On the feasibility of polyurethane based 3D dosimeters with optical CT for dosimetric verification of low energy photon brachytherapy seeds

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Purpose: To investigate the feasibility of and challenges yet to be addressed to measure dose from low energy (effective energy <50 keV) brachytherapy sources (Pd-103, Cs-131, and I-125) using polyurethane based 3D dosimeters with optical CT.

Methods: The authors' evaluation used the following sources: models 200 (Pd-103), CS-1 Rev2 (Cs-131), and 6711 (I-125). The authors used the Monte Carlo radiation transport code MCNP5, simulations with the ScanSim optical tomography simulation software, and experimental measurements with PRESAGE[®] dosimeters/optical CT to investigate the following: (1) the water equivalency of conventional (density = 1.065 g/cm³) and deformable (density = 1.02 g/cm³) formulations of polyurethane dosimeters, (2) the scatter conditions necessary to achieve accurate dosimetry for low energy photon seeds, (3) the change in photon energy spectrum within the dosimeter as a function of distance from the source in order to determine potential energy sensitivity effects, (4) the optimal delivered dose to balance optical transmission (per projection) with signal to noise ratio in the reconstructed dose distribution, and (5) the magnitude and characteristics of artifacts due to the presence of a channel in the dosimeter. Monte Carlo simulations were performed using both conventional and deformable dosimeter formulations. For verification, 2.8 Gy at 1 cm was delivered in 92 h using an I-125 source to a PRESAGE[®] dosimeter with conventional formulation and a central channel with 0.0425 cm radius for source placement. The dose distribution was reconstructed with 0.02 and 0.04 cm³ voxel size using the Duke midsize optical CT scanner (DMOS).

Results: While the conventional formulation overattenuates dose from all three sources compared to water, the current deformable formulation has nearly water equivalent attenuation properties for Cs-131 and I-125, while underattenuating for Pd-103. The energy spectrum of each source is relatively stable within the first 5 cm especially for I-125. The inherent assumption of radial symmetry in the TG43 geometry leads to a linear increase in sample points within the 3D dosimeter as a function of distance away from the source, which partially offsets the decreasing signal. Simulations of dose reconstruction using optical CT showed the feasibility of reconstructing dose out to a radius of 10 cm without saturating projection images using an optimal dose and high dynamic range scanning; the simulations also predicted that reconstruction artifacts at the channel surface due to a small discrepancy in refractive index should be negligible. Agreement of the measured with calculated radial dose function for I-125 was within 5% between 0.3 and 2.5 cm from the source, and the median difference of measured from calculated anisotropy function was within 5% between 0.3 and 2.0 cm from the source.

Conclusions: 3D dosimetry using polyurethane dosimeters with optical CT looks to be a promising application to verify dosimetric distributions surrounding low energy brachytherapy sources. © 2014 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4883779>]

Key words: LDR brachytherapy, 3D dosimetry, PRESAGE[®], Monte Carlo

1. INTRODUCTION

Seeds emitting low energy photons (effective energy <50 keV) are utilized in brachytherapy, including temporary eye plaque implants for optical melanoma¹ and permanent seed implants for prostate.² Commonly used low energy photon isotopes for permanent seed implant are Iodine-125 (I-125), Palladium-103 (Pd-103), and Cesium-131 (Cs-131).³⁻⁵

Dose surrounding these brachytherapy seeds is commonly modeled using the revised AAPM protocol for brachytherapy dose calculations,⁶ which relies on a prior characterization of the seed model. As is traditionally the case, characterization of these sources was performed using a combination of Monte Carlo simulations and point dose measurements using Thermoluminescent Dosimeters (TLDs).

Alternatively, 3D dosimetry has the ability to provide 3D spatial dose information at high resolution, which could assist in the source characterization process of future seed models by verifying Monte Carlo calculations in instances of complex and high gradient dose distributions. A number of methods have been developed to measure 3D dosimetric distributions, including radiochromic, polymer, and Fricke gel dosimeters.^{7–10} In these systems, the dose distribution is recorded by a large volume dosimeter, after which it can be analyzed with one of a number of imaging systems.^{11–13} However, application of many 3D dosimeters to brachytherapy is complicated by sensitivity to oxygen and/or loss of spatial resolution due to signal diffusion. For polymer gel dosimeters, the mobility of radicals have been shown to shift polymerized regions considerably even after irradiation, which has been shown to be problematic for application to brachytherapy.¹⁴

PRESAGE[®] is a solid and transparent polyurethane based dosimeter that has the advantages of being machinable, moldable, insensitive to oxygen exposure, negligible spatial diffusion of the measured signal,⁸ and minimal dependency of dose rate.^{15,16} These attributes are especially important for brachytherapy measurements with high dose gradients and requiring source placement within the dosimeter via a channel.¹⁷ PRESAGE[®] has already been applied to measure the dose distribution surrounding cesium-137 and iridium-192 brachytherapy sources^{18–20} in conjunction with the Duke Large field of view optical CT-scanner (DLOS).^{17,21} Recent advances in 3D dosimetry technology have potential to make feasible dose measurements surrounding low energy sources as well. This includes development of a new optical CT system that is capable of reconstructing dose distributions with resolution on the order of 50 μm ,²² which has potential to enable reconstruction of dose distributions surrounding these sources with minimal partial volume averaging of the high dose gradients. Another recent development is high dynamic range scanning²³ for optical CT reconstruction; this method utilizes multiple projection images at the same incident angle but with varied incident light intensity and increases the dynamic range of the projection images beyond the current 12-bit range. While these advances hold much promise for application to low energy brachytherapy sources, the fundamental challenges and logistics for polyurethane dosimeters with optical CT based evaluation of low energy sources have yet to be addressed. This includes water equivalency and energy sensitivity at very low photon energies, as well as challenges associated with reconstructing dose in the presence of large variations in dose signal and with a channel present within the dosimeter for source placement. The purpose of the current study was to investigate the feasibility of and current challenges for PRESAGE[®]/optical CT to measure dose surrounding low energy brachytherapy sources.

2. MATERIALS AND METHODS

2.A. Overview

For all analyses we used the following sources: models 200 (Pd-103),³ CS-1 Rev2 (Cs-131),⁵ and 6711 (I-125).⁴ Our

evaluation consisted of using Monte Carlo simulations, simulations with ScanSim,²⁴ which is a software tool developed in house that can simulate optical CT with various scanner and dosimeter geometries, and experimental measurements with PRESAGE[®]/optical CT. The Monte Carlo was first benchmarked by comparing the TG43 (Ref. 6) based radial dose function in water with prior published studies,^{3–5} after which following were investigated:

1. the water equivalency of conventional and deformable PRESAGE[®] formulations,
2. the scatter conditions necessary to achieve accurate dosimetry for low energy photon seeds,
3. the change in photon energy spectrum within the dosimeter as a function of distance from the source in order to determine potential energy sensitivity effects,
4. the optimal delivered dose to balance optical transmission (per projection) with signal to noise ratio (SNR) in the reconstructed dose distribution, and
5. the magnitude and characteristics of artifacts due to the presence of the channel.

The first three items were investigated using Monte Carlo simulations, while the latter two items were investigated using the ScanSim software and experimental measurements. Following is a brief summary of the rationale for investigating each.

1. *Water equivalency of polyurethane dosimeters for low energy radionuclides:* Many formulations of PRESAGE[®] have been developed including conventional (rigid) and deformable variations, with the main variables in the formulation being the choice of lucodye and the shore hardness.²⁵ Water equivalence of some formulations has been evaluated for various photon and particle beam energies,^{18,26} but has yet to be quantified for low energy brachytherapy sources.
2. *Backscatter during irradiation:* A prior Monte Carlo simulation²⁷ demonstrated that nearly 5 cm of backscatter material is required to achieve accurate dosimetry for many brachytherapy sources; their results showed that nearly 5 cm backscatter material is required to achieve full scatter conditions at a radius (r) of 10 cm for I-125 and Pd-103. In our initial experiments, we found that extended submersion of some PRESAGE[®] formulations in water had a substantial effect on the optical properties of some dosimeters. Because of this we implemented an alternative to achieve full scatter conditions, which consisted of wrapping the dosimeter in a tissue equivalent material. In order to minimize the required amount of backscatter material, we performed Monte Carlo simulations to further quantify the relationship between the amount of backscatter and phantom scatter conditions for the three low energy sources.
3. *Energy spectrum vs distance from source:* PRESAGE[®] has been applied for a wide energy range of photon^{16–18,26,28} and particle beams.^{29–31} These studies have included Monte Carlo simulations of photons

down to 50 kVp (Ref. 26) and experimental validation down to 80 kVp,^{16,28} however the brachytherapy sources evaluated here are below this range with mean photon energies <30 kV, which raises the question of whether an energy dependence of the sensitivity of optical density to dose exists within this lower energy range. Such a characterization at low energies is difficult; in this study we quantified the change in energy spectrum as a function of distance from the source to provide insight into the magnitude of change in photon energy spectrum throughout a dosimeter. We also compared measured signal in conventional PRESAGE[®] to expected dose from Monte Carlo as a function of distance from an I-125 source. Monte Carlo simulations for energy spectrum were performed using F4 tally on the transverse axis at 0.25, 0.5, and 1–20 cm with 1 cm increments.

4. *Optimal delivered dose to balance optical transmission and SNR:* When using polyurethane dosimeters with optical CT to measure a dose distribution, there is a tradeoff between achieving adequate optical transmission through the dosimeter and signal to noise in the reconstructed dose distribution. This is especially true for brachytherapy sources where the dose gradient causes a large range of dose values; high dose near the source results in low optical transmission, while areas distal from the source have low signal to noise due to low delivered dose. For relatively uniform dose distributions, the maximum dose that can be reconstructed while still achieving adequate optical transmission can be estimated using the dose sensitivity of the dosimeter formulation and the linear distance through the irradiated region. However for brachytherapy sources, calculating the optimal dose to be delivered is not straightforward due to the complicated dose distribution around the source.

Furthermore for brachytherapy sources, large variations in optical transmission occur across a given projection image which can cause image saturation. The nature and magnitude of these effects will depend on the dose delivered, sensitivity of optical density to dose, dosimeter outer radius, radius of channel, choice of brachytherapy source, and scanner geometry. High dynamic range scanning²³ is a recently developed technique for optical CT in which the dose is reconstructed using multiple projection images acquired with varying incident light intensity levels, and has potential to overcome saturation in projection images. Using simulations, we investigated projection image saturation and the potential benefit of high dynamic range scanning for various scenarios for dose delivery and scanning.

5. *Artifacts in reconstruction due to channel:* Differences in refractive indices of dosimeter and surrounding fluid are known to cause optical scattering and reconstruction artifacts during optical CT.²⁴ However, when the refractive indices are relatively close these artifacts are of little concern (which is the case for most

applications) because they occur at the fluid/dosimeter interface rather than at the area of interest (dosimeter center). In the case of low energy brachytherapy sources, the source is placed within a channel located at the center of the dosimeter; hence the interface between fluid and dosimeter occurs at the area of interest. We evaluated the nature and magnitude of reconstruction artifacts at the fluid/dosimeter interface of the channel that are caused by subtle differences in refractive index (n).

2.B. Monte Carlo simulations

The Monte Carlo radiation transport code MCNP5 was used for all the simulations.³² The photoatomic cross sections were based on EPDL97 (Refs. 33 and 34), and the photon spectra were from the National Nuclear Data Center (NNDC).³⁵ Dose distributions from each source model were simulated in 5 cm radius and 10 cm height cylindrical dosimeter which was simulated as being at the center of a 15 cm radius and 30 cm height right cylinder of air, water, or dosimeter (as shown in Fig. 1). Dose distributions were simulated in both transverse (X-Y) and longitudinal (X-Z) planes (mesh = 300×300). The *F4 FMESH tally estimator was used with tally grid of $0.1 \times 0.1 \times 0.1$ cm³ everywhere.^{36,37} The cutoff photon energy was set to 0.005 MeV and electron scoring was not performed since the range of secondary charged particles was small for the photon energies studied. The 2×10^9 histories gave a typical statistical uncertainty ($k = 1$) of $\sim 3\%$ at 6 cm on the transverse plane.

Many formulations of PRESAGE[®] have been developed and applied, with the main variables in the formulation being the choice of leucodye and the shore hardness. We simulated both a rigid (conventional, 1.065 g/cm³) and deformable (1.02 g/cm³) formulation of PRESAGE[®]. These dosimeters

FIG. 1. Monte Carlo simulation geometry.

TABLE I. Elemental composition of conventional (rigid) and deformable formulations.

Element	Composition (%)	
	Conventional	Deformable
C	62.9	63.6
H	9.2	9.2
N	4.9	5.0
O	21.8	21.8
Br	0.4	0.4
S	0.8	0.0

represent formulations that utilize the “o-MeO-LMG-DEA” leucodye with shore hardness of 80D and roughly 30–50 A, respectively.²⁵ Their respective elemental composition is shown in Table I. Radial dose function $g(r)$ and anisotropy function $F(r, \theta)$ from the TG-43 formalism were calculated from MC results in water and compared with published results to verify the MC technique. The $g(r)$ and $F(r, \theta)$ were then calculated for Pd-103, I-125, and Cs-131 in both X-Y and X-Z planes, and compared between conventional and deformable formulations.

2.C. ScanSim software simulations

ScanSim is a MATLAB (Mathworks, Natick, MA) based software tool developed inhouse that can simulate image formation in a variety of optical CT scanning configurations; various optical CT and dosimeter geometries can be simulated including brachytherapy irradiation.²⁴ A channel at the center of the dosimeter can be simulated for brachytherapy source placement, along with various brachytherapy source dose distributions. With knowledge of the dose sensitivity and geometry of a dosimeter formulation, the expected range of optical transmissions through the dosimeter may be calculated for a given delivered dose. We used this capability to estimate the optimal dose to be delivered to achieve adequate light transmission in the projection images. In addition, ScanSim is able to model effects of differences in the refractive index of materials. We used this capability to model differences in the refractive index of the fluid around the dosimeter and within the source channel during optical CT.

2.D. Experimental measurements with PRESAGE®/optical CT

For experimental measurements, cylindrical polymer based PRESAGE® dosimeters^{7,8,38} (5–8 cm diameter, 5–6 cm height) were manufactured using the conventional formulation (see Table I). The cylindrical dosimeters had a density of 1.065 g/cm³, a refractive index of 1.495. The dosimeter was manufactured with a channel (0.0425 cm radius) in the center to nearly midway through the dosimeter. An optical CT preirradiation scan of the dosimeter was acquired using the Duke midsize optical CT scanner (DMOS) (Ref. 39) with the channel being filled with and the entire dosimeter being

immersed in a refractive fluid consisting of a combination of Octyl Salicylate and mineral oil for the scan (with refractive index matched to dosimeter). For irradiation, the fluid was then removed and the source was inserted into the dosimeter channel. During irradiation the dosimeter was surrounded by 2 cm of backscatter material (bolus). A postirradiation optical CT scan was immediately acquired with the same refractive fluid inserted in the channel again and the dosimeter immersed. The 3D dose distribution was reconstructed with (0.02 and 0.04 cm)³ isotropic voxel size with no median filtering. All optical CT scans were acquired with 360 projections at 1° increments, and were flood and dark field corrected with an acquisition time of ~15 min. For analysis, the location and orientation of the source was determined using an inhouse registration algorithm that minimized dose variations at equal radius from the source. Dosimeter imperfections, stray voxels, and areas outside of the dosimeter were removed using morphological operators.⁴⁰

3. RESULTS

3.A. Water equivalency of PRESAGE® for low energy isotopes

In order to first verify the Monte Carlo simulations and code, the TG-43 based radial dose functions were calculated and compared to published data. Radial dose function $g(r)$ based on TG43 calculated from MC simulation results in water agreed with published^{3–5} values for Pd-103, I-125, and Cs-131 as shown in Fig. 2. Differences between the calculated radial dose function(s) from published values are expected to be from small differences in the modeled geometry of the sources.

The Monte Carlo simulations show that the conventional formulation overattenuates all three low energy sources (Fig. 3). Relative to water, $g(r)$ in the conventional formulation decreased linearly ~2.5%/cm within a radius of 1–6 cm for Pd-103, and 1–10 cm for I-125 and Cs-131. In contrast, the deformable formulation underattenuated Pd-103 but was water equivalent for I-125 and Cs-131, with $g(r)$ within 1% of water within a radius of 0–10 cm. Differences from water for both conventional and deformable formulations in the TG-43

FIG. 2. Radial dose function, $g(r)$, simulated in water and compared to published data (Refs. 3–5).

FIG. 3. Ratio of radial dose function, $g(r)$, in dosimeter and water calculated by Monte Carlo for source models of Pd-103 (Ref. 3) (a), Cs-131 (Ref. 5) (b), and I-125 (Ref. 4) (c). The dashed lines are ± 1 standard deviation. Between 0 and 10 cm from the source, the ratio of dose in the deformable dosimeter to that in water was constant to within 1% for Cs-131 and 5% for I-125.

based anisotropy function, $\Phi(r)$, were $<1\%$ for Pd-103 and Cs-131, and $<5\%$ for I-125.

3.B. Backscatter during irradiation

The difference in radial dose function, $g(r)$, that would be measured in a 5 cm cylindrical diameter dosimeter surrounded by air vs water is shown in Fig. 4 for Pd-103, I-125, and Cs-131. In this situation, a lack of backscatter causes underdose $>10\%$ at the dosimeter surface; at 0.5 cm from the discrepancy is 2.5%, 6.6%, and 8.3% for Pd-103, I-125, and Cs-131, respectively. However, the discrepancy decreases to $\leq 2\%$ at 0.6, 1.4, and 1.7 cm from the surface, respectively. The discrepancy falls to $\leq 1\%$ at a distance from the surface of 1.0, 1.9, and 2.2 cm, respectively. Hence ~ 2 cm backscatter material should be sufficient for agreement within 2%.

3.C. Energy spectrum vs distance from source

Figure 5 shows the energy spectrum calculated using Monte Carlo of each source within the conventional formulation at a radial distance of 0.25, 5, and 10 cm. Figure 6 shows the mean energy of the spectrum (error bars = 1 standard deviation of energy spectrum) as a function of radial distance from the source. It is apparent from Figs. 5 and 6 that I-125 is the source resulting in the most stable energy spectrum over

FIG. 4. Percent difference in radial dose function, $g(r)$, that would be measured in a 5 cm cylindrical diameter dosimeter surrounded by air vs water, as calculated by Monte Carlo simulations.

the entire range of a dosimeter with a 10 cm radius; over that range the mean energy of I-125 changes by less than 1 keV or 3.4%. The mean energy of Cs-131 decreases 1.2 keV (4.1%) within the first 5 cm, with little change between 5 and 10 cm. For Pd-103, the mean energy is relatively stable within the

FIG. 5. Energy spectrum in conventional dosimeter formulation at a radial distance of 0.25, 5, and 10 cm.

FIG. 6. Mean energy in conventional dosimeter formulation as a function of radial distance from the source. Also shown for each curve is the mean $\pm$ one standard deviation of energy spectrum.

first 5 cm, but considerable hardening of the spectrum occurs thereafter; the mean energy stays within 0.5 keV (2.3%) for the first 5 cm but changes considerably between 5 and 10 cm from the source (4.2 keV or 19.4%).

Since changes in energy spectrum as a function of distance from the source are minor for I-125, we anticipate effects from energy sensitivity on the relative dose to be minor. However other factors such as radiochromic sensitivity to dose rate may also affect the relative dose. We compared the dose surrounding an I-125 source measured using a conventional formulation with the expected dose from Monte Carlo [Fig. 3(c)]. For the PRESAGE[®] irradiation 2.8 Gy was delivered at 1 cm in 92 h. Figure 7 shows the average dose distribution surrounding the source measured by the dosimeter (a) and calculated using Monte Carlo (b), the dose distribution after dividing by the TG43 geometry factor for the dosimeter (c) and Monte Carlo (d), and the standard deviation (e), and number of averaged voxels (F) for the measurement. Figure 8 shows the measured and calculated radial dose function in the dosimeter out to 3 cm. The measured radial dose function generally agreed with Monte Carlo to within 5% over a range of 0.3–2.5 cm. Points away from the central axis of

FIG. 7. Mean dose distribution surrounding I-125 measured in PRESAGE[®] (a) and calculated using Monte Carlo (b), dose after division by TG43 geometry factor normalized at 1 cm for PRESAGE[®] (c) and Monte Carlo (d), standard deviation of dose measured in PRESAGE[®] (e), and number of samples measured in PRESAGE[®] (f).

the source define the anisotropy function of the source; the median difference in anisotropy function per radius was below 5% for radii between 0.3 and 2.0 cm, and below 10% for radii between 0.3 and 2.5 cm.

FIG. 8. TG43 radial dose function measured using PRESAGE[®] and calculated using Monte Carlo simulations. Also shown is ± 1 standard deviation of measurement.

FIG. 9. Profile across measured and simulated optical projection through dosimeter with a radius of 2.5 cm and a channel radius of 0.05 cm.

3.D. Optimal delivered dose to balance optical transmission and SNR

An experimental validation of the ScanSim simulations is shown in Fig. 9; illustrated is a comparison between a profile through a projection image simulated using the ScanSim software and measured in an actual dosimeter. For this case, 1.9 Gy was delivered to a dosimeter with a radius of 2.5 cm, and a channel radius of 0.0425 cm. The Pearson correlation coefficient between these two curves was 0.996, and the root mean square error after normalization was 3.5% of the average signal.

Select results are given in Figs. 10 and 11 from the ScanSim simulations regarding the optimal delivered dose to balance optical transmission and SNR. We used the ScanSim software to simulate irradiation of a dosimeter with a 10 cm radius. The irradiation was simulated using I-125, Pd-103, and Cs-131 sources located in a central channel with a radius ranging from 0.1 to 0.5 cm. Using this simulation, we determined the maximum dose that could be delivered without saturating the optical CT projection images. A projection image was considered saturated if the ratio of the maximum and minimum intensity was ≥ 81.92 ; this ratio was based on prior

FIG. 10. Maximum dose calculated from simulations that does not saturate a 12-bit projection image from optical CT.

FIG. 11. Maximum dose delivered from I-125 calculated from simulations that does not saturate a 12-bit projection image from optical CT using high dynamic range scanning.

scanning experience with the DLOS (Ref. 17) and DMOS (Ref. 39) and assumes a 12-bit imager (maximum counts = 4096) with the minimum acceptable counts being ≥ 50 . The simulations were performed assuming both standard and high dynamic range scanning with varied incident light intensity levels. Finally, we assumed a sensitivity of the dosimeter's optical density to radiation of 0.047 1/cm/Gy; this is a typical dosimeter sensitivity and the results are linearly scalable to other sensitivities.

Figure 10 shows the maximum dose at a radius of 1 cm that can be delivered without saturating the projection image as a function of channel radius. Figure 11 shows the maximum deliverable dose at 1 cm as a function of number of incident light intensity levels (high dynamic range scanning) for I-125. Without utilizing high dynamic range scanning, the maximum deliverable dose at 1 cm without saturating the projection image is approximately 4 Gy for all low energy isotopes and a 0.1 cm channel. Adding a second light intensity level increases the maximum deliverable dose by over twofold. Each additional light intensity level also increases the maximum dose by approximately the same amount (~ 4 Gy for 1 incident light intensity level).

3.E. Artifacts in reconstruction due to channel

Some conceptual observations can be made from the ScanSim simulations regarding subtle differences in index of refraction between the channel and the dosimeter during optical CT. When the index within the channel is lower than the surrounding dosimeter the channel causes light transversing it to diverge. Conversely, when the index of refraction within the channel is higher the channel causes the light to converge. Refraction is greatest near the edge of the channel. When simulating volumetric image reconstruction, the refraction caused major artifacts within the channel but had little effect on the accuracy just outside the channel; this effect is illustrated in Fig. 12 which shows simulations of reconstructed dose profiles across a dosimeter with a 0.1 cm radius channel.

FIG. 12. Profile across reconstructed dose distribution from a 3D dosimeter with various simulated discrepancies in refractive index (n) between dosimeter and fluid in 0.1 cm radius channel during optical tomography.

4. DISCUSSION

One advantage of 3D dosimetry for brachytherapy sources is illustrated in Fig. 11(f): the increasing number of sample points can partially offset decreasing signal with radial distance from the source. For a given position along the source, the number of sample points increases linearly with distance away from the source; for a $(0.04 \text{ cm})^3$ voxel size the number of sample points at a distance away of 0.3 and 2.5 cm was ~ 50 and ~ 400 , respectively. This is due to the assumed symmetry for all points at a given distance along and away from the source. Because of this, the signal to noise ratio will be greatest at the central axis of the source (distance along the source = 0) and will decrease with increasing distance along the source. In terms of the polar coordinate system used in the TG-43 formalism, signal to noise ratio is highest at $\theta = 90^\circ$ and poorest at 0° and 180° . A related challenge to measuring the anisotropy factor is the presence of the channel. Because of the channel the anisotropy factor is not measured at one extreme (0° or 180° depending on source orientation). However, this may be compensated using irradiation of multiple dosimeters with flipped orientation.

We found the deformable formulation to have attenuation characteristics similar to water for Cs-131 and I-125, but not for Pd-103. However because the deformable formulation underattenuated and the conventional formulation overattenuated compared to water, there is potential to create a new formulation with water equivalent attenuation characteristics. In this study, all measurements were performed using I-125 and the conventional formulation of PRESAGE[®]. One known difference between the conventional and early versions of the deformable formulations is that the radiochromic signal faded fairly quickly relative to the conventional formulation;²⁵ this raises the question of whether such a formulation with water equivalent properties would be adequately stable for a prolonged irradiation. However, recently semideformable formulations have been developed with stable radiochromic properties.

We observed a gradual hardening of the Pd-103 spectrum within the dosimeter that is noticeable at a distance of approximately 5 cm from the source. This is similar to the observation by Rivard *et al.*, who found that within water the high

energy photon contribution caused a more gradual falloff in the radial dose function beyond 11 cm for Pd-103.⁴¹

Our simulations indicated that differences in refractive index during reconstruction within the channel should have minimal effect on the reconstructed signal at the detector surface. However, our measured signal was lower than expected at distances closer than 0.3 cm from the source (see Fig. 8). We suspect two causes: (1) radiochromic sensitivity has been observed to be unreliable near the detector edge, and (2) some partial volume averaging may occur at the channel edge leading to lower than expected signal within the voxel.

5. CONCLUSION

The current deformable formulation has nearly water equivalent attenuation properties for Cs-131 and I-125, and there is potential to create a formulation with attenuation properties similar to water for Pd-103. However, further investigation is needed on whether the radiochromic response from these formulations would be sufficiently stable for irradiation by a low energy source with low dose rate. The energy spectrum of each source is relatively stable within the first 5 cm from the source especially for I-125, indicating that any relative effects from energy sensitivity may be small. Agreement of the measured with calculated radial dose function for I-125 was within 5% between 3 and 2.5 cm from the source, and the median difference of measured from calculated anisotropy function was within 5% between 0.3 and 2.0 cm from the source. The inherent assumption of radial symmetry in the TG43 geometry leads to a linear increase in sample points with distance away from the source, which partially offsets the decreasing signal. Simulations of dose reconstruction using optical CT showed the feasibility of reconstructing dose out to a radius of 10 cm without saturating projection images using an optimal dose and high dynamic range scanning; the simulations also predicted that reconstruction artifacts at the channel surface due to a small discrepancy in refractive index should be negligible. In summary, 3D dosimetry using polyurethane based dosimeters with optical CT looks to be a promising application to verify dosimetric distributions surrounding low energy brachytherapy sources.

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