Determination of exit skin dose for ^{192}Ir intracavitary accelerated partial breast irradiation with thermoluminescent dosimeters

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Purpose: Intracavitary accelerated partial breast irradiation (APBI) has become a popular treatment for early stage breast cancer in recent years due to its shortened course of treatment and simplified treatment planning compared to traditional external beam breast conservation therapy. However, the exit dose to the skin is a major concern and can be a limiting factor for these treatments. Most treatment planning systems (TPSs) currently used for high dose-rate (HDR) ^{192}Ir brachytherapy overestimate the exit skin dose because they assume a homogeneous water medium and do not account for finite patient dimensions. The purpose of this work was to quantify the TPS overestimation of the exit skin dose for a group of patients and several phantom configurations.

Methods: The TPS calculated skin dose for 59 HDR ^{192}Ir APBI patients was compared to the skin dose measured with LiF:Mg,Ti thermoluminescent dosimeters (TLDs). Additionally, the TPS calculated dose was compared to the TLD measured dose and the Monte Carlo (MC) calculated dose for eight phantom configurations. Four of the phantom configurations simulated treatment conditions with no scattering material beyond the point of measurement and the other four configurations simulated the homogeneous scattering conditions assumed by the TPS. Since the calibration TLDs for this work were irradiated with ^{137}Cs and the experimental irradiations were performed with ^{192}Ir , experiments were performed to determine the intrinsic energy dependence of the TLDs. Correction factors that relate the dose at the point of measurement (center of TLD) to the dose at the point of interest (basal skin layer) were also determined and applied for each irradiation geometry.

Results: The TLD intrinsic energy dependence for ^{192}Ir relative to ^{137}Cs was $1.041 \pm 1.78\%$. The TPS overestimated the exit skin dose by an average of 16% for the group of 59 patients studied, and by 9%–15% for the four phantom setups simulating treatment conditions. For the four phantom setups simulating the conditions assumed by the TPS, the TPS calculated dose agreed well with the TLD and MC results (within 3% and 1%, respectively). The inverse square geometry correction factor ranged from 1.023 to 1.042, and an additional correction factor of 0.978 was applied to account for the lack of charged particle equilibrium in the TLD and basal skin layer.

Conclusions: TPS calculations that assume a homogeneous water medium overestimate the exit skin dose for intracavitary APBI treatments. It is important to determine the actual skin dose received during intracavitary APBI to determine the skin dose-response relationship and establish dose limits for optimal skin sparing. This study has demonstrated that TLDs can measure the skin dose with an expanded uncertainty ($k=2$) of 5.6% when the proper corrections are applied. © 2010 American Association of Physicists in Medicine. [DOI: [10.1118/1.3429089](https://doi.org/10.1118/1.3429089)]

Key words: ^{192}Ir , accelerated partial breast irradiation, skin dose, thermoluminescent dosimeters, energy dependence

I. INTRODUCTION

Since the introduction of the MammoSite[®] radiation therapy system (RTS) (Hologic, Inc., Marlborough, MA) in 2002, high dose-rate (HDR) ^{192}Ir intracavitary accelerated partial breast irradiation (APBI) has become increasingly popular and several new applicators have been developed. The MammoSite RTS uses a balloon catheter with a single centered

source channel. To enhance dose-shaping capabilities, several new intracavitary APBI applicators with multiple source channels have been developed.^{1–4} One example is Contura[™] (SenoRx, Inc., Aliso Viejo, CA), which uses a balloon catheter with a central source channel surrounded by four peripheral source channels.²

Several radiation-induced skin effects have been reported

in the literature for MammoSite treatments.^{5–13} With the advent of several new intracavitary APBI devices that claim to modify the dose distribution and reduce the exit skin dose, it is particularly important to quantify the actual skin doses received. Most brachytherapy treatment planning systems (TPSSs) calculate dose using the formalism proposed by the American Association of Physicists in Medicine Task Group 43 (TG-43), which assumes a homogeneous water medium.^{14,15} They do not account for the absence of water around the breast and therefore overestimate the exit skin dose due to backscatter assumptions.¹⁶ Attempts have been made by several investigators to measure the exit skin dose received during intracavitary APBI treatments. The detectors have included thermoluminescent dosimeters (TLDs),^{9,17,18} radiochromic film,¹⁹ and metal oxide semiconductor field effect transistors.^{18,20,21} While all of these detectors are relatively thin and therefore well-suited for surface measurements, they all measure the dose at some distance above the skin. As demonstrated by Stathakis *et al.*²² for high energy photon beams, corrections that account for the dose gradient and electronic disequilibrium at the surface must be applied to determine the actual dose at the basal skin layer from measurements made at the surface. However, these corrections have been largely neglected in reported skin dose measurements for intracavitary APBI. Recently, Garnica-Garza²³ presented a set of Monte-Carlo-derived TLD cross-calibration factors for HDR ^{192}Ir intracavitary APBI to determine the skin dose from TLD measurements. However, those factors were developed for a particular geometry and combined several effects into a single cross-calibration factor which cannot be readily modified for application to different conditions. Also, no actual TLD measurements were included and the intrinsic energy dependence of the TLDs was not determined or included in their cross-calibration factor. In this work, a straightforward method of measuring the exit skin dose for intracavitary APBI with TLDs was developed and both patient and phantom studies were performed to compare the TLD measured and MC calculated skin dose to the TG-43 calculated skin dose currently used in brachytherapy treatment planning. The necessary TLD corrections have been broken down into modifiable components that can be readily applied to different conditions.

In this study, certain commercially available products are referred to by name. These references are for informational purposes only and imply neither endorsement by the University of Wisconsin Medical Radiation Research Center (UW-MRRC) nor that these products are the best or only products available for the stated purposes.

II. METHODS AND MATERIALS

Harshaw TLD-100 chips ($3.2 \times 3.2 \times 0.89$ mm³) (Thermo Scientific, Oakwood Village, OH) were used for all of the measurements in this work. The TLDs were annealed for 1 h at 400 °C, cooled for 20–30 min on an aluminum plate, annealed for 24 h at 80 °C, and then cooled to room temperature where they remained for at least 24 h before use. For the phantom measurements, an irradiation was per-

formed with a Hopewell G10 ^{137}Cs irradiator at the University of Wisconsin Radiation Calibration Laboratory (UWRCL) to determine individual TLD sensitivities. The TLDs were annealed for 24 h at 80 °C between the TLD sensitivity reading and the experimental irradiation. The TLDs were read at least 24 h post-irradiation with a Harshaw 5500 automatic hot gas reader. The same time-temperature profile was used to read all of the TLDs. The temperature of the gas was 50 °C at the start of each TLD reading, then increased by 15 °C/s to a maximum of 350 °C with an acquisition time of 26.7 s.

The Monte Carlo (MC) simulations in this work were performed using Version 5.1.40 of the Monte Carlo N-Particle code (MCNP5) developed by Los Alamos National Laboratory.²⁴ The mcplib04 (Refs. 25 and 26) and el03 (Ref. 27) libraries were used for the photon and electron cross sections. The photon energy cutoff was 1 keV and the electron energy cutoff was 10 keV. The *F8 energy deposition tally was used to score the absorbed dose and the forced collision variance reduction technique was used to increase the number of charged particles in the vicinity of the scoring regions.²⁴ While the F6 collision kerma tally has been shown to be equivalent to absorbed dose from ^{192}Ir sources at distances beyond the first few millimeters,^{28,29} the F6 collision kerma tally is not appropriate near the surface or at material boundaries where charged particle equilibrium does not exist.

II.A. Exit dose formalism

For each of the patient and phantom TLD measurements, the exit dose to water (in Gy) at the point of interest D_{POI} was calculated using the equation

$$D_{\text{POI}} = \frac{R \cdot N}{\eta(^{192}\text{Ir})} \cdot G \cdot k, \quad (1)$$

where R is the corrected TLD reading (in nC), N is the calibration coefficient (in Gy/nC), $\eta(^{192}\text{Ir})$ is the TLD intrinsic energy dependence correction factor, and G and k are both correction factors that relate the dose at the point of measurement (center of the TLD) to the dose at the point of interest. For all of the phantom and patient measurements in this work, the point of interest was the basal skin layer which was defined at a depth of 0.07 mm beneath the skin surface in accordance with the recommendations of the International Commission on Radiological Protection (ICRP) and the International Commission on Radiation Units and Measurements (ICRU).^{30,31} To obtain R , the raw TLD reading was multiplied by a photomultiplier tube (PMT) linearity correction and corrected for background by subtracting the average reading of a set of control TLDs. For the phantom measurements, the raw TLD reading was also corrected for individual TLD sensitivity. The patient TLD readings were not corrected for individual TLD sensitivity since the TLDs were taken from a large batch of sorted TLDs that were closely grouped according to their response. N was determined for each batch of TLDs by irradiating a set of calibration TLDs to a known dose to water with the UWRCL's ^{137}Cs irradiator.

$\eta(^{192}\text{Ir})$, G , and k will be discussed in more detail in the following sections.

II.A.1. TLD intrinsic energy dependence for ^{192}Ir relative to ^{137}Cs

Calibration TLDs for all of the HDR ^{192}Ir patient and phantom measurements were irradiated with the UWRC's ^{137}Cs irradiator traceable to NIST air-kerma standards. Performing routine calibrations with HDR ^{192}Ir is not practical due to the time-consuming methods required to achieve accurate air-kerma rate measurements and the limited area of uniform air-kerma rate available for irradiating multiple TLDs simultaneously. Since TLDs respond differently to different photon energies,³² a correction factor relating the TLD response to ^{192}Ir to the TLD response to ^{137}Cs was determined and applied. The absorbed dose to the TLD per unit dose to water for ^{192}Ir photons relative to ^{137}Cs photons can be calculated analytically using the mass energy-absorption coefficients for TLD and water at each energy.³² For ^{192}Ir and ^{137}Cs , the absorbed-dose energy dependence calculated using their approximate mean photon energies is 1.00079, which is negligibly small and therefore not included in Eq. (1). However, similar to the findings of Nunn *et al.*³³ for low-energy photon beams, it was found that there is an additional component to the energy dependence that is likely due to solid state processes in the TLDs that cannot be determined analytically. This additional component has been termed the intrinsic energy dependence. Nunn *et al.*³³ measured the intrinsic energy dependence of TLDs as a function of photon energy for 20–250 kVp photon beams relative to ^{60}Co photons. In this work, the same methodology was used to determine the intrinsic energy dependence of TLDs for ^{192}Ir photons relative to ^{137}Cs photons, termed $\eta(^{192}\text{Ir})$, using the equation

$$\eta(^{192}\text{Ir}) = \frac{M(^{192}\text{Ir})}{C_{\text{MC}}(^{192}\text{Ir})} = \frac{(R/D_{\text{TLD}})^{192}\text{Ir}}{(R/D_{\text{TLD}})^{137}\text{Cs}}, \quad (2)$$

which divides the TLD measured energy dependence $M(^{192}\text{Ir})$ by the MC calculated energy dependence $C_{\text{MC}}(^{192}\text{Ir})$. Dividing $M(^{192}\text{Ir})$ by $C_{\text{MC}}(^{192}\text{Ir})$ removes the effects of the different experimental setups used for the two beams (i.e., holders, etc.) as well as the absorbed-dose energy dependence that can be calculated analytically. Further explanation of M and C_{MC} can be found in the article of Nunn *et al.*³³ In Eq. (2), R is the measured signal from the TLD corrected for PMT linearity, background, and individual TLD sensitivity, and D_{TLD} is the MC calculated dose to TLD. For this work, in-air irradiations corresponding to a known air kerma were performed for ^{192}Ir and ^{137}Cs with Virtual Water™ (VW) (Med-Cal Inc., Verona, WI) TLD holders that were thick enough to provide charged particle equilibrium. For ^{192}Ir , the seven-distance technique described by Stump *et al.*³⁴ was used to determine the position and air-kerma rate of the VariSource VS2000 HDR ^{192}Ir source with an Exradin model A3 spherical ionization chamber. Then, the ionization chamber was replaced with the

TLDs in a VW holder for the TLD irradiations. Additional TLD irradiations were also performed with the back of the holder removed to determine whether the lack of backscatter affected the energy dependence. MC simulations were performed to determine the dose to TLD per unit air kerma for each experimental setup.

II.A.2. Geometry correction factor G

The geometry correction factor G accounts for the inverse square relationship between the dose at the point of interest and the dose at the point of measurement. Since the point of interest in this study (basal skin layer) was closer to the source than the point of measurement (center of the TLD), G was always greater than unity. For single dwell position patient measurements,

$$G = \left[\frac{d_{\text{TLD}}}{d_{\text{skin}}} \right]^2 = \left[\frac{d_{\text{surface}} + 0.45 \text{ mm}}{d_{\text{surface}} - 0.07 \text{ mm}} \right]^2 = \left[\frac{d_{\text{TLD}}}{d_{\text{TLD}} - 0.52 \text{ mm}} \right]^2, \quad (3)$$

where d_{TLD} is the distance from the source to the center of the TLD and d_{skin} is the distance of the source to the basal skin layer. For the measurements in this study, the form of the equation expressed in terms of the source to surface distance d_{surface} was used to calculate G because the distance from the source to the skin surface was measured in the TPS rather than d_{TLD} or d_{skin} . However, the equation expressed in terms of only d_{TLD} is included here because it simplifies the equation when it is extended to multiple dwell position treatments, as discussed below. It should be noted, however, that d_{TLD} was not measured directly but was instead determined from d_{surface} via the relationship

$$d_{\text{TLD}} = d_{\text{surface}} + 0.45 \text{ mm}, \quad (4)$$

where 0.45 mm is half the thickness of the TLD.

For treatments with multiple dwell positions, the calculation of G is slightly more complex as d_{TLD} for each dwell position must be accounted for and weighted in terms of the dwell time. Another way to express Eq. (1) is

$$D_{\text{POI}} = D_{\text{POM}} \cdot G \cdot k, \quad (5)$$

where D_{POM} is the dose to the point of measurement. Therefore, the total dose to the point of interest from n dwell positions ideally could be calculated using

$$D_{\text{POI}} = \sum_{i=1}^n D_{\text{POM},i} \cdot G_i \cdot k_i, \quad (6)$$

where G_i is calculated for each dwell position i using Eq. (3).

Since the contribution to the TLD dose from each dwell position is unknown, individual G values for each dwell position cannot be applied. Instead, a single G value, termed G_{T} , must be used. G_{T} is an overall geometry correction factor that combines the G factors from each of the dwell positions. A simple way to approximate G_{T} is to use a dwell time weighted average of the source to surface distances for each of the dwell positions as d_{surface} in Eq. (3). This simple ap-

proximation works quite well for situations in which the source to surface distances are roughly of the same order, which is the case for most intracavitary APBI treatments since the distance between dwell positions is generally small compared to the distance from each dwell position to the surface. However, a more accurate equation for calculating G_T , derived as part of this investigation, is

$$G_T = \frac{1}{\sum_{i=1}^n t_i / d_{\text{TLD},i}^2} \cdot \sum_{i=1}^n \frac{t_i}{(d_{\text{TLD},i} - 0.52 \text{ mm})^2}, \quad (7)$$

which explicitly accounts for the inverse square dose fall-off contribution from each of the n dwell positions, with t_i representing the dwell time for each position. Although approximating G_T using an average source to surface distance would work well for most intracavitary APBI cases, calculating G_T from Eq. (7) is a more robust way to account for the individual contributions to G_T from the different dwell positions and dwell times. However, it should be noted that Eq. (7) is still based on the assumption that the incident angle of radiation from all of the dwell positions is approximately normal to the TLD and skin. For this work, G_T was approximated using the average source to surface distance for the MammoSite patients and calculated with Eq. (7) for the Contura patients.

II.A.3. Additional correction factor k

An additional correction factor k is included in the exit skin dose equation to account for the lack of electronic equilibrium for measurements made without any backscatter material beyond the skin or TLD and any other deviations from the inverse square relationship between the dose at the point of interest (basal skin layer) and the dose at the point of measurement (center of the TLD). The inverse square relationship is valid in the absence of photon attenuation and scatter. It is also valid in situations where the photon attenuation and scatter cancel each other out, which is approximately the case for ^{192}Ir in water at close distances.³⁵ For MammoSite, the balloon size, contrast level within the balloon, and the source to surface distance alter the attenuation and scatter between the source and the point of interest. To determine whether any of these parameters affect the photon attenuation and scatter between the point of interest and the point of measurement, MC simulations were performed to calculate the basal skin dose and TLD dose for a wide range of treatment parameters. The breast was modeled as a cylinder with a radius of 8 cm and a height of 8 cm and was positioned on top of a cylinder with a radius of 15 cm and a height of 20 cm to approximate the patient's body (Fig. 1). The inner surface of the MammoSite balloon was modeled as a spherical shell with an inner radius of 2 or 3 cm and an outer radius of 2.05 or 3.05 cm for a balloon thickness of 0.05 cm. The TLD had dimensions of $3.2 \times 3.2 \times 0.89 \text{ mm}^3$ and was positioned along the transverse axis of the source on the flat upper surface of the breast cylinder. The basal skin layer had the same lateral dimensions as the TLD ($3.2 \times 3.2 \text{ mm}^2$) and extended from 0.04 to 0.10 mm beneath the surface. This thickness was chosen in light of

FIG. 1. Geometry for k factor MC simulation with a balloon radius of 2 cm and source to surface distance of 2.5 cm.

suggestions that the depth of the basal skin layer may be as shallow as 0.04 mm (Refs. 23 and 36) while maintaining an average depth of 0.07 mm in accordance with the ICRP and ICRU recommendations.^{30,31} The breast, body, and basal skin layer were modeled as water. The same VS2000 model and spectrum described later in Section II C were used for these simulations. The MammoSite balloon was modeled as pure silicone with a density of 2.33 g/cm^3 .³⁷ The composition and density for each contrast concentration was taken from Kassas *et al.*³⁸ The entire geometry was inside a sphere of dry air with a radius of 110 cm. The dose to the TLD D_{TLD} and the dose to the basal skin layer in the absence of the TLD D_{skin} were obtained in two separate simulations for each geometry with sufficient particles transported to achieve statistical uncertainties of less than 0.4%. The additional correction that accounts for the effects of electronic disequilibrium and other deviations from the inverse square relationship between the basal skin layer and the center of the TLD is defined as

$$k = \frac{D_{\text{skin}}}{D_{\text{TLD}} \cdot D_{\text{water}} / D_{\text{TLD}} \cdot G}, \quad (8)$$

where $D_{\text{water}} / D_{\text{TLD}} = 1.199$ is the ratio of the mass energy-absorption coefficient for water to that for LiF at 400 keV

(Ref. 39) and G is the geometry correction for a single dwell position defined in Eq. (3).

II.B. Patient measurements

Patient measurements were made at three collaborating clinics. Of the 59 patients, 56 were treated with MammoSite and three were treated with Contura. Different TLD-100 chips were used to measure the exit skin dose for each treatment fraction. The clinicians determined the TLD placement (normally at the estimated maximum skin dose point) and provided the TLD position, the source dwell position(s), the dwell time(s), and the TPS predicted skin dose for the point on the skin where the TLD was placed. During the five-day course of treatment (usually on the third day), a set of calibration TLDs was irradiated using the UWRCL ^{137}Cs irradiator. Along with the patient and calibration TLDs, two sets of control TLDs were used for each patient to account for background. One set of control TLDs was sent to the clinic and the other was kept at the UWMRRC. Once the patient and control TLDs were received from the clinic, they were read together with the UWMRRC calibration and control TLDs with a Harshaw 5500 automatic TLD reader.

II.C. Phantom measurements

A phantom exit dose experiment was performed to compare the TLD measured dose, the TPS predicted dose, and the MC calculated dose for well-defined experimental conditions. The phantom consisted of varying thicknesses of $30 \times 30 \text{ cm}^2$ slabs of VW. A 6 cm thick slab of VW was placed beneath the source and additional VW slabs were added above the source to achieve source to surface distances (d_{surface}) of 3, 4, 5, and 6 cm. A single TLD was used for each irradiation and six irradiations were performed for each d_{surface} . A Varian VS2000 HDR ^{192}Ir source located at the UWRCL was used for these irradiations and the dwell time for each irradiation was based on a prescription of 3.4 Gy at 3 cm, which is typical for intracavitary APBI treatment fractions. For each depth, there were three TLD measurements without backscatter and three with backscatter. For the measurements made without backscatter, the TLD was placed on the surface of the VW phantom [see Fig. 2(a)]. For the measurements with backscatter, a 1 mm thick layer of VW was added around the TLD and an additional 15 cm of VW was added above the TLD layer to approximate full backscatter conditions [see Fig. 2(b)]. Equation (1) was used to determine the TLD measured skin dose. MC simulations for each of the experimental setups were performed to calculate the dose to TLD. The MC dose to TLD results were multiplied by the ratio of the mass energy-absorption coefficient for water to that for LiF at 400 keV, $D_{\text{water}}/D_{\text{TLD}}=1.199$, and by the geometry correction G discussed in Sec. II A 2 to attain the dose to water at a depth of 0.07 mm beneath the phantom surface. For the geometries without backscatter, the additional correction factor of $k=0.978$ was applied to account for the effects of electronic disequilibrium. This correction was not applied to the geometries with backscatter since $k=1$ when electronic equilibrium exists. The MC mod-

FIG. 2. Phantom configurations for surface dose experiment (shown with d_{surface} of 6 cm): (a) Without backscatter; (b) with backscatter. The inset shows the TLD layer.

els included the VW phantom, the VS2000 source, and the TLD, all within a 110 cm radius sphere of air. The VW density and composition were taken from Murphy *et al.*⁴⁰ The TLD density and composition were taken from Nunn *et al.*³³ and the modeled TLD dimensions were $3.2 \times 3.2 \times 0.89 \text{ mm}^3$. The composition of dry air was taken from Berger *et al.*³⁷ and a density of $1.197 \times 10^{-3} \text{ g/cm}^3$ was used, which is consistent with standard environmental conditions for North America (temperature of 22 °C and pressure of 101.33 kPa). The dimensions and materials for the VS2000 source were based on the model of Angelopoulos *et al.*⁴¹ The photon spectrum was taken from the table of radioactive isotopes database.⁴² The TPS predicted dose for each d_{surface} was calculated using the TG-43 line source calculation with the VS2000 dose rate constant and radial dose function from Angelopoulos *et al.*^{14,41} A polynomial fit of the radial dose function data was used to determine $g_L(r)$ for a depth of 0.07 mm beneath the phantom surface (i.e., $r=d_{\text{surface}}-0.07 \text{ mm}$).

III. RESULTS AND DISCUSSION

III.A. TLD intrinsic energy dependence for ^{192}Ir relative to ^{137}Cs

The results of the energy dependence measurements and simulations are presented in Table I. The intrinsic energy

TABLE I. Results of energy dependence measurements and simulations.

TLD measured energy dependence, $M(^{192}\text{Ir})$	MC calculated energy dependence, $C_{\text{MC}}(^{192}\text{Ir})$	Intrinsic energy dependence, $\eta(^{192}\text{Ir})$
1.044	1.003	1.041

dependence of the TLDs for ^{192}Ir relative to ^{137}Cs , $\eta(^{192}\text{Ir})$, was determined to be $1.041 \pm 1.78\%$. This correction was applied to all of the TLD measurements in this work since all of the calibration TLDs were irradiated with ^{137}Cs . The energy dependence measurements performed without backscatter agreed within the stated uncertainty. Therefore, the same energy dependence value was used for all of the measurements, regardless of the presence or absence of backscatter. The value of $\eta(^{192}\text{Ir})$ determined in this work is consistent with the intrinsic energy dependence curve of Nunn *et al.*³³ although no photon beams with the effective energy of ^{192}Ir were included in their work.

III.B. Correction factors G and k

The results of the MC k factor simulations are presented in Table II along with the calculated values of G . For the geometries simulated, G ranged from 1.023 to 1.042 and k ranged from 0.972 to 0.983. While the value of k may have a slight dependence on geometry or contrast concentration, no clear trend is evident outside of the statistical uncertainties of the simulations. Therefore, the average value of $k=0.978$ was applied to all of the phantom and patient measurements made with no backscatter regardless of the source to surface distance, balloon size, or contrast concentration. The variation in k for the different geometries was taken into account in the uncertainty determination in Sec. III E. For treatment

TABLE II. Summary of MC simulations and resulting values for correction factor k .

Balloon radius (cm)	Source to surface distance (cm)	Contrast concentration (by volume) (%)	$D_{\text{skin}}/D_{\text{TLD}}$	G	k
2	2.5	0	1.222	1.042	0.979
		5	1.225		0.981
		25	1.221		0.978
2	3.5	0	1.207	1.030	0.978
		5	1.207		0.978
		25	1.212		0.981
2	4.5	0	1.192	1.023	0.972
		5	1.194		0.974
		25	1.198		0.977
3	3.5	0	1.209	1.030	0.979
		5	1.210		0.980
		25	1.213		0.983
3	4.5	0	1.196	1.023	0.975
		5	1.199		0.977
		25	1.199		0.978

FIG. 3. Percent difference between the TPS predicted dose and the average TLD measured dose for 59 intracavitary APBI patients from three different clinics. Results for patients with two different treatment plan estimates (20, 24, and 57) are presented as two different patients.

geometries outside of the range of parameters studied and for depths other than 0.07 mm, further simulations should be performed to determine the appropriate value of k .

III.C. Patient measurements

The TLD measured exit skin dose for each fraction was determined using Eq. (1) and the average over all fractions was compared to the TPS estimated skin dose for each patient. Three of the patients had two different TPS estimated doses. The treatment plan was changed during the course of treatment for one MammoSite patient (patient number 20 in Fig. 4) and one Contura patient (57), and the dose was measured at one point for three of the fractions and a different point for the other seven fractions for one MammoSite patient (24). The averaged results for each patient are presented in Fig. 3. The TPS overestimated the skin dose for all but two of the patients. The percent difference between the TPS and TLD determined skin doses ranged from -13% to 47% .

FIG. 4. Percent difference between the TPS predicted dose and TLD measured dose for 59 intracavitary APBI patients from three different clinics. Closed symbols represent percent difference for each fraction. Open symbols represent the average for each patient. Patients with two open symbols (20, 24, and 57) had two different treatment plans during the course of their treatment.

TABLE III. TLD measured dose, MC calculated dose, and TG-43 calculated dose for phantom configurations with backscatter, along with percent differences among them.

Source to surface distance (cm)	TLD measured dose (cGy)	MC calculated dose (cGy)	TG-43 calculated dose (cGy)	Percent difference MC vs TLD (%)	Percent difference TPS vs TLD (%)	Percent difference TPS vs MC (%)
3.0	333	342	342	3	3	0
4.0	193	192	192	0	-1	0
5.0	124	122	121	-1	-2	-1
6.0	85	84	83	-1	-2	-1

The average overestimation was 16%, but varied from clinic to clinic with an average overestimation for MammoSite patients of 22% at site A, 14% at site B, 14% at site C, and an average overestimation for Contura patients of 8% at site C. The results for each measured fraction are presented in Fig. 4 to illustrate the variation from fraction to fraction. The standard deviation of the overestimation over all fractions ranged from 2% to 19%, with an average standard deviation of 8% for site A, 7% for site B, 10% for site C MammoSite patients, and 6% for site C Contura patients. The wide variation in results from patient to patient is likely due in part to difficulties in registering the positions of the TLD and dwell position(s) within the TPS. The TLD position is marked with a reference BB during CT scanning. The resolution of the CT images limits the accuracy with which the dwell position(s) and the breast surface can be determined and adjusting the window and leveling of the planning image can also alter the perceived location of the breast surface. The variation in results from fraction to fraction is likely due in part to changes in the source position relative to the TLD. For some patients, different source to surface distances were reported for each fraction and were therefore taken into account when calculating G for the TLD measured skin dose. However, the TPS estimated skin dose is more sensitive to changes in source to surface distance than G and only one TPS estimated skin dose was provided for each data point in Fig. 4.

III.D. Phantom measurements

In Tables III and IV, the TLD measured dose, the MC calculated dose, and the TG-43 calculated dose for each of the phantom configurations are shown along with the percent differences between them. In Fig. 5, the same results are

plotted as the dose multiplied by d_{surface}^2 vs d_{surface} . The results are plotted this way so that the inverse square component of the dose falloff does not obscure the effect of backscatter on the overall dose. For all of the configurations, the TLD and MC results agreed within the expanded TLD uncertainty and did not exhibit a systematic offset. The agreement of the TLD and MC results supports the application of the TLD intrinsic energy dependence, $\eta(^{192}\text{Ir})$. If the intrinsic energy dependence of the TLDs had not been accounted for, there would have been an approximate 4% systematic offset between the TLD and MC results. For the experimental configurations with backscatter, both the TLD and the MC results agreed well with the TG-43 results (within 3% and 1%, respectively). However, for the configurations without backscatter, the TG-43 calculations overestimated the dose by 9%–15% for the studied distances relative to the TLD and MC results. This overestimation was expected since the dosimetry parameters used for the TG-43 calculations were determined for constant backscatter conditions. Also, this overestimation grows with increasing d_{surface} since backscatter contributes a greater fraction of the dose at further distances.

III.E. Uncertainty

The uncertainty budget for the patient and phantom measurements is shown in Table V. This is the uncertainty for the measured dose corrected to the point of interest directly below the TLD. The uncertainty budget is divided into six sections: One for each variable in Eq. (1), except for the calibration coefficient which has one section to represent the measured component and a second section to represent the MC calculated component. Sources of uncertainty for the

TABLE IV. TLD measured dose, MC calculated dose, and TG-43 calculated dose for phantom configurations without backscatter, along with percent differences among them.

Source to surface distance (cm)	TLD measured dose (cGy)	MC calculated dose (cGy)	TG-43 calculated dose (cGy)	Percent difference MC vs TLD (%)	Percent difference TPS vs TLD (%)	Percent difference TPS vs MC (%)
3.0	298	313	342	5	15	9
4.0	171	172	192	1	12	11
5.0	107	108	121	1	13	12
6.0	73	73	83	1	14	14

FIG. 5. TG-43 calculated dose, MC calculated dose, and TLD measured dose for phantom configurations, with and without backscatter, plotted as the product of dose and d_{surface}^2 as a function of d_{surface} . Error bars represent the total uncertainty ($k=1$) for the TLD measurements.

corrected TLD readings are the TLD reproducibility, the PMT linearity correction and the reader stability. For the TLD reproducibility, the Type A uncertainty was determined from the statistical uncertainty of the TLD calibrations. The uncertainty for the PMT linearity correction is greater for higher doses where larger corrections are applied. For the low dose energy dependence and calibration coefficient measurements, the uncertainty was determined by comparing measurements with and without the linearity correction applied. For the patient and phantom measurements, however, the PMT linearity correction was larger and the uncertainty was estimated by comparing the PMT linearity corrections for a clinically relevant dose (3.4 Gy) determined for the same TLD reader three years apart. Reader stability was determined to be a negligible source of uncertainty since it could not be determined independently from, and is assumed to be included in, the TLD reproducibility. For the calibration coefficient measurements, the sources of uncertainty were TLD reproducibility, air-kerma rate determination, TLD positioning, PMT linearity correction, reader stability, and field uniformity. For the calibration coefficient MC simulations, the sources of uncertainty were the MC statistical uncertainty, the photon spectrum and cross sections used, and the simulation geometry. The air-kerma rate determination components of the uncertainty budgets come from the UWRCL's uncertainty budget for each source. Several components make up the TLD intrinsic energy dependence uncertainty included in Table V. A detailed uncertainty budget for the energy dependence is shown in Table VI. The TLD measured and MC calculated components for ^{192}Ir and ^{137}Cs are added in quadrature since the energy response for ^{192}Ir is calculated relative to ^{137}Cs . An additional source of uncertainty has been added to Table V to account for the difference in energy dependence for the ^{192}Ir spectrum exiting the breast or phantom relative to the ^{192}Ir spectrum at the TLD for the energy dependence measurements described in Sec. II A 1. The G factor contribution to the uncertainty budget is the uncertainty in the distance between the source and the surface. The maximum deviation from the measured distance

TABLE V. Uncertainty budget for patient and phantom TLD measurements.

Parameter	Relative standard uncertainty (%)	
	Type A	Type B
<i>R-Corrected TLD readings</i>		
TLD reproducibility	0.87	
PMT linearity correction		0.87
Reader stability		0.00
<i>N-Calibration coefficient (measured component)</i>		
TLD reproducibility	0.87	
Air-kerma rate determination		0.85
TLD positioning		0.10
PMT linearity correction		0.10
Field uniformity		0.10
Reader stability		0.00
<i>N-Calibration coefficient (MC component)</i>		
MC statistical uncertainty	0.10	
Photon spectrum		0.04
Cross sections		0.01
Simulation geometry		0.35
<i>TLD intrinsic energy dependence</i>		
Combined uncertainty ^a	0.77	1.60
Effect of patient or phantom on spectrum		1.00
<i>G factor</i>		
TLD positioning (distance determination)		0.30
<i>k factor</i>		
MC statistical uncertainty	0.40	
Geometry variations		0.50
Combined uncertainty	1.50	2.35
Total uncertainty ($k=1$)		2.79
Expanded uncertainty ($k=2$)		5.58

^aFrom Table VI.

was estimated to be 4 mm and a rectangular distribution was assumed. The k factor contributions to the uncertainty budget were the MC statistical uncertainty and the variations in k due to the different simulation geometries detailed in Table II.

IV. CONCLUSION

Current brachytherapy TPS calculations based on TG-43 dosimetry overestimate the exit skin dose for intracavitary APBI. To date, skin sparing for intracavitary APBI has been based mainly on skin spacing and TPS isodose lines. While skin spacing recommendations made sense for the original MammoSite treatments with one central dwell position, they are not as practical for newer applicators with multiple dwell positions. Also, it is insufficient to rely on TG-43 isodose lines calculated for homogeneous conditions when it has been established that these dose estimates are inaccurate for realistic treatment conditions. With the development of new

TABLE VI. Uncertainty budget for TLD intrinsic energy dependence.

Parameter	Relative standard uncertainty (%)	
	Type A	Type B
^{192}Ir TLD measurements		
TLD reproducibility	0.63	
Air-kerma rate determination		1.32
TLD positioning		0.23
PMT linearity correction		0.10
Reader stability		0.00
Field uniformity		0.10
^{137}Cs TLD measurements		
TLD reproducibility	0.39	
Air-kerma rate determination		0.85
TLD positioning		0.10
PMT linearity correction		0.10
Reader stability		0.00
Field uniformity		0.10
^{192}Ir MC simulations		
MC statistical uncertainty	0.18	
Photon spectrum		0.04
Cross sections		0.01
^{137}Cs MC simulations		
MC statistical uncertainty	0.10	
Photon spectrum		0.03
Cross sections		0.01
Combined uncertainty	0.77	1.60
Total uncertainty ($k=1$)		1.78
Expanded uncertainty ($k=2$)		3.56

applicators and the increased use of low-energy miniature x-ray sources for intracavitary APBI, it is important to establish accurate dose limits for optimal skin sparing.

The inability of the TG-43 dose formalism to account for partial scatter conditions and accurately calculate the skin dose provides motivation for advances in TPS capabilities. Improved methods for calculating the skin dose should be benchmarked with actual measurements. Even with improved calculation methods that account for the lack of backscatter surrounding the breast, TPS skin dose estimates are limited by the resolution of the planning images and the ability to delineate the skin surface. Also, modern dose calculation methods based on collision kerma approximations of the absorbed dose will not be able to provide accurate dose calculations at the surface and near material boundaries where charged particle equilibrium does not exist. Due to their established utility in brachytherapy dosimetry and the ease of use for patient measurements, TLDs are appropriate dosimeters for intracavitary APBI skin dose measurements. This study has shown that TLDs can measure the skin dose with an expanded uncertainty ($k=2$) of 5.6% and has demonstrated the importance of correcting for the inverse square dose falloff and the effects of electronic disequilibrium.

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