Optical and X-ray computed tomography scanning of 3D dosimeters

K J Jordan1,2, M Hilts3 and A Jirasek4,5
1London Regional Cancer Program (London Health Sciences Centre)
2Dept. Medical Biophysics, Western University London, ON, Canada
3Medical Physics, BC Cancer Agency - Southern Interior, Kelowna BC
4Physics, I.K. Barber School of Arts and Science, University of British Columbia, Okanagan Campus, Canada
5Dept. Physics and Astronomy, University of Victoria, Canada

E-mail: kevin.jordan@lhsc.on.ca

Abstract. Optical computed tomography (CT) scanners can measure attenuation in large-volume radiochromic dosimeters. Scattered light within the dosimeters is the major factor limiting dynamic range. Scattered light limits the use of large volume polymerization gels for small field dosimetry with broad-beam scanning geometries. Planar and single ray scanning geometries generate and accept less stray light due to scatter in the dosimeter. Transparent, radiochromic dosimeters provide higher dynamic range for accurate transmission imaging. X-ray CT scanning is applicable for large volume polymerization gel dosimeters.

1. Introduction
Optical computed tomography (CT) is being developed for three-dimensional (3D) scanning of radiochromic hydrogels or plastics and radiation-induced polymerization hydrogels. Several scanner geometries have been reported by researchers and “research” versions are available commercially from MGS Research Inc. and Modus QA Inc. In contrast, X-ray CT scanners are present in radiotherapy clinics and can be used without modifications to scan polymerization hydrogels. Polymerization gel formulations are available from MGS Research Inc. (BANG®), radiochromic gels from Modus QA and radiochromic plastic (PRESAGE®) from Heuris Inc. Many researchers have developed both their own CT scanners and dosimeter materials in order to optimize a complete dosimetry system and provide excellent problems for students. This diversity of dosimeter materials, scanner geometries and 3D dosimetry applications has likely impeded commercial development [1]. To further complicate the decision process some of these materials can also be read with magnetic resonance imaging (MRI). But the optimized formulations and dose ranges may be different for optical, X-ray CT and MRI.

Throughout the DOSGEL and IC3DDose series of conferences a large amount of information related to CT scanning of 3D radiation dosimeters has been presented. Until there are edited text books for reference, those interested in efficiently learning more about this topic are encouraged to start with the review talks in the proceedings. Previously optical and X-ray CT have been addressed separately however there are many issues that are common to both. For those unfamiliar with the principles of computed tomography several books are available, for example Kak and Slaney’s text is a common reference [2].

Computed tomography is a diverse imaging field and many advances are due to the evolution of increased performance and decreased cost of computers [3]. Graphical processing units and parallel processing algorithms are now making 3D iterative reconstructions practical for clinical dosimetry QA.
processes. Optical CT evolved in many distinct applications, including: atmospheric [4], chemistry [5], radiation dosimetry [6] and developmental biology [7]. In the focus of this conference, 3D dosimetry, CT scanner development is still active as scan times are reduced and accurate dynamic range increases. Hardware modifications of clinical X-ray CT units has not been actively pursued since the emphasis has been on using available technology.

In order to select which dosimeter and scanner is optimum the dosimetry problem needs to be defined.

Is 3D dosimetry really needed? Could the problem be answered with point measurements with an ion chamber or planar measurements with film of 2D detector arrays? Is the integrated 3D dose distribution required? What is the maximum and minimum doses that must be accurately measured and what is the corresponding spatial resolution? Often, just knowing the maximum dose gradient can limit the system choices. The most challenging problems for external beam 3D dosimetry are small fields in large volumes, for example a 1 cm diameter field in a 15 cm diameter dosimeter. This problem is also challenging for CT readout of dosimeters since scattered visible light or X-rays will have a large impact on the accuracy of the reconstructions and the dose distribution is entirely due to the beam penumbra. In this refresher course the emphasis is on larger dosimeters, greater than 1 litre. Smaller dosimeters are likely to provide greater accuracy due to lower scatter contributions in projection images. Optical and X-ray CT are discussed sequentially but using similar formats.

2. Optical CT

2.1 Scanner geometries
Optical CT scanners can be categorized as broad-beam, planar and single-ray geometries. Stray light due to scatter, fluorescence, reflections and refractions within the hydrogel dosimeter and vessel is maximum for broad beam geometry. In contrast, single-ray geometries minimize “cross-talk” within the dosimeter by sampling different ray paths at different times. Camera based detectors in broad-beam systems can also have up to 5% stray light due to lens glare [8]. As the number of optical interfaces increases the potential for increased stray light also increases. There appears to be a trend of optical CT systems evolving to fewer optical interfaces. This approach should minimize stray light. Simpler systems may have more optical aberrations but computations can sort actual ray paths through dosimeters [9-13]. Lowering acceptance angles is an effective method to reduce stray light. But this increases the scanner’s optical path length. Practically, stray light becomes a larger problem as the size of the dosimeter increases. Optical elements such as mirrors, lenses and filters dramatically increase in cost as they become larger and methods to quickly scan larger 15 cm dosimeters is ongoing. In general, full 3D scan speed increases from broad-beam to single-ray. However, if additional scans are required to measure stray light then broad-beam and single-ray scan times may converge. In those geometries where a small detector moves during scanning, the advantages of both small acceptance angle stray light rejection and tolerance of ray deviations of an effectively larger acceptance angle can both be achieved. Based on symmetry, scatter results in a maximum at the optic axis. Depending on system geometry, reflection artifacts may also be maximum at the optic axis due to light travelling perpendicular to optical interfaces. Generally, the centre of the dosimeter is positioned at the optic axis. If there are temperature gradients, they will also be maximized with respect to the dosimeter centre. For all of these reasons, it is prudent not to place the reference dose point at the very centre of the dosimeter. Reference points displaced more than one cm from the optic axis generally agree better with independent point measurements.

There are alternatives to the “transmission CT” geometries. Laser sheet imaging is common for 3D optical microscopy and early results for gel dosimeters has been reported [14]. This geometry has practical advantages if the dosimeter must be rectangular in shape.

2.2 Spectral and source issues
All optical sources require spectral filtering to remove unwanted wavelengths from reaching the detector and increasing the transmission. This is especially important for obtaining high dynamic range. Generally, a narrow bandpass filter is inserted into the optical path at the source. Typically, 10 nm bandpass filters are sufficient, but this choice is coupled to the spectrum of the dosimeter. Preferably, the central probe wavelength corresponds to the peak of the dosimeter absorption spectrum. If the signal spectrum is rapidly changing over the bandwidth of the probe source then spectral changes can be expected and artifact non-linearities in dose responses will be introduced. Alternately, if the signal spectrum is broad the specific wavelength and bandwidth will have minor importance. The narrow linewidths of lasers are not required from a spectral argument and coherence may introduce interference effects which add noise to the measurements. Generally, the intense, narrow, collimated laser beams are nearly ideal for single ray scanning and there is currently no inexpensive X-ray beam of similar properties.

Light emitting diode (LED) sources are rapidly improving in brightness and choice of wavelengths. These sources are very stable when operated with appropriate heat sinks. Intensity drifts of less than 0.1% can be achieved with modest effort. Low power (<1 mW) continuous lasers are sufficient for optical CT scanning, especially for single ray scanning systems. Typical powers are 10 microwatts for dosimeters prior to irradiation. Generally, these lasers may drift up to 3% once they have reached thermal equilibrium and require some method of intensity correction. For example, normalizing profiles to signal in liquid outside the dosimeter is generally sufficient.

Users need to anticipate that all radiation dosimeters are also sensitive to ultraviolet light and minimize exposure. They should, test if photochromic responses are obtained by light levels of ambient lighting and the scanner during readout. In practice, minimize dosimeter exposure to light. Work in dim room lighting if possible and avoid exposure to daylight.

2.3 Polarization factors
While polarization is not an issue at x-ray wavelengths, it is a factor for quantitative imaging with visible light. The Fresnel coefficients describe polarization dependence of reflectivity at an optical interface. In the case of a hydrogel in a cylindrical vessel, incident unpolarized light becomes elliptically polarized due to preferential reflection of light polarized parallel to the rotation axis of the cylinder. The vessel may contain strains that introduces localized optical activity further polarizing the probe light. Gelatin is optically active and can be a further source of polarization artifacts. Polymerization to form plastic dosimeters can also introduce spatially unique optical activity [13]. The effects of polarization need to be evaluated for each dosimeter. In general avoiding rays that are not near perpendicular incidence to interfaces will minimize polarization sensitivity of scanner geometries. Systems with folded optics and high acceptance angles are likely to have polarization artifacts. While polarization sensitivity in 3D dosimeters has not been reported, the radiochromic film product EBT2 is very sensitive to polarization due to a preferentially orientation of the needle-like crystals of the diacetylene monomers. If EBT film is read with optical CT scanners assessment of polarization effects are required.

2.4 Dosimeters
Radichromic, transparent dosimeters are preferred for optical CT scanning. Absorption measurements are robust since ratio methods are used and signal is proportional to the concentration of the molecular product formed due to absorbed dose. Polymerization hydrogels have dose dependent increases in scatter which degrades linearity and spatial resolution as multiple scatter becomes a significant fraction for the attenuation. Changes in refractive index due to polymerization can create another problem for accurate 3D readout [15]. In principle, fluorescent dosimeters may have high dose sensitivity but quantitative 3D emission imaging is a more complicated measurement than transmission imaging. For the clinical dose range, 1 to 60 Gy, likely radiochromic and polymerization dosimeters will be sufficient.

All chemical dosimeters have multiple, temperature dependent reaction rates. Maintaining uniform temperatures is essential for quantitative 3D dosimetry. Reporter molecules such as dyes have high molar extinction coefficients and have much lower temperature sensitivity during absorption
measurements than weaker transitions such as ferric xylenol orange. Scanner designs need to minimize thermal gradients during readout. Ideally, a method to measure temperature throughout the dosimeter would be useful for accurate dosimetry.

Isotropic dose response is assumed in 3D dosimetry. However, this must be verified by measuring response to uniform doses. If dosimeters are found to be unique due to their manufacturing then individual, voxel-specific calibrations are required. This is analogous to current situation with radiochromic EBT films where pixel specific corrections are required for non-uniform thicknesses and variations in monomer crystal orientations.

2.5 Vessels
Hydrogel dosimeters are typically made just stiff enough not to flow if tipped horizontally. The vessel provides the mechanical stability. The vessel wall thickness is the dominant factor in how close to the inner wall optical transmission measurements can be obtained. The thinner the wall the more transmission data can be obtained. In the specific case of 15 cm diameter cylinders, 0.2 mm thick PET walls provide a good compromise between mechanical strength and missing projection data. At smaller diameters, 0.1 mm thick walls can also be used. Figure 1, shows reconstructed transverse slice from an 11 cm diameter vessel with 0.2 mm thick PET wall containing a carbon black micelle solution. The scanner was a modified Vista 10 scanner with a Fresnel lens convergent cone beam source [16].

![Figure 1.](image)

Because the vessel wall has a different refractive index from the hydrogel, no single refractive index liquid will provide minimum deflection of all transmitted rays at increasing radius [6, 17]. This occurs because the fraction of the chord length due to the gel decreases as the radius increases. The net effect of the wall is to bend rays away from the optic axis when the refractive index inside and outside the vessel is same. By lowering the refractive index outside the bending can be compensated and nearly all rays near the inner vessel wall can be sampled, see figure 2. In practice this is an iterative process for each gel formulation and vessel.

Polyethylene terephthalate (PET) is a commonly used polyester plastic in the food industry because of the following features: low oxygen permeability, strength, stiffness, high melting point and transparency. All of these features are preferred for hydrogel dosimeter vessels as well. Teflon formulations such as perfluoroalkoxy (PFA) are nearly transparent and have refractive indexes similar to many hydrogel formulations. From an optical perspective, Teflon is a good match, however it has a high Hounsfield unit for diagnostic energy x-rays, is relatively permeable to oxygen and is a soft plastic.
Also, extruded thin-walled tubes of PFA are optically active which will add uncertainty in polarization sensitive scanners geometries.

![Figure 2](image.png)

Figure 2. Schematic top view of selected optical ray paths through aquarium and gel. Refractive index liquid is less than gel dosimeter. Vessel wall is thin PET plastic cylinder. Translation of turning mirror and detector is coupled. Ray paths demonstrate: normal incidence (optical axis), refraction near wall (missing data), reflection, reference liquid path.

Generating ratio projection images requires that dosimeters can be reproducibly aligned with pixel resolution between the pre and post-irradiation scans. This is essentially a mechanical problem, how to mount the vessel reproducibly, but it may be possible to use image processing to perform subpixel alignments if required. Fiducial marks such as small dark ink spots of the vessel wall can be used for alignment as well as any sharp feature such as vessel edge or wall seam. Fiducial marks should provide adequate contrast for alignment but low enough contrast not to introduce streaking artifacts in the reconstruction. Optimal placement of fiducial marks may be problem specific.

As the acceptance angle of the scanner decreases, the optical quality of the vessel becomes more important. Stray light in certain scanner geometries essentially masked the effects of scratches, see [16]. By increasing the diameter of the source many optical artifacts such as missing primary rays near the vessel wall can be minimized. Many scanners have some degree of “source defocusing” in order to minimize high spatial frequency features such as scratches of the dosimeter or vessel surface. Laser beam scanners with large effective area detectors are insensitive to small variations in refractive index changes that can deviate the transmitted rays. A projection image of the vessel with a solution with absorption and scatter values similar to the intended hydrogel dosimeter will provide a measure of the optical quality of the entire system.

PRESAGE® is generally scanned with oil based refractive index matching fluid mixtures. These liquids are inconvenient to handle because of high viscosity and ability to interact with plastics. In order to reduce the volume of liquid a hybrid aquarium has been evaluated by [12]. In this approach, a matching solid aquarium is required for each dosimeter size. It is possible to scan PRESAGE® with low viscosity liquids than have much lower refractive indexes with specific scanner geometries that have large effective area detector and calibration of ray paths through the dosimeters [13].

2.6 Scanning strategies
Ratios of projections acquired pre and post irradiation [18], or ratios of dual wavelength post irradiation [19] are helpful for minimizing effects of optical imperfections such as scratches on the dosimeter surface which make each dosimeter optically unique (at this time). Pre and post irradiation scanning can introduce artifacts if the dosimeter is not repositioned with sub pixel accuracy [20]. Dosimeter transfers can introduce dust and debris into the aquarium and Langmuir-Blodgett films get deposited on the
dosimeter and aquarium windows due to the changing fluid levels. These sources of noise degrade the SNR of the 3D reconstruction. Post processing of projections is another approach filtering features related to bubbles and debris suspended in the liquid.

Dual wavelength scanning may be an effective alternative to pre and post-irradiation scans since the dosimeter is positioned only once in the scanner. This approach will be more convenient if comparable quality reconstructions can be obtained. Scanner systems that allow rapid wavelength switching may have a practical advantage. However, dual wavelength scanning may compromise the magnitude of the signal depending on the specific spectrum of the dosimeter and wavelengths available for scanning. If spectral scanning proves to be a significant advantage, then tunable sources could be introduced for optical CT scanning.

2.7 Data corrections
Optical aberrations result in non-uniform spacing of pixels in projection images. The most common aberration is due to refraction from non-matched refractive indexes, see vessel wall discussion above.

Examples of ray position correction have been reported by several authors [10, 12, 13, 20]. Scattered light in hydrogels is a common problem especially with broad beam geometries and small fields [17, 21-24]. In general a stray-light image for each transmission image should be measured and subtracted since they are coupled. The character of the stray light will be system dependent. But the major components are scatter within the dosimeter, reflections and refractions from the vessel and each optical interface in the system. This is most important for ‘broad-beam’ geometries and possibly a minor correction for ‘single-ray’ scanning geometries. Comparison of scan speeds should include time for additional scatter measurement scans if required. For example, if 3 scans are required for broad beam scanning with a structured illumination grid [25] then a slower single ray scanning approach may be equivalent [26]. Well-designed LED sources should be stable to 0.1% and laser sources to ~3% over typical scan times of one hour. Correction for source stability generally involves normalizing to transmission in liquid outside the dosimeter. Depending on system design there may be a settling time associated with switching wavelengths. Spatial filtering of projection images to remove spurious features such as scratches may be necessary. However, blemish-free optics and clean liquids are essential for low artifact images and maximum SNR of reconstructed images.

Some dosimeters, for example ferrous xylanol orange gelatin hydrogel, will darken with time at room temperature. This drift in absorption must be subtracted in order to calculate the dose dependent change. While other dosimeters may slowly fade following irradiation. Dosimeter specific thermal effects need to be measured and appropriate corrections applied. In general, fast responding dosimeters and fast 3D scanning, for example less than one hour can minimize magnitude of post irradiation corrections.

2.8 Performance tests
Uniform solutions that have refractive indexes, scatter and absorption similar to the dosimeter are important for assessing field of view and linearity. The uniform solutions allow testing of impact of missing data from near the vessel wall. Measuring small fields is another challenging geometry for optical CT. Thin-walled PFA Teflon tubes filled with absorbing aqueous solutions are a substitute for radiochromic hydrogels or custom “finger phantoms” [27]. If these uniform liquids and tubes can be accurately reconstructed then less challenging distributions such as an arc delivery can also be scanned. The scatter and absorption coefficients need to span the range that will be sampled by the irradiated dosimeter to allow interpolation. Spatial resolution QA phantoms have been assessed for 3D optical CT with PRESAGE® [28, 29].

2.9 Reconstruction algorithms
Incomplete projections are the most common problem in optical CT. The missing data generally occurs at grazing angles near the wall of the dosimeter or vessel. Refractive index optimization and thin walled vessels provide the maximum data. Iterative reconstruction algorithms generally result in lower
reconstruction artifacts compared to filtered back projection. Typically threefold improvements of signal
to noise ratio are observed. Graphical processor units (GPU) allow parallel processing of iterative
reconstructions of 512³ arrays in less than 30 minutes. Researchers can evaluate data sets with GPU
enabled open access iterative reconstruction packages such as ASTRA [30] and TIGRE [31].
Implementation of iterative reconstruction has been reported specifically for optical cbct [32] and
polymer gel dosimeter [33].

2.10 Dynamic range
Before irradiating the dosimeter, a calculation needs to be performed. The minimum transmission of
unknown dose distribution based is estimated from the calculated dose distribution and dosimeter
sensitivity. A suggested minimum transmission of 10% for broad beam scanners and 2% for single ray
scanners should be adequate for most systems.
Dynamic range can be increased by measuring and subtracting scatter components. In general,
is this a 3D problem coupled to the absorbing 3D dose image. This means a scatter image for each
projection is required [8]. Structured illumination using a spatially modulated grid and acquiring
multiple images is one approach to obtaining scatter subtracted broad beam images [25]. Dual or
multiple wavelength scanning is another method to correct for scatter. Finally, single ray scanning will
likely provide the lowest scatter projections since only the ray being sampled is illuminated.

3. Basic principles of X-ray CT gel dosimetry
The principle of utilizing x-ray Computed Tomography (CT) for imaging irradiated polymer gel
dosimeters was introduced at the first Dosgel meeting in 1999 and in a subsequent paper by Hilts et al [34].
The fundamental principles revolve around the fact that irradiated polymer gel dosimeters
exhibit a small physical density change upon irradiation with ionizing radiation from clinical linear
accelerator beams. As the density change occurring in irradiated gel dosimeters is small, on the order of
1%, CT polymer gel dosimetry is a low contrast imaging modality and, as such, much effort in CT
polymer gel dosimetry have been in the area of contrast resolution improvements. Comprehensive
review of the fundamentals of x-ray CT gel dosimetry have been published in the past [35, 36]. In this
short overview we highlight some of the main points for consideration in performing CT polymer gel
dosimetry.

3.1 Formulations of 3D dosimeters for use with x-ray CT
A primary limitation in CT polymer gel dosimetry is the small density change in irradiated dosimeters.
To this end, a significant effort has been expended in developing dosimeters with sufficient density
change per Gray (Gy) of irradiation so as to render the technique viable. Early investigations in CT gel
dosimetry (e.g. [34, 37]) worked with the standard acrylamide-based dosimeters (termed BANG
dosimeters) of Maryanski et al [38] and were based on a formulation consisting of acrylamide (3%), N-
N'-methelynebisacrylamide (3%), gelatin (5%), and water as remainder. However, three primary factors
propelled the further development of dosimeter formulations:

- the dose resolution of the BANG-based system was low (0.78 H/Gy, [33] and ref therein);
- the drive to simpler manufacturing protocols, specifically, the advent of “normoxic” formulations
  [39], or, dosimeters that could be manufactured on a bench top in normal atmospheric conditions
  as opposed to within a sealed and nitrogen-filled glove box;
- the toxicity of the BANG systems was moving research interest to consider formulations with
  reduced toxicity.

Unfortunately, initial moves from BANG dosimeters to normoxic formulations and to formulations with
reduced toxicity generally, with few exceptions, lowered the dose resolution in the overall system [37-
40-42]. However, Chain et al [43] reported on a formulation of normoxic, reduced toxicity dosimeter
with reasonable dose resolution (~0.8 - 1 H / Gy) and it is, largely, variants of this formulation that are
in use in current CT gel dosimetry. The formulation (termed NIPAM) consists of N-isopropylacrylamide (5-7% by weight), N-N’-methylenbisacrylamide (5-7%), gelatin (3-5%), tetrakis-hydroxymethyl-phosphonium chloride (THPC, antioxidant, 5-10mM), and deionized water as remainder. Figure 1 illustrates the progression of NIPAM-based formulation development.

![Figure 3](image-url) Dose response of NIPAM-based polymer gel dosimeter, as measured with x-ray CT. Shown is the effect of dosimeter composition on resultant dose response. From Chain *et al* [43].

3.2 Dosimeter containers

Not much work has gone into the specific investigation of dosimeter containers for CT polymer gel dosimetry. However, unlike in the case of optical CT imaging, the dosimeters vessel is, in most situations, of secondary importance. Most common dosimeter vessels are largely transparent to CT, however high density materials should be avoided as they can cause artifacts that are difficult to adequately remove [44]. One advantage is the general need for cylindrical symmetry that exists in optical CT is lifted in x-ray CT polymer gel dosimetry and an example of anthropomorphic phantom geometry can be found in the literature [45].

3.3 Scanning strategies

Concomitant with gel formulation development, the parameters used in CT imaging can affect image quality. While this is not a new finding (standard CT imaging theory applies here), the effect of imaging parameters on CT gel dose resolution have been studied [44, 46] and there exist robust recommendations for imaging protocol for CT gel dosimetry. Table 1 outlines the main parameters for consideration in CT imaging of polymer gel dosimeters and they can impact resultant dose measurements.

An optimal choice of imaging parameters is difficult to recommend as this will depend on the specific needs of a given application, considered balancing of noise and spatial resolution needs. For example, to improve contrast resolution it may make sense to scan with a slice thickness that matches the treatment plan you are intending to verify (e.g. 2 mm) rather than choosing the smallest available slice thickness (e.g. 1 mm) which would maximize spatial resolution. That said, typical scan values are listed in Table 1.

Recent work has investigated use of multi-slice CT scanners for CT gel dosimetry [47, 48]. This is a promising advance as multi-slice imaging has proven straightforward to implement for CT gel dosimetry and offers faster imaging time as well as reduced load on x-ray tube. As such, multi-slice imaging, when available, is recommended for CT gel dosimetry. An initial investigation has also looked at cone-beam CT for gel dosimetry [49]. While the dose response observed in this study was significantly poorer than that observed with conventional CT (less than half the sensitivity) this remains an interesting avenue to explore, particularly with any advances in CBCT technology.
Table 1: Image acquisition parameters for consideration in x-ray CT polymer gel dosimetry.

| Parameter          | Impact on Dose Measurement                                                                 | Typical Values |
|--------------------|---------------------------------------------------------------------------------------------|----------------|
| Tube current (mAs) | Increasing mAs decreases image noise and improves dose resolution. However tube overheating is a limiting factor. | 200 mAs        |
| Tube voltage (kVp) | Increasing kV decreases image noise and improves dose resolution. This outweighs any potential increase in sensitivity at low kV as no significant impact has been observed. | 120 or 140kV   |
| Scan time (s)      | Increasing scan time decreases image noise (combined with mA, above). Minimize to reduce scan time. | 1 s            |
| Slice thickness (mm)| Increasing slice thickness decreases image noise and improves dose resolution, but it reduces spatial resolution in “z”. | 2 - 5 mm       |
| Reconstruction     | Algorithms are designed for various options e.g. to smooth or enhance detail and can affect image noise dramatically. | Standard, Smooth |

3.4 Image reconstruction

One area that has received little attention is the effect of image reconstruction technique on resultant gel image quality, beyond a comparison of commercially available reconstructions on a single scanner system [44]. While image reconstruction analysis is a well-documented area of interest (e.g. [50]), little application has been made to the specific case of CT polymer gel dosimeters. In addition to the investigation of image reconstruction filters [51], the primary open area of investigation is in the area of iterative reconstruction techniques. While much effort has gone into the investigation of iterative CT reconstruction, scant literature exists in this area when concentrating on the specific case of polymer gel dosimetry. One such example [52], taken from the optical CT arena, suggests that further investigation in this area may be warranted for x-ray CT. A common issue in iterative CT reconstruction research for CT polymer gel dosimetry is the need for sinogram data output from CT machines. Typically, sinogram data is not provided as standard output on commercial scanner, thus impeding the progress of this line of investigation.

3.5 Image processing

3.5.1 Image averaging and background subtraction

Due to the low contrast images produced in CT polymer gel dosimetry, much work has gone into optimizing post-acquisition image processing. Early efforts [34] established a simple yet effective image averaging and background subtraction technique that is still employed in current research. In summary, recommended imaging protocols include:

- acquire 16 - 25 acquisitions / slice to create averaged images of irradiated dosimeter;
- acquire 16 - 25 acquisitions / slice to create averaged images of background (non-irr) dosimeter;
- subtract averaged background image set from averaged irradiated dosimeter image set.

Figure 2 illustrates the overall efficacy of the protocol.

3.5.2 Image filtering

Significant effort has gone into understanding the need and efficacy of image filtering in CT polymer gel dosimetry. Early efforts [51] involved studying a host of static (e.g. mean filter) and adaptive (e.g. adaptive mean) filtering techniques on resultant image noise reduction and concomitant spatial resolution degradation. Adaptive mean is easy to implement and a detailed investigation showed its efficacy for CT gel dosimetry over a range of dose distributions [53].

As with most instances of image filter analysis, summary recommendations possess a certain qualitative aspect, and hard recommendations are difficult to ascertain, primarily due to the fact that the
tradeoff between signal to noise ratio improvements and spatial resolution degradation are not universal across applications. Image signal to noise ratio and gel irradiation pattern, including the presence of steep dose gradients within the irradiation, all affect the choice and degree of filtering. Given the preceding significant caveat, we can, however, make a general conclusion that over most “typical” scenarios, an Adaptive Mean filter of 3x3 or 5x5 kernel size and two iterations has been shown to perform well over a range of dose gradients and irradiation patterns.

Figure 4. (a) Single CT image of irradiated polymer gel. (b) Image of averaged image of irradiated polymer gel. 25 averages acquired. (c) Averaged and background subtracted image. Numbers within image indicate beam dose (in Gy).

3.5.3 Remnant artefact removal
Even with reasonably heavy filtering it has been shown [54] that persistent, non-stochastic artefacts can remain in filtered and subtracted gel dosimetry images. These “remnant artefact” have been attributed to macroscopic gel structure and subtends local spatial ranges of several image pixels (i.e. mm range). Efforts to mitigate the image degradation from these remnant artefacts have involved the development of a “remnant artefact removal” algorithm. The algorithm is based on a signal removal set of algorithms that involve the selection of three fundamental parameters:

- filter span: the size of the window;
- filter degree: the degree of the fitting polynomial;
- filter iterations: the number of iterations to run the filter over the image.

As with the above comments on filter recommendations, hard recommendations on filter span, degree, and iteration are difficult. In general, the filter span should be on the order of the size of the macroscopic structure to be removed from the image. The polynomial degree is typically low (3-5), and the number of iterations typically run between 10-30. Higher iterations produce diminishing returns at the expense of increased computational time.

3.5.4 Zero scan method
A novel CT imaging method has been proposed for gel dosimetry, known as a “zero scan method”, in efforts to account for any dose resulting from CT imaging [55]. In its basic form, the zero-scan method monitors the pixel intensity within each mage pixel as images are acquired on a given slice. A linear function is fit to each set of pixel values, and a “zero scan” pixel intensity is extrapolated from the resultant fit to represent the “true” pixel intensity prior to CT imaging. The method has been shown to improve overall image noise, however, the effects of the zero scan method within high dose gradient regions remains an area of further investigation. A recent work has extended the zero-scan concept to multi-slice imaging [48].

3.6 CT Imaging Dose
Unlike with optical CT or MRI, in x-ray CT read-out of polymer gels ionizing radiation dose will be deposited into the gel as a result of the imaging process [56]. The effect of this imaging dose has been
observed and, in a novel study, an active polymer gel was used to measure the CTDI for a CT scanner [41, 57, 58]. However, the response of normoxic polymer gel dosimeters to kV photon energies, as are emitted in CT scanning, has been shown to be less than half that of MV irradiations and the additional CT contrast resulting from a typical imaging process, incorporating 16 to 32 image averages, is very low, < 0.2H [59]. Further, this effect was measured for fully active gels which were imaged a short time following manufacture. In reality, it is common practice in CT gel dosimetry to expose gels to atmospheric oxygen following polymerization and prior to irradiation which would further reduce the impact of CT dose on read-out by, at least partially, rendering the gel inactive.

3.7 Summary
In summary, the primary considerations that should be evaluated prior to undertaking a CT polymer gel dosimetry experiment include:

- choice of dosimeter;
- smart design of container;
- single or multi-slice CT;
- CT image acquisition parameters;
- image filtering protocol;
- remnant artefact removal protocol, if employed;
- decision on use of zero-scan method.

Suggested areas of further investigation in CT gel dosimetry include: iterative reconstruction techniques, the zero scan method, and development of clinical applications.

4. Conclusion
CT scanning of 3D dosimeters is a practical approaching to achieving accurate dosimetry. Scatter within the dosimeter is often the limiting factor for dynamic range. It is a bigger problem for broad beam geometries, small fields and for larger dosimeters. If the dynamic range is inadequate for the problem of interest then scatter must be measured and subtracted from projection data. Prefiltering of transmission images will reduce reconstruction artefacts associated with the vessel, such as scratches. Iterative reconstruction algorithms will provide higher SNR compared to filtered backprojection.

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