The use of radiographic film for linear accelerator stereotactic radiosurgical dosimetry

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The measurement of stereotactic radiosurgical dose distributions requires an integrating, high-resolution dosimeter capable of providing a spatial map of absorbed dose. Although radiographic film is an accessible dosimeter fulfilling these criteria, for larger radiotherapy photon fields the sensitivity of film emulsion exhibits significant dependencies on both depth in phantom and field size. We have examined the variation of film sensitivity over the ranges of depths and field sizes of interest in radiosurgery with a 6 MV photon beam. While for large (20 cm×20 cm) fields the potential error in dose due to the variation of the film response with depth reaches 15%, the corresponding maximum error for a 2.5 cm diameter radiosurgical beam is 1.5%. This uncertainty was observed to be comparable in magnitude to that produced by variation in processing conditions (1.1%) and by varying the orientation of the film plane relative to the beam central axis (1.5%). The dependence of emulsion sensitivity on field size has been observed to be negligible for fields ranging in diameter from 1.0 cm to 4.0 cm. The source of the dependence of film sensitivity has been illustrated by using an EGS4 Monte Carlo simulation for large fields to illustrate significant increases in the photon spectrum below 400 keV with depth in phantom. In contrast, relative increase of this low-energy component is negligible for radiosurgical photon fields. © 1999 American Association of Physicists in Medicine. [S0094-2405(99)00110-8]

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I. INTRODUCTION

Linear accelerator-based stereotactic radiosurgery is a treatment modality employing narrowly-collimated, 6–10 MV photon beams to deliver a high dose to intracranial lesions with a sharp dose gradient at the periphery of the treatment volume. Conventionally, a number of circular convergent beams or beam arcs are used to create an approximately spherical or ellipsoidal distribution of dose. Many lesions are highly irregular in shape, however, and recent advances in radiosurgery have focused upon improving the conformity between the target volume and the delivered dose distribution. While software treatment planning systems provide analytic, three-dimensional calculations of even complex dose distributions, the required spatial and numerical tolerances for dose delivery in radiosurgery are demanding (±1 mm and 5%, respectively1) and dose calculations must be verified through direct measurement. Dosimetric measurements are required both in commissioning the radiosurgical hardware and software and in the course of periodic quality assurance.

Dosimetry for conventional (circular-beam) radiosurgery is challenging2 and the criteria for an appropriate dosimeter become more stringent for measurement of irregular dose distributions. First, the required spatial accuracy in localizing and treating the lesion necessitates a dosimeter with high (sub-millimeter) spatial resolution. The sensitive volume of the detector must be sufficiently low in order to record the dose over the sharp fall-off region of the distribution while minimizing inaccuracies caused by volume averaging of the detector size3 and the presence of lateral electronic disequilibrium.4 Second, the dosimeter must provide a spatial map of the dose distribution rather than the dose at a single point, particularly for cases where it is necessary to verify conformal distributions. Finally, the dosimeter must integrate over time to reflect the total dose administered by a number of beams or beam arcs.

The requirement for a simultaneous measurement of a spatial distribution of dose alone precludes the use of common dosimeters such as ionization chambers, thermoluminescent dosimeters (TLDs), and diodes. In order to verify conformal radiosurgical dose distributions, we have evaluated the accuracy of radiographic film. Radiographic film is an accessible, integrating dosimeter capable of recording high spatial-resolution (up to 20 line-pairs per mm) dosimetric information in two dimensions. Radiographic film is affordable and the optical density recorded on the film is readily digitized using scanning densitometers found commonly in radiotherapy centers.

The use of radiographic film for the dosimetry of larger (i.e., >10 cm×10 cm) photon radiotherapy beams is complicated due to a dependence of film emulsion sensitivity on depth within the phantom. This dependence is caused by a relative increase with depth in the population of lower-energy, scattered photons in the spectrum and the subsequent photoelectric absorption of these photons by the film emulsion. This effect becomes most pronounced with increases in the photon population in the energy region below approximately 400 keV where the mass attenuation coefficient of
emulsion increases rapidly, approximately as $1/(h\nu)^3$. As the low-energy spectral component increases, the tissue equivalence of film emulsion diverges from that of tissue-equivalent materials due to an approximate $Z^3$ dependence of the photoelectric interaction cross-section.

In practice, this dependence causes the sensitometric calibration curve of the film to vary with the depth in phantom at which the film is exposed. To overcome this problem, Hale et al. have measured a series of sensitometric curves, one for each of a range of depths, and have achieved 2% accuracy by using the appropriate curve for dosimetric studies. Williamson et al. have parametrized film sensitivity as a function of depth in order to compensate for this dependence, and have obtained 3% accuracy with this technique. Rather than correcting measured optical densities for the effect of phantom scatter, Burch et al. have proposed a method to remove low-energy, lateral scatter by placing lead foils on either side of the film plane.

These types of corrections increase the overall complexity of film dosimetry with the necessity of an increased quantity of calibration data, special modifications to the experimental setup or post-processing of acquired data to correct for the depth dependence of emulsion sensitivity. While the data provided by Burch et al. demonstrate the potential inaccuracy of film dosimetry for larger (>6 cm×6 cm) radiotherapy fields, these results have also indicated that the magnitude of the depth dependence of film sensitivity is diminished as the field size is reduced. The present work draws upon these findings with the reasoning that because stereotactic radiosurgery employs the smallest field sizes produced commonly by clinical linear accelerators, the corrections required in converting the recorded film optical density to dose should be minimized. There are examples in the literature of the use of radiographic film for accurate measurement for narrow photon beams, but the depth and field size dependencies of film sensitivity have not been quantified for radiosurgical dosimetry.

The focus of the current work is the measurement of the variation of radiographic emulsion sensitivity over the range of depths (up to ~20 cm) and field sizes (1.0 cm to 4.0 cm diameter) of interest for stereotactic radiosurgery. This variation is quantified by comparing optical density-to-dose conversion curves as a function of depth in phantom and field size. Since the dependence of emulsion sensitivity on depth has been attributed to shifts in the photon spectrum, we have interpreted experimental results by using the EGS4 FLURZ Monte Carlo simulation to examine the variation of the spectrum with depth in phantom. In particular this simulation has permitted a comparison of spectral increases in the energy region below 400 keV for both large (i.e., radiotherapy-sized) and small (radiosurgical-sized) beams.

Additional experiments were conducted to examine whether emulsion sensitivity depends on the film orientation relative to the central axis of the beam. This possible dependence becomes important when measuring dose distributions created by multiple beams or beam-arcs, where the orientation of the beams relative to the film plane may be arbitrary. Finally, the reproducibility of film dosimetry was assessed by comparing identically-exposed films.

II. METHODS AND MATERIALS

A. Film exposure technique

In order to examine possible dependencies of emulsion exposed by radiosurgical beams, a series of sensitometric curves was established for a range of depths and field sizes. For each exposure a 10 in.×12 in. sheet of Kodak X-Omat V film from a single batch (#194 05 2) was sealed within a light–tight Solid Water (Gammex RMI, Inc.) cassette. The cassette consists of two 2 cm-thick slabs of solid water sealed around three edges by nylon screws and a rubber O-ring. In order to minimize the occurrence of air gaps, the film was removed from its envelope and paper liner under safelight before insertion into the unsealed edge cassette. On the linear accelerator couch, 10 cm of solid water phantom material was added both above and below the cassette to simulate the amount of scattering material that would be adjacent to the film if it were placed at the center an anthropomorphic head phantom.

The 6 MV photon beam used for all exposures was identical to that generated routinely for radiosurgical treatment. The linear accelerator (Clinac 2100C/D, Varian Associates) was used with titanium alloy tertiary collimators (BrainLAB, GmbH), which are tapered according to beam divergence and range in size to produce fields from 1.0 cm to 4.0 cm diameter at the isocenter. The linear accelerator secondary collimator jaws were fixed at 5 cm×5 cm for all tertiary collimator sizes.

B. Film processing and measurement of optical density

To minimize possible variations due to film processing conditions, a Kodak X-Omat RP processor was used for which the throughput is very high and quality assurance is performed daily. Developer temperature fluctuated by less than ±0.5 °F between processing sessions. Optical density was measured using a scanning densitometer (Wellhofer WP102, Schwarzenbruck, Germany). The densitometer infrared light source beam width is approximately 0.8 mm in diameter and is mounted on a translation stage capable of approximately 0.2 mm precision. Base-plus-fog optical density was subtracted from scanned optical densities for each film.

C. Depth-dependence of film sensitivity

In order to measure dose in phantom along the depth axis, the gantry was rotated to 90° from the vertical and room lasers were used to align the beam central axis and plane of the film to within approximately ±0.05 cm. All films were exposed using the median stereotactic collimator field size of 2.5 cm. To acquire data to establish sensitometric curves in sufficient detail, 14 films were exposed with different monitor unit (MU) settings resulting in maximum doses to phantom ranging between 5 and 385 cGy. Although the goal of
this series of exposures was to establish the sensitometric curves over a dose range of approximately 0–140 cGy, the higher dose settings used were required in order to obtain a sufficient range of dose at greater depths in phantom. At least three films were exposed for each dose setting in order to minimize variations due to processing and scanning. This procedure was then repeated using a 20 cm×20 cm field size for comparison of small-field and large-field results.

For each exposure made with the film in this “parallel” orientation, a two-dimensional profile of optical density is recorded along the depth dimension of the phantom. In scanning the film, it is necessary to ensure that the densitometer travels along the recorded location of the central axis (i.e., so that the film is not tilted relative to the direction of scanning). To align the film, the densitometer was first used to record beam profiles (perpendicular to the central axis) at depths of 5 cm and 10 cm, and the geometric centers of each profile were calculated from the scanned data. A discrepancy between the calculated location of these profile centers indicated that the film was angled relative to the scanning direction. Once aligned, the film was scanned down the central axis location and the optical density was measured along the depth axis from the surface to a depth of 20.0 cm in 0.1 cm increments.

Each scan produces a series of optical density values recorded at each of 200 depths. In order to relate these optical densities to absolute doses, the dose to water corresponding to each of these depths and for each MU setting was calculated from

\[ D(d,A) = \frac{\text{MU} \cdot \text{PDD}(d,A) \cdot S_t(A)}{100}, \]

where \( D(d,A) \) is the dose at depth \( d \) for field size \( A \) at the phantom surface, MU is the number of monitor units given, PDD\((d,A)\) is the percent depth dose and \( S_t(A) \) is the total scatter factor. Both \( S_t(A) \) and PDD\((d,A)\) were measured in a water phantom (model WP 600 Wellhöfer, Schwarzenbruck, Germany) using a p-type silicon electron diode (Scanditronix, Uppsala, Sweden). The PDD was measured in 0.1 cm increments to a maximum depth of 25.0 cm and was linearly interpolated when required to obtain values at arbitrary depths.

After scanning all 14 films the resultant set of data consisted of 14 (optical density, dose) pairs at each of 200 depths in phantom. These data were therefore sufficient to generate sensitometric curves at each of 200 different depths in phantom, with each curve defined by 14 points.

D. Field size-dependence of film sensitivity

Similar sets of sensitometric curves were established, as described above, for both the minimum (1.0 cm-diameter) and maximum (4.0 cm-diameter) field sizes. These data facilitate the examination of variation between the sensitometric curves, at any arbitrary depth, for the range of field sizes used routinely in radiosurgery.

E. Effect of film orientation

In order to quantify a possible dependence of film orientation, exposures were also made with the film plane perpendicular to the central axis at a single depth of 10 cm in phantom. Five separate films were exposed with given doses ranging from 5 to 385 cGy. At a depth of 10 cm this dose range extends into the saturation region of the film, but it was useful to define the sensitometric curve over as wide a range as possible for accurate curve-fitting. For each exposure the optical density was measured at the center of the spot pattern recorded on the film. Using Eq. (1) to calculate absolute doses for each exposure, a sensitometric curve was established for this “perpendicular” geometry and compared with the sensitometric curve for the “parallel” geometry. Films for both orientations were developed in the same batch to minimize the effect of processing variability.

F. Film reproducibility

During each experimental session a minimum of three “control” films were exposed. The control exposure was made using 2.5 cm-diameter collimator by administering a maximum dose of 87 cGy to a film oriented parallel to the central axis. The optical density curves obtained by scanning the film along the depth dimension at the location of the central axis were compared. Because these control exposures were repeated during a single session and also over the course of approximately six months, realistic measures of both film-to-film and session-to-session variation were obtained. In addition to processing variations, these values also include contributions such as drift and fluctuation of the densitometer signal and nonuniformity of the film emulsion.

G. Monte Carlo simulation

To examine the variation of the photon spectrum with depth and to compare large- and small-field sizes in terms of this variation, the EGS4 Monte Carlo FLURZ code was used to model a cylindrical water phantom as shown in Fig. 1. The phantom is 40.0 cm in diameter and 40.0 cm deep. Both large (25 cm-diameter) and small (2.5 cm-diameter) beams were modeled for comparison. A published 6 MV incident spectrum selected was used to be representative of that used in the experiments. The full photon spectrum was scored along the depth axis at 1.0 cm increments within a 1.0 cm-diameter region at the center of the phantom. 15×10⁶ incident histories were recorded for each beam. The PRESTA algorithm was used to model electron transport, and the electron cut-off (ECUT) and photon cut-off (PCUT) parameters were set to 0.521 MeV and 0.010 MeV, respectively.

III. RESULTS

A. Depth-dependence of film sensitivity

Figure 2(a) shows the sensitometric curves obtained for the large (20 cm×20 cm) photon beam at depths of 1.0 cm,
For each depth a curve was fitted to the measured sensitometric data using the single-target/single-hit theory equation,

$$\text{OD} = \text{OD}_{\text{sat}} (1 - 10^{-\alpha D})$$

where OD and D are the measured optical density and given dose, respectively. The saturation density of the film, OD_{sat}, was estimated by delivering a large dose (500 cGy) to a film in phantom, and was held constant in the fitting algorithm. The fitting parameter \( \alpha \) represents emulsion sensitivity and was allowed to vary. For this large field, the curves diverge markedly, indicating that using a sensitometric curve that is not depth-specific would introduce significant error in converting optical density to dose. In order to minimize this error to below approximately 10\%, for example, it is necessary to confine the dose range to less than 80 cGy. In contrast, Fig. 2(b) shows sensitometric curves corresponding to the same depths for the radiosurgical (2.5 cm-diameter) field. For this small field the curves agree to within the reproducibility of film development and scanning (discussed below).

The fitted curves using the single-hit/single-target equation correspond to the data closely, in all cases resulting in correlation coefficient values greater than 0.9957. By determining the value of \( \alpha \) for each curve, film sensitivity was obtained as a function of depth as illustrated for the 20 cm \( \times \) 20 cm and 2.5 cm-diameter fields (Fig. 3). As expected from the disparity in the sensitometric curves in Fig. 2(a), the film sensitivity for the large (20 cm \( \times \) 20 cm) field increases systematically with depth. While small fluctuation of the values of \( \alpha \) is apparent for the radiosurgical field, no systematic variation of sensitivity with depth is apparent.
B. Field-size dependence of film sensitivity

Figure 4 illustrates the measured sensitometric curves for the range of field sizes used typically for linear accelerator stereotactic radiosurgery. These data are shown for a single depth of 10.0 cm. Curves fitted using Eq. 2 are congruent, indicating that there is no significant variation in sensitivity with field size over this range. This congruence was observed for all depths in phantom ranging from 1.0 cm to 20.0 cm.

C. Film reproducibility

A subset of the film reproducibility data is shown in Fig. 5 which compares the optical density recorded on six films exposed identically and processed in three batches of two films, where each batch was processed on a separate day over the course of approximately six months. At a typical depth for radiosurgery of 10.0 cm, the “intra-session” maximum variation in optical density between films processed in one batch is 0.85% (where “variation” has been quantified by calculating $\frac{OD_{\text{max}} - OD_{\text{min}}}{OD_{\text{mean}}}$). The maximum “inter-session” variation in optical density between films processed on separate occasions is 2.3%. Using the fitted sensitometric curve shown in Fig. 4, this translates into intra-session and inter-session variations in the calculated dose of 1.1% and 2.9%, respectively. To determine the component of this fluctuation resulting from the film digitization process, a single film was scanned repeatedly with the densitometer. This source of variation was found to be negligible compared to the total variation in optical density between control films. For lower OD values, however (\(<\sim 0.5\)), the presence of noise became apparent.

D. Effect of film orientation

Figure 6 compares the sensitometric curves obtained with the film positioned parallel and perpendicular to the central axis at a depth of 10.0 cm in phantom. Over this wide dose range, the maximum discrepancy between these curves is 1.5%, slightly greater than the inter-session variation due to processing.

E. Monte Carlo simulation

The photon spectra computed using EGS4 FLURZ are shown in Fig. 7 and serve to explain the source of the increase in emulsion sensitivity with depth for larger fields, as well as the comparatively low depth-dependence for radiosurgical fields. Figure 7(a) shows the total (primary plus scattered) photon spectra for a large (25 cm-diameter) field on central axis at depths of 1.0 cm, 5.0 cm, and 20.0 cm. Each spectrum has been normalized to its own maximum to illustrate relative spectral changes. These data depict a consistent increase in the lower-energy (\(<400 \text{ keV}\)) component of the spectrum with depth. Since the mass attenuation coefficient for film emulsion increases rapidly in this region, this would result in an increase with depth in the proportion of absorbed photons and therefore an enhanced sensitivity. Changing only the field size in the simulation to a radiosurgical-sized diameter of 2.5 cm produces the spectra
illustrated in Fig. 7. In contrast to the large-field case, the spectra for the radiosurgical field show a negligible increase in the low-energy population of photons.

IV. DISCUSSION

Although film dosimetry is most problematic for lower-energy photon beams (e.g., cobalt-60 and 4 MV) due to their lower primary-to-scatter ratio, Fig. 2(a) emphasizes the importance of corrections in converting scanned optical density to dose even for the 6 MV beam. For this 20 cm × 20 cm field, using a depth-unspecific curve may introduce an error in dose of up to 15% for the dose range examined in this study. This error is confined to below 10% only for doses below 80 cGy. For the 2.5 cm-diameter radiosurgical field, using a sensistimetric curve established for a depth of 1.0 cm to calculate a dose at 20.0 cm depth would result in a maximum error of 1.5%, which is comparable in magnitude to the intra-session variability. This invariance of the calibration curve with depth greatly simplifies the task of film dosimetry for 6 MV radiosurgical beams in two respects. First, corrections to compensate for a depth-dependence of the film emulsion sensitivity are not required. Second, since a single calibration curve can be used over a range of depths, the quantity of data required in calibrating the film is greatly reduced.

As depicted by the data in Fig. 7, the photon spectra for both the large (25.0 cm-diameter) and small (2.5 cm-diameter) fields exhibit variation with depth in phantom. These results agree with the simulation findings by Heydarian et al.,2 which indicate that the mean photon energy of 6 MV radiosurgical beams increases with depth due to beam hardening, but for large beams (20 cm-diameter) the mean photon energy decreases with depth up to 15 cm due to an increase of phantom scatter. In terms of the potential over-response of film emulsion, however, only spectral changes below approximately 400 keV are crucial. Figure 8 shows the 6 MV spectrum on the same energy scale as the mass attenuation coefficients for water, film base and film emulsion. The shaded region of this incident spectrum indicates the small population of photons (≪2% of the total spectrum) with energies below 400 keV. As indicated by the Monte Carlo results in Fig. 7(b) for radiosurgical fields, this low-energy tail does not increase significantly with depth in phantom. Over the range of depths of interest in radiosurgery (to 20 cm) the vast majority of photons in the spectrum exist in the energy range between 400 keV to 6 MeV. It should be noted that this simulation was performed only for the median radiosurgical collimator size of 2.5 cm, and potential increased softening of the spectrum in phantom for the larger radiosurgical collimators was not examined. However, the recent Monte Carlo simulation of a linac radiosurgery unit by Verhaegen et al.17 suggests that the mean photon energy at depth of dose maximum decreases by only 0.45 MeV when the aperture diameter is increased from 0.5 cm to 5.0 cm.

Figures 2 and 4 show that sensitometric curve is approximately linear for doses below 40 cGy. As indicated by Evans and Schreiner,18 restricting the dose to this region offers the advantage of direct conversion of optical density to dose. As illustrated by the data in Fig. 2(a) for the larger radiotherapy field, this constraint would also minimize the error caused by the variation of film sensitivity with depth, and thus would alleviate the need to acquire a family of calibration curves over a range of depths. In measuring radiosurgical dose distributions, however, in many cases it may be difficult to con-
fine the total dose to the linear region of the sensitometric curve. For example, for multiple convergent-arc techniques, a minimum achievable monitor unit-per-degree rate is imposed by the linear accelerator, and thus the total dose may necessarily exceed 40 cGy. Similarly, treatments involving multiple, static conformal fields would necessitate a very low dose per field, and thus the uncertainty inherent in the dose per beam would become appreciable. Therefore, to facilitate dosimetry over a larger dose range, we generally acquire sufficient calibration data to extend the curve-fitting over the nonlinear region. It should be noted, however, that the nonlinear shape of the sensitometric curve is specific to Kodak XOMat-V film. Cheng and Das and Cadman have reported on a relatively new type of silver-halide radiographic film (CEA TVS) which is characterized by a linear sensitometric curve up to 90 cGy, followed by abrupt saturation. While Cheng and Das have shown that the relationship between optical density and dose does not vary for 4–18 MV photon spectra for measurements made at the depth of maximum dose in phantom, possible variation of sensitivity with depth in phantom has not been examined.

The low variation of emulsion sensitivity with depth for small (<4.0 cm-diameter) 6 MV beams will offer an advantage in terms of dosimetric accuracy only if the variation due to processing conditions does not prevail. In this study, although the developer temperature was held constant to within ±0.5°F and the replenishment rate was kept as continuous as possible, the data in Fig. 5 suggest that in order to achieve a variation of <2%, it is necessary to process recalibration films in the same batch as the dosimetric films.

V. CONCLUSION

This work was motivated by a need for an accessible, high-resolution dosimeter for stereotactic radiosurgery, and in particular, for small conformal fields. The results indicate that radiographic film, while providing the required spatial resolution, is particularly applicable to small 6 MV photon beams. Dependencies of emulsion sensitivity on depth and field size which limit film dosimetry for larger photon fields are sufficiently minor for radiosurgical fields to warrant the use of a single sensitometric curve. Monte Carlo results indicate that even at depth in phantom, the majority of photons in the spectrum have energies above 400 keV, and therefore film does not greatly compromise tissue equivalence. Other potential sources of uncertainty, such as the variability of processing conditions and dependence of film sensitivity with orientation do not preclude the use of radiographic film for the measurement of radiosurgical dose distributions produced by a 6 MV beam.

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