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Measurement of radiotherapy x-ray skin dose on a chest wall phantom

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Sufficient skin dose needs to be delivered by a radiotherapy chest wall treatment regimen to ensure the probability of a near surface tumor recurrence is minimized. To simulate a chest wall treatment a hemicylindrical solid water phantom of 7.5 cm radius was irradiated with 6 MV x-rays using $20\times20\,\mathrm{cm}^2$ and $10\times20\,\mathrm{cm}^2$ fields at 100 cm source surface distance (SSD) to the base of the phantom. A surface dose profile was obtained from 0 to 180°, in 10° increments around the circumference of the phantom. Dosimetry results obtained from radiochromic film (effective depth of 0.17 mm) were used in the investigation, the superficial doses were found to be 28% (of $D_{\max}$) at the 0° beam entry position and 58% at the 90° oblique beam position. Superficial dose results were also obtained using extra thin thermoluminescent dosimeters (TLD) (effective depth 0.14 mm) of 30% at 0°, 57% at 90°, and a metal oxide semiconductor field effect transistor (MOSFET) detector (effective depth 0.5 mm) of 43% at 0°, 62% at 90°. Because the differences in measured superficial doses were significant and beyond those related to experimental error, these differences are assumed to be mostly attributable to the effective depth of measurement of each detector. We numerically simulated a bolus on/bolus off technique and found we could increase the coverage to the skin. Using an alternate ''bolus on,'' ''bolus off'' regimen, the skin would receive 36.8 Gy at 0° incidence and 46.4 Gy at 90° incidence for a prescribed midpoint dose of 50 Gy. From this work it is evident that, as the circumference of the phantom is traversed the SSD increases and hence there is an inverse square fluence fall-off, this is more than offset by the increase in skin dose due to surface curvature to a plateau at about 90°. Beyond this angle it is assumed that beam attenuation through the phantom and inverse square fall-off is causing the surface dose to reduce. © *2000 American Association of Physicists in Medicine.* [S0094-2405(00)00107-3]

Key words: radiotherapy, skin, radiochromic film, thermoluminescent dosimeter, MOSFET

I. INTRODUCTION

Skin dose assessment to the chest wall (breast) region is important to ensure that sufficient dose is given to structures at risk from near surface recurrence[1] (e.g., scars, dermal lymphatics) while not exceeding the tolerance level required for acceptable skin cosmesis.[2] The superficial layers of interest include the dermal lymphatics which extend to about 1 mm depth and the basal cell layer at about 70 μm.[3]

Patients undergoing radiotherapy treatment receive an inherent skin sparing effect due to the range of electrons set in motion from megavoltage x-ray beam interactions.[4,5] An additional, skin dose component is due to electron contamination, the magnitude of which depends on parameters such as beam modifiers, field size, and air gap.[5–10] The dependence of surface dose on angle of incidence has been measured in slab phantoms[11–14] and *in vivo* at a few discrete points including the entry and exit position.[6] The effect of a curved contour with overshooting beams in a configuration resembling two chest wall tangent fields is the subject of this report. The study also outlines the impact of adding 1 cm of bolus.

II. MATERIALS AND METHODS

To simulate a chest wall (post mastectomy) breast treatment a hemicylindrical solid water phantom,[15] of 7.5 cm radius was irradiated with 6 MV x-rays from a Varian 2100C at 100 cm source surface distance (SSD) (see Fig. 1). Three radiation detector types were used in the investigation, radiochromic film (Gafchromic MD-55-2/Lot No. 97011) thermoluminescent dosimeters (TLDs) (solon/Harshaw, USA) of

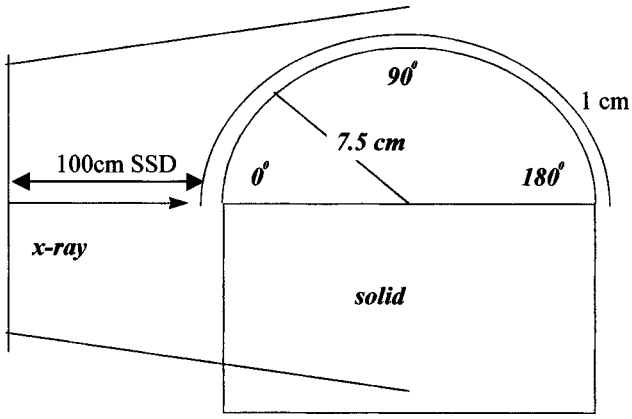

FIG. 1. The hemicylindrical chest wall phantom used to simulate a tangential breast treatment. A $20\times20\,\mathrm{cm}^2$ x-ray beam with 2.5 cm overshoot was set incident at 100 cm SSD, for some of the radiochromic film readings 1 cm of bolus was added.

three different nominal thicknesses, and a MOSFET [Radfet (Courtesy of Rem, Oxford, UK)]. Radiation measurements were undertaken for all detectors using a $20\times20\,\mathrm{cm}^2$ field at nominal SSD, the point where the hemicylinder starts. This field size ensured detector measurements could be normalized to $d_{\max}$ in a central axis beam position. Tangent fields are generally rectangular to ensure less lung dose, therefore a more clinically typical field of $10\times20\,\mathrm{cm}^2$ half-jaw was also tested using the radiochromic film detector.

A. Radiochromic film

For the $20\times20\,\mathrm{cm}^2$ field the hemicylindrical solid water phantom was marked from 0° (i.e., beam entry) to 180° (i.e., beam exit) in 10° increments. A strip of radiochromic film with the dimensions of 1 cm$\times$16 cm was placed around the arc on the surface of the hemicylinder. Measurements were made at the central axis from the entrance to the exit beam location. Because the 0° and 180° positions for the $10\times20\,\mathrm{cm}^2$ field were in the penumbra, only measurements from 10° to 170° were undertaken. The optical density of the irradiated film was read using a 660 nm modified densitometer.[16] These readings were then converted to dose using a calibration curve. This was performed by irradiating films from the same batch to known doses at $d_{\max}$ in a solid water (Radiation Measurements Inc. RMI, USA) phantom. Doses of between 0 Gy and 20 Gy were delivered in 2 Gy increments. A calibration curve was then created using a third order polynomial fit to the data.[17] The radiochromic film was determined as having an effective measurement depth of 0.17 ± 0.03 mm. This thickness was determined by comparison to the Attix chamber. Note the thickness of the outer (protective) layer of the film is about 0.012 g/cm^2 according to the manufacturer.[18–26]

B. Thermoluminescent dosimeters

Three different thicknesses of TLD chips were used. Lithium fluoride (LiF) crystals doped with magnesium and titanium were used with nominal thicknesses [0.14 mm (extra thin), 0.39 mm (thin), and 0.89 mm (normal)]. The details of how these nominal thicknesses were determined are outlined in a previous paper by Kron *et al.*[14] The TLD's from Harshaw Chemicals (Solon/Harshaw, USA) were used. Thin and normal chips were made from LiF with the natural mixture of lithium (TLD 100). The extra thin TLDs were however only available as TLD 700 material containing the isotope ^{7}Li enriched. This reduces the TLD response to neutrons but does not alter the TLD properties in x-ray fields.

The chips were read out in a NE Rialto TLD reader in a two step read out cycle. The readout temperature was 300 °C. All TLD chips were annealed in a dedicated annealing oven at 400 °C for 1 hour followed by fan forced cool down to 100 °C which was held for 2 hours. For each TLD, readings were normalized to the relative response of that dosimeter at 100 cGy. Each chip was characterized by an individual sensitivity value which was obtained as outlined elsewhere.[27]

C. MOSFET dosimeter

A MOSFET was also used in the investigation. The detector consisted of a source, drain and gate placed on a substrate. The dimensions of the active detector are 0.55 mm $\times$1.10 mm attached to the tip of an alumina substrate with epoxy. Other details of the construction of the detector can be found in the paper published by Gladstone *et al.*[28] Many MOSFET devices have leads underneath with a metal casing which would have been difficult to mount for these experiments. By placing the detector on the surface of the hemicylindrical phantom in 10° increments. All the data obtained from each dosimeter was converted to show surface dose profiles. The MOSFET detector was estimated to have an effective measurement depth of 0.5 mm by scaling the 0.3 mm thickness of the epoxy coating with a density of 1.5.[29–31] All MOSFET detector response values were normalized to a standard dose of (20 cGy) at $d_{\max}$ for a $10\times10\,\mathrm{cm}^2$ square field at 100 cm SSD. Directional dependence of this device was tested by placing it at 10 cm depth in a cylindrical phantom. In equilibrium conditions at 7.5 cm depth variations in dose recorded from angle 0° to 60° was $\pm3\%$.

The experiment was repeated using a 1 cm bolus (MedTec, USA) with radiochromic film measurement only. By combining the results and normalizing to a prescribed midpoint dose of 50 Gy (i.e., from two tangent fields), using an alternate ''bolus on,'' 'bolus off' regimen, patient skin dose was calculated from the measurements.

III. RESULTS

The results of measurements, taken at 10° increments, using radiochromic film, MOSFET detectors, extra thin, thin and normal TLD are presented in Fig. 2. Uncertainties are expressed as $\pm$ two standard deviations and they are as follows: radiochromic film $\pm3\%$, MOSFET $\pm3\%$, extra thin TLD $\pm3\%$, and normal TLD $\pm2\%$.[13,14,30] Note the characteristic shape of the curve indicated how the near superficial dose increases with the effective depth of measurement of the detectors. Note that entrance percentage superficial dose (%Dose) was 30% and exit was 39% for the extra thin TLDs. For the thin TLDs the results were 36% and 47%, respec-

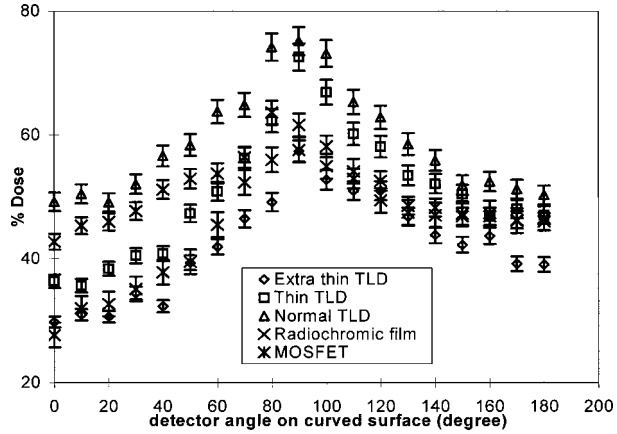

FIG. 2. Percentage near surface dose obtained using various detectors including radiochromic (Gafchromic MD 55-2) film, TLDs (extra thin, thin and normal), and a MOSFET detector. Normalized to incident dose at d_{max} for each detector.

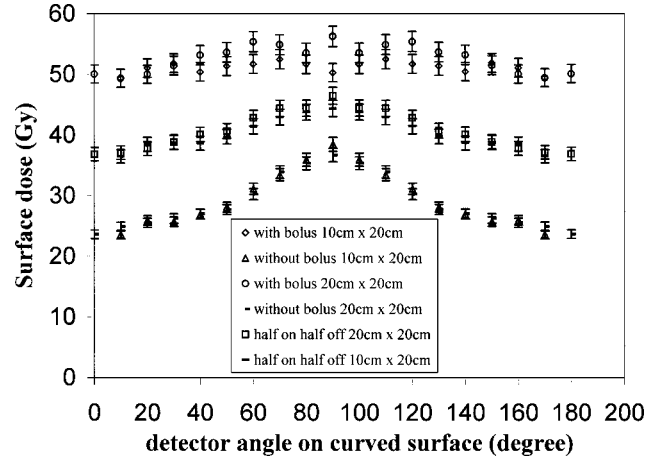

FIG. 4. Combined surface dose vs angle on a hemicylindrical surface with bolus on for all fractions, bolus off for all fractions, and bolus on for half the fractions. All data normalized to 50 Gy midline dose.

tively. For the normal thickness TLD's the readings were 49% and 50%. The subsequent results were 28% and 47% for the radiochromic film, and 43%, 46% for the MOSFET, respectively.

At 90°, the peak of the percentage superficial dose for all three dosimeters were 57% (extra thin TLD), 73% (thin TLD), 75% (normal TLD), 58%(Radiochromic film), and 62% (MOSFET).

Figure 3 shows the incident percentage dose measured using radiochromic film for $20\times20\,\mathrm{cm}^2$ and $10\times20\,\mathrm{cm}^2$ tangent fields, with and without 1 cm of bolus present. The smaller field size shows less incident surface dose particularly at the angles 0° to 60° without bolus. Adding bolus causes a large increase (350%) in dose at shallow incident angles (<60°) but at steep incident angles (>140°) the incident dose is only increased by 10% with the bolus. The increase in dose on the entrance side is due to the increase in buildup material. The increase in dose on the exit side is due to an increase in back scatter material. The data presented in Fig. 3 can be used to predict the surface dose for a parallel opposed treatment where 50 Gy is delivered to the midline of the breast. Figure 4 presents this data with and without bolus as well as with bolus for half the fractions.

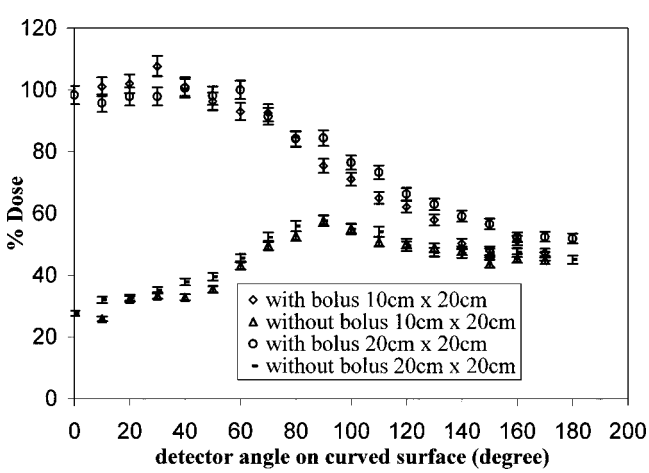

FIG. 3. Percentage dose at the phantom surface for an x-ray field with and without 1 cm of bolus for $20\times20\,\mathrm{cm}^2$ and $10\times20\,\mathrm{cm}^2$ field size as measured using radiochromic film. Normalized to incident dose at d_{max}.

IV. DISCUSSION AND CONCLUSION

In the interests of normalizing dose to a central axis reference position a $20\times20\,\mathrm{cm}^2$ field was used. It is acknowledged that a half-jaw $10\times20\,\mathrm{cm}^2$ field is more clinically realistic. The differences showed in Fig. 4 for measurements without bolus (e.g., between 26% and 32% at 10°) indicate that the main difference at shallow angle of incidence is due to less electron contamination for the smaller rectangular field size. To confirm this the electron contamination of a $10\times20\,\mathrm{cm}^2$ and $20\times20\,\mathrm{cm}^2$ was measured using an Attix model 449 parallel plate ionization chamber and a 6% (i.e., as percentage of D_{max}) difference was found.

In clinical setup it is common for part of one edge of the beam to pass through lung. In general the lack of attenuation in a low density medium would lead to an increase in exit dose.[32,33] However disequilibrium in these regions can lead to penumbral flaring and a deficit of dose in lung which then has a secondary buildup.[34] This phantom does not attempt to measure dose enhancement or attenuation when lung is involved.

When clinicians are concerned about a near surface recurrence they may use bolus continuously or on alternate days of treatment. The minimum dose for each scenario was at the beam entry position angle (i.e., 0°). For a 50 Gy prescribed midline dose the dose at this position would be 23.6 Gy without bolus, 50 Gy with bolus, and 36.8 Gy with bolus applied for half the fractions. The maximum dose for each scenario was at the most oblique beam angle (i.e., 90°). The dose at this position was 36.6 Gy without bolus, 56.2 Gy with bolus, and 46.4 Gy with bolus applied for half the fractions.

Note that the use of a thicker bolus placed on alternate days probably gives a more homogeneous final dose distri-

bution because the bolus is close to the d_{max} thickness where dose gradients are reduced. There is a potential radiobiological difference between the dose delivered in this manner on alternate days vs the same dose per fraction each day. However using the linear quadratic model and typical parameters,[5,35,36] this difference is relatively small. For example at 0° the biological effective dose (BED) for tumor control was $42.9 Gy_{10}$ for the uneven dose fraction delivery vs $42.2 Gy_{10}$ if the dose were delivered using even fractions. For late effects the difference is only a little more. The BED is $57.1\,Gy_3$ (bolus on/off) vs $54.9\,Gy_3$ (half-bolus). If proliferation is considered the BED changes but the offset between each regimen does not.

Radiochromic film is potentially a useful tool as an integrative skin dosimeter in radiotherapy. It can measure a surface dose profile at a physical effective depth of 0.17 mm, hence superficial dose can be determined. The film is light weight and can be cut to any size or shape to be used clinically. Radiochromic film can be used in the dose range from 10 to 100 Gy. The film can be used a multiple number of times until a total dose of 100 Gy is accumulated. Its lack of sensitivity is in-turn a potential disadvantage for use on patients. Unlike radiochromic film, TLD detectors can only be used to check discrete points, *in vivo* to determine the dose received. However at this time TLDs still maintain a more clinically useful sensitivity range than radiochromic film.

MOSFET detectors provide instant read out and can store previous readings. The construction of the detector is compact, of minimal weight, and can be used on any surface. The detector itself does not involve a connection to a high voltage terminal. The spatial resolution (1 μm) is extremely good for these detectors. However as previously discussed in the methods section there is some directional sensitivity probably due to the packaging at very oblique angles.[31]

The investigation has outlined different detectors with different responses to surface dose measurement. These differences are likely to be mainly due to the effective depth of measurement for each detector. Radiochromic film proved to be adequate for this work, but its low sensitivity precludes its use as an *in vivo* dosimeter for single fraction dose monitoring. It would be clinically more useful if a more radiation sensitive type was available.[37] TLD still provides a good clinical skin dose measurement despite the somewhat high work input required to anneal and calibrate these detectors. MOSFET dosimeters are very dose sensitive (i.e., 10 cGy) and thus could be placed *in vivo* for the first part of a dose fraction to assess skin dose, MOSFETs do however currently have a finite lifetime which is proportional to the bias voltage applied. The continued use of these detectors probably depends somewhat on the low cost of providing replacement detectors.

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