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High throughput film dosimetry in homogeneous and heterogeneous media for a small animal irradiator

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ABSTRACT

Purpose: We have established a high-throughput Gafchromic film dosimetry protocol for narrow kilovoltage beams in homogeneous and heterogeneous media for small-animal radiotherapy applications. The kV beam characterization is based on extensive Gafchromic film dosimetry data acquired in homogeneous and heterogeneous media. An empirical model is used for parameterization of depth and off-axis dependence of measured data.
Methods: We have modified previously published methods of film dosimetry to suit the specific tasks of the study. Unlike film protocols used in previous studies, our protocol employs simultaneous multi-channel scanning and analysis of up to nine Gafchromic films per scan. A scanner and background correction were implemented to improve accuracy of the measurements. Measurements were taken in homogeneous and inhomogeneous phantoms at 220 kVp and a field size of 5×5 mm^2. The results were compared against Monte Carlo simulations.
Results: Dose differences caused by variations in background signal were effectively removed by the corrections applied. Measurements in homogeneous phantoms were used to empirically characterize beam data in homogeneous and heterogeneous media. Film measurements in inhomogeneous phantoms and their empirical parameterization differed by about 2%–3%. The model differed from MC by about 1% (water, lung) to 7% (bone). Good agreement was found for measured and modelled off-axis ratios.
Conclusions: EBT2 films are a valuable tool for characterization of narrow kV beams, though care must be taken to eliminate disturbances caused by varying background signals. The usefulness of the empirical beam model in interpretation and parameterization of film data was demonstrated.

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Introduction

The last decade has seen considerable advances in pre-clinical radiotherapy research. Several devices for small animal radiation research have been developed, allowing the delivery of conformal doses in a precise and reproducible fashion [1]. Until recently, small animal radiotherapy research relied on relatively crude methods such as broad-field and whole-body irradiations of test subjects. However, with new small animal radiation therapy platforms, it has become possible to better mimic modern treatment modalities commonly found in the human clinic, such as image guidance, 3D-conformal, and arc deliveries. This is an important step forward in bringing pre-clinical research closer to clinical application.

Modern pre-clinical systems often employ X-rays in the kV energy range for therapy [1]. While this is necessary to minimize penumbrae and build-up effects, which can extend several millimetres for photons in the MV range, it comes with a practical advantage, as X-ray tubes are much cheaper and treatment rooms require less shielding. However, precise dose calculation becomes much harder, as determining water-equivalent thickness for the different tissue types at this energy range is more complex than for MV beams due to a stronger dependence of absorption on tissue composition. Hence, established pre-clinical dose calculation systems capable of providing accurate tissue-dependent dose calculations for small animal radiotherapy treatment using kV beams have yet to be established [1].

Accurate dose calculations for small animal treatment planning and experimental verification measured doses can be performed using Monte Carlo simulations. Unlike most conventional dose calculation engines, Monte Carlo directly includes cross sections for tissue heterogeneities, thereby considerably improving the

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accuracy of the final dose distribution [2]. Monte Carlo approaches have been chosen by research groups, at John Hopkins [1,3], Maastricht [4], Toronto [5], and Stanford [6,7]. The last group managed to achieve a statistical uncertainty of 1%, but a computation time of up to 100 h was needed [6]. Much larger uncertainties would be acceptable for animal radiotherapy treatment. Nevertheless, considerable computation times render Monte Carlo an inconvenient tool for dose calculation, especially in a small-animal laboratory setting. Currently, Monte Carlo in this setting has only involved fairly simple treatment planning, as of now falling short of its potential as a powerful dose calculation tool [1]. Therefore, for practical applications, many groups rely on analytical methods based on the standard beam model to calculate point dose and dose distributions [3,8], but none of these approaches consider tissue or phantom inhomogeneities. Recently, preliminary results for a convolution-superposition dose calculation system have been reported [9,10], but they do not contain sufficient experimental verification in heterogeneous media, which is not an easy task.

Verifying a new treatment planning dose calculation model experimentally is especially challenging for small animal radiotherapy, since field beam diameters usually range from 0.5 to 15 mm, making it impossible to rely on more common dosimeters such as ion chambers. Therefore, EBT and EBT2 Gafchromic films have become generally accepted as a means to measure 2D dose distributions under these conditions [3,11], as they provide the high resolution required for this task. Their high acceptance was increased as they have previously been shown in some studies to be energy-independent for energies ranging from 75 kVp–6 MV [12], which includes the energy ranges used for small animal radiotherapy, though more recent data suggest otherwise [13–15]. However, Gafchromic film dosimetry for narrow beams (1 mm–100 mm size) is typically associated with relatively large pixel-to-pixel noise, film-to-film variability and poor dynamic range (in measurement of in- vs. out-of-field, shallow vs. deep depths). For this reason, film data analysis requires a beam model-based approach to guide interpretation of the data, especially for heterogeneous media. Until now, empirical beam data in heterogeneous media for extremely narrow kV beams is scarce [3] and there are no empirical models that would help with its characterization.

In view of the above, the aims of this study are: (1) to modify the existing film dosimetry protocol and make it more accurate and efficient in simultaneous processing of many films, and (2) parameterize the obtained data using an empirical beam model for simultaneous homogeneous and inhomogeneous media. For the first goal, we modified the current EBT2 film dosimetry approach for the SARRP [3] to include the new dosimetry approach introduced in Micke et al. [16], which was further studied by other groups [13,17], with additional modifications of our own, and to increase the number of processed films while removing uncertainties due to scanner non-uniformity and film inhomogeneities. This protocol was then used to generate an extensive film dose database employed for the derivation of beam model parameters. Relative and absolute dose measurements were taken in inhomogeneous media of various settings and were compared against an empirical beam model for a 5×5 mm^2 field. The model was then determined by experimentally determining the attenuation and scatter of different materials and determining the radiological pathlength through water-equivalent thicknesses.

Material and methods

Phantom material and setup

An overview of the phantoms used in this study can be seen in Fig. 1. Solid water slabs measured $60 \times 60 \times 5$ mm^3 and were water-equivalent for kV beams according to the manufacturer (CIRS, Norfolk, VA). Bone-equivalent material (Gammex Inc., Middleton, WI) was cut to measure $50 \times 10 \times 2$ mm^3; inflated lung-equivalent material (Gammex Inc., Middleton, WI) had the dimensions $50 \times 20 \times 10$ mm^3. The material densities were 1.85 g/cm^3 for bone-equivalent material and 0.26 g/cm^3 for the inflated lung-equivalent material. EBT2 Gafchromic films for PDD measurements were placed between the slabs as indicated in Fig. 1. For phantoms (a), (d), (e), and (f), a commissioning jig similar to that described previously [3] was employed, for the other set-ups, a custom-made phantom jig was used. To minimize setup errors, phantoms were marked and aligned to lasers corresponding to the beam axis.

Irradiations were performed using a Small Animal Radiation Research Platform (SARRP™) from Xtrahl, Ltd. (Surrey, UK) with a collimator that resulted in a field size of 5×5 mm^2 at the isocenter. Photon energy was set to 220 kV with a tube current of 13 mA and a total irradiation time of 1.5 min. Unless stated otherwise, the phantoms were setup at a source-to-surface distance (SSD) of 340 mm. The gantry angle was set to $\theta = 0°$ for all phantom setups, except for phantoms (g) and (i), where it was set to $\theta = -15°$ and $\theta = 15°$, respectively.

EBT2 film dosimetry

Our film dosimetry protocol largely followed the one described previously in Tryggestad et al. [3] which had since been adjusted to apply multi-channel dosimetry [16]. In addition, we included an improved scanner correction that allowed for faster processing of multiple films and a film background correction to reduce the impact of off-sets found between different deliveries of films of the same LOT number.

EBT2 films (LOT A08161005A) were cut into 60×60 mm^2 pieces using a cutting template, and scanned in landscape orientation as 48-bit RGB image in transmission mode using an Epson Perfection V700 Photo scanner. All colour corrections provided by the scanner software were disabled. The resolution of the scanned images was 400 dpi, and the images were saved as TIFF files. To ensure the scanner lamp was sufficiently warmed up, the warm-up routine was run and two preview scans were performed. Nine films were scanned in a scanning frame that allowed reproducible setup, which could hold nine films of 60×60 mm^2 simultaneously. Films were scanned prior and approximately 12 h after irradiation, with values defined as the average over three consecutive scans to reduce random scanner noise. Each time, it was ensured that the scanning direction for each film was consistent in regard to the orientation of the uncut film.

To correct for non-uniformity in scanner response, lateral scanner correction based on the approaches described in previous publications was performed [18,19]. For this, films were exposed to five different dose levels (0, 100, 200, 300, 400 cGy at 6 MV in solid water) to achieve increasing levels of darkness and scanned in the different positions of the frame. The separate scans were patched together in Matlab (MathWorks, Inc., Natick, MA, USA). In contrast to previous publications, this scanner correction can only be applied to certain regions of interest, which correspond to the locations the films on the scanning frame. This was done to ensure that lack of scatter due to the missing plastic frame would not affect the scanner response. Each film location was divided into smaller areas for which an average pixel value for each dose level was determined. These were then normalized to the average of all pixels in the regions of interest at each dose level to determine a correction factor for this area.

As the previous publications and our own preliminary analysis have shown, lateral scanner response perpendicular to the scanning direction (y) can vary considerably while longitudinal scanner