On the feasibility of optical-CT imaging in media of different refractive index

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Purpose: Achieving accurate optical-CT 3D dosimetry without the use of viscous refractive index (RI) matching fluids would greatly increase convenience.

Methods: Software has been developed to simulate optical-CT 3D dosimetry for a range of scanning configurations including parallel-beam, point, and converging light sources. For each configuration the efficacy of three refractive media was investigated: air, water, a fluid closely matched to PRESAGE[®], and perfect matching (RI = 1.00, 1.33, 1.49, and 1.501 respectively). Reconstructions were performed using both filtered backprojection (FBP) and algebraic reconstruction technique (ART). The efficacy of the three configurations and the two algorithms was evaluated by calculating the usable radius (i.e., the outermost radius where data were accurate to within 2%), and gamma (Γ) analysis. This definition recognizes that for optical-CT imaging, errors are greatest near the edge of the dosimeter, where refraction can be most pronounced. Simulations were performed on three types of dose distribution: uniform, volumetric modulated arc therapy (VMAT), and brachytherapy (Cs-137).

Results: For a uniformly irradiated dosimeter the usable radius achieved with filtered backprojection was 68% for water-matching and 31% for dry-scanning in air. Algebraic reconstruction gave usable radii of 99% for both water and air (dry-scanning), indicating greater recovery of useful data for the uniform distribution. FBP and ART performed equally well for a VMAT dose distribution where less dose is delivered near the edge of the dosimeter. In this case, the usable radius was 86% and 53% for scanning in water and air, respectively. For brachytherapy, the usable radius was 99% and 98% for scanning in water and air, respectively using FBP, and a major decrease was seen with ART. Point source geometry provided 1%–2% larger usable radii than parallel geometry. Converging geometry recovered less usable dosimetry data (up to 10% reduced usable radii) than point and parallel geometries. A further disadvantage of converging geometry was an increased requirement on detector size by up to 18°.

Conclusions: For applications where dose information is not required in the periphery of the dosimeter, some dry and low-viscous matching configurations may be feasible. For all three dose distributions (uniform, VMAT, brachytherapy) the point source geometry produced slightly more favorable results (an extra 1%–2% usable radii) than parallel and converging. When dosimetry is required on the periphery, best results were obtained using close refractive matching and ART. A concern for water or dry-scanning is the increase in required detector size, introducing potential cost penalties for manufacturing. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4798980>]

Key words: 3D dosimetry, optical-CT, PRESAGE[®], imaging, verification

I. INTRODUCTION

Immersion of dosimeters in a refractively matched fluid is a ubiquitous technique to minimize refraction artifacts in optical-CT 3D dosimetry.^{1–5} Radiochromic dosimeters such as PRESAGE[®] (Ref. 6) have high refractive index (RI) of ~1.50. Transparent solutions with this magnitude RI tend to be viscous, and require regular cleaning as they accumulate dust and particles from dosimeters. Obtaining an accurate match can be labor intensive, and this match may need to be changed depending on the RI consistency of different dosimeters. Although refractive matching with fluids is generally considered highly effective, it does introduce a substan-

tial inconvenience factor. Early work reported the feasibility of dry-scanning for small volumes with a laser microbeam CT (Refs. 5 and 7). Recently several works have begun to explore the feasibility of dry-scanning for larger volumes.^{8–10} The present work builds on these efforts by simulating optical-CT imaging of several relevant dose distribution models including uniform, volumetric modulated arc therapy (VMAT), and brachytherapy. These distributions span a range of clinical interest in 3D dosimetry as seen in recent publications.^{11,12} Simulations are also conducted for four different media and three different scanning configurations. The RIs of these media were chosen as: a PRESAGE[®]-matched fluid (1.501); a slightly mismatched fluid, e.g., mineral oil in DLOS

telecentric scanner¹³ (1.49); water (1.33); and air (1.00). These media were considered because they are either commonly encountered in a parallel-beam optical-CT lab or of great interest to dry-scanning research. Five main geometries of the incident light source were considered, which included point, parallel, and converging sources. The simulated geometries cover the range of available optical-CT scanning configurations in current use.^{2,13-15}

The overall goal is to investigate the achievable accuracy of optical-CT 3D dosimetry for the various combinations of refractive media, scanning configuration, and reconstruction method, when reconstructing the three dose-distribution scenarios. Simulations use a geometric ray-tracing model that obeys the laws of refraction (Snell's law) and transmission (Fresnel losses).

II. MATERIALS AND METHODS

II.A. Scanning configurations and modeled dose distributions

The controlled variables of the simulations were RIs of the media surrounding the PRESAGE[®] dosimeter and the incident ray geometry. A total of five ray geometries were investigated to simulate point, parallel, and converging ray optical-CT scanning configurations. Point and converging sources were each examined at focal lengths, f , of ± 200 and ± 500 mm, where focal length was defined as the distance between the point of ray convergence and the dosimeter center, as shown in Fig. 1. These focal lengths were chosen to sample the range of physical configurations of existing commercially available optical-CT scanners [cone-beam Vista[™] (Ref. 14) and parallel-beam He-Ne laser scanning OCTOPUS[™] (Refs. 15 and 16)] and the DLOS telecentric system,¹⁷ as well as explore potential geometric configurations for a novel water- or dry-scanner. For each geometrical configuration, simulations were run with four different RI values of the matching fluid surrounding the PRESAGE[®] dosimeter: 1.501, 1.49, 1.33, and 1.00. The purpose of these values was to represent ideal, minor-mismatch, water and air media, respectively. Our goal is to determine which geometrical configuration will best overcome the losses incurred by mismatched RI. Figure 1 contains examples of different incident ray geometries.

Three idealized dose distributions were simulated, (a) uniform 5 Gy to the entire dosimeter; (b) VMAT dual-arc treatment, with 10 Gy across PTV of radius 10 mm; and (c) Cs-137 brachytherapy, 8 Gy prescribed at $r = 10$ mm from source. The VMAT dose falloff outside PTV was modeled as $1/r$. For the brachytherapy case, dose falloff is described by geometry function $G_L(r, \theta)$ as given by Eq. (1):

$$G_L(r, \theta) = \frac{2 \arctan(L/2r)}{Lr} \quad \text{for } \theta = 90^\circ. \quad (1)$$

II.B. Dosimeter properties

The PRESAGE[®] formulation is a radiochromic dosimeter that, when irradiated, changes optical density (OD) causing more light attenuation rather than increasing light scatter.¹⁸ This dosimeter property greatly simplifies the simulation, especially in a telecentric system¹⁷ because scattered light can be ignored. To ensure relevant simulations, ScanSim is set up to simulate a PRESAGE[®] dosimeter typically used in clinical RPC head phantoms.¹⁹ These fixed dosimeter parameters were radius, $r_0 = 50$ mm, RI = 1.501, OD sensitivity = $0.023 \text{ cm}^{-1} \text{ Gy}^{-1}$, and unirradiated PRESAGE[®] background OD = 0.003 cm^{-1} . Realistic dosimeter modeling allows the software to be used to predict ideal dosimeter OD values prior to irradiation, assisting in current dosimetry experiments using the parallel beam RI matched laboratory optical-CT scanner.¹⁷

II.C. ScanSim

The ScanSim software⁹ was built in Matlab and is used to simulate optical-CT of cylindrical dosimeters with a range of useful customizations. The code essentially functions as an advanced ray-tracing tool, which records both (a) the path inside the dosimeter and in the surrounding medium, and (b) the relative intensity loss, due to transmission losses at boundaries and attenuation according to dosimeter's optical density (OD). The total number of rays traced through the dosimeter is controlled by the user input *ray spacing*.

Several fundamental physics principles are involved in this ray-tracing process. In accordance with Snell's law (2), a light ray of incident angle θ_i on a RI mismatched boundary ($n_1 \rightarrow n_2$) changes direction due to refraction, exiting with

FIG. 1. Three examples of scanning configuration showing incident ray geometry, dosimeter, and definition of focal length, f , and detector angle, $\Delta\theta$. (Left) Point source geometry, modeling current cone-beam Vista[™] optical-CT system. (Center) Parallel ray geometry, modeling telecentric DLOS system, and OCTOPUS[™] He-Ne laser scanner. (Right) Converging source geometry, exploring potentially new scanner geometries.

FIG. 2. Physics properties of our ray tracing model. A light ray incident on a cylindrical dosimeter immersed in a bath of RI mismatched fluid. The ray experiences *refraction*, the change in direction from incident angle θ_i to transmitted angle θ_t according to Eq. (2), and *intensity loss*, the drop in intensity from I_0 to $I = T \times I_0$ according to Eq. (5).

transmitted angle θ_t :

$$\theta_t = \arcsin\left(\frac{n_1 \sin \theta_i}{n_2}\right). \quad (2)$$

In addition, the ray will lose intensity due to partial transmission and reflection at the boundary, quantified by the Fresnel equations. The fraction of transmitted intensity T for unpolarized light is given by Eq. (5). A graphical representation of these variables is shown in Fig. 2:

$$R_s = \left| \frac{n_1 \cos \theta_i - n_2 \cos \theta_t}{n_1 \cos \theta_i + n_2 \cos \theta_t} \right|^2, \quad (3)$$

$$R_p = \left| \frac{n_1 \cos \theta_t - n_2 \cos \theta_i}{n_1 \cos \theta_t + n_2 \cos \theta_i} \right|^2, \quad (4)$$

$$T = 1 - \frac{R_s + R_p}{2}. \quad (5)$$

Finally, light is attenuated by the irradiated areas of the dosimeter, at which points the OD has increased due to the radiochromic response of PRESAGE[®]. The attenuation of each ray is determined by the line integral in Eq. (6), which is solved analytically where possible (uniform dose or VMAT), otherwise numerically using Simpson's Rule (Cs-137 brachytherapy).

$$\log_{10}\left(\frac{I}{I_0}\right) = - \int_{s_1}^{s_2} \text{OD}(r) \cdot ds. \quad (6)$$

ScanSim models rays according to a user specified ray spacing, geometry and focal length, dosimeter and fluid properties, and simulated dose distribution, $\text{OD}(r)$. The resulting intensity, angle, and distance from center are recorded in a sinogram for projections over 360° . For our simulations, all rays were declared detectable regardless of refracted angle or loss of intensity. Simulation's pre- and postscan intensity sinogram values are converted to OD, and image subtraction removes the effects of transmission losses. Once OD images are reconstructed they are converted to dose by normalizing to the true dose at a known point ($D_{\max}$ for all cases except brachytherapy, where $D_{1\text{ cm}}$ is used for normalization).

Optical density images are reconstructed using filtered backprojection (FBP) or algebraic reconstruction technique (ART). Filtered backprojection requires a complete, evenly spaced projection set (sinogram). Therefore, when the

dosimeter is not completely sampled at the edges due to refraction, interpolation must be used to provide estimates of these intensity values, and therefore quantitative accuracy of the FBP image is lost (i.e., the reconstruction must be normalized to a known value). In our simulations we use linear interpolation to estimate the unsampled sinogram and the rect $\times$ ramp filter for backprojection. The attractiveness of ART over FBP comes from the fact that it does not require a complete sinogram to reconstruct, so the interpolation is avoided. In addition, with ART quantitative accuracy is always conserved, even on an incomplete data set.

II.D. Algebraic reconstruction technique

Image reconstruction using ART, originally proposed in 1970 by Gordon *et al.*,²⁰ was revisited specifically for CT reconstruction by Andersen²¹ in 1989. This method is gaining popularity in x-ray CT reconstruction because it does not require a complete projection set and therefore may be used to produce reliable results outside the completely sampled field of view (FOV). Further, prior knowledge of the photon path (e.g., scatter, or detector point-spread-function) can be incorporated into a matrix [w_{ijmn} in Eq. (7)] that contains the photon path probability in units of length, and therefore accounted for during reconstruction. In optical-CT without the use of an RI matching fluid, refraction results in undersampling in the radial direction at the edges of the dosimeter. Therefore, ART was investigated as an alternative reconstruction method in our simulations.

Algebraic reconstruction works on the principle of iterative corrections to the reconstructed image, based on the differences between the measured sinogram and the sinogram calculated from the current reconstructed image. The expected sinogram value $\log_{10}(I/I_0)_{mn}$ of ray n and projection m is given by

$$p_{mn} = \log_{10}\left(\frac{I}{I_0}\right)_{mn} = \sum_i \sum_j w_{ijmn} \cdot \text{OD}_{ij}, \quad (7)$$

where w_{ijmn} is a 4D matrix with $n \times m$ slices of image size $i \times j$ that correspond to the path of the n th ray in the m th projection through the dosimeter. Each pixel ij of w_{ijmn} contains the expected path length (in cm) of ray n in projection m . At each iteration, the difference between p_{mn} and the measured sinogram P_{mn} provides an additive correction to the reconstructed image OD_{ij} which is multiplied by a relaxation factor λ for slower, more accurate convergence.

Calculating the w_{ijmn} matrix involves finding the exact path length through each pixel ij . Siddon²² proposed a fast algorithm for calculating the exact radiological path, which was surpassed in speed by Jacobs *et al.*'s algorithm,²³ which was used in our simulations. Table I provides some understanding of the required computational time for calculating w_{ijmn} and performing a single ART iteration on a standard lab desktop computer (Intel[®] Core[™] i7 975 3.33 GHz, 12 GB RAM). Recent optical-CT dosimetry simulations using ART were limited by high demand for computational resources.¹⁰ However, in the present work, significant time and memory storage was saved by making use of sparse matrix coding techniques.

TABLE I. Dimensions of ART image reconstructions, numbers of rays and projections used for ray tracing, and computational time requirement on lab desktop Intel® Core™ i7 975 3.33 GHz, 12 GB RAM.

Image size (pixels)	128 × 128	256 × 256
Field of view (mm)	110	110
Ray spacing (mm)	0.5	0.4
Number of rays, n	308	386
Number of projections, m	622	779
Time to compute w_{ijnm} (s)	170	540
Time to compute 1 iteration (s)	25	110
Memory required (GB)	1.2	2.5

Future implementation of ART could benefit from a significant speedup by creating and optimizing a dedicated reconstruction algorithm code, parallelizing the reconstruction process to make use of multicore CPUs, and making use of subset techniques such as OS-SART.²⁴

II.E. Metrics for evaluation

To evaluate the effectiveness of each scanning and RI configuration, we defined the term *usable radius*, r_u , as the distance (mm) from the center of the dosimeter at which measured dose and true dose differ by less than 2% (or 3% for Cs-137). Reference dose for percentage difference is $D_{\max}$ for uniform and VMAT distributions and $D_{1\text{ cm}}$ for brachytherapy simulations, the latter criteria reflecting the precipitous falloff from brachytherapy sources. In addition, for the brachytherapy evaluation, usable radius was measured outwards from the $r = 0.5$ cm point, due to the noise associated with high-frequency portions of the image near the Cs-137 seed.¹¹ An example measurement of usable radius r_u is shown in Fig. 3 (bottom row).

For dimensionless comparison, the *usable radius* can be expressed as the ratio of usable radius to dosimeter radius, r_u/r_0 . To quantify the required detector size, the exit angle, $\Delta\theta$, was quantified as the angular variation of all rays leaving the dosimeter. Finally, to measure the overall dose agreement, a 2%/1 mm Γ -index was applied to quantify *local* variation at each point in the measured dose distribution. The 1 mm margin is to account for the partial volume effect present in any discretized simulations. For Cs-137 brachytherapy, a more re-

laxed passing criteria of 3%/1 mm was applied due to local dose being very small for large r .

III. RESULTS

Initially, the irradiated dosimeter was modeled as a 5 Gy uniform dose distribution, and a total of 20 combinations of geometric configuration and surrounding RI were simulated at high resolution (ray spacing of 0.2 mm). For speed, FBP with linear interpolation and standard ramp + rect filter was used for image reconstruction. The usable fraction r_u/r_0 and exit angle $\Delta\theta$ of these simulations are given in Table II.

These initial results showed that converging geometry usable fractions are on average 3% lower than point source configurations of the same focal length. For point source geometry, $f = 200$ mm is an average of 1% higher than $f = 500$ mm. Our second set of simulations included the ART reconstruction technique, which requires large computational time and resources. For this reason, these simulations were performed on a limited geometry set. Chosen geometries included the highest performing point source ($f = 200$ mm), which models cone-beam systems, and the parallel ray setup, which models the telecentric system and parallel laser scanning systems. Point source geometries with focal lengths shorter than 200 mm were not considered for practical reasons, such as allowing room for lenses and casing, and retaining the ability to image larger dosimeters. Additionally, we used these configurations to explore the clinically applicable dose distributions VMAT and Cs-137. These two promising geometries are explored for uniform, VMAT, and brachytherapy dose models, with results shown in Tables III–V, respectively. Results are included for both FBP and ART reconstructions. For these simulations, ray tracing and intensity sinograms were calculated with 0.5 mm ray spacing (i.e., using fewer rays) to decrease computational time. Reconstructed ART images were 128 × 128 pixels, with the following parameters: number of rays, $n = 308$ (number of rays within dosimeter, $n' = 198$); number of projections, $m = \pi n' = 622$; number of iterations = 400, smoothed with a 1 pixel full width at half maximum (FWHM) Gaussian filter. Of these numerous simulations, Fig. 3 depicts three examples that illustrate the incident ray geometry, dose distribution, RI of surrounding media, FBP

TABLE II. Data from simulations of a 5 Gy uniform dose on an $r_0 = 50$ mm PRESAGE® dosimeter. For each geometrical setup, incident ray geometry is described by focal length, f . Line profile across the reconstructed measured dose distribution gave the usable fraction r_u/r_0 , and the exit angle $\Delta\theta$ was recorded to estimate the required detector size. The RI column represents the refractive index of the surrounding media during optical-CT scanning. Line integrals were calculated via ray tracing with 0.2 mm ray spacing, and reconstructions performed using FBP with linear interpolation and standard ramp + rect filter.

5 Gy uniform (FBP)										
RI, n	Point source				Parallel source		Converging source			
	$f = 200$ mm		$f = 500$ mm		$f = \pm\infty$		$f = -500$ mm		$f = -200$ mm	
	r_u/r_0 (%)	$\Delta\theta$ (deg)	r_u/r_0 (%)	$\Delta\theta$ (deg)	r_u/r_0 (%)	$\Delta\theta$ (deg)	r_u/r_0 (%)	$\Delta\theta$ (deg)	r_u/r_0 (%)	$\Delta\theta$ (deg)
1.501	99	28	99	12	99	0	99	12	93	28
1.49	99	22	98	10	97	14	97	24	91	36
1.33	68	78	68	92	66	90	66	102	62	98
1.00	31	160	31	174	31	174	31	178	29	176

FIG. 3. Three example simulations of optical-CT dosimetry with different dose distributions, scanning configurations, and RI matching. Reconstructions using both FBP and ART are shown. Each column contains (from top to bottom): reconstruction of simulated dose distribution using FBP; reconstruction of simulated dose distribution using ART; radial line profile through each reconstruction showing measured dose vs true dose and percent difference $|\Delta D/D|_{(\max \text{ or } 1 \text{ cm})}$ to visualize reconstruction artifacts and quantify the usable fraction, r_u ; ray diagram showing incident light geometry and refracted path through PRESAGE[®] ($n = 1.501$) dosimeter. (Left column a) VMAT irradiated dosimeter, parallel telecentric geometry, RI-mismatched fluid ($n = 1.49$) surrounding the dosimeter. (Center column b) Uniformly irradiated dosimeter, converging geometry ($f = -200 \text{ mm}$), water ($n = 1.33$) surrounding the dosimeter. (Right column c) Cs-137 brachytherapy, point source geometry ($f = 200 \text{ mm}$), air ($n = 1.00$) surrounding the dosimeter (dry-scanning).

TABLE III. Analysis of simulations of a 5 Gy uniform dose delivered across an $r_0 = 50$ mm PRESAGE[®] dosimeter. Two promising geometrical setups were considered: point ($f = 200$ mm) and parallel incident ray geometry. Usable fraction r_u/r_0 and Γ (2%/1 mm) index are shown for each combination of RI, geometry, and reconstruction method. Reconstructions were performed with FBP (odd rows, italic) and ART (even rows, bold).

5 Gy uniform (FBP and ART)				
RI, n	Point source ($f = 200$ mm)		Parallel source ($f = \pm\infty$)	
	r_u/r_0 (%)	Γ (2%/1 mm) (%)	r_u/r_0 (%)	Γ (2%/1 mm) (%)
1.501	98	97	97	95
	98	96	97	94
1.49	97	93	95	94
	97	96	96	94
1.33	68	51	64	46
	99	91	98	93
1.00	31	13	31	10
	98	98	99	98

and ART reconstructed dose distributions, and the resulting line profile plot of measured vs true dose and measurement of r_u .

IV. DISCUSSION

The simulation results for the uniformly irradiated dosimeter in Table II demonstrate that a fraction of the dosimeter remains usable for all scanning configurations. Optical-CT scanning in a water medium (RI = 1.33) may recover a usable fraction of up to 68% with point source geometry, or 66% with parallel ray geometry. Optical-CT of a dosimeter in a dry medium (air, RI = 1.00) was shown to recover a usable fraction of up to 31%. Edge effects currently seen in RI-matched optical-CT scanners limit the usable radius to the order of ~ 46 mm (or 92%) (Ref. 25) and, as expected, a greater RI mismatch eliminates more of this usable radius. The detector size $\Delta\theta$ is determined by the combination of RI match

TABLE IV. Analysis of simulations of a 10 Gy VMAT delivery, PTV radius = 10 mm, assuming $\sim 1/r$ falloff, on a $r_0 = 50$ mm PRESAGE[®] dosimeter. Two promising geometrical setups were considered: point ($f = 200$ mm) and parallel incident ray geometry. Usable fraction r_u/r_0 and Γ (2%/1 mm) index are shown for each combination of RI, geometry, and reconstruction method. Reconstructions were performed with FBP (odd rows, italic) and ART (even rows, bold).

10 Gy dual-arc VMAT (FBP and ART)				
RI, n	Point source ($f = 200$ mm)		Parallel source ($f = \pm\infty$)	
	r_u/r_0 (%)	Γ (2%/1 mm) (%)	r_u/r_0 (%)	Γ (2%/1 mm) (%)
1.501	98	97	99	95
	100	97	99	96
1.49	98	94	98	97
	98	97	99	94
1.33	86	72	84	70
	87	75	85	70
1.00	53	32	53	33
	52	30	47	27

TABLE V. Analysis of simulations of a Cs-137 brachytherapy source of prescription 8 Gy at $r = 1$ cm, on an $r_0 = 50$ mm PRESAGE[®] dosimeter. The two most promising geometrical setups were considered: point ($f = 200$ mm) and parallel incident ray geometry. Percentage constraints were loosened because the dose distal from the source was relatively small; usable radius r_u was calculated as the distance at which $\Delta D/D_{1\text{ cm}} > 3\%$ (considering only points for $r > 1$ cm) and Γ -index passing conditions were relaxed to 3%/1 mm. Usable fraction and Γ -index are shown for each combination of RI, geometry, and reconstruction method. Reconstructions were performed with FBP (odd rows, italic) and ART (even rows, bold).

8 Gy at 1 cm Cs-137 brachytherapy (FBP and ART)				
RI, n	Point source ($f = 200$ mm)		Parallel source ($f = \pm\infty$)	
	r_u/r_0 (%)	Γ (3%/1 mm) (%)	r_u/r_0 (%)	Γ (3%/1 mm) (%)
1.501	100	79	100	75
	100	84	100	80
1.49	100	76	100	77
	100	84	100	80
1.33	99	80	98	80
	95	65	86	57
1.00	98	68	97	66
	38	26	45	17

and incident ray geometry. In some cases, generally when RI of surrounding media is much less than RI of the dosimeter, point-source geometry is advantageous by requiring a smaller detector (e.g., for $n = 1.00$, $\Delta\theta = 178^\circ$ for converging and 160° for point). For these reasons, parallel and $f = 200$ mm point source geometries were established as the most favorable configurations.

These two favorable geometries were used in simulations of uniform, VMAT, and Cs-137 dose distributions in Tables III–V, respectively, reconstructed with both FBP and ART. For the uniformly irradiated dosimeter, ART recovered 99% of usable fraction for both water- and dry-scanning optical-CT. This is largely due to the nature of the dose-distribution, for which OD is constant and does not vary with r .

The VMAT simulation results (Table IV) recovered a usable fraction of 87% (ART) and 86% (FBP) for scanning in a water medium, and 52% (ART) and 53% (FBP) for dry-scanning. This dose distribution gives a meaningful comparison between ART and FBP reconstruction methods, with both methods producing comparable usable fractions and Γ -index passing rates. The clear advantage of FBP here is reconstruction time (compared to ART in Table I), whereas ART holds the advantage of standalone quantitative accuracy, i.e., no need to normalize to a measured dose if the dosimeter sensitivity is accurately known. Since the iterative process of ART relies on an initial estimate of the dose distribution, advanced reconstructions using a combination of FBP and ART should be explored.

Brachytherapy distributions differ from external beam therapy due to the geometrical falloff of fluence (approximately $1/r^2$ vs $1/r$, respectively) within tissue. For this reason, the analysis of the Cs-137 brachytherapy dose distribution involved loosening the usable radius and Γ -index evaluation metrics to 3% and 3%/1 mm, respectively. Overall, FBP

showed positive results, with usable radius of 99% for water- and 98% dry-scanning. The high usable radius values are due to the fact that brachytherapy dose distributions deposit a wide range of dose within the dosimeter, yet our metric was calculated by normalizing to dose at $r = 1$ cm, $D_{1\text{ cm}}$. The Γ -index, which was calculated according to *local* dose, gives a better indication of the errors induced by the refractive mismatch between dosimeter and surrounding media (Table V). The poor usable radii recovered by ART are due to the fact that iterative reconstruction algorithms are inherently noisy and at low OD this noise is comparable in magnitude to the signal.

Pronounced variation between measured and true dose distributions is seen near the dosimeter edges when there is a RI mismatch between the dosimeter and surrounding fluid as shown in Fig. 3 (center). The artifacts increase in radial width and magnitude for $n = 1.33$ and again for $n = 1.00$ surrounding media. The cause is optical undersampling due to refraction; incident rays are diverted toward the dosimeter center, leaving the simulation with no line integral attenuation data near the dosimeter edges. Larger RI mismatch results in a larger section of the dosimeter being incompletely sampled. In addition to edge effects, this also makes absolute dose measurements unobtainable when reconstructing with FBP. To overcome this, our reconstructed dose distributions were normalized to the true dose D_{max} . In practice, a point of known dose would be needed in the dosimeter if reconstructing with FBP.

It is apparent that FBP performs better when dose is low toward the edge of dosimeter (e.g., Cs-137), due the nature of backprojection. Each point along the backprojected line essentially depends on every OD value on that entire line, including those that were not sampled. In contrast, ART reconstruction performance does not require a complete dataset to maintain quantitative accuracy. For this reason, ART produced the most desirable results for dose distributions with little variation in magnitude (e.g., uniform). Finally, the success of ART reconstruction depends heavily on a starting estimate of the image. Performance of ART in the VMAT and Cs-137 cases may be improved in future by establishing good starting-point estimates of the dose distribution, perhaps using the FBP reconstruction itself as an initial estimate.

Noise was not modeled in these simulations; severely refracted rays at the dosimeter edges suffer large intensity losses due to reflection at the RI boundary, therefore a reduction in SNR is expected to add to the degrading effects of RI mismatching. Further simulations are may be used to quantify this effect. However, longer scan times, more powerful light sources, or more efficient detection may be used to limit these.

V. CONCLUSIONS

We have simulated optical-CT of three scanning geometry configurations: parallel, point, and converging. For each geometry, we reconstructed the measured dose using four different media surrounding the dosimeter with refractive indices of 1.501, 1.49, 1.33, and 1.00. Our results have shown that optical-CT using a water medium ($\text{RI} = 1.33$) may re-

cover a usable fraction r_u/r_0 of up to 99% (49.5 mm) and 87% (43.5 mm) for uniform and VMAT dose distributions respectively. Scanning the dosimeter in a dry medium (air, $\text{RI} = 1.00$) was shown to recover a usable fraction of up to 99% (49.5 mm) and 53% (26.5 mm), respectively. Accordingly, optical-CT in a water medium may be suitable for some clinical applications of external beam 3D dosimetry where accurate dosimetry is not required near the periphery. One possibility is using larger dosimeter and sacrificing information at the dosimeter edges.

For brachytherapy, our Cs-137 simulations showed that 99% (49.5 mm) and 98% (49 mm) usable radius is recovered in water and air using FBP, with Γ -indices of 80% and 68%, respectively. These relatively high usable radii are accredited to the normalization of the metric to $D_{1\text{ cm}}$ and the steep dose falloff of brachytherapy sources ($\sim 1/r^2$). Dry or water scanning shows potential for brachytherapy distributions due to this dose distribution shape.

A significant concern for dry- and low-viscous matching fluid configurations is the required detector size $\Delta\theta$ increases substantially (more than a factor 6 in some cases, see Table II). This would likely add a cost penalty.

Using an appropriate reconstruction method, optical-CT in water or air surrounding medium may be used to measure dosimeter OD distributions for dosimetry where the information at the edges of the dosimeter can be sacrificed. Other artifacts not simulated here, such as noise (transmission losses) or imperfect cylindrical symmetry (knowledge of refracted angle) could pose potential challenges to the implementation of a real RI-mismatched optical-CT scanner.

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