How effective can optical-CT 3D dosimetry be without refractive fluid matching?

L Rankine and Oldham
Department of Radiation Oncology,
Duke University Medical Center, Durham, USA

E-mail: leith.rankine@duke.edu

Abstract. Achieving accurate optical CT 3D dosimetry without the use of viscous refractive index (RI) matching fluids would greatly increase convenience. 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 3 refractive media were investigated: air, water, and a fluid closely matched to Presage (RI = 1.00, 1.33 and 1.49 respectively). The results revealed that the useable radius of the dosimeter (i.e. where data was within 2% of truth) reduced to 68% for water-matching, and 31% for dry-scanning in air. Point source incident ray geometry produced slightly more favourable results, although variation between the three geometries was relatively small. The required detector size however, increased by a factor six for dry-scanning, introducing cost penalties. For applications where dose information is not required in the periphery, some dry and low-viscous matching configurations may be feasible.

1. 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 [6] like PRESAGE® [7] have high refractive index (RI) of ~1.51. 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 substantial inconvenience factor [8]. Early work reported the feasibility of dry-scanning for small volumes with a laser microbeam CT [5, 9]. Recently several works have begun to explore the feasibility of dry-scanning for larger volumes [10-12]. The present work builds on these efforts, and presents simulations of various optical-CT configurations and refractive matching scenarios.

2. Method

The ScanSim software [11] was used to simulate optical CT imaging of a PRESAGE® dosimeter typically used in clinical RPC head phantoms [13]. Fixed dosimeter parameters were radius, r_0 = 50 mm, RI = 1.501, sensitivity = 0.023 OD/cm/Gy, and background un-irradiated Presage optical density (OD) = 0.003 /cm. The simulated *true* dose distribution was a uniform 5 Gy to the entire dosimeter.

Attenuation was calculated using integrals on rays that were spaced 0.2 mm apart on incidence, and all rays were assumed detectable, regardless of refracted angle. Filtered back-projection (FBP) was used to reconstruct the *measured* dose distribution from the simulation's pre- and post-scan intensity sinograms. The reconstructed image gives values in optical density (OD), which are then converted back to dose by normalizing to the true dose at a known point, in this case D_{max} at the center of the dosimeter.

The controlled variables of the simulations were RI 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; both point and converging sources were examined at focal lengths, f, of ±200 mm and ±500 mm. 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 1.0. These values represented ideal, minor-mismatch, water and air media respectively.

To evaluate the effectiveness of each scanning and RI configuration, we defined the term useable radius of the PRESAGE® dosimeter, r_u, for which the optical-CT measured dose is within 2% of the true dose. The usable fraction of the radius of the dosimeter is thus r_u/r_0.

3. Results

A total of 20 combinations of geometry and RI were simulated and the usable fraction r_u/r_0 and required detector size $\Delta\theta$ are given in table 1. Figure 1 shows illustrative simulations of parallel, converging, and point incident ray geometries. Reconstructed images corresponding to each scanning configuration are shown, together with line profiles comparing measured (simulated) and true dose.

Table 1: Data from simulations of a 5 Gy uniform dose on r_0 = 50 mm PRESAGE® dosimeter. For each setup, incident ray geometry is described by focal length, f, and the useable fraction r_u/r_0 and required detector size $\Delta\theta$ are recorded. The RI column represents the refractive index of the media surrounding the dosimeter during optical-CT scanning.

	Point Source				**Parallel Source**		**Converging Source**			
RI	f = 200 mm		f = 500 mm		f = ±∞		f = -500 mm		f = -200 mm	
	r_u/r_0	$\Delta\theta$	r_u/r_0	$\Delta\theta$	r_u/r_0	$\Delta\theta$	r_u/r_0	$\Delta\theta$	r_u/r_0	$\Delta\theta$
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°

4. Discussion

As shown in figures 1(a) and 1(b), pronounced variation between measured and true dose distributions is seen near the dosimeter edges when there is an RI mismatch between the dosimeter and surrounding fluid. These artifacts are caused by undersampling due to refraction; rays are diverted towards the dosimeter center, leaving the simulation with no attenuation line integral 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 affects any attempt to make an absolute dose measurement, due to the nature of 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).

The simulation results in table 1 demonstrate that an inner fraction of the dosimeter remains useable 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 useable fraction of

up to 31%. As expected, the detector size $\Delta\theta$ is dependent on 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.

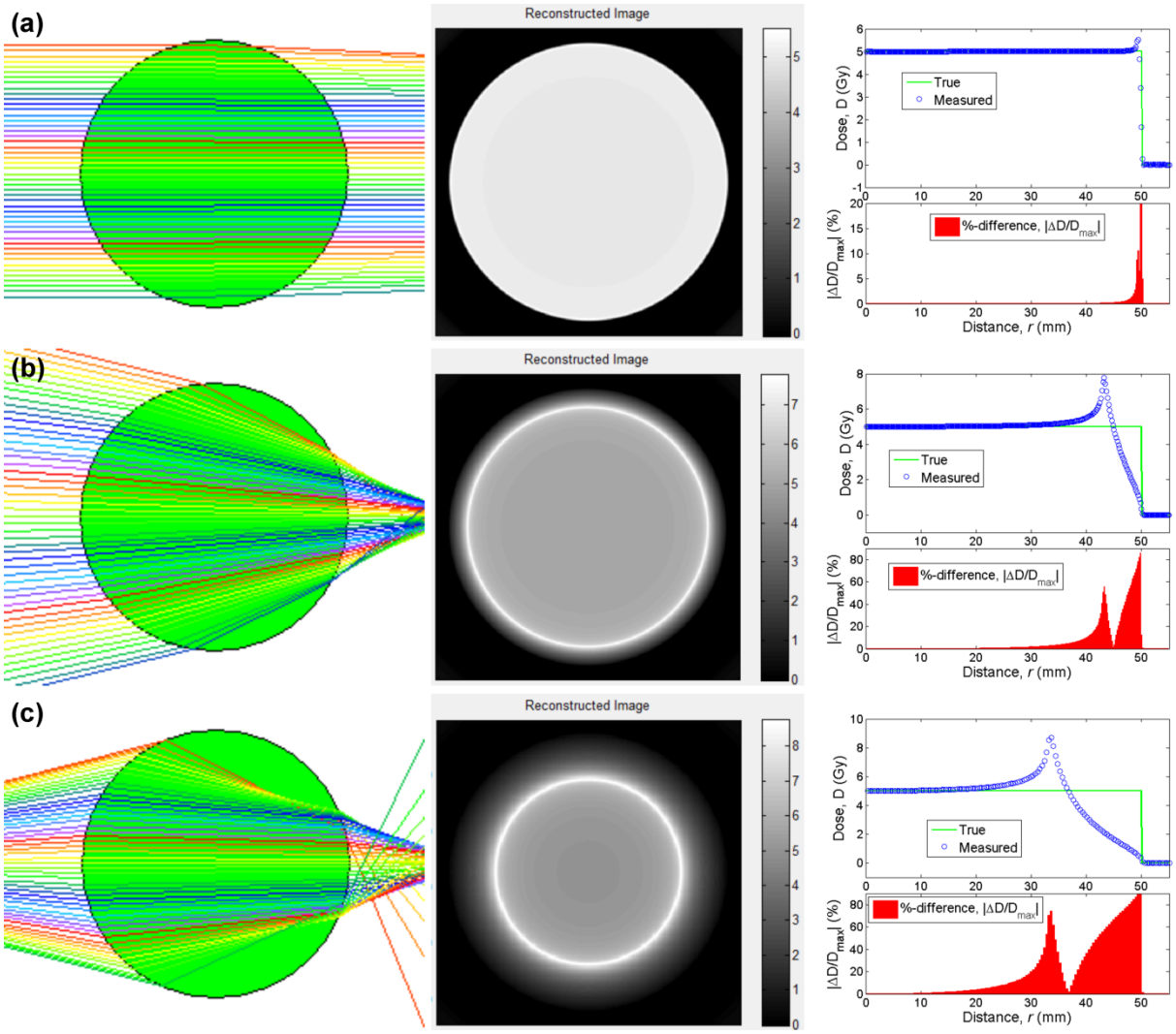

Figure 1: (a) Parallel telecentric geometry with surrounding media of RI = 1.49; **(b)** Converging light ray geometry (f = -200 mm), surrounding RI = 1.33; and **(c)** Point source geometry (f = 200 mm), surrounding RI = 1.00. [Left column] Ray diagram showing incident light geometry and refracted path through PRESAGE® (RI = 1.501) dosimeter. [Center column] Measured (simulated) dose distribution (Gy). [Right column] Radial line profile through the reconstruction. Plot includes measured dose (blue) vs. true dose (green), with difference $|\Delta D/D_{max}|$ shown beneath to visualize reconstruction artifacts and quantify useable radius fraction, r_u/r_0.

5. Conclusion

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 of a uniformly irradiated dosimeter, using a water medium (RI = 1.33), may recover a usable fraction r_u/r_0 of up to 68%. This may be suitable for some clinical applications of 3D dosimetry where accurate dosimetry is not required near the periphery. Scanning the dosimeter in a

dry medium (air, RI = 1.00) was shown to recover a useable fraction of only up to 31%. 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 – table 1). This would likely add a cost penalty.

6. References

[1] Gore J C *et al* 1996 *Phys. Med. Biol.* **41** 2695-2704
[2] Doran S J *et al* 2001 *Phys. Med. Biol.* **46** 3191-213
[3] Conklin J *et al* 2006 *J. Phys.: Conf. Ser.* **56** 211
[4] Wuu C *et al* 2004 *J. Phys.: Conf. Ser.* **3** 297
[5] Maryanski M J and Ranade M K 2001 *Proc. SPIE* **4320** 764
[6] Baldock C *et al* 2010 *Phys. Med. Biol.* **55** R1-63
[7] Adamovics J and Maryanski M J 2004 *J. Phys.: Conf. Ser.* **3** 172
[8] Bosi S *et al* 2007 *Phys. Med. Biol.* **52** 2893-2903
[9] Maryanski M J, *et al* 1998 *Med. Phys.* **25** A107
[10] Papadakis A E *et al* 2010 *IEEE Trans. Med. Imaging* **29** 1204-12
[11] Oldham M 2010 *J. Phys.: Conf. Ser.* **250** 012064
[12] Doran S J and Yatigammana D N B 2012 *Phys. Med. Biol.* **57** 665
[13] Oldham M *et al* 2012 *Int. J. Radiat. Oncol. Biol. Phys.* (in press)