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Implementation and evaluation of a transit dosimetry system for treatment verification

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ABSTRACT

Purpose: To evaluate a formalism for transit dosimetry using a phantom study and prospectively evaluate the protocol on a patient population undergoing 3D conformal radiotherapy.
Methods: Amorphous silicon EPIDs were calibrated for dose and used to acquire images of delivered fields. The measured EPID dose map was back-projected using the planning CT images to calculate dose at pre-specified points within the patient using commercially available software, EPIgray (DOSIsoft, France). This software compared computed back-projected dose with treatment planning system dose. A series of tests were performed on solid water phantoms (linearity, field size effects, off-axis effects). 37 patients were enrolled in the prospective study.
Results: The EPID dose response was stable and linear with dose. For all tested field sizes the agreement was good between EPID-derived and treatment planning system dose in the central axis, with performance stability up to a measured depth of 18 cm (agreement within −0.5% at 10 cm depth on the central axis and within −1.4% at 2 cm off-axis). 126 transit images were analysed of 37 3D-conformal patients. Patient results demonstrated the potential of EPIgray with 91% of all delivered fields achieved the initial set tolerance level of Δ_D of 0 ± 5-cGy or $\%\Delta_D$ of 0 ± 5%.
Conclusions: The in vivo dose verification method was simple to implement, with very few commissioning measurements needed. The system required no extra dose to the patient, and importantly was able to detect patient position errors that impacted on dose delivery in two of cases.

Introduction

It is now recommended by various national and international organisations that in vivo dosimetry monitoring should be undertaken by each radiotherapy centre [1–3]. In vivo dose measurements when compared to planned doses can spot errors in patient set-up, data transcription, machine fault or anatomical changes which may lead to over- or under-dosage of the tumour volume and unplanned toxicity. Point detectors such as diodes and thermoluminescent dosimeters are commonly used for these measurements. However, point detectors are sensitive to positioning errors particularly in highly modulated fields [4]. Such detectors are placed on external contours of patients, but it is known that external displacements can differ from internal tumour displacements up to approximately 2 cm [5,6].

Within the radiotherapy physics community EPID dosimetry is widely seen to have the potential to become an accurate and efficient means of large-scale patient specific dose verification [7–10]. A recent paper reported that between 2005 and 2009 the treatment plans of 4337 patients were verified using in vivo EPID dosimetry, among which 17 serious errors were detected [11]. Of these, 9 would have been missed if no treatment verification had been performed, thereby highlighting the importance of the method.

Current EPID technology for transit dosimetry is based upon passive, amorphous silicon (a-Si) flat panel imagers. Previous approaches have considered CCD camera based systems and liquid filled ionisation chamber matrixes. a-Si has proven popular due to its superior image quality and dose characteristics (linearity, uniformity, dose-rate dependence, field size dependence, relative dosimetry) [12–14].

Although of great use, EPID in vivo verification technology is still being developed and is not routinely available. A number of departments have implemented in vivo EPID dosimetry using

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in–house solutions [15–17]. The availability of a new commercial software EPIgray (DOSIsoft, France) has enabled us to use EPID devices as a portal dosimeter for in vivo dose verification. In this work the algorithm was implemented to calculate the point dose within any pre-specified location(s) within the patient, based on a recent publication which presents a method for back-projection of measured EPID fluence map to the plane of the patient [18]. A series of phantom measurements were undertaken to verify the algorithm's performance under a range of beam conditions. A review board approved study was undertaken at the Royal Berkshire NHS Foundation Trust to prospectively evaluate the algorithm's performance on a patient population, giving important test conditions that phantom studies cannot (namely anatomical changes including weight loss/gain, organ motion, tumour size changes etc.). This study was used to gain insight into clinical issues that come into play upon clinical implementation of EPID-based in vivo dose verification.

Materials and methods

Commissioning EPIgray

EPIgray version 2.0.3 software was used along with Elekta Synergy® and Precise linear accelerators running Integrity (Elekta, Crawley, UK) equipped with portal imaging system iViewGT a-Si panels (PerkinElmer, USA), and motorised 60° physical wedge. The sensitive layer consists of 1024 × 1024 pixels with a pitch of 400 μm, resulting in an active area of 409.6 × 409.6 mm^2 [19]. All measurements were performed with a nominal dose rate of 400 MU/min. Three linear accelerators were used for the patient measurements: the machines were matched to within Elekta specifications; for photons this is 1% for depth doses and output factors and 2% for profiles, although for most of the tested clinical fields upon commissioning, the match was found to be better than this. The treatment planning system (TPS) was Eclipse version 10.0 (Varian MS, Palo Alto, USA) with Analytic Anisotropic Algorithm (AAA). IMPAC/Mosaiq record and verify system version 2.41 (Elekta, Crawley, UK) was used to transfer plan parameters to the linear accelerator control system. Image datasets of patients and phantoms used were acquired with a GE LightSpeed 16 CT Scanner.

Prior to EPIgray commissioning the EPIDs were prepared for dosimetric measurement, according to manufacturer guidelines to ensure that all pixels have a similar response to a given irradiation and mechanical accuracy within 1 mm. The dosimetric properties of iViewGT EPIDs have been extensively investigated elsewhere [20].

The commissioning of EPIgray required configuration data for the CT, linear accelerator and EPID device. The HU to electron density calibration curve of the CT scanners was required. In order to model the linear accelerator the following were incorporated into the EPIgray beams library for each energy: beam profiles, percentage depth dose and quality index of open and wedged fields. A 10 × 10 cm^2 calibration image was taken for each energy with zero phantom thickness (through the treatment couch) for 100 MU to obtain a calibration factor (CF) that converted EPID signal to dose in water. A diagonal dose profile was input into the beams library to perform an equalisation correction for the EPID image, and linearity correction factors calculated to correct the nonlinearity of the EPID with MU variation. EPID measurements were performed to acquire linearity factors at 10–500 MU, 10 × 10 cm^2 field, zero phantom thickness and calculated as:

$$\text{Correction}_{\text{MU}} = \left[\frac{\text{image value}_{\text{MU}}}{\text{image value}_{100\ \text{MU}}}\right] \times \left[\frac{100\ \text{MU}}{\text{MU}}\right] \quad (1)$$

Couch transmission correction was commissioned per manufacturer guidelines which involved (i) measurement of an open field at 0 gantry angle through 20 cm thick solid water, (ii) measurement of the same field through the same phantom thickness at 90 gantry angle. The latter measurement provided a couch transmission value through the entire width of the couch. An average couch transmission factor derived by the manufacturer from data from (i) and (ii) was applied to all 3D conformal fields as per the EPIgray algorithm.

Finally, finite Tissue Maximum Ratio (fTMR) measurements were acquired and input into the EPIgray library, as guided by the manufacturer. fTMR is the ratio between two doses measured in a phantom at d_{max}. The numerator is the dose measured in the presence of an absorber of thickness '*t*' and the denominator is the dose measured in the same conditions without an absorber [18]. fTMR is measured in the presence of an absorber of finite thickness as opposed to TMR where the phantom dimensions are infinite. Commissioning fTMR measurements consisted of ion chamber dose measurement at source to chamber distance of 160 cm (at EPID position) within a water tank at depth d_{max} for each energy. fTMR measurements were performed for three overlying absorber solid water set-ups: with SAD of 100 cm to (i) top, (ii) centre and (iii) bottom of the solid water. Commissioning measurements were over five square field sizes (2 × 2 cm^2 to 25 × 25 cm^2), each for six solid water thicknesses (0–40 cm) across the three aforementioned solid water SDDs. In addition EPID images were acquired at each solid water/field size/energy configuration for set up (ii).

For all EPID images a dicom image and log file were needed; the latter provided the pixel scaling factor (PSF) for each image; this value held the dosimetric information of the pixel values, as iViewGT performed a greyscale normalisation to ensure that an image acquired with 5 MU has approximately the same grey level as a 500 MU image. Un-normalised pixel values $s(x,y)$ were obtained from normalised values ($s^*(x,y)$) as in Eq. (2):

$$s(x,y) = \frac{(2^{16}-1) - s^*(x,y)}{\text{PSF}} \quad (2)$$

EPIgray dose calculation process

The calibration factor CF measured during commissioning converted EPID dose to dose in water at d_{max}. Dividing by the appropriate finite tissue maximum ratio (fTMR) removed the effect of the overlying patient. Within the software fTMR values were expressed in a look up table of coefficient values listed for patient thickness (5–40 cm), field size defined at the EPID (3.2 × 3.2 cm to 40 × 40 cm). An inverse square law correction was then used to recover dose at the position of the pre-specified dose calculation point P (at depth of maximum dose d_{max}) from dose at the EPID (SDD = 160 cm). A TMR correction shifted the reconstruction point from d_{max} to point P. The fluence matrices for TMR and fTMR were calculated by individually computing the primary and scatter component. The primary component was yielded by attenuation correction of the fluence matrix (modelled using PDD and fluence profile) using attenuation coefficients derived from the quality indices provided during commissioning. The software contained a look up table of computed scatter factors as a function of field size of subsector and depth of point P. Peak scatter factor (PSF) values set within the software data tables were used to calculate the overall scatter factors; PSF describes dose to scatter only (defined as: dose in tissue at d_{max}/dose at d_{max} due to primary radiation only). PSF values for our centre were provided by the manufacturer and obtained from the quality indices of each beam energy provided

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