Published in final edited form as:

Phys Med Biol. 2011 June 7; 56(11): 3403–3416. doi:10.1088/0031-9155/56/11/014.

A method to correct for spectral artifacts in optical-CT dosimetry

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Abstract

The recent emergence of radiochromic dosimeters with low inherent light-scattering presents the possibility of fast 3D dosimetry using broad-beam optical computed tomography (optical-CT). Current broad beam scanners typically employ either a single or a planar array of light-emitting diodes (LED) for the light source. The spectrum of light from LED sources is polychromatic and this, in combination with the non-uniform spectral absorption of the dosimeter, can introduce spectral artifacts arising from preferential absorption of photons at the peak absorption wavelengths in the dosimeter. Spectral artifacts can lead to large errors in the reconstructed attenuation coefficients, and hence dose measurement. This work presents an analytic method for correcting for spectral artifacts which can be applied if the spectral characteristics of the light source, absorbing dosimeter, and imaging detector are known or can be measured. The method is implemented here for a PRESAGE[®] dosimeter scanned with the DLOS telecentric scanner (Duke Large field-of-view Optical-CT Scanner). Emission and absorption profiles were measured with a commercial spectrometer and spectrophotometer, respectively. Simulations are presented that show spectral changes can introduce errors of 8% for moderately attenuating samples where spectral artifacts are less pronounced. The correction is evaluated by application to a 16 cm diameter PRESAGE[®] cylindrical dosimeter irradiated along the axis with two partially overlapping 6×6 cm fields of different doses. The resulting stepped dose distribution facilitates evaluation of the correction as each step had different spectral contributions. The spectral artifact correction was found to accurately correct the reconstructed coefficients to within ~1.5%, improved from ~7.5%, for normalized dose distributions. In conclusion, for situations where spectral artifacts cannot be removed by physical filters, the method shown here is an effective correction. Physical filters may be less viable if they introduce strong sensitivity to Schlieren bands in the dosimeters.

1. Introduction

The first optical-computed-tomography (optical-CT) scanners developed for 3D dosimetry were designed to image polymer gels, where optical contrast arises from radiation-induced production of light scattering polymer micro-particles (Gore *et al* 1996, Maryanski *et al* 1996). The presence of significant scatter restricted the design of these first scanners to a single monochromatic laser beam that was raster-scanned across the polymer gels to build up projection images. Scatter rejection was effectively achieved by illuminating a single line integral of interest and limiting the size of the aperture to the photodiode at each scanning position (Gore *et al* 1996). Recently, radiochromic 3D dosimeters have been introduced

which exhibit light absorbing optical contrast with minimal scatter (Adamovics and Maryanski 2006, Kelly *et al* 1998). The problems associated with scattered light are greatly reduced in these transparent dosimeters, and the potential emerges for new and very fast broad-beam scanning configurations (Campbell *et al* 2010, Olding *et al* 2010b, Dejean *et al* 2001, Jordan and Battista 2006, Krstajic and Doran 2006, Sakhalkar and Oldham 2008, Wolodzko *et al* 1999). Several proposed systems employ either a single or a planar array of light-emitting-diode (LED) sources. In contrast to the laser systems, these sources are distinctly polychromatic, which can introduce new artifacts arising from non-uniform spectral absorption as a function of wavelength. The absorption spectrum of the dosimeter (PRESAGE[®] (Adamovics and Maryanski 2006, Sakhalkar *et al* 2009a, 2009b) in this work) can vary significantly over the spectrum of an LED light source. Preferential absorption of photons will occur at the dosimeters peak absorption wavelength leaving a higher ratio of photons at both longer and shorter wavelengths as light is transmitted through the dosimeter. This problem is not unique to 3D dosimetry; it is a common issue in several fields with the propagation of light through tissue. This phenomenon is similar to that of ‘beam hardening’ encountered in x-ray CT, except that in the latter preferential absorption occurs on only one side of the spectrum—for the lower energy photons. It should be noted that these artifacts present similarly to stray light artifacts (Babic *et al* 2008), so all measurements were acquired with a 5 mm high planar slit arrangement in order to minimize stray light, thereby isolating spectral effects.

In the optical case considered here, the spectrum of the beam does not become ‘harder’, but instead it becomes depressed corresponding to the shape of the absorption peak of the dosimetry material the beam passes through. The main consequence in optical-CT dosimetry is that an artificially high number of photons may be detected when compared to a monochromatic source of the same intensity. This in turn may give rise to artificially lower attenuation coefficients, and inconsistencies between different projections, which experience different attenuation differences due to pathlength and dose distribution variation through the dosimeter. Spectral changes are a source of error in the sinogram that are carried through to the reconstruction, and can create artifacts. An obvious solution to spectral changes is to introduce a band-pass filter (1–10 nm FWHM) to the optical chain immediately following the light source to prevent the introduction of already documented artifacts (Jordan 2004). In practice, however, this can produce a high sensitivity to Schlieren bands within the dosimeter, which can lead to high noise. The presence and challenges associated with these bands in PRESAGE[®] have been reported by another group (Krstajic and Doran 2006, 2007a, 2007b), and can lead to non-negligible dosimetry artifacts.

In this work we present a novel alternative method to overcoming spectral-induced artifacts, which do not use band-pass filters, and therefore reduce any artifacts associated with Schlieren bands. This method is distinctly different from algorithms developed to reduce beam hardening in x-ray CT. In the latter case, methods typically fall into two categories, pre-corrections and iterative segmentation algorithms (De Man *et al* 2001, Elbakri and Fessler 2002, Hopkins *et al* 2004, Idris and Fessler 2003, Peng *et al* 2007). Beam hardening filters are also in place on kV imaging CT machines to reduce errors. Errors both in magnitude of the reconstruction x-ray attenuation coefficients and suppression or ‘cupping’ artifacts of highly attenuated regions have been well characterized (Rueggsegger *et al* 1978, Joseph and Spital 1978). The methods developed for x-ray-CT could potentially be used to correct for spectral artifacts, but do not readily adapt to the unique requirement of optical-CT dosimetry which requires both a pre-irradiation and post-irradiation scan in order to determine the radiation-induced change in optical density (ΔOD). The essence of the proposed method is to use known spectral responses of the light source, detector and dosimetry material, to remove spectral artifacts on a pixel by pixel basis in each acquired projection image. The corrected projections then correspond to what would have been

acquired with an incoherent monochromatic light source at the peak wavelength of PRESAGE[®] absorption (633 nm). The correction methods presented here are applicable to any combination of light source and radiochromic dosimeter, and are particularly useful for scanner geometries that would require large, expensive band-pass filters.

2. Methods

The methods section is split into three subsections. Section 2.1 introduces the DLOS scanner (Duke Large-field-of-view Optical-CT Scanner). The spectral correction was developed and implemented for this scanner. This section also includes an example of the problem of strong sensitivity to Schlieren bands which may limit the use of band-pass filters. Section 2.2 introduces the formalism of the spectral correction and the pre-requisite measurements of spectral characteristics of system components. Section 2.3 contains two subsections. Section 2.3.1 illustrates and investigates the spectral artifacts through an analytic simulation which uses the measured spectral characteristics of the DLOS system, but applied to a simulated PRESAGE[®] dosimeter with idealized dose distribution. Section 2.3.2 demonstrates application of the correction to a real PRESAGE[®] dosimeter irradiated with a simple two-field irradiation.

2.1. The DLOS scanner

The spectral correction was developed for the DLOS scanner shown schematically in figure 1. This system includes a matched telecentric light source and imaging lens (Opto-Engineering, Italy). Parallel light travels through the sample, where it may be partially absorbed, and a projection image is formed by the lens from light that is nearly parallel to the optic axis. The system can image the 3D dose in the dosimeter by back-projecting a set of projection images acquired as the dosimeter rotates, as described in Sakhalkar and Oldham (2008). The telecentric lens has a narrow tolerance angle manufacturer specification of 0.1° for accepting light, and any stray light is rejected from the image as illustrated in figure 1. In principle, this is ideal for tomographic imaging, where each pixel of the image should correspond to the line-integral of attenuation without any corrupting scatter contribution. In practice, as will be discussed below, the system can be very sensitive to any pronounced Schlieren bands in the dosimeter.

Schlieren bands are visible streaks resulting from small variations in density presumably caused by a non-ideal mixing of the dosimeter creating slight differences in refractive index. This research group has used PRESAGE[®] for a number of years and has seen Schlieren bands in great enough frequency to warrant the solution described within. As the manufacturing and quality control of PRESAGE[®] improves this will become less of an issue, but the occurrence of Schlieren bands still persists at this time. Ideally the spectral problem would be solved using a band-pass filter centered about the dosimeters peak response wavelength. Often times this approach works well. However, as shown in the figure 2, placing a band-pass filter in the optical chain can result in making Schlieren bands more enhanced since fewer wavelengths are present. Fewer wavelengths increase the chance that when small refractive index in-homogeneities are present that light will be refracted more than the narrow 0.1° acceptance angle of the imaging lens creating artificially high absorption measurements making their existence more pronounced. Prominently displayed Schlieren bands result in reconstructions with Schlieren artifacts reconstructed to levels equal to the radiation-induced ΔOD , a large an unacceptable source of noise to an image and gave reason for a correction to be introduced and applied to the data where the Schlieren bands are better masked, those without a chromatic filter.

2.2. Correction method

The rationale of the method is, in essence, to acquire Schlieren-free polychromatic projection images, and from these calculate the projection images that would have been achieved with a monochromatic source. Polychromatic images are less sensitive to Schlieren bands as each pixel in the image now corresponds to a larger group of possible ray-paths through the dosimeter, which deviate slightly from the ideal parallel ray line-integral. The deviation is limited by the minor changes in the refractive index in the dosimeter, and by the strict acceptance angle of the imaging lens (0.1°), but is enough to reduce the Schlieren artifact. The cost is a theoretical reduction in contrast and spatial resolution, although this has not been of sufficient magnitude to be observed in experiments to date. The loss of resolution and contrast is of negligible practical consequence as the system has an inherent spatial resolution that is much higher than that required for clinical 3D dosimetry.

2.2.1. Spectral information—The method relies on accurate data regarding the spectral characteristics of the light source, the detector response, the absorption of the fluid and the absorption of the dosimeter (PRESAGE®). The LED spectrum was measured with a Thorlabs compact CCD spectrometer (CS100) and is shown in figure 3. The absorption characteristics of PRESAGE® responses to deposited dose were determined with a spectrophotometer from optical cuvettes or PRESAGE® irradiated to different doses (Guo *et al* 2006, Krstajic *et al* 2004). Three 1 cm pathlength cuvettes that were irradiated to 0.5, 2 and 4 Gy over the LED spectrum (610–655 nm) in 1 nm increments were measured with a Genesys 20 spectrophotometer from Thermo Scientific. The ΔOD was calculated by subtracting the unirradiated spectrum from the irradiated spectrum and dividing by the dose delivered. The absorption characteristics of the matching fluid and unirradiated PRESAGE® were also measured. Finally, the spectral response of the camera was obtained from the manufacturer's reported quantum efficiency. Figure 3 overlays the spectral information for these five spectra. It is clear from the figure that as light moves through irradiated PRESAGE®, the LED spectrum will be preferentially absorbed at 633 nm, depressing the central peak. It should be noted the response curve of PRESAGE® (green) is a $\Delta OD \text{ cm}^{-1} \text{ Gy}^{-1}$, while the unirradiated PRESAGE® curve (black) is simply $OD \text{ cm}^{-1}$, so the OD upon irradiation is the addition of the absorbed dose times the PRESAGE® response curve and the unirradiated PRESAGE® curve.

2.2.2. Formalism of the spectral correction—Figure 4 illustrates the principle measurement comprising optical-CT with the DLOS system. Projection images are acquired where each pixel in the image corresponds to a line integral of attenuation through the dosimeter. Each pixel's response to irradiance at each wavelength can be determined by multiplying the LED power spectrum, the wavelength-specific quantum efficiency of the detector and a Beer's law factor that incorporates the amount of attenuation over any given line integral. Summing over all the wavelengths and normalizing gives an equation for normalized transmission:

$$S = \frac{\int_{\lambda} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f\lambda} \cdot x_f - \mu_{d\lambda} \cdot x_d - \mu_{\Delta d\lambda} \cdot x_{\Delta d} \cdot D} \cdot d\lambda}{\int_{\lambda} I_{\lambda} \cdot \eta_{\lambda} \cdot d\lambda} \cdot \left(\frac{\int_{\lambda} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f\lambda} \cdot x_{fT}} \cdot d\lambda}{\int_{\lambda} I_{\lambda} \cdot \eta_{\lambda} \cdot d\lambda} \right)^{-1} \quad (1)$$

$$S = \frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f\lambda} \cdot x_f - \mu_{d\lambda} \cdot x_d - \mu_{\Delta d\lambda} \cdot x_{\Delta d} \cdot D}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda}} \cdot \left(\frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f\lambda} \cdot x_{fT}}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda}} \right)^{-1} \quad (2)$$

where S is the normalized transmission through the dosimeter at the detector along a pathlength, x . I_λ is the irradiance of the light source for wavelength λ . η_λ is the quantum efficiency of the detector for λ . μ_λ is the wavelength-specific attenuation coefficient along the pathlength, x . The subscripts f , fT , d and Δd are for fluid, fluid tank, dosimeter, and irradiated dosimeter, respectively. D is the dose delivered. The factor in the parenthesis is commonly referred to as a flood correction, but is present to give the transmission. The numerators of the two factors are summations over all wavelengths allowing for the calculation of the transmission measurement using Beer's law with the detector, source and dosimeter spectral information. The denominator is a summation over the source and detector components over all wavelengths to normalize out these components.

The quantity of interest in dosimetry is the radiation-induced ΔOD , which is known to be directly proportional to dose for PRESAGE® (Adamovics and Maryanski 2006, Guo *et al* 2006). A map of ΔOD for each slice of the dosimeter can be reconstructed by dividing the sinogram of the pre-irradiation scan by that of the post-irradiation scan, taking the logarithm, and feeding into a filtered back projection algorithm. Here we refer to a sinogram as $\log(S_{\text{pre}}/S_{\text{post}})$, thus representing a change in attenuation and not strictly attenuation.

For a monochromatic light source, the signal from each pixel on the detector (sinogram) is modeled as follows from equation (2):

$$S_{\text{pre}} = I_0 e^{-\mu_f(x_f - x_{fT}) - \mu_d x_d} \quad (3)$$

$$S_{\text{post}} = I_0 e^{-\mu_f(x_f - x_{fT}) - \mu_d x_d - \Delta\mu_{\Delta d} x_{\Delta d}} \quad (4)$$

where S_{pre} and S_{post} are the signals collected from the pre-irradiation and post-irradiation scans, respectively. I_0 is the initial beam irradiance μ_f , x_f , μ_d , x_d are the attenuation coefficients and pathlengths of the fluid and unirradiated dosimeter, and $\Delta\mu_{\Delta d}$ and $x_{\Delta d}$ are the change in attenuation coefficient and pathlength for the irradiated portion of the dosimeter. Note the lack of dosimeter response terms in S_{pre} as the dosimeter has not undergone irradiation yet. Using the previously mentioned definition of a sinogram leaves:

$$\text{Sino} = \ln\left(\frac{S_{\text{pre}}}{S_{\text{post}}}\right) = \ln\left(\frac{I_0 e^{-\mu_f(x_f - x_{fT}) - \mu_d x_d}}{I_0 e^{-\mu_f(x_f - x_{fT}) - \mu_d x_d - \Delta\mu_{\Delta d} x_{\Delta d}}}\right) = \ln(e^{\Delta\mu_{\Delta d} x_{\Delta d}}) = \Delta\mu_{\Delta d} x_{\Delta d} \quad (5)$$

In this equation, the sinogram simplifies to an expression independent of the fluid and initial dosimeter conditions.

Conversely, for a *polychromatic* light source, equation (2) yields

$$S_{\text{pre}} = \frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_\lambda \cdot \eta_\lambda \cdot e^{-\mu_{f\lambda} x_f - \mu_{d\lambda} x_d}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_\lambda \cdot \eta_\lambda} \left(\frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_\lambda \cdot \eta_\lambda \cdot e^{-\mu_{f\lambda} x_{fT}}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_\lambda \cdot \eta_\lambda} \right)^{-1} \quad (6)$$

$$S_{\text{psot}} = \frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f,\lambda} x_f - \mu_{d,\lambda} x_d - \Delta\mu_{\Delta d,\lambda} x_{\Delta d}}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda}} \left(\frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f,\lambda} x_f}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda}} \right)^{-1} \quad (7)$$

Inserting the values into the sinogram we obtain

$$\text{Sino} = \ln \left(\frac{S_{\text{pre}}}{S_{\text{post}}} \right) = \ln \left(\frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f,\lambda} x_f - \mu_{d,\lambda} x_d}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f,\lambda} x_f - \mu_{d,\lambda} x_d - \Delta\mu_{\Delta d,\lambda} x_{\Delta d}}} \right) \quad (8)$$

Equation (7) does not simplify to an expression independent of the fluid or the initial dosimeter conditions. These extra terms will lead to inaccuracies of reconstruction values making absolute dosimetry unfeasible and may lead to reconstruction artifacts if unaccounted for.

2.2.3. Correction for spectral artifacts—Equations (5) and (8) can be relabeled for monochromatic and polychromatic sources:

$$\text{Sino}_{633} = \Delta\mu_{\Delta d} \cdot x_{\Delta d} \quad (9)$$

$$\text{Sino}_{\text{LED}} = \ln \left(\frac{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f,\lambda} x_f - \mu_{d,\lambda} x_d}}{\sum_{\lambda=610 \text{ nm}}^{\lambda=655 \text{ nm}} I_{\lambda} \cdot \eta_{\lambda} \cdot e^{-\mu_{f,\lambda} x_f - \mu_{d,\lambda} x_d - \Delta\mu_{\Delta d,\lambda} x_{\Delta d}}} \right) \quad (10)$$

Equations (9) and (10) are the predicted sinogram values for the polychromatic LED source and an ideal source at 633 nm. With these equations and the spectral characteristics presented in figure 3, correction factors can be calculated to convert a measured polychromatic sinogram value to the one that would have been measured with an ideal monochromatic source ($\Delta\lambda \leq 1 \text{ nm}$). The correction factors are calculated by considering the ratio of predicted Sino_{LED} (equation (10)) and Sino_{633} (equation (9)):

$$\text{Sino}_{\text{cor}} = \text{Sino}_{\text{LED}}^{\text{meas}} \cdot \text{CF}_{\text{attn}}^{x_d, x_f} \quad (11)$$

$$\text{CF}_{\text{attn}}^{x_d, x_f} = \left(\frac{\text{Sino}_{633}}{\text{Sino}_{\text{LED}}^{\text{meas}}} \right)_{\text{attn}}^{x_d, x_f} \quad (12)$$

where Sino_{cor} is the corrected sinogram, $\text{Sino}_{\text{LED}}^{\text{meas}}$ is the measured sinogram from the DLOS and CF_{attn} is the attenuation-specific correction factor for a sinogram reading with an average dose and a pathlength, d . attn is the average dose along a dosimeter pathlength multiplied by the dosimeter pathlength. The ratio of these predicted values will change with the amount of dose deposited, dosimeter pathlength and fluid pathlength as all other variables in equation (10) will remain constant for each measurement. Different pixels in a projection image correspond to ray paths with potentially very different pathlengths through the dosimeter, and fluid. To implement the correction efficiently, a look-up table was therefore created to enable a fast look-up of pre-calculated correction factors for a wide

range of pathlengths of different materials. The table was created by incrementing the dosimeter pathlength from zero to the dosimeter diameter by 2 mm, and by incrementing an average dose from zero to the maximum dose delivered by 0.1 cGy. Since the summation of fluid and dosimeter length will always equal the aquarium length, together they represent one parameter. The table is then used by selecting the appropriate dosimeter pathlength and matching the transmission measurement, $\text{Sino}_{\text{LED}}^{\text{meas}}$, with the Sino_{LED} value in the table and selecting the corresponding correction factor. For clarity, figure 5 shows the general use of the look up table.

2.3. Application of the spectral correction

2.3.1. Simulation of spectral correction—To investigate and illustrate the effect of spectral artifacts, Matlab software was written to simulate optical-CT imaging with the DLOS telecentric system. The main variables of the simulation were the spectral characteristics of the light source. Simulations with a monochromatic beam represent the gold standard data, free of spectral warping artifacts. Simulations with the measured polychromatic spectrum from the DLOS scanner enabled investigation of the magnitude of spectral warping for this system. All simulations used the measured spectral absorption profile of PRESAGE[®] (figure 3) and the camera response from the manufacturer. The simulation presented here represents a simple single beam irradiation ($6 \times 6 \text{ cm}^2$ field) irradiated centrally along the axis of a 15 cm diameter dosimeter to a uniform dose of 10 Gy. A dose of 10 Gy typically gives an attenuation value in PRESAGE[®] of 0.2 cm^{-1} . Thus the signal variation with a monochromatic light source would range from -0 dB in the matching fluid to -50 dB through the center of the dosimeter. A model of the attenuation was created using a square grid with 1 mm pixel sizes. To create sinograms for reconstruction, transmission values for each light source were calculated along a projection angle, the model was rotated and the process repeated until there were sinogram data sets consisting of 360 projections over 360° with 1 mm pixel sizes along each projection. The sinograms were then reconstructed using a parallel beam iradon transform with a Ram-Lak filter. The reconstructions were then compared for differences. The images generated from the sinograms represented a change in attenuation coefficients per pixel for the specific wavelengths in the simulated LED and laser scans.

2.3.2. Test of the spectral correction—A rigorous test of the spectral-warping correction was performed by application to a real PRESAGE[®] cylindrical dosimeter without the presence of Schlieren bands. The dosimeter was scanned in the DLOS scanner three times: first without any filter (full polychromatic light source, and maximum spectral warping artifact), second with a 10 nm FWHM filter (moderate spectral warping) and third with a 3 nm filter (minimal spectral warping). Corresponding pre-irradiation scans with the same filter settings were acquired. The filters were centered on the maximum absorption wavelength of PRESAGE[®] at 633 nm. Each optical-CT scan consisted of 360 projections taken 1° apart with 20 images taken at each projection angle (for averaging) and reconstructed to 2 mm isotropic voxel lengths. The cylindrical dosimeter was 16 cm in diameter, and was irradiated with 2 partially overlapping $6 \times 6 \text{ cm}^2$ square fields. These fields were of different strength, 400 and 800 MU, respectively, creating three distinct dose levels in the dosimeter of $\sim 4, 8$ and 12 Gy covering a $6 \times 9 \text{ cm}^2$ region. Stray light and spectral warping present artifacts in the same fashion; both show a depression of overall reconstruction values and dipping artifacts in homogenous regions (Bosi *et al* 2009, Olding *et al* 2010a). To avoid any confusion of the two effects, stray light was eliminated by a slit collimator of width 5 mm. The pre-and post-projection data were processed and placed into respective sinograms. A correction factor was applied to each pixel on the unfiltered, 10 nm and 3 nm sinograms and each was reconstructed using an iradon transform with a Ram-Lak

filter. The resulting reconstructions were compared to determine the relative magnitude of spectral warping artifacts in the three scenarios.

3. Results and discussion

3.1. Magnitude and effects of spectral artifacts as determined through simulation

The consequence of spectral artifacts in the simplest of simulated dose distributions can be seen in figure 6. The figure shows optical-CT reconstructions of a single axial slice through a simulated cylindrical PRESAGE[®] dosimeter of diameter 15 cm irradiated with a single square beam ($6 \times 6 \text{ cm}^2$) incident along the central axis to a dose of 10 Gy. Figure 6(A) shows the reconstruction obtained from a simulated optical-CT scan using monochromatic light representing a laser scanner. Figure 6(B) shows the corresponding reconstruction for a polychromatic DLOS light source. Figure 6(C) compares line profiles through the two reconstructions along the dashed line as indicated in figure 6(A).

As one may have speculated from inspection of figure 3, spectral artifacts depress the reconstruction signal, or cause the dosimeter to appear more transparent, but more dramatically along the heaviest attenuating line integrals; in this example, along the diagonals of the square field. The effect, 7% in magnitude shown here (3–14% for typical irradiations), could potentially play a large role in typical dosimetry analysis techniques such as dose difference or gamma criteria, with common thresholds at 3%. The magnitude of any depressions in the reconstruction values should be limited to this 25% assuming the dynamic range of the system is on the order of ~60 dB.

3.2. Spectral correction applied to a real dosimeter

Figure 7 shows 2 mm thick axial slice reconstructions through the Schlieren free PRESAGE[®] dosimeter irradiated with two overlapping fields. The top row (figures 7(A)–(C)) corresponds to uncorrected optical-CT scans of the three scanning scenarios: no-filter (maximum spectral effects), 10 nm filter and a 3 nm filter (minimal spectral effects). The middle row (figures 7(D)–(F)) shows the corresponding corrected images. Profiles along the indicated dashed line show all curves in good agreement except for the unfiltered uncorrected profile, which exhibits a marked reduction of ~20% in reconstructed attenuation values. This 20% is entirely attributed to spectral artifacts. It is apparent that the application of either the 10 nm or 3 nm filter dramatically removes this artifact without the need for a correction. The dipping artifact observed in the simulation, was not noticeable here because the irradiation only deposited a dose of ~12 Gy to a smaller region which did not provide enough attenuation through the projections to have a significant dip relative to the noise.

It is clear from figure 8, that applying a calibration curve (attenuation-coefficient to dose) on the uncorrected attenuation values for the no-filter scenario, would cause substantial error in the reported dose estimate. In this case, a linear calibration curve was obtained though cuvettes measurements, and yielded a response of $0.0203 \Delta\text{OD cm}^{-1} \text{ Gy}^{-1}$. Using this result, the measured dose estimate in the central region of the high dose plateau was 7.8 and 10.0 Gy for the uncorrected and corrected unfiltered reconstruction values, respectively. The calculated dose from the Eclipse's AAA algorithm gave 10.4 Gy indicating good agreement with the corrected data, but a difference of 26% the uncorrected data. This demonstrates the importance of accounting for spectral warping when calibrating to absolute dose in this way.

If there is no desire to calibrate the data and it is to be used as a relative dosimeter another problem is presented. A simple normalization of the uncorrected data will not be sufficient since each point of the sinogram has a different correction factor, i.e. the correction factor is larger for stronger attenuating line integrals. Figure 8 shows the same line profiles as figure 7(G), but normalized to a value of 1. The figure also includes a calculation from the

treatment planning system (Eclipse) as an independent verification of the actual dose profile. This indicates that even if absolute dosimetry is not the goal, spectral artifacts may still be present, and in this case up to 8% off in the mid to low dose regions. This suggests that when using a polychromatic light source even a simple normalization may yield undesirable results with non-obvious spectral artifacts.

Even if no calibration is applied, and the dosimeter is used in relative mode, problems still present. A simple normalization of all the profiles in figure 7 is presented in figure 8. The figure illustrates that relative dosimetry of uncorrected data also shows spectral warping artifacts arising from the fact that the correction factor is larger for stronger attenuating line integrals. In this case the relative errors were ~8% in the low dose region. This shows that when using a polychromatic light source even a simple normalization may yield inaccurate results due to spectral artifacts.

3.3. Uncertainty in the correction algorithm

The accuracy of the correction algorithm is dependent upon the accuracy of spectral absorption characteristics of the light source, fluid and dosimeter. The spectrometer used to measure the spectrum of light source has an advertised SNR of 2000 for the settings used, so uncertainty due to noise is negligible, but the uncertainty in the accuracy of the measurements is larger, up to 1%. The quantum efficiency of the camera is estimated to be accurate within 2%. The uncertainty in the spectrophotometer measurements can be estimated within 0.002 for OD measurements of the fluid and dosimeter properties. Carrying the propagation of error through from the raw sinogram to the corrected sinogram reveals an uncertainty of 4.5%. This would translate into maximum dose error of ~4.5%. This is consistent with the discrepancy when trying to calibrate the dosimeter in the previous section where a discrepancy of 4% was noted.

4. Conclusion

This work demonstrates that it is possible to achieve quantitatively accurate dosimetry images utilizing radiochromic dosimeters and a polychromatic light source without resorting to physical filters. The introduction of narrow band-pass physical filters can introduce undesirable noise arising from Schlieren bands in PRESAGE[®] dosimeters. In this situation, the correction method presented here is a viable and preferable solution. The magnitudes of spectral artifacts are significant (up to 25% in the example shown here) and must be accounted for either by use of filters or by analytic correction. The correction is easy to implement and just requires spectral characteristics of the source, dosimeter and camera. We note that recent improvements in the manufacturing processes of PRESAGE[®] have led to reduced Schlieren bands. This may enable spectral artifacts to be accounted for more routinely with band-pass filters.

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Figure 1. Schematic of the DLOS telecentric scanner (Duke Large field-of-view Optical-CT Scanner). The red dashed lines indicate simplified light paths. When the band-pass filter is in place it is located immediately following the LED.

Figure 2.

(A) An OD projection image of an unirradiated dosimeter with a 3 nm chromatic filter centered about 633 nm. (B) An OD projection image of the same dosimeter without a chromatic filter masking the Schlieren structures present from the manufacturing of the dosimeter.

Figure 3.

Spectral characteristics for the five key components affecting spectral artifacts. Preferential absorption of photons occurs at 633 nm with less radiation-induced attenuation at longer shorter wavelengths. This leads to an artificially high number of photons collected as compared to a monochromatic laser light source at 633 nm. CCD efficiency from manufacturers guide, all other spectra are measurements from commercial spectrometers or spectrophotometers.

Figure 4.

Simplified schematic of a line integral for a projection image without the optics. The collimated light source is sent through the aquarium with index matching fluid (μ_f), portions of unirradiated dosimeter (μ_d) and portions of irradiated dosimeter ($\mu_{\Delta d}$).

Dosimeter Pathlength 5.2cm				Dosimeter Pathlength 5.4cm			
Dose	$Sino_{LED}$	$Sino_{633}$	CF	Dose	$Sino_{LED}$	$Sino_{633}$	CF
10	.070	.086	.814	10	.071	.086	.829
20	.027	.037	.761	20	.028	.037	.770
30	.011	.016	.706	30	.011	.016	.711
40	.004	.007	.650	40	.004	.007	.652
50	.002	.003	.593	50	.002	.003	.594
D_{max}	...	...	...	D_{max}	...	...	...

LED Measurement

Corrected LED Measurement

Red arrows point from the circled values .028 and .037 in the table to the corresponding pixels in the sinograms.

Figure 5.

Example of the look up table. The table is pre-generated by calculating values for $Sino_{LED}$ and $Sino_{633}$ for several discretized dosimeter pathlength and dose values. For each pixel in the sinogram, upon appropriate selection of the dosimeter pathlength has been chosen, a match between the $Sino_{LED}$ and $Sino_{LED}^{meas}$ is made. At this point the corresponding correction factor can be applied to create a corrected sinogram.

Figure 6.

Simulated optical-CT reconstructions of a 10 Gy, $6 \times 6 \text{ cm}^2$ field irradiation delivered to a 15 cm diameter dosimeter in a 20 cm aquarium scanned with (A) a monochromatic (633 nm) light source and (B) the measured polychromatic LED light source and other spectral characteristics from the DLOS scanner (figure 3). (C) A line profile through the middle row of the reconstructions show an approximate 7% dipping spectral artifact in the uniform dose region and a 25% depression is observed in the overall reconstructed attenuation values. The dashed line in (A) represents the line profile in (C).

Figure 7.

Top row: uncorrected optical-CT reconstructions using the projection data acquired with (A) no chromatic filter, (B) a 10 nm chromatic filter and (C) a 3 nm chromatic filter. Second row: corrected optical-CT reconstructions using the projection data acquired with (D) no chromatic filter, (E) a 10 nm chromatic filter and (F) a 3 nm chromatic filter. Bottom row (G): line profiles taken of reconstruction values representing the calculated attenuation coefficient for each data set. The data set affected most dramatically by the spectral correction is that without a chromatic filter.

Figure 8.

(A) The same line profiles as presented in figure 6(G) with the maximum value normalized to 1 and the addition of a normalized treatment planning system (Eclipse) calculation. The high dose region looks fairly consistent for all data sets although significant discrepancies are present in the lower dose regions when the correction is not applied. (B) A magnified view of the 4 Gy region in (A) showing an 8% difference in the spectrally accounted and calculated values versus the uncorrected data set acquired with no band-pass filter.