Achieving accurate radiochromic optical-CT imaging when using a polychromatic light source

A Thomas*, M Pierquet, M Oldham
Dept of Radiation Oncology, Duke University Medical Center, Durham, NC, USA

*Corresponding author: ast5@duke.edu

Abstract. Optical-CT performed with a broad spectrum light source can lead to inaccurate reconstructed attenuation coefficients (and hence dose) due to ‘spectral warping’ as the beam passes through the dosimeter. Some wavelengths will be attenuated more strongly than others depending on the absorption spectrum of the radiochromic dosimeter. A simulation was run to characterize the error introduced by the spectrum warping phenomena. Simulations of a typical dosimeter and delivered dose (6cm diameter, 2 Gy irradiation) showed reconstructed attenuation coefficients can be in error by >12% when compared to those obtained from a monochromatic scan. A method to correct for these errors is presented and preliminary data suggests that with the correction, polychromatic imaging can yield imaging results equal in accuracy to those of monochromatic imaging. The advantage is that polychromatic imaging may be less sensitive to prominent scleraring artefacts that are often observed in telecentric optical-CT scanning systems with tight bandwidth filters applied.

1. Introduction
An accompanying abstract introduces the DLOS scanner, a novel large field-of-view telecentric optical-CT scanner for fast 3D radiochromic dosimetry. The light source employed in this system is a red LED, and the spectrum, although narrow (~ 25nm FWHM), can introduce errors into the measured dose through the phenomena of ‘spectral warping’. 3D radiochromic dosimetric materials have been designed for use with extremely narrow bandwidth light sources typical to that of a laser (< .1nm) corresponding to the materials peak absorption wavelength upon irradiation. The absorption spectrum [1, 2] of the dosimetric materials used in 3D radiochromic dosimetry can vary drastically over the spectrum of an LED light source, thus introducing differences in readouts of a laser system to that of an LED system. Spectral warping arises due to the preferential absorption of photons at the dosimeters peak absorption wavelength leaving a higher ratio of photons at both higher and lower wavelengths as light is transmitted through the dosimeter. This process leaves an artificially high number of photons to be detected for each projection leading to errors in the sinogram and reconstruction; a process similar to spectral hardening or beam hardening in X-ray imaging, although here the spectrum is not getting harder. This area has been addressed extensively in X-Ray imaging [3-9] although does not translate directly to dosimetry. One other common difference between area and laser scanning imaging can be light scatter [10]; the aim of this work, however, was to characterize the magnitude of the spectral warping effect and develop a method to overcome it.

2. Methods
All the measurements and corrections made within are specific to a PRESAGE Radiochromic dosimeter and an in-house area based CCD telecentric lens optical-CT scanning system with a 3W Red LED light source. Section 2.1 describes the process of collecting spectral information used in determining the magnitude of the
spectral warping. In order to better understand the spectral warping, section 2.2 details the signal received at the detector for a traditional laser source and the LED source used in the DLOS system. Also included in this section is a method for making corrections to LED projection data to mimic what the laser projection would return. These sections are followed by a simulation of a uniformly irradiated dosimeter measured by a laser source and an LED source to show the errors introduced, and the implementation of the correction on a two field plan delivered to a 16 cm dosimeter.

2.1. Spectral Information

The 1st step in obtaining a correction for an LED light source system is to understand the spectrum. Spectral information was required for the unattenuated LED, the dosimeters absorption spectrum and the detectors spectral response. The LED spectrum was measured with an Ocean Optics HR 2000 spectrometer and can be seen in figure 1. This information determined the range of wavelengths to consider when looking for the absorption response characteristics of the radiochromic dosimeter. In order to determine the spectral response of the dosimeter, three 1 cm thick cuvettes were irradiated to .5, 2 and 4 Gy and the change in optical density ($\Delta OD$) over the LED spectrum (580 - 690 nm) was measured with a Genesys 20 spectrophotometer from Thermo Scientific. The spectral response of the camera was obtained by using the manufacturer’s documentation on the chips quantum efficiency.

![Figure 1 – LED, quantum efficiency and dosimeter absorption spectrum. It can be seen from analysis of the plot that the LED spectrum will be preferentially distinguished at 633 nm and less so at higher and lower energies causing a spectrum deformation giving an artificially high signal collected as the photons at higher and lower wavelengths are not being attenuated as strongly.](image)

2.2. A Correction: Laser v LED projections

Once the spectral information has been gathered one can begin to develop a correction for the LED light source. Next, an understanding of the signal collected for the laser v the LED is needed. With the dosimeters absorption spectrum, the LED’s power spectrum and the cameras quantum efficiency spectrum from the manufacturer, calculations can be made to determine the signal collected from the different light sources by the following equation:

$$S_{LED} = \sum_{\lambda=590nm}^{\lambda=680nm} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{\lambda} \cdot D \cdot d}$$

(eqn 1)

Where $S_{LED}$ is the signal at the detector for an image projection through the setup in figure 1. $I_{\lambda}$ is the intensity of the light source for wavelength $\lambda$. $\eta_{\lambda}$ is the quantum efficiency of the detector for $\lambda$. $\mu_{\lambda}$ is the attenuation coefficient for $\lambda$. $d$ is the pathlength travelled through the dosimeter and $D$ is the dose deposited. Similarly, the laser’s signal can be written as:

$$S_{633} = e^{-\mu_{633} \cdot D \cdot d}$$

(eqn 2)

Where $S_{633}$ is the detected signal and $\mu_{633}$ is the attenuation coefficient of the dosimeter at 633 nm.
Calculations were made with the above data for different levels of attenuation (ie \( D \cdot \alpha \)) assuming a uniform attenuation along the pathlength. Comparisons could then be made to relate the ideal 633nm signal to that of the LED spectrum. Knowing this relationship allows for a correction factor to be applied to each point on the sinogram, thus taking the raw data of an LED spectrum and converting it to what would have been measured by a tighter HeNe light source. The specific correction factor can be written as:

\[
CF_{\text{attn}} = \left( \frac{S_{633}}{S_{\text{LED}}} \right)_{\text{attn}}
\]

(eqns 3)

Where \( CF_{\text{attn}} \) is the attenuation specific correction factor for a pathlength with an average dose, \( D \), and a pathlength, \( d \).

2.3. Spectral warping effects via simulation

To show the effect of spectral warping in practice a Matlab simulation was run mimicking the response of a uniformly circularly irradiated 6cm diameter dosimeter to 2Gy. Two sinogram data sets were created, both based on imaging the \( \Delta OD \), one mimicking the data acquired from an LED light source and the other from a laser light source. The sinograms consisted of 180 projections over 180 degrees with .1mm pixels.

The images generated from the sinograms represented attenuation coefficients per pixel for the specific wavelengths in the simulated LED and laser scans. These data were forward projected and summed over the appropriate wavelengths to produce projection data. The reconstructions were then compared for differences.

2.4. Application of correction

A 16cm diameter dosimeter was irradiated with two orthogonal square fields to a dose of 2Gy and later imaged in an in house LED light source system (DLOS) with and without a 1nm bandwidth laser line filter and in the MGS commercial scanner with a HeNe laser light source. 360 projections were taken 1 degree apart with 25 images taken at each projection angle for the LED based images and 360 projections over 360 degrees were also done with the laser based scanner. The projection data was processed and placed into respective sinograms. The correction factor was applied to each point on the unfiltered sinogram. The sinograms were reconstructed and the resulting images compared for differences.

3. Results & Discussion
3.1. Simulation results
The simulation gave results very similar to what one would expect. Figure 2c shows the dramatic difference returned in attenuation coefficient values. Differences of greater than 70% for just a 2 Gy irradiation over a 6 cm diameter were seen for the LED and laser sources. Any calibration attempts for moving to absolute dosimetry would need to consider these discrepancies if using a broad band light source such as an LED.

Figure 2a & 2c also exhibit a common imaging artefact present in optical-CT imaging, a depression in the center of the dosimeter. In this case the depression was 12% below the edges of the dosimeter. While there have been several causes of such depressions reported, “spectral warping”, the preferential distinction of photons at 633 nm one of them to consider. Figures 2b & 2c show how utilizing only one wavelength in the reconstruction can return results on par with the designed expected results of the dosimeter & scanner system as a whole. The reconstruction returns values that have the expected shape and value such that the potential for calibration exists in shifting to absolute dosimetry.

3.2. Application results
Similar to the simulated results, the images reconstructed with the LED based system have the general shape of the reconstruction, but are not consistent with the laser based scanners in terms of accuracy and contrast. Figure 3 shows 4 sinograms where the post irradiation sinogram was divided pixel by pixel by the pre irradiation sinogram collected from (a) an uncorrected LED system, (b) a corrected LED system, (c) a HeNe laser system, and (d) a LED system with a 1 nm filter centered at 633 nm in the optical chain.

Comparisons of these images show very striking resemblances in the Corrected LED sinogram and the HeNe sinogram. Figure 4 displays the correlating image reconstructions in a, b, c & d with a line profile of each in e. Again there is good correlation between the corrected LED image and the HeNe image with the major difference being the HeNe...
image has more noise associated with it although the values are very similar. Looking at the 1 nm FWHM bandwidth LED image it is evident that the filter makes the system much more sensitive to any schlering bands present in the dosimeters during the curing process.

4. Conclusion
This work looked at the magnitude of error introduced by ‘spectral warping’ when performing optical-CT scans utilizing a polychromatic LED light source, and described a correction method to overcome it. This correction has the ability to eliminate discrepancies previously seen between laser scanning imaging and LED light source imaging. The correction is highly dependent upon accurate input data for the detectors quantum efficiency, the LED spectrum, and the dosimeters spectral absorption response to radiation over the spectrum of the LED. Use of these corrections should lead to more accurate dosimetry with results comparable to the laser scanning optical-CT scanning systems.

Acknowledgements: Supported by NIH R01 CA100835

References

[1] Kelly, R.G., K.J. Jordan, and J.J. Battista, Optical CT reconstruction of 3D dose distributions using the ferrous-benzoic-xylenol (FBX) gel dosimeter. Med Phys, 1998. 25(9): p. 1741-50.
[2] Adamovics, J. and M.J. Maryanski, Characterisation of PRESAGE: A new 3-D radiochromic solid polymer dosemeter for ionising radiation. Radiat Prot Dosimetry, 2006. 120(1-4): p. 107-12.
[3] Ruegsegger, P., et al., Standardization of computed tomography images by means of a material-selective beam hardening correction. J Comput Assist Tomogr, 1978. 2(2): p. 184-8.
[4] Peng, G.H., et al., [Hardening correction model of energy spectrum for X-ray TICT in testing composites workpiece]. Guang Pu Xue Yu Guang Pu Fen Xi, 2007. 27(4): p. 823-6.
[5] Joseph, P.M. and R.D. Spital, A method for correcting bone induced artifacts in computed tomography scanners. J Comput Assist Tomogr, 1978. 2(1): p. 100-8.
[6] Idris, A.E. and J.A. Fessler, Segmentation-free statistical image reconstruction for polyenergetic x-ray computed tomography with experimental validation. Phys Med Biol, 2003. 48(15): p. 2453-77.
[7] Hopkins, F., et al., Analytical corrections for beam-hardening and object scatter in volumetric computed tomography systems, in 16th World Conference of Nondestructive Testing. 2004: Montreal, Canada.
[8] Elbakri, I.A. and J.A. Fessler, Statistical image reconstruction for polyenergetic X-ray computed tomography. IEEE Trans Med Imaging, 2002. 21(2): p. 89-99.
[9] De Man, B., et al., An iterative maximum-likelihood polychromatic algorithm for CT. IEEE Trans Med Imaging, 2001. 20(10): p. 999-1008.
[10] Jordan, K. and J. Battista, Linearity and imageuniformity of the VistaTM optical cone beam scanner, in 4th International Conference on Radiotherapy Gel Dosimetry. 2006: Sherbrooke, Canada.