Dosimetric characterization of round HDR ^{192}Ir AccuBoost applicators for breast brachytherapy

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Purpose: The AccuBoost brachytherapy system applies HDR ^{192}Ir beams peripherally to the breast using collimating applicators. The purpose of this study was to benchmark Monte Carlo simulations of the HDR ^{192}Ir source, to dosimetrically characterize the round applicators using established Monte Carlo simulation and radiation measurement techniques and to gather data for clinical use.

Methods: Dosimetric measurements were performed in a polystyrene phantom, while simulations estimated dose in air, liquid water, polystyrene and ICRU 44 breast tissue. Dose distribution characterization of the 4–8 cm diameter collimators was performed using radiochromic EBT film and air ionization chambers.

Results: The central axis dose falloff was steeper for the 4 cm diameter applicator in comparison to the 8 cm diameter applicator, with surface to 3 cm depth-dose ratios of 3.65 and 2.44, respectively. These ratios did not considerably change when varying the phantom composition from breast tissue to polystyrene, phantom thickness from 4 to 8 cm, or phantom radius from 8 to 15 cm. Dose distributions on the central axis were fitted to sixth-order polynomials for clinical use in a hand calculation spreadsheet (i.e., nomogram). Dose uniformity within the useful applicator apertures decreased as depth-dose increased.

Conclusions: Monte Carlo benchmarking simulations of the HDR ^{192}Ir source using the MCNP5 radiation transport code indicated agreement within 1% of the published results over the radial/angular region of interest. Changes in phantom size and radius did not cause noteworthy changes in the central axis depth-dose. Polynomial fit depth-dose curves provide a simple and accurate basis for a nomogram. © 2009 American Association of Physicists in Medicine.

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I. INTRODUCTION

The AccuBoost® peripheral breast brachytherapy system (Advanced Radiation Therapy of Billerica, MA) provides a means of delivering a partial breast irradiation treatment regimen noninvasively using mammographic image guidance.^{1–3} The patient's breast is immobilized within the mammography system and is imaged with a radiopaque coordinate grid. The radiation oncologist determines the isocenter coordinates and appropriate applicator size and shape based on this image. Without decompressing the breast, collimating HDR ^{192}Ir brachytherapy applicators are positioned on either side of the breast (Fig. 1) along a common axis (the so-called “dual-vector” approach) facing inward. Brachytherapy dose is delivered from these parallel-opposed beams from a stepped HDR ^{192}Ir source on the common imaging/treatment platform. This process is repeated, typically, along an orthogonal axis to distribute the entrance dose in accordance with a 4-field box approach. Selection of the applicator shape, either round or D-shaped, is based on the shape and position of tissue to be treated. For lumpectomy cavities that tend to be spherical or slightly ovoid in shape, the round applicator described in this text is most commonly

used. For cavities in close proximity to the chest wall, the D-shaped applicator⁴ is more commonly used as it better approximates the tissue being targeted. The purpose of this study is to characterize the dose distribution from the round collimating applicators for clinical use. This is the first dosimetric study of the round AccuBoost applicators (Fig. 2).

II. MATERIALS AND METHODS

II.A. Brachytherapy applicator

An HDR ^{192}Ir brachytherapy source enters the applicator via a 6F plastic (nylon) catheter. This catheter fits within a 2 mm square groove milled into a 4.5 mm thick polycarbonate (1.2 g cm^{-3}) disk opposite the open surface of the applicator. The round applicators range in size (inner diameter) from 4 cm (model SBA-4) to 8 cm (model SBA-8) in 1 cm diameter increments. The number of active dwell positions ranges from 12 to 24 based on internal circumference. To minimize HDR ^{192}Ir source anisotropy effects, the applicator design positions the source long axis normal to the round applicator central axis and 24 mm from the applicator open surface (Fig. 2). The applicator wall is 6 mm thick and is made of tungsten alloy (17 g cm^{-3} , 90% tungsten, 6% nickel, and 4%

FIG. 1. Application of the AccuBoost system illustrating a breast positioned between the compression paddle and the magnification stand. The round AccuBoost applicators (arrows) with white catheter strain reliefs are positioned to direct HDR ^{192}Ir radiation in a parallel-opposed, mediolateral alignment.

iron by mass). For the dual-vector geometry, the applicator is in contact with a 3.67 mm thick polycarbonate compression paddle that positions the HDR ^{192}Ir source 27.67 mm from the skin surface. The coordinate system defines the lateral radial position from the round applicator central axis with depth d into the phantom relative to the compression paddle: surface interface.

FIG. 2. Monte Carlo geometry for the 6 cm diameter AccuBoost applicator illustrating parallel-opposed positioning of two applicators around a compressed breast phantom (80 mm thick) bounded by two 3.67 mm thick polycarbonate compression paddles. The tungsten alloy applicator encloses a polycarbonate catheter guide for the HDR ^{192}Ir source (shown at the upper-right in the top applicator). Air fills the applicator and the surrounding environment. The upper-right inset shows a magnified drawing of the HDR ^{192}Ir brachytherapy source with the encapsulation and iridium core with compositions and dimensions described in the text.

II.B. Monte Carlo and measurement techniques

The Monte Carlo methods were described in Ref. 4. However, the paddle thickness has increased from 2.75 to 3.67 mm due to the introduction of the dual-vector system having an integrated radiopaque grid for applicator localization. Also, the right cylindrical phantoms, made of polystyrene (PS) or ICRU 44 breast tissue, were modeled with radii R of 8 cm by Yang and Rivard and are now 15 cm. Unlike for the D-applicator simulations, cylindrical symmetry of round applicators allowed simulation of only a single HDR ^{192}Ir dwell position due to advantageous toroidal volume averaging. Consequently, sampling volumes were square torii voxels with cross sections 0.5 mm on a side, whereas Yang and Rivard used cubic voxels of 1.0 mm on a side based on a mesh tally. The influence of the number of dwell positions on circumferential surface dose uniformity was analytically examined to justify using square torii voxels. Dose to points located directly beneath a source dwell position was estimated and compared to dose to points at the same radius located between dwell positions. For 12–24 dwell positions (10 mm step size) for the 4–8 cm diameter applicators, respectively, circumferential surface dose uniformity was within 0.1% of the optimal, i.e., if there were infinite dwell positions such as a ring source. Consequently, surface dose uniformity in the radial direction was examined in 0.5 mm increments using the aforementioned voxel geometry. The cylinder thickness t was either 40 or 80 mm to evaluate the influence of radiation scatter (along with varying cylinder radius from 8 to 15 cm). For each combination of phantom composition and applicator size, 2×10^9 photon histories were simulated so that the dose estimation statistical uncertainties were below 0.5% and kept significantly less than the Monte Carlo systematic uncertainties.

Measurements were performed in PS using radiochromic film (model EBT, International Specialty Products, Wayne, NJ) and Farmer (model N23333) and Markus (model N23343) air ionization chambers (PTW-Freiburg GmbH, Freiburg, Germany), as described by Yang and Rivard.⁴ The chambers had NIST-traceable $N_{D,W}$ calibration factors, and the EBT film was calibrated using NIST-traceable linac-produced 6 MV photons. Figure 3 depicts the experimental setup for the film and ion chambers. Chamber positioning uncertainties for depth-dose measurements were estimated to be <1 mm laterally and <0.5 mm in depth relative to the applicators due to the precision-machined PS plates accommodating the chambers. Film positioning uncertainties for dose profile measurements were similar to the chamber depth-dose values due to film orientation parallel to the PS plates. By excluding the 3.67 mm paddle, the film edge abutted the open edge of the round applicators and also provided a rigid geometry to generate film localization uncertainties similar to the other orientation. Ion chamber volume averaging effects and dosimetric uncertainties in the current study are similar to that of Yang and Rivard given the identical methods.

To validate the Monte Carlo results, dose distributions in water were simulated for the Nucletron HDR ^{192}Ir V2 source

FIG. 3. Experimental setup for measurements using EBT radiochromic film and ionization chambers. The rectilinear PS phantom (25×25 cm² plates) was positioned beneath the round applicators to measure depth-dose data and dose profiles at a fixed depth using film. Ion chambers (Markus chamber is shown) were used to measure depth-dose data. Methods for both detectors are described in detail by Yang and Rivard (Ref. 4).

(model 105.002) to compare to TG-43 brachytherapy dosimetry parameters^{5,6} from Ref. 7. For this benchmarking sub-study, a total of 2×10^9 photon histories was also simulated and the sampling volumes were similar to that described by Rivard, in which 1 mm radial bins (except for $r < 3.5$ mm, where 0.5 mm bins were used to $r = 0.5$ mm) and 1° polar angle bins for θ were used.⁸ However, this sub-study differed from Rivard's paper in that the phantom spherical radius was 40 cm and the ^{192}Ir spectrum was taken from the National Nuclear Data Center.⁹ Only photon transport was performed (versus coupled photon:electron transport) since tissue was at least several millimeters from the ^{192}Ir source or high-Z materials.¹⁰ The brachytherapy source was modeled as a right cylinder iridium core (22.4 g cm⁻³) 0.65 mm in diameter and 3.6 mm long. The source had encapsulation com-

posed of 316L stainless steel (8.0 g cm⁻³) with 0.90 mm outer diameter, 4.5 mm total length, and spherical endwelds 0.45 mm in radius. For use in round applicator modeling, the sensitivity of Monte Carlo dose simulations to the design of the endwelds and delivery cable was minimized due to applicator collimation along the source long axis so that line of sight within the applicator restricted polar angles to $\theta > 55^\circ$ and $\theta > 41^\circ$ for the 4 and 8 cm applicators, respectively.

III. RESULTS

III.A. HDR ^{192}Ir source benchmarking

Data from Table I of Daskalov *et al.* were plotted adjacent to $g_L(r)$ simulation results in Fig. 4(a). Agreement was within 0.5% of unity on average (range of 0.997–1.023) for $0.1 < r < 14$ cm after correcting for the phantom size.¹¹ The $F(r, \theta)$ data were also plotted for comparison; Fig. 4(b) shows agreement within 1% for all radii/angles except for $r = 0.25$ cm where results from the current study are higher by about 4% for $\theta < 10^\circ$. Daskalov *et al.* used a slightly different ^{192}Ir photon spectrum, and did not specify their Monte Carlo detector volume. The current study averaged r between 0.249 and 0.251 cm, but may have oversimplified the source and capsule endwelds. Daskalov *et al.* calculated a dose rate constant $_{\text{MC}}\Lambda$ of 1.108 cGy h⁻¹ U⁻¹ versus $_{\text{MC}}\Lambda = 1.109 \pm 0.003$ cGy h⁻¹ U⁻¹ for $89.5^\circ < \theta < 90.5^\circ$ for this study, with the uncertainty based on the combined statistical uncertainties of the $\dot{d}(r_0, \theta_0)$ and s_K simulations.

TABLE I. Monte Carlo depth-dose curves along the central axis for 4, 5, 6, 7, and 8 cm diameter round applicators in polystyrene and ICRU 44 breast fitted to sixth-order polynomials of the form $\dot{D}(d, \text{diameter}, \text{medium}) = a_0 + a_1 d + a_2 d^2 + a_3 d^3 + a_4 d^4 + a_5 d^5 + a_6 d^6$, where the depth d is specified in millimeters and dose rate $\dot{D}$ has units cGy h⁻¹ U⁻¹.

	Polystyrene				
	SBA-4 cm	SBA-5 cm	SBA-6 cm	SBA-7 cm	SBA-8 cm
a_0	$1.021\,51 \times 10^{-1}$	$8.683\,80 \times 10^{-2}$	$7.257\,84 \times 10^{-2}$	$6.085\,76 \times 10^{-2}$	$5.159\,55 \times 10^{-2}$
a_1	$-5.185\,74 \times 10^{-3}$	$-3.981\,84 \times 10^{-3}$	$-2.817\,08 \times 10^{-3}$	$-2.100\,63 \times 10^{-3}$	$-1.684\,61 \times 10^{-3}$
a_2	$1.508\,68 \times 10^{-4}$	$1.101\,69 \times 10^{-4}$	$5.840\,74 \times 10^{-5}$	$4.149\,25 \times 10^{-5}$	$3.706\,63 \times 10^{-5}$
a_3	$-2.841\,24 \times 10^{-6}$	$-2.206\,37 \times 10^{-6}$	$-7.187\,93 \times 10^{-7}$	$-6.401\,64 \times 10^{-7}$	$-7.430\,80 \times 10^{-7}$
a_4	$3.295\,42 \times 10^{-8}$	$3.045\,54 \times 10^{-8}$	$4.464\,11 \times 10^{-9}$	$8.189\,00 \times 10^{-9}$	$1.117\,96 \times 10^{-8}$
a_5	$-2.082\,45 \times 10^{-10}$	$-2.469\,35 \times 10^{-10}$	$-3.783\,72 \times 10^{-12}$	$-6.973\,38 \times 10^{-11}$	$-9.567\,40 \times 10^{-11}$
a_6	$5.352\,12 \times 10^{-13}$	$8.550\,08 \times 10^{-13}$	$-6.713\,82 \times 10^{-14}$	$2.610\,78 \times 10^{-13}$	$3.352\,33 \times 10^{-13}$
	ICRU 44 breast				
	SBA-4 cm	SBA-5 cm	SBA-6 cm	SBA-7 cm	SBA-8 cm
a_0	$1.054\,58 \times 10^{-1}$	$8.968\,54 \times 10^{-2}$	$7.493\,98 \times 10^{-2}$	$6.283\,48 \times 10^{-2}$	$5.326\,21 \times 10^{-2}$
a_1	$-5.347\,40 \times 10^{-3}$	$-4.107\,18 \times 10^{-3}$	$-2.899\,59 \times 10^{-3}$	$-2.157\,37 \times 10^{-3}$	$-1.715\,94 \times 10^{-3}$
a_2	$1.557\,31 \times 10^{-4}$	$1.132\,30 \times 10^{-4}$	$5.996\,95 \times 10^{-5}$	$4.191\,38 \times 10^{-5}$	$3.619\,44 \times 10^{-5}$
a_3	$-2.940\,16 \times 10^{-6}$	$-2.243\,19 \times 10^{-6}$	$-7.359\,27 \times 10^{-7}$	$-6.172\,06 \times 10^{-7}$	$-6.812\,99 \times 10^{-7}$
a_4	$3.415\,39 \times 10^{-8}$	$3.043\,88 \times 10^{-8}$	$4.536\,90 \times 10^{-9}$	$7.378\,04 \times 10^{-9}$	$9.777\,44 \times 10^{-9}$
a_5	$-2.154\,14 \times 10^{-10}$	$-2.423\,51 \times 10^{-10}$	$-3.524\,13 \times 10^{-12}$	$-5.945\,08 \times 10^{-11}$	$-8.137\,74 \times 10^{-11}$
a_6	$5.494\,02 \times 10^{-13}$	$8.257\,28 \times 10^{-13}$	$-6.932\,47 \times 10^{-14}$	$2.146\,89 \times 10^{-13}$	$2.803\,87 \times 10^{-13}$

FIG. 4. (a) Comparison of $g_L(r)$ data for the Nucletron HDR ^{192}Ir V2 brachytherapy source. The data of Daskalov *et al.* were calculated in a 30 cm diameter liquid water sphere (Ref. 7) and corrected (as indicated by an asterisk) to approximate an unbounded sphere using the methodology of Granero *et al.* (Ref. 11). While both data sets were calculated for radii beyond those depicted, only data at common radii are shown. (b) Comparison of $F(r, \theta)$ data for the Nucletron HDR ^{192}Ir V2 brachytherapy source. Differences were typically $<1\%$ except for $r=0.25$ cm, where differences in Monte Carlo methods became pronounced.

III.B. Round applicator Monte Carlo simulations

A factor of $1360 \text{ cGy h}^{-1} \text{ U}^{-1} / \text{MeV g}^{-1} \text{ sp}^{-1}$ was used to correlate dose rate to PS or breast with MCNP5 results. The Monte Carlo depth-dose curves along the central axis for all five applicators in PS and breast were fitted to sixth-order polynomials with coefficients provided in Table I for use in a clinical nomogram. Differences between the polynomial fits and the Monte Carlo results were within $\pm 1.4\%$ ($k=1$) with the largest differences (4.2%) within the 5 mm region at the furthest depth from the HDR ^{192}Ir source for both PS and breast phantoms where the opposing applicator dose would dominate. Dose to breast was higher than dose to PS by 3.4%–4.0% for $d=0$ mm and $d=80$ mm, respectively.

III.C. Round applicator film and ion chamber measurements

For all five applicators, film and ion chamber depth-dose data in PS normalized to the surface agreed within 3.8% and

FIG. 5. (a) Dosimetric comparison of radiochromic EBT film and MCNP5 calculated dose distributions in polystyrene at 0 mm and 30 mm deep. (b) Radiochromic film profiles show dose distributions at 30 mm (left hand side) and 0 mm deep (right hand side) with color legend specified in cGy and x/y -axis scale shown in centimeters. Prior to normalization, the absolute dose between the EBT film and MCNP5 results agreed within 6%.

2.4% to Monte Carlo depth-dose data for $0 < d < 80$ mm. These variations were within the dosimetric measurement uncertainties ($k=1$), and the corresponding differences in absolute doses of 6% and 3%, respectively. This level of agreement was expected based on the estimated uncertainties in the measurement and simulation techniques.⁴ The biggest variations between the measured and the simulated results were observed at large d . Figure 5(a) depicts dose profiles at $d=0$ mm and $d=30$ mm for Monte Carlo and radiochromic film, with 99% of the points estimated to be within 0.3 mm or 1.4% up to 50 mm beyond the aperture (lateral distance of 80 mm). The central axis dose falloff was steeper for the 4 cm diameter applicator in comparison to the 8 cm diameter applicator, with surface to 3 cm depth-dose ratios of 3.65 and 2.44, respectively. Figure 5(b) shows the cylindrical dose

FIG. 6. Parallel-opposed radiochromic film isodose distribution (color legend specified in cGy) for a dual-vector treatment using two 6 cm diameter AccuBoost applicators applied to a 6 cm thick polystyrene phantom with x/y -axis scale shown in centimeters.

uniformity for two separate films, with the $d=30$ mm film irradiated three times longer to account for the decreased dose rate. The change in the beam profile as a function of depth indicates that the useful applicator aperture (90% of central axis dose) is about 1 cm less than the nominal applicator diameter on a single-axis basis. When the orthogonal dose is delivered to the same target volume, regions of lower dose from the first axis are coincident with regions of higher dose from the second axis, substantially mitigating the effect and expanding the uniform treatment zone. This effect has not yet been quantified and is the subject of ongoing research. For the 4–8 cm round applicators within this aperture, the minimum surface dose in the radial direction relative to the central axis dose ranged from -7.6% to -0.1% , respectively. Similarly, the maximum surface dose ranged from $+0.0\%$ to $+4.4\%$. A dose summation as measured with radiochromic film for two parallel-opposed beams is shown in Fig. 6.

IV. DISCUSSION

Validation of the Monte Carlo simulations was assisted by agreement obtained for $g_L(r)$, $F(r, \theta)$, and M_{MC} for the HDR ^{192}Ir source model with published results.⁷ While there are no published Monte Carlo studies on the round applicator dose distributions other than the current study, agreement obtained with the measured results served to validate the simulation results. To further corroborate these results, a comparison of Monte Carlo-derived central axis depth-dose in PS between the 6, 7, and 8 cm diameter applicators was made to the depth-dose of the published D45, D53, and D60 D-shaped applicators.⁴ The justification for these comparisons is based on similarity (within 7%) of the applicator aperture area: 28.27, 38.48, and 50.27 cm² for the 6, 7, and 8 cm diameter round applicators versus 26.40, 36.62, and 46.94 cm² for the respective D-shaped applicators. For 0

$< d < 80$ mm, the average depth-dose ratios were 1.009 ± 0.010 , 1.021 ± 0.015 , and 1.023 ± 0.021 , respectively.

For generalization of these results for a variety of breast compression thicknesses t , simulations were also performed in PS and breast for $t=40$ mm. The central axis dose rate at $d=10$ mm was only 0.5% less compared to $t=80$ mm, but decreased by 3% when in contact with the opposing paddle ($d=40$ mm). Less important, decreasing phantom radius from $R=15$ cm to $R=8$ cm, decreased dose by 0.1% and 0.6% at $d=10$ mm and $d=40$ mm, respectively. Given the dominant dose contributions from the opposing applicator at this location (about a factor of 4 higher) due to the dual-vector treatment delivery, simplification of the breast to $t=40$ mm and $R=8$ cm in comparison to the standardized geometry of $t=80$ mm and $R=15$ cm produced central axis total dose errors $< 2\%$ for parallel-opposed HDR ^{192}Ir beams. While total dose errors were larger off axis near the phantom periphery, the dose rate at these locations was much lower, especially outside the applicator aperture where the dose rate was at least a factor of 10 less than the central axis surface dose rate.

Other than the cylindrical sampling voxels for the round applicator reducing the statistical uncertainties, there were no differences between the uncertainty analysis of the current study and the prior D-applicator study.⁴ Specifically, the statistical standard uncertainties ($k=1$) for all applicators and both phantom materials were approximately 0.1%, 0.2%, and 0.4% at $d=0$, 30, and 70 mm on the central axis, respectively. Several centimeters outside the applicator aperture, the statistical uncertainties at these same depths were nearly constant at $\sim 0.3\%$. Smaller statistical uncertainties were observed than in Table I of Yang and Rivard⁴ due to the cylindrical sampling voxels used herein. However, the current overall Monte Carlo dosimetric uncertainty of 3.2% ($k=1$, which includes type B systematic uncertainties) was not significantly different from that of Yang and Rivard.

V. CONCLUSION

Monte Carlo benchmarking simulations of the HDR ^{192}Ir source using the MCNP5 radiation transport code indicated agreement within 1% of the published results over the radial/angular region of interest. Additional simulations and radiation dose measurements using radiochromic EBT film and air ionization chambers of the AccuBoost round breast brachytherapy applicators have been performed with methods previously established for the AccuBoost D-applicators. When applying these methods to the round applicators, dosimetric agreement within a few percent was obtained along the central axis between measurements and simulations in PS. Polynomial fit depth-dose curves provide a simple and accurate basis for a nomogram. Additional Monte Carlo and dosimetry measurement studies of this new treatment device should be researched by other investigators.

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