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NOTE

IMRT verification with a camera-based electronic portal imaging system

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Abstract. An evaluation of the capabilities of a commercially available camera-based electronic portal imaging system for intensity-modulated radiotherapy verification is presented. Two modifications to the system are demonstrated which use a novel method to tag each image acquired with the delivered dose measured by the linac monitor chamber and reduce optical cross-talk in the imager. A detailed performance assessment is presented, including measurements of the optical decay characteristics of the system. The overall geometric accuracy of the system is determined to be $\pm 0.2$ mm, with a dosimetric accuracy of $\pm 1.25$ MU. Finally a clinical breast IMRT treatment, delivered by dynamic multileaf collimation, is successfully verified both by tracking the position of each leaf during beam delivery and recording the integrated intensity observed over the entire beam.

1. Introduction

With widespread interest in the introduction of intensity modulated radiotherapy (IMRT) delivery techniques in clinical practice, the verification of novel delivery techniques has become an area of significant current importance. Intensity modulated beams can be delivered in a number of different ways (Webb 1997), from the use of physical compensating filters and multiple-static multileaf collimator (MLC) fields to dynamic MLC and ‘tomotherapy’ type deliveries such as those delivered by the NOMOS MIMiC collimator.

The term ‘verification’ can include a number of different types of measurement that it is important to distinguish between (Burman et al 1997, Tsai et al 1998). Possible measurements include:

(a) Commissioning of new techniques and equipment.
(b) Pre-treatment dosimetric checks made either in air or in phantom.
(c) Post-treatment comparison of the integrated fluence delivered with some predicted distribution.
(d) Real-time, on-line comparison of in vivo measured collimator positions with those prescribed.

For accurate type (b) verification, the dose delivered by beams customized for a particular patient must be calculated in-phantom. This is not therefore a true measure of the dose delivered to the patient, but nevertheless it is a valuable quality assurance test of the planning
and delivery process. The analysis of accurate in vivo measurements of integrated fluence is complicated by the effects of changes in patient position or anatomy between planning and delivery, making it hard to distinguish between subtle errors in IMRT beam delivery and patient movement. The importance and complexity of each of these steps will depend, of course, upon the complexity and the degree of confidence in the delivery technique overall.

For many of the dynamic MLC methods currently being clinically implemented, novel MLC hardware and software control systems, highly complex leaf sequencing and inverse planning software are all being applied for the first time. It is therefore desirable to have a verification system that monitors collimator trajectories throughout beam delivery, giving on-line feedback in real-time. Complications also arise with the dosimetry of dynamic MLC beams, with the effects of head scatter, transmission through collimators, leaf penumbras and the small-field output factors all affecting the absolute dose delivered. It is therefore also desirable to have an accurate dosimetric record of the delivery to compare to pretreatment predicted values.

Several types of IMRT verification measurement have been reported in recent years. Commissioning measurements of a dynamic MLC delivery system using film are presented by Wang et al. (1996) and Chui et al. (1996) who also proposed a routine quality assurance protocol using film.

The Varian PortalVision electronic portal imaging system was evaluated for pretreatment checks of dynamic MLC fields by Lutz et al. (1996) (also Chang et al. 2000). They compared integrated fluence images with portal dose maps (Wong et al. 1990), concluding also that an acquisition rate of at least one frame per second was required for accurate results. Other pre-treatment verification measurements are presented by Ma et al. (1997) who used the Wellhöfer BIS to derive a global correlation coefficient from comparing fluence maps; Curtin-Savard and Podgorsak (1999) who show step-and-shoot verification with PortalVision, but claim that it cannot be used for dynamic MLC verification; Pasma et al. (1999) who demonstrate dynamic MLC verification with the Philips SRI-100; and Murmann et al. (1999) who show an agreement of 3–6% between PortalVision measurements and treatment planning system calculations.

Measurements of the collimator trajectories during dynamic MLC beam delivery were presented by Partridge et al. (1998), Paul et al. (1999a, b) then demonstrated the use of an amorphous silicon flat-panel detector (a-Si:H) to verify dynamic MLC delivery by ‘stamping’ the cumulative dose, taken from the linac monitor chamber, on images at 1 MU intervals. These images were also summed to show integrated intensity images. James et al. (2000) presented a working clinical real-time on-line dynamic MLC verification system based on the SRI-100. A series of ‘snap-shot’ images taken at known percentages of the total prescribed dose are used to track the moving collimators and alert the operator if any are out of tolerance. Finally, the performance of the Partridge et al. (1998) and James et al. (2000) imaging systems are compared by Partridge et al. (2000) with a third real-time collimator-tracking system developed at The Netherlands Cancer Institute. The verification of absolute dose delivery, such as that presented by Papatheodorou et al. (2000), is outside of the scope of this work.

In this note, we present an evaluation of the capability of the TheraView camera-based electronic portal imaging system (Cablon, NL) for dynamic MLC treatment verification. Collimator tracking measurements are demonstrated, with digital dose signals taken from the linac monitor chamber encoded on each image with a pattern of light-emitting diodes (LEDs). To improve the dosimetric properties of the as-supplied commercial system, an optical antiscatter grid was added to the imager (Partridge et al. 1999). A detailed study of the geometric and dosimetric properties of the modified system is then presented, including measurements of geometric distortion, linearity and afterglow in the scintillation screen/camera system, long- and short-term camera stability and field size effects. The modified system has
possible uses for both machine quality assurance and on-line verification of treatment delivery (by leaf-tracking and the comparison of integrated intensity maps with portal-dose images). Finally the system is demonstrated with the verification of a clinical breast case.

2. Materials and methods

The work described by this paper is concerned with determining the accuracy with which dynamic MLC beam delivery on Elekta SL series linacs can be verified using the TheraView camera-based electronic portal imaging system. All measurements presented are from an SL15 and were recorded using 6 MV photon beams.

2.1. Imager modifications

The standard TheraView imager, employing a Plumbicon tube camera, is described by Munro et al (1990) and is the starting point for this work. Two simple modifications were made to the standard imager. The first is the addition of an optical antiscatter grid (3M light control film, 3M Corporation). Multiple reflections between the mirror and scintillation screen produce a field-size-dependent glare signal that is strongly peaked towards the centre of the image for square fields (Partridge et al 1999). If the system is dosimetrically calibrated using a large-area field to cover the whole imager (which is a common practice), the optical scatter signal is a maximum. Subsequent images of smaller fields then have reduced scatter contributions and therefore a smaller fluence is seen than would be expected from the calibration set. The optical antiscatter grid minimizes multiple diffuse reflections between the mirror and scintillation screen, thereby making dosimetric calibration simpler. The grid is placed flat on the surface of the scintillation screen, sandwiched between a thin transparent PTFE membrane, to prevent it from sticking to the screen, and a mylar window glued to an aluminium frame for mechanical
support (see figure 1). Alternative methods of reducing the optical scatter contribution of the image have been presented by de Boer et al (2000) who use a ‘high elbow’ version of the imager, or Pasma et al (1998) who use a mathematical deconvolution technique. When suitable antiscatter measures are taken similar signals should be observed for all field sizes. To test this, a series of images of a uniform slab of water-equivalent plastic were taken for different sized fields. The images were then corrected using the quadratic calibration method (Morton et al 1991) and line profiles taken to check for flatness in the AB and GT directions (parallel to and normal to the plane of gantry rotation respectively). The average value over a small central region of interest was taken to check the absolute accuracy.

The Elekta dynamic MLC prescriptions are composed of a set of control points. These points specify the position of each leaf at a set of known percentages of the total number of monitor units to be delivered for each beam. To be able to verify the position of any given leaf in a image, the number of monitor units delivered when that image was acquired must therefore be known. The monitor chamber circuitry of the Elekta linacs produces a digital pulse every 1/64th of a monitor unit (MU). This pulsed signal is the fundamental dosimetry signal of the treatment machine and was therefore chosen to ‘tag’ each camera frame during image acquisition. Without access to the imager’s data acquisition hardware or software, some method of tagging each image with this dose signal is required. The method chosen here was to count the number of digital dose pulses and display the cumulative dose on a counter mounted inside the imager along the very edge of the scintillation screen. The count value in each of four decades is indicated by one of ten LEDs being lit. In this way each image is tagged with the cumulative dose by a series of dots running down one edge. The decade display allows correct decoding even when the exposure is several dose counts long (this would not, in general, be the case using a conventional numeric display). The dot patterns are decoded automatically in software. The pulse count can be reduced using a divider, programmable in powers of 2, to increase the range of the display.

2.2. System characterization—mechanical

For accurate geometric and dosimetric verification, the basic operation of the imaging system has to be carefully characterized. A measure of the intrinsic accuracy of static measurements made with the imager is essential if the accuracy to which a dynamic beam delivery can be verified is to be estimated. The first characteristic to be measured is the spatial resolution of the system. This was carried out using the method reported by Shalev et al (1997). Pairs of images of a phantom containing a series of bar targets are acquired and used to calculate the 50% cut-off point of the relative modulation transfer curve.

Geometric distortion was measured using a specially constructed phantom consisting of a 300 × 300 × 10 mm steel slab with an array of 49 12 mm holes on a regular square grid. The phantom was aligned at the isocentre, exactly perpendicular to the beam, with the axes of the hole-grid parallel to the room lasers and the gantry at 0, 90, 180 and 270°. To check repeatability and backlash, the 270° measurement was repeated, approaching 270° from each direction. The centroids of the images of the holes were compared with a perfect rectangular grid using an automatic image processing routine and the resulting shift vectors calculated. A note of caution should be added about the motorized arm supporting the imager. Errors in the indicated source to surface distance of the imager lead to small errors in magnification. Perhaps more seriously, errors in the extension of the imager (GT direction) lead to a misplaced isocentre and, because of changes in the camera to scintillation screen distance, larger changes in magnification. If care is taken to make independent checks of the imager position, these errors should be negligible.
The effect of any motion of either the head of the linac or the imager with respect to a fixed isocentre during gantry rotation (gantry sag), was investigated by placing a pointer at the isocentre, fixed with respect to the treatment room, and taking images of the pointer at 0, 90, 180 and 270°.

2.3. System characterization—dosimetric

To test the dosimetric accuracy of the imager, a point-by-point quadratic calibration was carried out (Morton et al 1991) by taking a series of images of water-equivalent plastic blocks with well characterized water-equivalent thicknesses \( t \) ranging from 4.56 to 27.36 cm. The quadratic coefficients for a given pixel of intensity \( I \) are determined by fitting the following equation, converting to radiological thickness \( t \):

\[
\frac{I}{I_0} = \exp(-\beta t - \gamma t^2)
\]

where \( I_0 \) is the corresponding pixel intensity in an open field image. Patient images are converted to radiological thickness by inverting the above equation, taking account of changes in output factor due to field size by using a local open field image. This calibration method relies on the imaging system showing good absolute short term stability, i.e. no changes in system response between taking the open field and patient images, and good relative long-term stability (between acquiring the calibration set and patient image). These two parameters were measured by taking a series of images of a uniform slab of water-equivalent plastic with corresponding open field images and converting each to radiological thickness. Images were taken over a period of 90 min to check short-term stability and 3 months to check long-term stability. Care should be taken to avoid the ‘warm-up’ period when the camera has just been switched on (Glendinning and Bonnett 2000). For the system presented here, the camera is never normally switched off.

The ‘afterglow’ in the imaging system is a measure of how the observed signal in a given frame is affected by previous frames from a combination of persistence in the phosphors of the scintillation screen and the camera tube. This was measured in two separate ways (a) by preirradiating the system with 50 MU, waiting a variable period of time and then acquiring an image and (b) by preirradiating the system with differing numbers of monitor units, waiting a fixed period of time, and then acquiring an image. If the afterglow is dominated by the persistence of a single element of the imaging system, the contribution from the primary signal at a given instant to subsequent frames is expected to follow a simple exponential decay. For the second case both the excitation of the phosphor and the rate of decay have to be taken in to account and the system behaviour is described by the following differential equation

\[
\frac{dI}{dt}_{\text{excitation}} + \mu I_{\text{decay}} = C
\]

where \( C \) is constant and \( \mu \) is the rate of decay. Considering the case where the system is excited by irradiation for time \( \tau \), the radiation interrupted for some other time \( t \) and an image acquired, then the expected intensity, from equation (2), becomes

\[
I(t) = \left(\frac{C}{\mu} + D e^{-\mu \tau}\right) e^{-\mu t}
\]

where \( D \) is constant.
2.4. Sample delivery—IMRT breast field

As the final test of the system an intensity-modulated beam was taken from a clinical tangential breast treatment case and delivered using the ‘sliding window’ dynamic MLC technique. The resulting delivery was imaged using a constant integration time of 0.25 s, determined by the dynamic range of the camera, with frames recorded every second. The intensity-modulated beam was delivered at a dose rate of 200 MU min\(^{-1}\) resulting in maximum leaf speeds of about 6 mm s\(^{-1}\) in the plane of the isocentre.

3. Results

All results presented in the following section were acquired using a source to detector distance of 150 cm and a 6 MV photon beam at a dose rate of 200 MU min\(^{-1}\). The integration time of 0.25 s was set to use the maximum dynamic range of the camera. The calibration was performed using six blocks with a water-equivalent thickness of 4.56 cm, placed with the exit surface at a distance of 82 cm from the source. All field sizes are quoted in the plane of the isocentre.

3.1. Spatial resolution and distortion

The imager spatial resolution is not significantly degraded by the presence of the antiscatter film. The value for the 50% cut-off of the relative MTF curve, or \(f_{50}\) point with the antiscatter film in place, is 0.267 line pairs/mm, compared with a value of 0.276 without (Partridge \textit{et al} 1999). These values compare well with a published value for this imaging system of 0.231 ± 0.11 (Shalev \textit{et al} 1997). It should be noted that this TheraView system, installed on Elekta linacs, has a slightly shorter camera to scintillation screen distance than for the Shalev published value (≈120 cm compared with ≈140 cm), making a direct comparison impossible.

The major geometric distortion in the system was found to be a rigid body rotation of 1.2 ± 0.4° about the optic axis of the system. This rotation was found to be independent of gantry angle. The residual distortion present after rotation correction is less than 0.5 mm in the central 10 × 10 cm of the field of view and less than about 1 mm over the rest of the image. These results agree show broad agreement with the findings of Glendinning and Bonnett (2000) who estimate geometric distortion to be less than ±2 pixels (roughly ±1 mm) for a similar system. For the purposes of IMRT verification, this level of accuracy was considered to be acceptable. The AB and GT defections due to gantry sag were found to be relatively repeatable and independent of rotation direction. They can therefore be approximated by rigid body transformations that are simple sinusoidal functions of gantry angle. Following systematic corrections for image rotation and translation, the overall geometric accuracy of the system was estimated to be in the region of ±2 mm.

3.2. Dosimetric resolution

The digital dose signal used for tagging each image with the cumulative dose proved to be ideal in many ways. The acquisition of the first TheraView frame is triggered by waiting for the imager to ‘see’ a signal; this inevitably introduces some uncertainty as to the exact elapsed time and delivered dose at the very start of each image sequence. However, because the dose tagged onto each image is derived from counting digital pulses from the monitor chamber, exactly the same signal as is used by the MLC control computer, there should be no systematic error introduced (except for the ‘reaction time’ of the MLC control system). A small random error is, however, introduced by the effect of the integration time of the imager
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Figure 2. Stability of the imaging system over (a) short periods and (b) long periods of time. Error bars indicate 1 SD on experimental measurements, and the broken lines mark the ‘true’ radiological thickness.

The dosimetric accuracy of the imaging system itself was determined by performing a full quadratic calibration of the system, and then using this calibration set to convert images of uniform water equivalent plastic blocks to radiological thickness using equation (1). Figure 2 shows the stability of the calibration over short periods of time (upper panel) and long periods of time (lower panel). For the short-term data, a single $I_0$ value is used, taken from the start of the time series. For the long-term data, a separate $I_0$ is used for each data point. A single quadratic calibration set is used for all data (corresponding time $t = 0$ for both panels). The thicknesses quoted are averages over a $4 \times 4 \text{ cm}$ region of interest centred on the isocentre, with
error bars showing one standard deviation. The first thing to note from figure 2 is that random noise in the camera is quite large, and effectively limits the dosimetric resolution to about ±5 mm water (±2.5% dose). The camera is, however, quite stable and shows no significant drift within the random errors over both the short- and long-term data sets. Again these results are in good agreement with the findings of Glendinning and Bonnett (2000) who show good camera stability. The gradual linear reduction in sensitivity they report of 0.04% day⁻¹ is not observable for the long-term data shown in figure 2 due to the large random error on these measurements.

Figure 3 shows the effect of the optical scatter on dosimetric calibration. The imager was calibrated for a 26.6 × 26.6 cm field, covering its field of view. Images of a uniform water-equivalent block for a series of smaller field sizes were converted to radiological thickness using the calibration set derived from the large-area field. Corresponding $I_0$ values were measured.
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Figure 4. Decay of afterglow signal. 50 MU were delivered to the imaging system, the radiation interrupted for the period of time shown on the x-axis and an image taken with a short 3 MU irradiation. The imager response is normalized to an effectively infinite time delay (i.e. no afterglow).

for each different field size and radiation scatter corrections were performed using the methods described by Swindell and Evans (1996). The upper panel shows the resulting values for the modified imaging system, with every point being identical to the true radiological thickness (dotted line) within the errors on the measurement. This is in contrast to the case in the lower panel, where no measures are employed to counter the effects of optical scatter, resulting in thickness errors of up to 2 cm (15% dose). The optical scatter rejection measures clearly work well, giving a system response that is independent of field area. Line profiles taken in the AB and GT directions from the calibrated images with scatter rejection also show the images to be flat to within the accuracy of the measurement.

The afterglow measured in the system is significant, as seen in figure 4 with the observed signal decay (error bars) following a simple exponential model (broken curve). The experimentally determined decay constant \( \mu \) is 83 ms\(^{-1}\) (i.e. decay time of 12 s), compared with a value for a simple Gd\(_2\)O\(_2\)S screen of 3 ms, indicating that the major contribution to the afterglow is probably a function of persistence in the camera tube rather than scintillation screen. Figure 5 shows similar behaviour, with the observed signal gradually building up with preirradiation of the system, and hence excitation. The decay constant calculated from the data shown in figure 4 is used in equation (3) to produce the broken curve fit on figure 5, illustrating that the two sets of results are consistent. Values of the constants \( C \) and \( D \) are 50.1 s\(^{-1}\) and \(-58.5\) respectively. Although the afterglow signal can be large, contributing up to 30% total measured signal, if all images are acquired using exactly the same camera integration times and frame rates, it should be constant, and therefore will be taken into account by the standard calibration described earlier.

3.3. Sample delivery

Figure 6 shows the measured and prescribed trajectories of one of the central MLC leaf pairs (leaf pair 20) for a clinical IMRT breast field. Leaf positions are extracted from images
Figure 5. Rise of afterglow signal with preirradiation. A dose given as shown by the x-axis was delivered, the irradiation interrupted for 15 s, and an image taken with a short 3 MU irradiation. The imager response is normalized to infinite preirradiation (i.e. equilibrium conditions).

Figure 6. Leaf trajectories of the central leaf pair measured for a clinical IMRT breast treatment beam. Error bars show the experimentally measured points, the full curves show the prescribed leaf trajectories.

recorded every second; the error bars shown are the ±1.25 MU and ±2.0 mm limits derived above. Clearly, from the figure, excellent agreement is seen at all times. Because the accuracy of each point measurement is independent of the frame rate, exactly analogous results are observed for the lower frame rates.

Integrated intensity measurements, summed over an entire frame sequence, are presented in figure 7. At 1 frame/s, with a 0.25 s integration time, the duty cycle is 25%. A horizontal line section through the centre of the integrated intensity image (i.e. along the axes of leaf pair 20) is presented in figure 8 with the ideal intensity-modulated beam overlaid for reference, normalized
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Figure 7. Integrated intensity image of a clinical IMRT breast treatment beam acquired with a 0.25 s integration time and a frame rate of 1 frame/s.

Figure 8. Horizontal line section through the centre of the integrated intensity image shown in figure 7. The ideal beam is taken directly from the planning system and normalized at the isocentre. The good agreement shown between the measurement and prescription indicate that the integrated intensity is a good approximation of the incident fluence. Note that significant sampling errors begin to be seen for frame rates of less than 1 frame/s, but further discussion of this is outside of the scope of this note.

4. Discussion

For the work presented in this note, the authors had no access to the imager’s acquisition hardware and software. Simple modifications to the scintillation screen were carried out to reduce optical scattering and therefore make dosimetric calibration simpler. This visual display of dose signal, ‘tagging’ each image using digital pulses from the linac monitor...
chamber, is straightforward to implement and was shown to give an accurate and robust signal. These techniques are readily applicable to any camera-based imaging system. The processing presented was carried out off-line using hardware and software independent of the imaging system. Construction of a real-time on-line system based on the TheraView imager has been shown to be feasible in principle, but would require collaboration with the manufacturer in practice.

For the clinical case presented, simultaneous verification by both leaf-tracking and integrated intensity measurement are successfully demonstrated. For the integrated intensity measurements, the system must be operated at its maximum frame rate of 1 frame/s, this is agrees with the findings of Chang et al (2000), who also recommend a frame rate of 1 frame/s. The spatial and dosimetric resolution of the system of ±2 mm/±1.2 MU permits verification to within the limits of ±3% dose or ±3 mm displacement of an isodose line, whichever is lower, as used by Wang et al (1996), based on the study by Masterson et al (1991). It should be noted that integrated intensity measurement can also be used to verify non-dynamic MLC IMRT deliveries such as multiple static fields or the use of physical compensators. The dosimetric resolution of the imaging system itself (rather than the dose readout from the monitor chamber) is limited by the camera sensitivity to about ±5 mm water (±2.5% dose) for the acquisition parameters used in this study. A careful assessment of the dosimetric properties of the TheraView system is presented by Glendinning and Bonnett (2000), including measurements of geometric distortion, gantry sag and afterglow similar to those reported here.

5. Conclusions

A detailed study of the IMRT verification capabilities of the TheraView electronic portal imaging system has been presented, including measurement of the optical decay characteristics of the system. Two modifications were performed to the system. The first was shown to effectively eliminate multiple optical scattering within the imager, making dosimetric calibration straightforward. The second uses a novel method to tag each image with the cumulative dose as recorded by the linac monitor chamber, thus allowing verification of absolute leaf positions at any time during beam delivery. The TheraView system is demonstrated to be capable of verifying collimator positions to an accuracy of ±2.0 mm and ±1.25 MU and is therefore suitable for making IMRT verification by leaf tracking. Simultaneously acquired integrated intensity images from a clinical breast case are also presented, showing good agreement with prescribed primary fluences if frame rates of 1 frame/s are used.

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