert optical densities into doses.

Charge-coupled device (CCD) camera based EPIDs are a promising tool for verification of DMLC due to their high data acquisition rate and capability to measure simultaneously in all points of the treatment field.23–26 Balter et al. showed preliminary results of a method to derive leaf positions in each camera frame acquired during treatment and to compare them with a table of prescribed leaf positions.26 Leaf positions could be determined with an accuracy of 0.6 mm and a duty cycle of less than 1 s. A similar approach was implemented by Partridge et al.25 They used a custom-made EPID with image acquisition synchronized to the accelerator magnetron current pulse production, with one CCD camera frame acquired per accelerator pulse. Data were presented for a 6 MV beam. The accuracy of the leaf position measurements was 2 mm. Due to a limitation of the camera triggering hardware the pulse rate of the linac had to be reduced. Ma et al. calculated normalized reference images from MLC leaf sequencing files and compared these with normalized images measured with a fluorescent beam imaging system (BIS, Wellhöfer Dosimetrie, Schwarzenbruck, Germany) for a 6 MV photon beam.24 This imaging system can be fastened to the blocking tray holder of a linear accelerator. It was especially designed for quality control tasks. For the calculation of the reference images a measured portal image of a large open field was used. The reference images were therefore not only related to the prescribed fluence, but also contained optical distortions in the EPID system. A global correlation coefficient was used to compare the calculated reference image with the image measured with the BIS system. This method can be used to verify whether the leaf sequencing files have been transferred correctly to the linac control computer and whether the treatment can be correctly executed without machine faults. Transmission through the leaves (both intraleaf and interleaf) was not taken into account and a simple empirical method was used to model extrafocal scatter. Despite these limitations, they concluded that it was possible to detect uncertainties of less than 0.5 mm in leaf position during DMLC.

The above described methods to verify leaf motion cannot be used to check whether the calculated leaf trajectories do indeed generate the absolute beam fluence profiles used in treatment planning. In this paper a new method is presented for a pretreatment verification of these absolute beam fluence profiles utilizing a commercially available CCD camera based fluoroscopic EPID. In the absence of the patient, EPID images are acquired for all beams produced with DMLC. These images are then converted into two-dimensional dose distributions and compared with calculated dose distributions. The calculations are performed with a pencil beam algorithm as implemented in a commercially available treatment planning system (TPS) using the same absolute beam fluence profiles as used in the TPS for calculation of the patient dose distributions. In this paper results are presented for intensity modulated 10 and 25 MV photon beams; both model cases and real clinical cases were studied. Absolute dose profiles measured with the EPID were also compared with ionization chamber measurements. Preliminary results on measurements of absolute dose distributions in IM fields produced with DMLC have been reported.23

II. MATERIALS AND METHODS

A. EPID and ionization chamber dose measurements

The applied EPID was a Philips SRI-100 (Philips Medical Systems, Crawley, UK), which basically consists of a fluorescent screen, two mirrors, and a CCD camera. The fluorescent screen is a 1.65-mm-thick stainless steel plate coated with a layer of gadolinium oxysulphide. To reduce the detection of high-energy electrons generated in patients, an extra 1-mm-thick stainless steel slab has been mounted on the standard fluorescent screen.23 The added slab hardly affects the image quality.27 The acquired images may be used both for setup verification and for in vivo dosimetry.28 The EPID has a fixed focus to fluorescent screen distance of 160 cm. Technical details of the EPID have been described by Visser et al. and Althof et al.29,30

Image acquisition is performed with a procedure written in the macro command language that comes with the system. The integration time on the CCD chip was set to 240 ms and 120 camera frames were accumulated in the frame store memory. The readout time needed to transmit a frame from the CCD to the frame grabber (during which no signal is accumulated) is 80 ms. The final image is the mean of the integrated camera frames, corrected for the dark current measured prior to the irradiation. In the original procedure image acquisition starts automatically when the measured pixel values in the center of the camera frame exceed a threshold, i.e., when the beam is switched on. For the DMLC measurements described in this paper image acquisition was started manually, since with the sliding window technique the pixels in the center of the image are blocked by the leaves when the irradiation starts. The procedure is discussed in detail in another paper.23
Due to sagging of the EPID structure the field center can shift slightly. To correct the image for this shift the position of the field center was derived using the position of the field borders in an EPID image of a (static) square field, which is stored in a lookup table. The raw EPID images of 512×256 pixels were then resampled to arrays of 64×64 elements; each element represents a region of interest with an area of 0.5×0.5 cm² projected at isocenter. Conversion of these arrays into absolute dose distributions was performed in three steps. Acquired images were first corrected for the nonlinear response of the system.23,31 Subsequently, the image was corrected for the optical “cross talk” by deconvolving it with a point spread function.23,32 Finally, the resulting array was divided by an array that accounts for relative EPID sensitivity.23 The system was calibrated once. The observed day-to-day variation of the EPID response per unit of delivered dose is 0.4% (1σ).32 Due to radiation damage to the CCD chip the EPID response gradually decreases (~3%/yr).33 This decrease was carefully monitored and corrected for using the daily acquired images for the quality control of the absolute output and field flatness of the scanning photon beams of the MM50.34 and the two weekly output checks with an ionization chamber.

Dose measurements were also performed with a N31002 ionization chamber (PTW, Freiburg). The ionization chamber was inserted in a polystyrene miniphantom35 at a depth of 2.0 cm for the 10 MV beam and at 2.5 cm for the 25 MV beam. For those depths it was experimentally found that the variations in the on-axis response of the EPID [EPIX pixel value/portal dose measured with the ionization chamber (GID)] were minimal for field sizes ranging from 3×3 up to 18×18 cm² and polystyrene absorber thicknesses ranging from zero up to 35 cm. For a 25 MV photon beam the mean of the standard deviations of the EPID response for all field sizes was 0.4%.23 The miniphantom was scanned in an empty RFA-300 water phantom (Scanditronix Medical AB, Uppsala, Sweden) with the center of the ionization chamber positioned at a distance of 160 cm from the focus, which is equal to the fixed focus to fluorescent screen distance of the EPID.

B. Calculation of the dose distribution at the detector

The starting point for the calculation of the dose distribution at the fluorescent screen of the detector is the optimized beam fluence $F_{\text{opt}}(x,y)$ [in monitor units (MU)] to be delivered at position $(x,y)$, which is the result of a computer optimization or inverse treatment planning. A fluence of 1 MU corresponds to the fluence due to an irradiation that results in a dose delivery of 1 cGy at a depth of dose maximum in water with a source surface distance (SSD) of 100 cm in a static 10×10 cm² field. Using an iterative algorithm described by Dirkx et al., leaf trajectories are then calculated taking into account collimator scatter and the effective leaf transmission, which is the sum of the transmission through the leaves and the extra focal radiation under the moving leaves.15 The algorithm fully avoids tongue-and-groove underdosage effects.14 Generally less than ten iterations are necessary to minimize the difference between optimized [$F_{\text{opt}}(x,y)$] and realized fluence profiles [$F(x,y)$], which are used for calculation of the final dose distribution in the patient with the CadPlan 3D TPS (Varian-Dosetek, Espoo, Finland).

The expected absolute dose distribution in the plane of the fluorescent screen of the EPID, $D_{\text{p,0}}(x,y)$, is calculated from the realized fluence profile using the pencil beam algorithm as implemented in the CadPlan TPS.36,37 In the current implementation (CadPlan v2.7.9) the penumbra width is a linear function of the SSD. As a result the predicted dose distributions would become inaccurate for the SSD of the detector (160 cm), which is much larger than the SSDs clinically used. Therefore, the dose distribution is calculated at 100 cm from the focus by enlarging the field with a factor of 1.6 (160/100). The dose distribution is then normalized using the calculated on-axis dose in a static 16×16 cm² field for 150 MU. Finally, the absolute dose distribution is calculated using the measured cGy/MU value at the detector for a 10×10 cm² field. Dose calculations and the calibration measurement are performed at a water depth of 2 cm for the 10 MV photon beam and at a depth of 2.5 cm for the 25 MV beam, equal to the effective measuring depths of the EPID, as discussed in Sec. II A.

C. Realization and verification of fluence profiles

Measured and predicted dose distributions $D_{\text{p,0}}(x,y)$ were compared for the 10 and 25 MV photon beams of the MM50 racetrack microtron (Scanditronix Medical AB, Uppsala, Sweden). The dose rate was 200 MU min⁻¹ for the 10 MV beam and 300 MU min⁻¹ for the 25 MV beam. The microtron produces 200 radiation pulses per second (pps). The unit is equipped with a double-focused multileaf collimator with 32 leaf pairs. Projected at isocenter the leaf width is 1.25 cm. The maximum leaf speed is 1 cm s⁻¹. For all measurements the leaves moved from left to right parallel to the x axis. During DMLC, every 50 ms the accelerator control system compares the actual leaf positions, measured with potentiometers, with the prescribed positions. If the deviation between a prescribed and a measured position is more than 0.2 cm during three subsequent checks, the irradiation is interrupted. Realization of IM fluence profiles with this unit is discussed in detail elsewhere.15

The developed method was tested for a range of fluence profiles, both model cases and real clinical cases were studied. Calculated absolute dose distributions (Sec. II B) were compared with dose distributions derived from EPID images and with dose profiles measured with an ionization chamber (Sec. II A). The axes for comparisons in the leaf direction were chosen at the center of each leaf pair $(y = \pm 0.6, \pm 1.9, \pm 3.1,...)$. Throughout the paper, positions and distances are defined at the plane normal to the beam axis at 100 cm from the focus. The reported differences are the mean deviation in percent and the corresponding standard deviation in percent (mean ± 1σ [%]).
III. RESULTS

In Figs. 1 and 2 data are presented for intensity modulated 10 MV beams. Within the treatment field the leaf trajectories were identical for all leaf pairs; the presented data are for \( y = 0.6 \) cm. Outside the penumbra, there is an excellent agreement between EPID and ionization chamber measurements for both beams: 0.3 ± 0.6% (Fig. 1) and 0.1 ± 0.8% (Fig. 2). The actual agreement may even be slightly better, since each ionization chamber measurement required the complete irradiation to be repeated. The short term reproducibility of the absolute dose delivery with DMLC at the MM50 racetrack microtron is 0.2%.38 The deviations between the calculated dose profile and the profile measured with the EPID are ±2.1 ± 1.2% and ±0.9 ± 0.8%, respectively.

The data presented in Fig. 3 are for a beam fluence profile that was also used to generate the data in Fig. 2, but now realized with the 25 MV beam. Again deviations between EPID and ionization chamber measurements are small: ±0.2 ± 1.3%. The deviation between the predicted dose profile and the profile measured with the EPID is ±1.0 ± 0.7%. Similar results were found for the profile presented in Fig. 4: ±0.2 ± 1.0% and ±1.0 ± 1.4%, respectively.

In Figs. 5(a) and 5(b) acquired EPID images (512×256 pixels) for two nonsquare IM 25 MV beams are shown. The fluence decreases in the \( y \) direction; the dose at the top of the image is a factor of 2.2 lower than at the bottom. Due to the synchronization of leaf trajectories of adjacent leaves, underdosages (lower pixel values) do not occur in the overlap regions.14 The small overdosages (higher pixel values) in the overlap regions of adjacent leaves are due to the interleaf leakage of about 2%.15 These overdosages can be avoided using partial synchronization.16 Figures 5(c) and 5(d) show cross sections along the \( y \) axis (normal to the axis along which the leaves move) of the two-dimensional dose profile derived from the EPID images shown in Figs. 5(a) and 5(b). Corresponding predicted dose profiles and dose profiles measured with an ionization chamber are included. For the first field [Fig. 5(a)] the deviation between EPID and ionization chamber measurements was 0.2 ± 1.1% and for the second field [Fig. 5(b)] 0.4 ± 2.0%. The deviations between the predicted profile and the EPID measurements were ±0.9 ± 1.7% and ±1.4 ± 1.7%, respectively. Standard deviations are slightly increased due to the interleaf leakage that was measured but not taken into account in the calculations. Under the center of the leaves the deviations are ±0.7 ± 1.2% and ±0.4 ± 1.6%, respectively.

In Fig. 6 results are presented for a two-dimensional IM profile designed for treatment of a prostate cancer patient. The deviation between the predicted profile and the profile measured with the EPID is 0.5 ± 1.1%.

IV. DISCUSSION AND CONCLUSIONS

A procedure for pretreatment verification of absolute beam fluence profiles realized with DMLC was developed.
and tested. The time required to verify an IM beam is about 2 min, which is much shorter than any other dosimetric technique. The EPID system only has to be calibrated once. The agreement between calculations and EPID measurements and between EPID and ionization chamber measurements was within 2% (1σ). The procedure allows an overall verification of (i) the leaf trajectory calculation (including the models to incorporate collimator scatter and leaf transmission), (ii) the correct transfer of the leaf sequencing file to the treatment machine, and (iii) the mechanical and dosimetrical performance of the treatment unit. It is not always possible to distinguish between these types of errors using only portal images. Previously published DMLC verification methods with EPIDs only verified leaf motion or relative dose profiles. The excellent agreement between EPID and ionization chamber measurements in all cases shows that the readout time of 80 ms (during which no signal is collected) has no detectable effect; the data acquisition rate is sufficiently high.

In the near future the developed method will be extended to enable verification of DMLC during patient treatment. Acquisition of portal images suitable for dosimetric verification of DMLC can be fully integrated into existing imaging routines for patient setup verification, without introducing an increase in the overall treatment time. Preliminary results on measurements of portal dose images (PDI), i.e., the dose distribution behind a patient in a plane normal to the beam axis, in an IM beam have been reported. The calculation of a PDI for a patient irradiated with an IM beam is a relatively simple extension of existing methods. The relation is $$D_p(x,y) = D_{p,0}(x,y) T(x,y),$$ with $$D_p(x,y)$$ the predicted portal dose at position $$(x,y)$$ beneath the patient, $$D_{p,0}(x,y)$$ the predicted portal dose in absence of the patient (as described in Sec. II B), and $$T(x,y)$$ the predicted transmission through the patient using the planning CT data. The method for calculation of these transmission functions has been described elsewhere. A potential problem is to distinguish between deviations in predicted and measured PDIs due to machine faults and differences due to deviations between the patient anatomy during acquisition of the planning CT scan and during treatment. For prostate cancer patients we have observed that deviations in patient anatomy introduce large local differences between predicted and measured PDIs. Machine faults are likely to produce a constant difference over the whole irradiation field. In case of malfunctioning of
a single leaf, the difference will be limited to the beam’s eye view of that leaf.

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