CCD-based optical CT scanning of highly attenuating phantoms

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Abstract. The introduction of optical computed tomography (optical-CT) offers economic and easy to use 3-D optical readout for gel dosimeters. However, previous authors have noted some challenges regarding the accuracy of such imaging techniques at high values of optical density. In this paper, we take a closer look at the ‘cupping’ artefact evident in both light-scattering polymer systems and highly light absorbing phantoms using our CCD-based optical scanner. In addition, a technique is implemented whereby the maximum measurable optical absorbance is extended to correct for any errors that may have occurred in the estimated value of the dark current or ambient light reaching the detector. The results indicate that for absorbance values up to 2.0, the optical scanner results have good accuracy, whereas this is not the case at high absorbance values for reasons yet to be explained.

1. Introduction

As has been described in previous DOSGEL conferences [1, 2], optical CT is an attractive imaging readout modality for 3-D dosimetry experiments, because of its speed, spatial resolution and low cost. However, the scanners currently available are the result of only a limited development effort by a small number of academic teams and this is in marked contrast to the 30 years of major investment by multinational companies that is the case with Magnetic Resonance Imaging, the main competing modality.

One result of this is that a number of artefacts may still be present in optical CT images and, while careful image acquisition can usually eliminate these, their existence may reduce confidence in the technique. In this study, we examine further the “cupping” artefact described by De Jean et al. at the 2006 DOSGEL meeting [3]. Figure 1 of their paper demonstrates the effect with data acquired on the commercial Vista cone-beam scanner. The authors comment:

\textit{As expected from the multiple scattering conditions in polymer systems, gels irradiated to uniform but high doses with significant opacity exhibit cupping in CT numbers as one crosses the phantom... Gels that are irradiated with distributions that preserve transparency throughout most of the phantom... show well behaved dose response.}

The suggestion of De Jean et al. is that the effect is caused by multiple scattering, and thus represents, effectively, a breakdown in Beer’s Law (an implicit assumption of the CT method). However, this is not the only possible explanation for this type of observation. In this paper, we set out to reproduce the effect and discuss other potential causes.
2. Materials and methods

A uniform and stable water phantom was created, whose optical attenuation was varied either by the addition of an absorbing blue dye (food colouring), or a scattering agent (Dettol™) [4]. Two containers with different refractive indices were investigated. The first was clear plastic jar of approximately 7 cm diameter made of polyethylene terephthalate (PET, $n=1.575$), the material used for the Vista scanner. The second container was formed from a transparent, thermoplastic sleeve made of Teflon® FEP (DuPont, $n=1.341$) approximately 7.9 cm in diameter. The bottom of the sleeve was sealed to an aluminium base plate using a thin layer of a warm gelatin mixture and subsequently cooled in a refrigerator. Both containers were filled with deionised water (Milli-DI® systems, Millipore), then inserted in the scanning tank and surrounded by deionised water. Optical CT images were acquired using the method and apparatus described in [5]. For the second series of scans, the Teflon container was filled with deionised, degassed water, to which varying concentrations of blue dye or Dettol™ were added, and optical CT images acquired.

As will be shown in the following section, these images demonstrated cupping artefacts. Since the effect appeared with both scattering and absorbing phantoms, we eliminated multiple scattering as a cause of the cupping in our case (although this may have been the origin in the case of De Jean et al.). It was hypothesised that this effect was caused by the low detected signal at the centre of the phantom, which is only slightly above the level of the dark-current image. As shown in [6], the dynamic range of the CCD should have been sufficient to measure successfully the expected absorbance range encountered. However, any errors in the estimated value of the dark current, or ambient light reaching the detector not accounted for, would cause corresponding errors in the measured absorbance values.

In order to discover whether this was the cause of the discrepancy, the dynamic range extension method of Krstajić and Doran ([6]) was employed. Briefly, this consists of the following procedure.

(a) Acquire one set of projection images ($P_1$) with the normal light intensity level, chosen such that the largest signal value is close to the detector maximum, but no saturation occurs. (For highly absorbing phantoms, the minimum transmitted light intensity will be very close to the detectable threshold — this is the problem that we are setting out to solve.)

(b) Acquire a second set of images ($P_2$) with increased light intensity. The low absorption regions of the projection will now be saturated but bona fide data will now be detected in the centre.

(c) Construct composite projections ($P_{\text{comp}}$) pixel-by-pixel, using data from $P_1$, wherever the signal is above a pre-determined threshold (related to the dark current) and suitably scaled data from $P_2$ where the data fall below this threshold. Scaling may be accomplished either by adjusting the light intensity such that certain regions of the projection contain valid data for both $P_1$ and $P_2$ or by acquiring an additional light-field image with a calibrated neutral density filter in the beam path.

(d) Reconstruct projections $P_{\text{comp}}$ as normal.

Figure 1: Results of image acquisitions to demonstrate the effect of wall artefacts. Profile across sample with image inset for (a) Teflon® FEP and (b) PET containers.
If the cause of the problem was indeed an erroneous estimation of the ambient light or the dark image, then projections $P_2$, which are obtained with a much higher light level than $P_1$, should be less affected by the artefact.

3. Results and discussion

Figure 1 demonstrates the importance of using a container whose refractive index is close to that of its contents and surroundings. Wall effects in the reconstructed slices in Figure 1(a) for PET are significantly worse than those of Figure 1(b) for Teflon FEP. In the former case, the field-of-view over which reliable quantitative measurements can be made is reduced to less than 50% of the phantom diameter, compared with around 80% for Teflon FEP. Although in a practical measurement on a dosimeter, a pre-irradiation scan of the phantom [7] will correct for much of the error, it is still better to minimise the potential problems by good data acquisition practice and match the refractive index of the container to that of the gel dosimeter.

Figure 2 shows the results of increasing the concentration of the attenuating additive. In each case, profiles have been corrected by dividing by the zero-concentration scan. It is seen that, as the concentration increases, the shape of the absorbance profile changes from a smooth arc of form

$$A(x) = \mu y = 2\mu(r^2 - x^2)^{1/2}$$

into a profile with much steeper sides and a flatter top (Figure 2(a)). This, in turn, leads to an image, whose edges are much brighter than expected and whose middle is depressed (Figure 2(b)). If one plots the image intensity at the centre, one sees that the expected linear relationship between dye concentration and reconstructed optical density breaks down (Figure 2(c)).

In implementing the extended dynamic range method, we expected to encounter the problem of blooming of the CCD and Figure 3 shows graphically how serious this can be. When the CCD receives a light signal that is too high, i.e., too many photons are incident on a given pixel on the chip, the signal output saturates and the charge-accumulating well “overflows”. Photo-electrons are detected in adjacent wells (pixels). Although the method described above expects the saturation, the corruption of data in the pixels surrounding the saturated region is a problem. Furthermore, in extreme cases the output amplification stage of the camera may be overloaded causing complete breakdown and zero recorded signal, as illustrated in Figure 3(b). The solution is straightforward for this experiment, with its simple cylindrical geometry — in the second scan we simply masked the regions in which signal overflowed — but this would not be possible in the general case and this would limit severely the
application of this method in the general case. (This issue is not of course a problem with the fast laser scanner [8]).

Figure 4(a) analyses the results of an imaging experiment with normal light intensity level for a highly absorbing phantom. Owing to the quality of the CCD, low noise data are obtained even at high values of absorbance. However, the systematic error causing the wings in Figure 2(b) is evident. Data are expected to satisfy Eq. [1] (smooth green line), but this is patently not the case. Initially, we hypothesised that the cause of this was an incorrect dark image level used in the correction of the data. In order to obtain our values for absorbance we use the following equation [6]:

\[
A = -\log_{10}\frac{S - D}{S_0 - D} = -\log_{10}\frac{\text{Raw data} - \text{Dark image}}{\text{Correction data} - \text{Dark image}}
\]  

[2]

It can be shown that if \(D\) is in error by an amount \(\delta\), then the corresponding error in absorbance \(A\) is \(\delta / (S-D)\). If \(S \gg D\), then this poses little problem. However, for highly attenuating samples, \(S\) approaches \(D\) (i.e., the contribution made to the signal by the light we are actually trying to measure becomes small). Under these circumstances, any slight error in the estimation of \(D\) (or equivalently, failing to take account of changes in ambient light) becomes very significant. The blue (middle) line in Figure 4(a) demonstrates how, by adding an appropriate (~10%) constant to the value of \(D\) in Eq. [2] before performing the data processing, we “improve” the fit dramatically. The high light field experiment would confirm this was the correct thing to do, because if the hypothesis were true, the new signal at the centre of the phantom should be much less sensitive to inaccuracies in the estimation of \(D\). — i.e., a good fit without including a \(\delta\) term, since \(D\) is inherently much greater than \(S\).
An independent measurement of the optical density value of the neutral density filter was performed by filling the scanning tank with de-ionized water and increasing the light intensity until the detected signal was just below saturation level. A single projection was taken, then a reflective neutral density filter (Edmund Optics NT43-815) of nominal absorption 1.0 and tolerance ±5% was placed in front of the LED, and another projection taken. The experimental optical density 0.95165 (SD=0.005, SEM=5×10^{-5}) was used to correct the data in Figure 4(b) and it should be noted that there are no adjustable parameters in the scaling between the overlying red and black lines. The increase in light intensity leads to data that have higher SNR, as expected [6], but demonstrates categorically that our hypothesis concerning the cause of our incorrect measurement of high absorbance values is incorrect.

4. Conclusions

We have reproduced an artefact very similar to that demonstrated in [3] using different apparatus. Analysis has shown that the incorrect image profile is not specific to a scattering sample but occurs in other cases where high absorbance values are incorrectly measured by the scanner. The hypothesis that the cause of the effect is an incorrect measurement of the dark current or unaccounted for ambient light has been proved to be incorrect. Further investigations are ongoing.

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