Optical-CT scanning of polymer gels

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Abstract

The application of optical-CT scanning to achieve accurate high-resolution 3D dosimetry is a subject of current interest. The purpose of this paper is to provide a brief overview of past research and achievements in optical-CT polymer gel dosimetry, and to review current issues and challenges. The origins of optical-CT imaging of light-scattering polymer gels are reviewed. Techniques to characterize and optimize optical-CT performance are presented. Particular attention is given to studies of artifacts in optical-CT imaging, an important area that has not been well studied to date. The technique of optical-CT simulation by Monte-Carlo modeling is introduced as a tool to explore such artifacts. New simulation studies are presented and compared with experimental data.

1. Introduction

1.1. Origins of optical-CT polymer gel dosimetry

In 1996 Gore et al [1] introduced a new method of 3D dosimetry using optical-computed-tomography, or optical-CT, to scan tissue equivalent gels that exhibited radiation-induced polymerization when exposed to high energy ionizing radiation. A dose response amenable to optical measurement was reported, arising from the production of light scattering micro-particles from radiation induced polymerization of acrylic monomers dispersed in the gel. Unirradiated gel was virtually transparent, but irradiated gel became increasingly opaque with dose as the number density of scattering micro-particles increased. The partial transparency of the gel lends itself to optical scanning by optical-CT, a technique analogous to first generation x-ray CT, except that a visible light source is used instead of an x-ray source. Gore et al proposed a simple optical-CT instrumental set-up reproduced below in figure 1. In figure 1 the radiation sensitive polymer gel is enclosed in a cylindrical transparent flask immersed in an optically matched water bath to minimize refraction effects at the surface of the flask. The 632nm HeNe laser beam is stepped across the flask by means of synchronous moving mirrors to acquire the transmission data for each projection. At each step of the projection the intensity of transmitted laser light was measured by a photodiode detector. The flask was rotated by a small amount between each projection such that projections were acquired for full 360 degree views around the flask. Image data for different slices could in principle be obtained by advancing the flask vertically and re-acquiring a full projection set. 2D maps of optical attenuation coefficients are then obtained using standard CT reconstruction algorithms (e.g. filtered backprojection) as illustrated in figure 2. 3D dosimetry can then be achieved by creating an image stack from scanning multiple slices. Gore et al suggested that strong refraction artifacts associated with significant refraction of the laser when incident near the edge of the flask, could be minimized by limiting the scan and irradiation to within 90% of the diameter of the flask. This simple method works well to minimize the otherwise significant wall artifacts, but has the limitation that radiation fields cannot enter through the side walls of the flask. Under these conditions an accuracy of within 5% was claimed for doses in the range 0–10 Gy, with a spatial resolution of better than 2 mm. The minimum detectable dose was estimated as 10 cGy with 5 cGy standard deviation. These bold claims, and the impressive images and dose
response in figures 1 and 2, inspired considerable interest from radiation dosimetry workers. The paper did not address in detail certain aspects of scanner performance (e.g. geometrical distortion, imaging artifacts, and the effect of scatter on measured distributions), and there was no quantitative comparison of optical-CT measured doses with independent dosimeters (e.g. film).

In a companion paper to Gore et al, Maryanski et al 1996 [2] presented investigations into the optical properties of BANG polymer gels. Several important distinctions were noted. First the absence of absorption bands in turbidity spectra of gel irradiated to different doses was interpreted as indicating the gel is a predominantly scattering medium, with negligible absorption (figure 3).

Measurement of the refractive index of samples of gel irradiated to different doses revealed a slight correlation as reproduced here in figure 4. Refractive index changes in irradiated gel could cause significant artifacts in optical-CT imaging and so the implications of these changes warrant careful study (more in section 2).

Maryanski et al 1996 were also able to estimate the maximum particle sizes of the polymer micro-particles in gels irradiated to different doses. The method involved taking the difference in turbidity spectra for gel samples irradiated to different doses and fitting the resulting difference curves using Mie theory for monodispersive spherical scattering particles. Maximum micro-particle sizes in the range 440–680 nm were reported.

1.2. Gel dosimetry by optical-CT or MRI scanning?

The two papers of Gore et al and Maryanski et al generated considerable interest in the potential and feasibility of high resolution 3D dosimetry by optical-CT scanning of polymer gels. Prior to these works MR imaging of polymer gels was the only available method for imaging the dose distribution recorded in polymer gels [3-6]. A fundamental question thus arose as to what were the relative merits of these two imaging methods. Oldham et al [7] investigated this question by applying both imaging methods to the same polymer gel samples that had been irradiated by spatially separated radiosurgery beams to various doses. Optical-CT was performed with an in house scanner. Figure 5 shows comparison dose maps of the same slice through the same gel dosimeter obtained by MRI and optical-CT imaging. To render a meaningful comparison Oldham et al required that both imaging techniques should meet as closely as possible an RTAP criteria (Resolution, Time, Accuracy, and Precision). In practice this meant dose maps should be produced with 1×1 mm$^3$ spatial resolution within 1 hour of imaging time (standard MR time slot) and with accuracy within 3% and noise within 1%.

This figure illustrates the primary advantages of the optical-CT technique in that high spatial resolution and low noise can be achieved because of the high sensitivity of photodiode light detection technology. Optical-CT can provide a low cost and attractive alternative to MRI scanning of polymer gels for many applications. It was noted that MRI gel-dosimetry retains unique abilities in that it can image both arbitrary shaped gel-phantoms and phantoms containing opaque features. These situations may cause significant optical artifacts through reflection, refractions and absorbance.

1.3. Clinical optical-CT polymer gel dosimetry

A clinical application of optical-CT polymer gel dosimetry was presented by Oldham et al [7] in the context of verification of a radiosurgery delivery. A cylindrical polymer gel flask of 80 mm diameter was placed inside a water-filled anthropomorphic head-phantom and taken through the entire radiosurgery treatment planning and treatment procedure at William Beaumont Hospital, MI, USA. A 3 isocentre lesion was simulated inside the gel, and treated
with 15 arcs of 6 MV radiation from a Varian Clinac 2100C. Comparison of the gel-measured and planning isodose lines showed excellent agreement within 1 mm (figure 6).

1.4. High resolution 3D dosimetry

High resolution 3D dosimetry using optical-CT scanning of polymer gels was described in Oldham et al [8]. 1 mm^3 dosimetry was achieved over a cylindrical volume of BANG polymer gel of dimensions 8 cm diameter and 2 cm thick. The gel had been irradiated with a 40–80 keV narrow focused x-ray beam (i.e. max diameter < 1 cm). The dosimetry of narrow focused low-energy x-ray beams represents a significant dosimetric challenge, non-feasible with conventional dosimeters, and the results (figure 7) demonstrate the unique potential of 3D optical-CT gel dosimetry.

1.5. Optical-CT polymer gel dosimetry pros and cons

The brief review of optical-CT scanning of polymer gels outlined above illustrates the main advantages and challenges of the technique.

Advantages

- Polymer gels are relatively optically stable post irradiation. There is some post irradiation increase of polymerization with time, but diffusion, the overriding problem associated with Fricke gel dosimetry is not a significant problem [9].
- Optical-CT polymer gel dosimetry is a very sensitive technique. Doses as low as 5 cGy can be measured. An important paper by Xu et al 2003 [10] demonstrated the feasibility and benefit of modifying and matching the sensitivity of polymer gels and the delivered dose to the dynamic range of the scanner.
- The dynamics of polymer gel dosimetry is now fairly well studied and understood [11,12]. Investigations have concentrated on reactions times, linearity of response, tissue equivalence, dose response, manufacture repeatability, uniformity of response etc.
- While many researchers manufacture their own Polymer gels, both the polymer gels and optical-CT scanners are now commercially available. The recent development of dry kits has significantly reduced the commercial cost of polymer gel.

Challenges

- Despite the advances in our understanding of polymer gel dosimetry, it remains non-trivial to make reproducible high quality polymer gels with uniform response and high optical clarity.
  - Since their introduction in 1993 continuous improvements in gel formulations occur and there continue to be developments to this day. Standardization remains elusive.
- Gel dosimetry can be time consuming and expensive, and quality optical phantom materials which present an effective oxygen barrier can be difficult to procure.
- The technique of optical-CT brings some unique challenges. In particular
  - Optical reflection, refraction and scattering may cause inaccurate dose reconstruction. These issues are examined in more detail below and in section 4.
2. Characterizing and improving optical-CT scanner performance

These early papers contained limited in-depth characterization of optical-CT scanning techniques. In particular there had been little attempt to characterize the gel and scanning systems independently. For optical-CT gel-dosimetry to reach its full potential, in-depth and independent characterization and optimization of the performance of optical-CT scanning system and the optical properties of the polymer gels was required.

2.1. Geometrical Distortion

Experiments to characterize the performance of an in-house optical-CT scanner were recently reported by Oldham et al [8,13]. A series of phantoms were designed to generate known baseline conditions that could be used to verify the performance of the scanner. ‘Needle phantoms’ were developed to investigate potential geometrical distortion. A needle phantom consists of a gelatin gel in an optically transparent flask rigidly supporting a geometrical pattern of optically opaque steel needles. True needle positions were determined in the central axial slice by x-ray CT scanning. Distortion in optical-CT images was then determined by registration and superposition of the x-ray CT and optical-CT images of the same plane of the same phantom. Negligible distortion of needle positions was observed (i.e. < 0.25 mm) over a phantom of 11 cm diameter when the water-bath was well matched to the refractive index of the gel. When the water bath was poorly matched, a radial compression distortion was observed the magnitude of which was linear with increasing refractive index of the waterbath [13].

In addition to x-ray CT/optical CT comparisons, polymer-gel radiation sensitive needle phantoms were constructed and imaged by optical-CT both pre and post irradiation. A comparison of needle positions in the pre and post images then enabled a determination of distortion associated with radiation induced refractive and scattering changes in the gel. Two extreme irradiation geometries were investigated in an attempt to determine worst case radiation induced geometrical distortion (figure 8).

The overlay pre and post-irradiation optical-CT images in figure 8 illustrated good matching of the pre and post irradiation needle positions. The agreement (within .3 mm) demonstrates negligible distortion within this phantom of 5.5 cm diameter and doses of ∼4 Gy.

Another approach to studying distortion arising from refractive index changes in the gel is illustrated in figure 9.

‘Slab’ phantoms were created consisting of clear gelatin gel in a cylindrical flask subtending a slab of gelatin which had known modified refractive and or attenuation properties. In figure 9a, the slab had the same refractive index as the surrounding gelatin, but had increased attenuation by virtue of blue food coloring dye. The outline and shape of the slab are clearly visible in the reconstruction and no significant artifacts are present. In figure 9b, the modified gelatin slab contained the same concentration of blue food coloring dye but the refractive index of the slab was 2% higher through doping with clear omnipaque CT dye. The reconstructed image shows significant refractive artifacts which are most prevalent at the inner and outer edges of the slab. At these interfaces laser light experiences the greatest degree of refraction in certain projections. The grainy features visible in both slabs are real features arising from the nature of the dye interaction with the gelatin.

2.2. Accuracy of reconstruction and dependence on scanning parameters

The accuracy of optical-CT reconstruction, and the influence of varying reconstruction parameters were investigated using finger phantoms by Oldham et al [8]. Finger phantoms contain gel ‘finger’ regions of known attenuation or scattering power within an otherwise
transparent gel (figure 10 a, b). The finger phantoms were optically-CT scanned and comparisons were then made between the reconstructed and known parameters of the fingers. Comparison of optical attenuation measured by the optical-CT scanner and a spectrophotometer showed excellent linearity and absolute agreement to within 3–4%. A wide range of reconstruction and imaging parameters were also evaluated with respect to their effect on image quality, including optical laser step size, number of projections, projection angular increment, and number of ADC readings to average. A nominal set of operating conditions was identified consisting of laser-step-size=1 mm, 120 projections at 1.5 degree intervals, and each point on a projection was the average of 50 ADC readings. An example of the effect of varying the laser step size is illustrated in figure 10 c, d.

2.2.1. Scatter artifacts—Several authors have noted the potential for artifacts in optical-CT images due to the detection of scattered light [10,14-16]. Very little is known however of the specifics of how scattered light might perturb 3D dose maps obtained from optical-CT. Oldham et al [13] performed a basic study of one aspect of this effect by optical-CT imaging a series of BANG gel samples, uniformly irradiated to different doses. Each sample was imaged multiple times under conditions of varying scatter-rejection achieved by varying the diameter of a scatter-rejecting collimator positioned close to the photodiode detector. A plot of reconstructed attenuation coefficients determined with the collimator fully open to 2 cm diameter and closed to 3 mm revealed a consistent 13% difference. This study illustrates how scatter can cause a global depression on reconstructed attenuation coefficients in uniform samples. A more complicated challenge is to determine the effect of scatter on the reconstructed dose in non-uniformly irradiated and high scattering gel samples. Xu et al [10] and Islam et al [14] noted a cross shaped artifact that appeared in optical-CT images of high-dose square field irradiations where the irradiated gel was highly scattering (figure 11). The cross shaped depression of reconstructed attenuation values was only observed once the optical-density of the gel had increased above ∼2.5 (Xu et al). Interpretation of these artifacts is non-trivial in many cases but the effects are amenable to Monte Carlo optical-CT simulation (discussed further in section 4).

2.2.2. Artifacts due to reflection and refraction—Two of the most significant sources of artifacts in optical-CT imaging arise from reflection and refraction of light at the walls of the gel-flask and within the gel itself. Early attempts to minimize these effects were suggested by Gore et al [1], which involved limiting the range of projection data to exclude regions close to the edges of the flask. This method can give useful results if the radiation does not extend beyond about 90% of the diameter of the flask. For many radiation deliveries however this is not the case as beams impinge on the flask in an axial manner. Several correction techniques to minimize refractive wall artifacts have since been proposed [7,13,17]. One method involves forming subtraction images of post and pre-irradiation optical-CT images of the same gel phantom, to subtract out common artifacts (e.g. wall artifacts, and any bubble or spec artifacts). As many of the same refraction and reflection artifacts are present in both scans, the difference image reveals maps of the changes in optical attenuation that were induced by the radiation. This method can also yield useful results but is limited in the amount of information that can be retrieved close to the walls of the flask. In depth discussions of these artifacts is discussed in Kelly et al [17] and Oldham et al [13].

3. Commercial optical-CT scanner

The field of 3D gel dosimetry by optical-CT has recently become more accessible through the introduction of the first commercially available OCTOPUS™ optical-CT scanning system from MGS Research. The scanner is an extension of the design illustrated in figure 1, and is capable of high-resolution 3D dosimetry of flasks of up to 20 cm diameter and 15 cm
Two recent publications describe initial experiences commissioning and using the scanning system in combination with BANG polymer gels. Islam et al [14] reported very similar dose distributions for simple field arrangements, including a single field IMRT delivery, between the gel and radiochromic film. Xu et al [10] also reported good agreement between 3D gel measured IMRT distribution and film for isodose lines greater than 30%. Discrepancies were noted at lower isodose lines, close to the walls of the flask where a degree of oxygen contamination may have occurred through the walls of the flask.

4. Optical-CT Monte-Carlo simulation

The use of optical Monte-Carlo modeling to investigate and optimize various aspects of optical-CT scanning of scattering polymer gels was introduced by Oldham et al [8]. A simple light scattering measurement was performed to study the broadening of a laser beam as it passed through 32 mm thick slabs of BANG gel that had received different uniform doses. The measured scatter profiles were compared with optical Monte Carlo simulations obtained from the LightTools software (Optical Research Associates, Pasadena, CA). The LightTools simulation assumed Mie scattering from monodisperse particles of uniform number density within the irradiated region of the slab. Particle size estimates were determined from dynamic-light-scattering (DLS) and were in broad agreement with independent estimates based on fitting measured turbidity spectra with standard Mie theory [2]. These preliminary simulations illustrated that optical Monte Carlo could model light scattering in simple geometries and had potential as a tool for studying optical effects in optical-CT scanning. Here we present more complex simulations that extend this work to investigate the ‘cross’ artifact (figure 11), independently reported by Xu et al [10] and Islam et al [14], when optical-CT scanning square field irradiations under high scatter conditions.

The simulation geometry, defined within the LightTools optical Monte Carlo software, to simulate a full optical-CT scan acquisition, is shown in figure 12. A 1 mm thick PET transparent flask of inner radius 8.6 cm is positioned within a square water-bath of side 20 cm. The flask contains unirradiated gel and a square region of irradiated gel which contains light scattering particles. Laser rays travel from the source (right side of images) through the simulated optical-CT scanner to strike a 2×2 cm detector on the opposite side of the waterbath. During the simulation the laser and detector are stepped across the full length of the flask in 2 mm increments (vertically in figure 12a). At each step a number of photon histories are run (e.g. 30,000) to determine the line integral attenuation for that step in that projection. Between each projection, the square field is rotated by a small amount and the laser stepped back across the flask to simulate data for the next projection. In the simulations presented here, 100 projections were simulated each consisting of 92 laser positions at 2 mm increments, and 3.6 degrees rotations between projections.

An initial simulation was performed where the square field was set to be purely absorbing with Mie scattering turned off (figure 13).

Profiles taken across the reconstructed square field indicate a very uniform attenuation. The reconstructed attenuation was 0.352 per cm (0.0704 per pixel), which agreed well with the true attenuation 0.356 per cm entered into the simulation. The agreement validates this implementation of optical-CT Monte Carlo model in non-scattering conditions. The simulation was extended (figure 14) to study optical-CT scanning of highly scattering gels by switching the Mie scattering option on in the LightTools software. Mono-dispersive particles were assumed of radius 475 nm. This figure is reasonably consistent (although slightly larger) than corresponding measured estimates for BANG1 gel given high dose [2,8]. A number density of micro-scatterers of 40,000 per cubic mm, and refractive index 1.54 was assumed. The mono-dispersivity assumption is reasonable given recent advances in
radiation-induced polymerization of acrylamide [18,19]. The number density was estimated empirically to give strong attenuation for this simulation geometry (e.g. reconstructed attenuation of \( \sim 0.25 \) per cm).

Despite the fact that the scattering properties of the irradiated gel were uniform across the square field in the simulation geometry, the reconstructed scan shows non-uniform attenuation. This discrepancy is evidence of a scatter artifact that occurs under high-scatter conditions in optical-CT scans of square field arrangement. The similarities between the depressed reconstructions in the center of the square field in the simulation and in measurements by other investigators is striking.

The diagonal profile indicates extended depression of the reconstructed attenuation when compared with the horizontal and vertical profiles through the field. This is consistent with the cross artifact of figure 12, and is not seen when the gel is non-scattering (compare figure 14 and figure 13). A simple interpretation is that the diagonal crosses of low attenuation correspond to line projections generating the most scattering. The corresponding increase in scattered light reaching the detector acts to make the cross regions appear more transparent to laser light. These preliminary simulations clearly indicate the potential of optical Monte Carlo simulation to study artifacts in optical-CT and to develop methods to optimize both the gels and the technique.

5. Conclusions

There is a growing body of work characterizing optical-CT scanning systems and compatible gel dosimeters. The performance of the systems are generally promising with high resolution, high sensitivity and low noise being particularly attractive features of optical-CT. The first generation scanning systems (single stepped laser-photodiode configurations) have been shown to be relatively insensitive to geometrical distortion although refractive and scattering effects have been observed. The presence of scattered light has been shown to introduce artifacts in highly scattering gels. These artifacts can be kept negligible by limiting the dose to the gel. Lowering the dose incurs a trade-off with sensitivity and noise and it remains to be seen what accuracy can be achieved with optical-CT gel-dosimetry of polymer gels.

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Figure 1.
A schematic diagram of a prototype optical-CT scanner. The mirrors translate left to right to obtain projections of the optical attenuation through the gel as described in the text. (From Gore et al 1996)
Figure 2.
Optically scanned 2D dose distribution of irradiated polymer gels (a) the calculated dose map of a cylindrical sample of radius 10cm in which four rectangular fields of different doses were placed; (b) the relationship of optical attenuation to dose. (From Gore et al 1996)
Figure 3.
Optical density (turbidity) spectra of BANG gels irradiated to different doses. This gel formulation contained 50% weighting of Bis to acrylamide. (From Maryanski et al 1996)
Figure 4.
The dose dependence of the refractive index of BANG polymer gels. This gel formulation contained 50% weighting of Bis to acrylamide. (From Maryanski et al 1996)
Figure 5.
Axial images through the same plane of the same gel-flask irradiated with 3 radiosurgery beams (a) MRI image of the R2 distribution, (b) optical-CT image of optical attenuation coefficient. The corresponding profiles along the dashed lines are shown in (c) and (d) respectively. (From Oldham et al 2001)
Figure 6.
Illustration of the use of polymer gel-dosimetry in conjunction with optical-CT scanning to verify a 3 isocenter radiosurgery treatment. The left image is the computer modeled dose distribution (20, 50 and 80% isodose lines). The right images shows the corresponding dose map obtained from the optical-CT scanner. (From Oldham et al 2001)
Figure 7.
Optical-CT images of the relative dose-distribution delivered in an 8 cm diameter polymer gel after irradiation by focused 40–80 keV x-rays. The in-plane resolution was 1×1 mm$^2$ and the images were taken 1 mm apart. The comprehensive 3D measurement presented here would be very difficult to achieve with conventional dosimeters because of the low energy of the radiation and the small dimensions of the field. (From Oldham et al 2003)
Figure 8.
Superposition of pre- and post-irradiation optical-CT images of two polymer gel needle phantoms. (A) a high dose of radiation was delivered by a circular 1.5 cm diameter field of 6 MV photons axially through the flask. (B) a rind of dose was delivered by a 5 cm circular field containing a 2 cm central circular lead block. The irradiated regions appear only lightly attenuating because the image is windowed to show needle positions. In reality the irradiated gel was highly attenuating. Irradiated regions are highlighted with dashed line. (From Oldham et al 2004)
Figure 9.
(a) Curved slab has same refractive index as surrounding gel, (b) uniform refractive index,
(b) curved slab has 2% higher than the refractive index of the surrounding gel.
Figure 10.
(a,b) 12 cm diameter gelatin finger phantom containing four variably attenuating dyed-blue gelatin fingers. (c,d) optical-CT images of the same plane through the finger phantom but acquired with different laser step spacing across each projection. Both scans consisted of 100 projections at 1.8 degrees. The laser step sizes across each projection were 2 mm (c) and 0.5 mm (d). When the laser is stepped coarsely, the fingers appear narrower and less attenuating in places due to under-sampling. (From Oldham et al 2003)
Figure 11.
Reconstructed image for 6×6 cm² field irradiation. The gel appears less attenuating in a cross shaped region aligned with the diagonals across the field. The cross only appeared in highly scattering gel where the maximum OD was >2.5. Xu et al discuss the possibility that the artifact might arise from the detection of scattered light although the specific mechanisms were not addressed. (From Xu et al 2003)
Figure 12.
Top and angled view of the optical-CT simulation geometry. The laser source (right side) and 2x2 cm detector (left side) are visible at opposite sides of the water bath (green square cube of side 20x20 cm). A PET cylindrical gel-flask of radius 8.6 cm is visible submerged in the water-bath. The blue cube located centrally in the flask represents the region of gel irradiated with a square 5x5 cm field of 6 MV radiation. The refractive indices of water-bath fluid, PET flask, unirradiated gel, and irradiated gel were 1.358, 1.52, 1.36, 1.36. Scattered light for 100 photon histories at an arbitrary laser position incident centrally on the waterbath are illustrated. The dimensions and materials are given in the text.
Figure 13.
(a) Optical Monte Carlo simulation of an optical-CT gel-dosimetry scan of a 5×5 cm square field irradiation inside a cylindrical BANG gel-dosimeter of radius 8.6 cm. The simulation geometry is given in figure 12. The gel was set to be purely absorbing with no scattering. The projections were truncated to 15 cm to minimize wall effects. The greybar scale indicates reconstructed attenuation values per pixel. (b) profiles across the square field in (a) showing excellent reconstruction of the uniformly attenuating square field. The blue profiles are horizontal and vertical profiles through the center of the field. The red outer profile is along the diagonal as illustrated.
Figure 14.
(a) Optical Monte Carlo simulation of an optical-CT gel-dosimetry scan of a 5×5 cm square field irradiation inside a cylindrical BANG gel-dosimeter of radius 8.6 cm. This time full Mie scattering was assumed in the irradiated square field portion of the gel. (b) profiles across the square field in (a) showing non-uniform reconstruction of attenuation coefficients. As in figure 13, the inner blue profiles are horizontal and vertical profiles through the center of the field. The red outer profile is along the diagonal. Scattering artifacts are clearly visible as the depressed coefficients centrally in the field.