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A comparison of electronic portal dosimetry verification methods for use in stereotactic radiotherapy

Anthony E. Millin *, Rebecca S. Windle, D. Geraint Lewis

Physics Department, Velindre Cancer Centre, Velindre Road, Cardiff CF14 2TL, UK

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ABSTRACT

Three methods of transit dosimetry using Electronic Portal Imaging Devices (EPIDs) were investigated for use in routine in-vivo dosimetry for cranial stereotactic radiosurgery and radiotherapy. The approaches examined were (a) A full Monte Carlo (MC) simulation of radiation transport through the linear accelerator and patient; (b) Calculation of the expected fluence by a treatment planning system (TPS); (c) Point doses calculated along the central axis compared to doses calculated using parameters acquired using the EPID. A dosimetric comparison of each of the three methods predicted doses at the imager plane to within ±5% and a gamma comparison for the MC and TPS based approaches showed good agreement for a range of dose and distance to agreement criteria. The MC technique was most time consuming, followed by the TPS calculation with the point dose calculation significantly quicker than the other methods.

Introduction

Stereotactic radiotherapy has been used in the treatment of tumours both intra-cranially and extra-cranially for some years. The increased spatial accuracy of such techniques allows treatments with higher fractional doses and smaller fields than would generally be given in conventional radiotherapy. Due to their small dimensions, these fields may be made up of largely penumbral effects and have large dose gradients across their extent. Therefore, any dosimeter placed in the field will exhibit increased measurement uncertainties due to errors in the positioning of the detector and a potentially uneven dose gradient across it.

In order to reduce the possibility of errors in the treatment process, in addition to standard independent monitor unit calculations, in-vivo dosimetry techniques have been implemented that measure the dose entering the patient or exiting the patient while on treatment. In this way any errors in dose calculation or in the transfer or data to the machine may be identified and if discovered early enough in the process rectified. Recent publications such as "Towards Safer Radiotherapy" [1] have recommended the use of in-vivo dosimetric techniques for all patients undergoing radical radiotherapy. Although stereotactic techniques are not explicitly included in this the use of high doses per fraction may make it desirable for centres to include stereotactic radiotherapy amongst the techniques benefitting from in vivo dosimetry.

Conventional in-vivo dosimetric techniques, such as those involving diodes and thermoluminescent detectors (TLDs), are likely to suffer large uncertainties due to the spatial issues outlined above when applied to stereotactic radiotherapy and dosimetric verification would therefore benefit from an alternative approach, in which some or all of these uncertainties are removed. The application of EPIDs to transit dosimetry has previously been reviewed [2], with the conclusion that many authors have been able to show good agreement between dose measured using an EPID with that predicted using a variety of approaches applied to conformal therapy, IMRT and VMAT. Little work however has been done in adopting these techniques to stereotactic radiotherapy, where the difficulties in using conventional in-vivo dosimetric techniques may be more pronounced. Therefore the adoption of one of these EPID based methods for dose verification of stereotactic techniques may be seen as a solution to the problem of introducing routine in-vivo dosimetry.

Any such system requires the ability to compare the dose measured using the dosimeter with that expected from the dose calculation. Using EPID based techniques, several approaches to achieve this have been described by several authors including [2–6], which can be summarised as:

(1) Full Monte Carlo transport of dose through the patient to the imager plane with possible back projection to the plane of the isocentre.
(2) Calculation of the expected dose at the imager plane using a clinical treatment planning system with possible back projection to the plane of the isocentre.
(3) Calculation of the dose at a single point at the imager plane using an equivalent depth method.

* Corresponding author. Physics Department, Velindre Cancer Centre, Velindre Road, Cardiff CF14 2TL, UK. Tel.: +44 (0)29 20615888; fax: +44 (0)29 20316293.
E-mail address: tony.millin@wales.nhs.uk (A.E. Millin).

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Each of these methods has strengths and weaknesses; the full MC transport of the dose through the treatment head and patient to the imager plane is completely independent of the planning system calculation and with a good model of the accelerator and patient is likely to achieve a high degree of accuracy. However these simulations are time consuming in preparation of the electronic phantoms used in the simulation and in the simulation itself. Those using the TPS to calculate the expected dose are likely to be quicker to calculate but may suffer inaccuracies due to weaknesses in the calculation algorithms generally used for treatment planning purposes. A single point method is likely to be the quickest of the three approaches but only calculates the dose at a single point, which may not be desirable.

A study was therefore undertaken to compare these three approaches for use in in-vivo dosimetry in cranial stereotactic radiotherapy.

Methods and materials

Portal dosimetry

Cranial stereotactic radiotherapy was delivered at our centre using 6 MV beams using a Varian[1] 600c linear accelerator with a tertiary BrainLAB[2] M3 μMLC and planned using the BrainLAB iPLAN treatment planning system. Portal images were taken with the Varian aS500 amorphous silicon EPID incorporated into the machine. Images were taken on every fraction at a standard source to detector distance of 1400 mm.

During routine calibration of the Varian EPID a *dark field* and *flood field* are acquired in order to account for temporal drift of the electrometers attached to the imager and individual sensitivities of active elements of the detector array. The dose obtained from the imager is therefore obtained using the following equation:

$$Dose_{epid}(x,y) = \frac{Epid_{signal}(x,y) - dark_{field}(x,y)}{flood_{field}(x,y)} \times flood_{mean} \times d_{corr}(x,y) \quad (1)$$

where

- $Dose_{epid}(x,y)$ is the dose measured by the detector at a point (x,y) on the imager panel
- $Epid_{signal}(x,y)$ is the signal measured by the EPID at (x,y)
- $dark_{field}(x,y)$ is the dark field measured by the detector with no incident on the detector at (x,y)
- $flood_{field}(x,y)$ is the flood field measured with the largest possible field incident on the detector (x,y)
- $flood_{mean}$ is the mean value of the flood field image.
- $d_{corr}(x,y)$ is the EPID signal to dose correction factor

The factor $d_{corr}(x,y)$ was obtained by irradiating the imager with known doses, calculated from previously measured tables with the EPID placed at the standard source to detector distance of 1400 mm, used in subsequent data acquisition. This ensures that effects due to back scatter of radiation from the detector arm into the sensitive part of the imager (as described by [7]) were constant.

In order to compare the methods three sample patients were considered. Each of these received cranially stereotactic radiotherapy for brain metastases using 5 or 6 coplanar static fields according to a plan devised using the BrainLAB iPLAN planning systems. All patients were immobilised using the BrainLAB mask system and received a dose of 28 Gy in 4 fractions. On each fraction portal images of each beam were acquired over the entire size and duration of the treatment field and compared with the predicted doses from each of the three methods. A total of 16 beams were used for each method in the comparison.

Method I – full Monte Carlo transport

A Monte Carlo (MC) model of our stereotactic facility incorporating a model of the accelerator head and a bespoke model of the BrainLAB μMLC was developed and validated [8,9] using the BEAMnrc [10] and DOSXYZnrc [11] Monte Carlo codes based on the general EGSnrc code [12]. Examples of the agreement between measured and simulated commissioning results are shown in Fig. 1. In this case results were obtained by simulating the radiation transported through a 98 mm diameter circle μMLC shape shown in Fig. 1a, incident on a water phantom at 900 mm focus to skin distance (FSD). Dosimetric comparisons (Fig. 1b) show excellent agreement between the two data sets even in the penumbra of the beam, for example, in the profile, 50 mm off axis of Fig. 1b.

In order to simulate each clinical case, a virtual phantom was derived for the relevant patient from the stereotactic planning CT scans, which were used as the basis for the MC simulation. Stereotactic localisation was achieved by scanning the patient in a stereotactic frame, from which stereotactic coordinates were generated by the planning system. This frame was not in place for treatment, therefore using software methods, the image of the stereotactic localisation frame used in the planning scans was removed and the patient data extracted and centred around the isocentre derived from the clinical plan. The phantom was then rotated around the isocentre by the relevant gantry angle. A model of the EPID (derived from confidential data provided by the manufacturer) was then added to the scan at the predefined distance to be used in the acquisition of the image during treatment. The support arm was modelled using a slab of water added to the phantom in a method identical to that described by [13]. The rotated patient scan and EPID assembly can be seen in Fig. 2.

An input file containing the treatment beam parameters was then generated and used, together with the phantom information submitted to the RTGrid [14]. This is a web based portal, designed to distribute radiotherapy MC simulations amongst networked computing resources. In order to achieve a balance of the size of the computing task and to reduce total simulation time, the phantoms were re-sampled from the resolution of the raw image (0.5 mm–2 mm depending on image size) to a fixed resolution of 1 mm by 1 mm in the x-y direction. In the z direction (normal to the image plane in Fig. 2), a resolution of 1 mm was used within the phantom, a single voxel for the air gap between the phantom and the imager plane (approximately 300 mm) and voxels of identical dimension to the imager component thickness within the EPID itself. For each beam on each patient the simulation was split into 50 jobs with a total of 10^8 particles used. Approximately 200 CPUs were available when the simulations were run enabling uncertainties of approximately 1.5% in about 48 hours of real time per plan. Following simulation the dose per incident particle was converted to dose to the imager plane according to the equation

$$MCDose_{epid}(x,y) = \frac{MCEpid_{simulated}(x,y)}{MCflood_{simulated}(x,y)} \times MCflood_{mean} \times MCd_{corr} \quad (2)$$

where

- $MCDose_{epid}(x,y)$ is the dose simulated in the detector at a point (x,y) on the imager panel

[1] Varian Ltd, Palo Alto, USA.
[2] BrainLAB GmbH, Feldkirchen, Germany.

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