The history and principles of optical computed tomography for scanning 3-D radiation dosimeters: 2008 update

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Abstract. The current status of optical CT for 3-D radiation dosimetry is reviewed. The technique is first placed in its historical context, pointing out the relationship with other methods of optical imaging and showing how optical-CT has emerged independently in several different fields and under different names. The theoretical background of the method is described briefly and this provides the foundation for an explanation of the different types of scanner. The relative advantages and disadvantages of instruments based on scanned lasers and pixelated (area) detectors are presented. The latest generation of “fast laser scanners” is described and the review is concluded with a discussion of the different radiation-sensitive materials used as samples in optical CT.

1. Introduction
It is now more than a decade since the first optical computed tomography (CT) data for a polymer gel radiation detector were published [1, 2]. Since that date, there have been considerable advances in both 3-D imaging techniques and dosimeter formulation. Developments have continued apace since the last reviews of the area in 2006 [3, 4] and this paper is an updated version of [3]. Whilst it is not possible to discuss fully all the latest contributions in the limited space available, the reference list has been updated with the intention that it can provide a definitive record of the activity in the area.

Of course, the roots of 3-D imaging via optical CT may be traced back much further than 1996. The first part of this review will place the current work in its historical context and also explain how it is related to other methods of optical imaging. An expanded theoretical description of the principles of optical CT will then be given, which will allow the reader to appreciate the differences between the various types of scanner currently being developed and also to give an idea of the different types of data processing that are employed. The article will discuss in detail the relative merits of scanners based round pixelated (area) detectors, compared with those using a single laser beam, and will compare the latest crop of so-called “fast laser scanners”. Central to our ability to image dose distributions has been the development of novel materials whose optical properties change in response to radiation, and a survey of these will be given. Finally, I will speculate briefly on the future of the technique.
2. General overview

Although 3-D radiation dosimetry using magnetic resonance imaging (MRI) was already well established by the time of Gore and Maryanski’s seminal 1996 paper [1], the idea of detecting radiation changes optically has, in fact, a much longer history than the use of MRI. Chemical dosimeters date back at least as far as 1927 [5] and, at the outset, used UV spectroscopy as the readout method. Two factors contributed to the relatively late appearance of optical CT on the scene for gel dosimetry. Firstly, whilst MRI (like X-ray CT) was developed primarily for medical applications and then applied as a mature imaging modality to 3-D radiation dosimetry, there was no parallel medical development of optical CT, because the human body is opaque at visible wavelengths. Secondly, it has been only since the late 1990’s that the technology needed for such measurements has been available. The consumer imaging boom of the past decade has given the impetus for the production of high quality, cheap CCD’s; high performance photodiode detectors and receiver electronics have also become readily accessible; and sophisticated image reconstruction algorithms have been imported from other branches of medical imaging. It may be argued that as soon as the conditions were right, the technique was “discovered”.

Figure 1: Examples of work from the three different fields in which optical CT emerged independently: (a) gel dosimetry, reproduced from Fig. 5 of [1] (b) 3-D microscopy, reproduced from Fig. 3 of [7]; (c) visualization of chemical waves, composite of Figs. 1, 4 and 5 of [6].
This is borne out by the fact that optical CT emerged independently in at least three completely different fields. Contemporaneously with Gore and Maryanski’s invention, Winfree presented [6] what appears to have been the first CCD-based optical-CT scanner, in an elegant experiment to investigate self-organising chemical structures. In 2002, Sharpe [7] published the first optical micro-CT images, and patented his approach, which he termed optical projection tomography (OPT), apparently unaware of the prior developments. More recently, optical CT has been re-introduced with the name optical transillumination tomography in studies of tissue-engineered blood vessels [8-11].

Optical CT is part of a family of different techniques of optical imaging. Each is applicable to only a certain range of samples and has different advantages and disadvantages.

Confocal microscopy [12], optical coherence tomography (OCT) [13] and diffuse optical tomography (DOT) [14] are techniques that are all commonly used to image biological tissues. Laser scanned confocal microscopy is capable of producing high-resolution (1µm in-plane and 5µm axially) images at video frame rates via a process known as optical sectioning, in which only the region of the object close to the focal plane of the lens system is imaged efficiently. Other 3-D microscopy techniques are reviewed by Steltzer [15], but not discussed further here. OCT is an interferometric technique that combines back-scattered light from tissue with a reference beam. A succession of high-resolution (potentially sub-micron) real-time, 2-D images may be reconstructed from different depths using methodology that is analogous to ultrasound pulse-echo imaging. Both confocal microscopy and OCT are limited (as used conventionally) to a maximum penetration depth of the order of a few mm by the highly scattering nature of tissue samples. However, given that 3-D radiation dosimeters scatter relatively weakly in comparison with tissue, it is probable that variants of OCT could be used successfully for dosimetry. DOT is used for larger (up to tens of cm) samples that are highly scattering. Using sophisticated mathematical techniques for solving what is a very complex inverse problem, both optical absorption and scattering coefficients may be reconstructed, though with a spatial resolution only of order 1 cm.

The configuration used for our parallel-beam optical CT scanner [16-19] has much in common with schlieren and shadowgraphy methods [20], which find application in areas as diverse as the optics of flames and particle velocimetry. The distinction of parallel-beam optical-CT from these techniques is that, while they aim specifically to image inhomogeneities in refractive index, we would rather avoid them.

3. Principles of computed tomography

3.1. Beer’s Law

The starting point for our discussion is Beer’s Law, which describes how light and X-rays are attenuated as they pass through a medium. In a uniform substance of linear attenuation coefficient μ, the light intensity, as measured by a detector placed at depth d is given by

\[ I(d) = I_0(d) \exp(-\mu d), \]

Note in passing that the applicability of Beer’s Law is something that needs to be formally demonstrated for each chemical system and is not a given [21]. To our knowledge the limits of applicability in 3-D dosimeters have not yet been investigated, but one may anticipate that, for the range that can be probed via CT with current technology (up to 14-bit ADC, i.e., integrated optical density up to ~4), then Beer’s Law will hold for absorbing media. However, for systems such as polymer gel dosimeters, where contrast is obtained via scattering, then Beer’s Law will “break down” much earlier.
where $I_0$ is the intensity measured at depth zero. Suppose we now consider a set of $N$ blocks of different material, each of width $\Delta y$, as shown in Figure 1(a). The X-ray intensity measured at the exit of the set of blocks is

$$I = I_0 \exp \left\{ - \sum_{i=1}^{N} \mu_i \Delta y \right\}.$$

For the limit $\Delta y \to 0$, $N \to \infty$, this becomes

$$I = I_0 \exp \left\{ - \int \mu(y) \, dy \right\}.$$

3.2. The Radon transform and sinograms

As shown in Figure 2(b), the laser-scanned CT apparatus (often described as a first-generation system) consists of a source and detector, placed on either side of the object to be imaged. These slide along in tandem. Consider the intensity of the attenuated laser beam received by the detector when the source-detector assembly is at position $x$:

$$I(x) = I_0 \exp \left\{ - \int \mu(x, y) \, dy \right\}.$$
where $\mu(x,y)$ is now the 2-D distribution of optical attenuation coefficient. $\mu$ is related to the optical density (OD)\(^2\) by the relation $\mu = \text{OD} \ln 10$. In a medical X-ray scanner, it would be normal for the source-detector track to rotate around the sample. However, a major design simplification where the object to be scanned is inanimate is to make the sample rotate instead. (This makes no difference to the principles involved, but the mathematical formalism below must be modified in a minor way from that seen in many textbooks by adding a negative sign to the rotation angle.)

As the sample rotates by a positive angle $\phi$ in Figure 1(c), the function describing the attenuation coefficient changes to $\mu'(x,y)$ by the following transformation of co-ordinate system:

$$\mu'(x',y') = \mu(R_x \phi, x', y'),$$

(5)

where $R_x \phi$ is the standard 2-D matrix for a rotation through angle $-\phi$, such that

$$x' = x \cos \phi + y \sin \phi$$

$$y' = -x \sin \phi + y \cos \phi.$$

(6)

To take a concrete example, suppose the sample has rotated through $+30^\circ$. A point now at $x = 1, y = 0$ and thus contributing to the $I_{30^\circ}$ profile at $x = 1$ used to be at $(\sqrt{3}/2, -0.5)$ at the start of the experiment.

The “profile” or “projection signal” when the sample has rotated through angle $\phi$ is

$$I_\phi(x) = I_0 \exp\left\{-\int \mu(x,y) \, dy\right\} = I_0 \exp\left\{-\int \mu(x',y') \, dy\right\},$$

(7)

We define the Radon transform — often the word “projection” is used interchangeably — as

$$P_\phi(x) = \int \mu(x',y') \, dy = -\ln\left(\frac{I_\phi(x)}{I_0}\right)$$

(8)

and create a new “space”, called Radon space, in much the same way as one defines the reciprocal Fourier domain in MRI\(^3\). As shown in Fig. 3, Radon space has two dimensions, $x$ and $\phi$, and, at the general point $(x, \phi)$, we “store” the result of the projection $P_\phi(x)$. Taking lots of projections at a complete range of $x$ and $\phi$ “fills” Radon space with data in much the same way that one fills Fourier space with 2-D MRI data, and these may be presented as a sinogram.

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\(^2\) Note that for an optical component such as a film or neutral density filter, we often talk about the optical density of the component as a whole and define OD as $-\log_{10} T$, where $T$ is the transmittance $I/I_0$. In that case, OD has no units. However, in this article, we use the definition found at [http://en.wikipedia.org/wiki/Optical_density](http://en.wikipedia.org/wiki/Optical_density): $\text{OD} = -(\log_{10} T) / \text{sample length}$, in which case OD does have units and these may be expressed in terms of cm\(^{-1}\) or mm\(^{-1}\) as appropriate. Note that this makes sense, because OD then becomes an intensive property of the sample ([http://en.wikipedia.org/wiki/Intensive_and_extensive_properties](http://en.wikipedia.org/wiki/Intensive_and_extensive_properties)), that is susceptible of being imaged on a voxel-by-voxel basis. This is exactly the same distinction as that between absorbed dose and total absorbed energy.

\(^3\) Strictly speaking, Radon space is not exactly like Fourier space in the sense that a Fourier transform may be exactly inverted to return to the original data — the data and its transform are simply two different representations of the same information. The back-projection reconstruction technique does not allow us to get back exactly to the original attenuation coefficient distribution.
Consider what the sinogram looks like for a sample consisting of a single point in real (image) space. For a given sample rotation angle $\phi$, all positions $x$ on the profile lead to $\lambda(x) = 0$, except the one coinciding with the point to which $(x_0, y_0)$ has been rotated. Thus, all points in the Radon space corresponding to the single-point object are zero, except along the track $x = x_0 \cos \phi - y_0 \sin \phi = R \cos(\phi + \phi_0)$ (9)

where $R = (x_0^2 + y_0^2)^{1/2}$ and $\phi_0 = \tan^{-1} (y_0/x_0)$. If we have a composite object, then the filled Radon space is simply the sum of all the individual points making up the object (i.e., multiple sinusoids, with different values of $R$ and $\phi_0$). See Figure 3 for an illustration of this.

### 3.3. Reconstruction via backprojection

The aim of optical CT is to obtain the optical attenuation of the sample (which is related to the absorbed dose) as a function of spatial position by acquiring a set of projections. In other words, given the complete set of projection data $P_{\phi}(x)$, i.e., a full Radon space, we wish to reconstruct the function $\mu(x,y)$. This is performed by a process known as back-projection (see Figure 4), which works as follows:

1. Consider one row of the sinogram, corresponding to angle $\phi$. Note how in Figure 4, the value of the Radon transform $P_{\phi}(x)$ is represented by the grey level of the pixel. When we look at a single row (i.e., a 1-D set of data), we can draw this as a graph.
2. Place the sinogram row an angle $\phi$ in real space. Then “smear it out” evenly all the way along the perpendicular direction. This is called back-projecting the data.

3. Repeat steps 1 and 2 for all the lines in the sinogram — see Figure 4. Where the back-projections overlap, the signal adds constructively to give high-intensity image regions. Try out the excellent “Reconstruction Demo” in the medical section of the IDL distribution (http://www.ittvis.com/idl/).

3.4. Mathematical description of filtered back-projection
This is not quite the whole story. It turns out that the image that is produced by this method is blurred. To get the right representation of the object, we need an additional mathematical “trick” called filtering. In order to understand what filtering is mathematically and how the complete filtered backprojection technique works, we must first see the relationship between the Fourier transform of the image and those of its projections.

For the simple case of an un-rotated object, the Fourier transform of the projection is:
If there were an extra factor \( \exp(-i k_y y) \) after the integral sign, then we would have exactly the expression for the 2-D Fourier transform of the image. Instead, we can imagine that there is an extra multiplying factor of 1 “hidden” in this equation. Furthermore, we know that 1 = \( \exp(0) = \exp(-i k_y y) \) whatever \( k_y \) might be. Hence, we can deduce the result that

\[
\tilde{P}_0(k_x) = \int_{-\infty}^{\infty} P_0(x) \exp(-i k_x x) dx = \int_{-\infty}^{\infty} \mu(x, y) dy \cdot \exp(-i k_x x) dx
\]  

(10)

What this means is that the 1-D Fourier transform of the projection at zero degrees is simply the central line of the 2-D Fourier transform of the image. This idea will be very familiar to any MRI scientists, for whom it is a well known fact that the Fourier transform of the zero phase-encode step of an MR scan gives the projection of the sample in the phase-encoding direction.

Since the co-ordinate system can be chosen arbitrarily, it follows intuitively — and can easily be proved formally — that an equivalent result applies for projections taken at any angle. The 2-D Fourier transform of \( \mu \), evaluated along the line at angle \( \phi \) in \( k \)-space is the same as the 1-D Fourier transform of the projection at angle \( \phi \).

This important result is known as the Fourier slice theorem and it is the basis of backprojection reconstruction. Let us now write the desired final image in terms of its Fourier transform:
\[
\mu(x, y) = \frac{1}{2\pi} \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \tilde{\mu}(k_x, k_y) \exp(+ik_x x + ik_y y) \, dk_x \, dk_y , \tag{12}
\]

It is very useful to write the expression on the right hand side of the equation in terms of a polar rather than Cartesian representation of k-space. Thus \(k_x = k \cos \phi\) and \(k_y = k \sin \phi\), yielding
\[
\mu(x, y) = \frac{1}{2\pi} \int_{0}^{2\pi} \int_{0}^{\infty} \tilde{\mu}(k, \phi) \exp(+ik \left[ x \cos \phi + y \sin \phi \right]) \, k \, dk \, d\phi . \tag{13}
\]

We note that the expression \(\tilde{\mu}(k, \phi)\) is the same as \(\tilde{P}_\phi(k)\), by the Fourier slice theorem, and that the expression inside the square brackets above is exactly what we wrote down earlier for \(x'\). So we end up with the following expression directly linking the measured projections and the image that we are trying to reconstruct:
\[
\mu(x, y) = \frac{1}{2\pi} \int_{0}^{2\pi} \left\{ \tilde{P}_\phi(k) \cdot \exp(+ikx') \right\} k \, dk \, d\phi . \tag{14}
\]

Notice that there is an \(x\) on the LHS and an \(x'\) on the right of Eq. (14). Going back to the explicit expression for \(x'\) in terms of \(x\) and \(y\) and looking at exactly which point on the profile contributes to the reconstructed \(\mu\) at position \((x, y)\), it should be apparent that this is the mathematical formulation of the back-projection operation itself. In this “continuous version” of the formula, to which our acquired data are a discrete approximation, the final image, \(\mu\), is made up from a differential contribution (i.e., weighted by \(d\phi\)) of the projection from each angle between zero and \(2\pi\).

Again, we see that the contents of the integral in the braces are almost a Fourier transform. Without the \(k\), the result of this integral would simply be the original projection \(P_\phi\). The \(k\) term modifies or “filters” the Fourier transform by giving more weight to the higher spatial frequencies. Our final formula for reconstructing images is thus:
\[
\mu(x, y) = \frac{1}{2\pi} \int_{0}^{2\pi} Q_\phi(x') \, d\phi , \tag{15}
\]

where \(Q_\phi\) is the filtered projection, which can be obtained from the measured data.

In order to arrive at a practical reconstruction algorithm, it is important to take account of a number of other considerations that are beyond the scope of this review. More detailed descriptions of the back-projection process may be found in [22] and [23]. Finally, it is worth pointing out that filtered back-projection is not the only method of image reconstruction. A family of methods based on the algebraic reconstruction technique (ART) is also available. These are potentially more flexible, allowing one to cater for various deficiencies in the raw projection data and different models for how the projections are formed\(^4\), but they are more time-consuming to run. See Ch. 7 of [22] for a general description and [8] for an application in optical CT.

\(^4\) Examples of the type of problems that might be solved in this way are reconstruction from a limited number of projections [8]; dealing with refractive index inhomogeneities in the sample; and accounting for the fact that projections from a CCD scanner are not correctly modeled by line integrals of the form shown in Eq. (4), but, rather, are the result of absorption over a “double cone”, as described in [17].
4. Optical CT in 3-D radiation dosimetry

4.1. First generation laser scanners
The first-generation scanner geometry illustrated in Fig. 2(b) was the one employed by Gore et al. [1]. Fig. 6(a) gives a more detailed schematic of their apparatus, which developed into what was marketed under the name OCTOPUS by MGS Inc., for a number of years the only commercial optical-CT scanner. At approximately the same time, Tarte et al. were scanning gel sections using a single laser beam [24] and, in the years that have followed, a number of other groups have published work of increasing sophistication using first-generation scanners [25-52].

Excellent reviews by Jordan [53-55], presented at previous DOSGEL conferences, have discussed in outline a number of the optical problems to be addressed when designing such systems. These include: minimisation of interference effects and stray light; scatter from optical components and the radiochromic gels themselves [56]; reflection; dynamic range; wavelength selection; wall corrections [57]; the plasma discharge from lasers; temperature changes; and the characterisation of detectors. Further detailed work on characterising a custom-built optical-CT laser system was performed by Oldham et al. [36, 39].
First generation scanners tend to produce high quality images and, in the absence of a systematic comparison of all the different scanner types on standard samples, the OCTOPUS scanner has, on occasion, been referred to as a “gold standard” [52].

However, a major disadvantage of first-generation systems is their slow scanning speed. In order to obtain a profile, the laser beam and photodiode must be stepped mechanically in parallel across the sample. A typical performance figure for the original OCTOPUS system is found in [35], where slices of 128 × 128 pixels were acquired at a rate of 12 minutes per slice. To obtain multiple slices, the sample was raised or lowered mechanically an appropriate distance. True-3D scans, with isotropic high resolution and a large field-of-view in the slice direction are not feasible using this methodology and even images with fairly modest numbers of slices take many hours. More recently, improved imaging times of order 5 minutes per slice have been demonstrated using an improved version of OCTOPUS scanner [48], but there is a limit as to how far this technology can be pushed and a full 3-D scan still took of order 16 hours.

4.2. Scanners based on pixelated (area) detectors

4.2.1. Parallel- and cone-beam geometries. A faster alternative is the use of scanners based on CCD or CMOS area detectors. These chips are now extremely widespread because of their use in digital cameras. Whereas laser systems acquire data in a point-by-point fashion, imaging detectors allow us to obtain a complete 2-D projection in one go — see Figure 2 of [53]. Each 2-D projection gives the data required for creating a row in the sinogram for every slice in a 3-D reconstruction. A modern scientific CCD camera will have a matrix size of typically 1000 × 1000 pixels and so it is easy to see that speed gains of more than two orders of magnitude might result if 3-D data are needed. (This is of course a big “if”, since large numbers of slices and an ultra-high spatial resolution are not always required.) In practice, the speed when using a CCD-based system is often limited by the data-throughput rate. Particular bottlenecks can occur in transferring data off the CCD chip to the camera and then in the transfer process out of the camera to the host computer.

Two sub-classes of CCD scanner have so far been presented in the gel dosimetry literature: parallel-beam and cone-beam.

Our group has developed scanners based on the parallel-beam geometry [16-19, 58, 59], as illustrated in Fig. 6(c). Although conceived entirely independently, these are similar to the early design of Winfree [6], which pre-dates them. A key element of the parallel-beam design is the use of telecentric optics and this can be achieved using either a combination of a large converging lens close to the matching tank and a standard camera lens [17] or a specialist, but more expensive, telecentric lens [52]. The parallel-beam scanner created by Oldham et al. is a dual-purpose instrument, capable of imaging both in absorption and emission modes and it can be applied equally to the areas of radiotherapy dosimetry and of biological microscopy [60-63], as popularized by Sharpe [7, 64-66].

The cone-beam geometry (Fig. 6(b)) was first introduced by Wolodzko et al. [67] and later pursued further by Jordan et al. [68, 69]. A commercial scanner is now marketed by Modus Medical Devices Inc. under the trade name Vista™. Since its release in 2004, a number of preliminary evaluations of this product have been made [70-74].

To the best of this author’s knowledge, there are at present no comparative data in the literature to determine whether either of the two geometries is more advantageous. However, several practical comparisons may be drawn. (i) The cone-beam design is generally cheaper and is easily scaleable [69]. It has the advantage of being able to scan larger objects without recourse to expensive optical
components. However, whilst large optics are expensive, they still represent only a relatively small fraction of the typical purchase price of a scanner. (ii) The cone-beam scanner is more compact. (iii) The parallel-beam scanner uses less light power (single LED vs. 2-D array of LED’s behind a diffuser), suggesting that it may be easier to guard against stray light. (iv) The original parallel-beam scanner [18] was unsuitable for imaging scattering dosimeters, but the most recent design [17] has a more controllable and, potentially, better scatter rejection than the cone-beam design.

It should also be noted that a considerable body of work has also been performed using so-called “stacked-gel” detectors, together with 2-D CCD imaging, rather than a full 3-D readout [75-84].

4.2.2. Particular considerations for the use of pixelated detectors
(i) A general disadvantage of scanners based on pixelated detectors together with a wide beam is the possible introduction of artefacts by refractive index inhomogeneities (schlieren). Such variations in refractive index result, typically, from incomplete mixing of the various components of either the dosimeter itself or the matching liquid, or from local temperature gradients, which lead to convection currents. If these inhomogeneities are present in the dosimeter, as it cools during the manufacturing process, then they are “frozen in”. By contrast, the effects in the matching liquid can be minimised by leaving the liquid for some time after inserting the dosimeter, so that settling occurs.

Figure 7: (a) Illustration of the principle behind the generation of schlieren effects; (b) an extreme example of schlieren and other effects that degrade the quality of projection images. The dark region in the centre is a brachytherapy irradiation; (c) an extreme example of the problems schlieren can cause in the final reconstructed image and profile.
As light rays pass through the schlieren, they are deviated and the effect on the projections is two-fold. Firstly, in some cases, this causes several rays to overlap, leading to bright regions of the image, whilst the regions away from which the light has been refracted are dark. This is precisely the mechanism exploited in shadowgraphy [20]. Fig. 7(a) illustrates the phenomenon schematically, whilst Fig. 7(b) is an extreme example that we observed. The consequence for the final image is difficult to predict \textit{a priori}, and manifests itself as a structured, but complex artifact, as in Fig. 7(c).

The second effect is more subtle: light rays no longer travel in straight lines through the sample and this means that Eq. (4), on which the whole imaging experiment is predicated, is invalid. The only work to have been performed in investigating the implications of this source of systematic error appears to be that of Oldham and Kim [39], who observed no deviation in the measured positions for their needle phantom in the presence of radiation-induced refractive index inhomogeneities. However, the importance or otherwise of this effect for quantitative dose images has yet to be assessed and, since Oldham and Kim’s work used a laser scanner, further work is needed to clarify the effects for area scanners.

(ii) \textit{Ring artefacts} are generated when a feature not associated with the dosimeter sample is present in the same place in all projections. A typical cause might be a bubble or scratch on the wall of the tank containing the matching liquid. Whilst such artefacts have been studied both in the context of laser scanning [39] and CCD scanning [18], an additional problem for CCD scanners, particularly those using low-end chips is a non-uniform pixel response to incident light and/or isolated “dead” pixels. Additional calibration or other steps must be taken. In our case, purchase of a high-quality CCD (the ORCA BT-1024G from Hamamatsu) has largely eliminated this problem, although cleaning of the scanning tank must be very thorough. Recently, an algorithm has been developed in the context of X-ray CT [85] to ameliorate any residual image artefacts.

(iii) \textit{Calibration} for a possible non-linear response of the CCD chip is necessary to ensure that the correct ratio $I / I_0$ is calculated in Eq. [8]. Some CCD’s have a non-linear response. It is also necessary to subtract the value of the CCD “dark current” from pixel values in both the light-field and dosimeter images [86]. An allowance needs to be made for the transmission properties of the lens systems and the quality of the beam. This may be done conveniently by acquiring a “light-field” image prior to insertion of the sample. Alternatively, a complete dataset may be acquired from the dosimeter prior to irradiation. This has two important advantages: firstly, the reconstructed image contains only the changes induced by the radiation and no background; and secondly, effects of residual refractive index mismatches at the walls are greatly reduced, leading to a much larger useable volume of dosimeter.

(iv) \textit{Dynamic range}: A potential concern for the design of any optical CT system to be used for accurate metrology is the range of optical densities that may successfully be reconstructed and the artefacts that will be generated if the integrated optical density across the sample exceeds this. In many cases, the dynamic range of the photodiode detector used in laser scanning systems is superior to that of a CCD, which might be as low as 8-bit. Given the presence of the CCD dark-current, it may well be difficult to image successfully optical densities greater than 2 (i.e., $I / I_0 = 0.01$). This question is discussed in depth in [86]. The ORCA BT-1024G has a dynamic range of approximately 20000:1 and has a 14-bit ADC and thus allow us to overcome these problems.

(v) \textit{Image signal-to-noise ratio and accuracy}: When describing the performance of an imaging CCD, manufacturers often quote the \textit{full-well capacity} (i.e., how many electrons a pixel on the chip can hold) and the \textit{RMS readout noise} in electron units. The dynamic range of the camera is defined as the ratio of these two numbers. For our camera, these figures are 80,000 and 4, giving a dynamic range of 20,000 : 1. However, it is important to realise that this is not the same as the image signal-to-noise ratio (SNR). In an excellent didactic article [87], Healey and Kondepudy describe the roles of shot
(vi) **Type of sensor:** Currently, CCD’s are superior to CMOS detectors in terms of noise performance and dark current. However, future developments of pixelated detectors are likely to favour CMOS devices, which will be quicker and cheaper, with the capability of detecting greater OD dynamic range and better resolution. CMOS is a standard lithographic technique for microchips, so adding more logic for each pixel is easy. This in turn enables dual mode operation (linear or logarithmic). Our focus should be on the logarithmic mode, since it should allow us to obtain a more uniform OD resolution across the full OD dynamic range. Extending the dynamic range of CCD’s is possible by programming multiple acquisitions with different exposure times [86], but this is in many cases time consuming and runs the risk of images being corrupted by *blooming* [88].

4.3. **Fast laser scanners**

For some while, research has been ongoing to try and create a novel design of scanner that combines the advantages of a single-beam laser scanner with the speed of a CCD scanner. A first step was taken by Maryanski [89], with a design the contained two novel features. Firstly, the laser was reflected by a rotating mirror, which allowed the beam to be scanned very rapidly across the sample in a single
dimension. Secondly, there was no matching tank and the sample was used to refract the beam in order to obtain a parallel set of beams through the sample. Both of these features are present in the latest scanner from MGS, the IQ-SCAN. Van Doorn’s group [90] and Conklin et al. [91] have also demonstrated the feasibility of a rotating mirror approach. Two of these scanners are illustrated in Figure 8. Perhaps the most sophisticated fast laser scanner built so far was introduced by Krstajic and Doran [92] and makes use of ideas drawn from confocal microscopy. Instead of using simple rotating mirrors, this instrument (described in detail in [92] and characterized further in [93]) manipulates the laser beam using a pair of galvanometer controlled mirrors. This allows the beam to be scanned in 2-D, with the potential for control of the beam using adaptive optics to reduce errors. It is well known in the microscopy and OCT communities that this type of arrangement can be used to output images at video frame rates, leading to the potential (not yet realized in 3-D dosimetry) of 3-D scans in under a minute.

5. Dosimeter materials
As described earlier, chemical dosimetry was initially based around so-called Fricke solution, which absorbs in the UV (304 nm). Pioneering work investigating how to shift this absorption into the visible range was performed by Gupta, who authored some 20 papers on the ferrous xylene-orange system (e.g., [94]). More recently, since the advent of optical CT in 3-D radiation dosimetry, a large number of other authors have investigated different properties of ferrous xylene-orange gels (FXG) [58, 84, 95-104]. FXG is easy to manufacture and its performance is not hindered by the presence of atmospheric oxygen (in fact, this positively helps). However, its major disadvantage is the mobility post-irradiation of the ferric ion complex. Diffusion causes an unacceptable blurring of the dose pattern within a few hours of irradiation and this has led many groups to move away from Fricke gels to other chemical systems. Nevertheless, Babic et al. have demonstrated that, once the problems involved are well understood and controlled, then excellent results may be obtained [46, 105].

Polymer gels were developed to overcome the diffusion problem. Initially developed for MRI-based dosimetry, their potential in optical CT studies was obvious from early on [2]. Whilst contrast in FXG is developed by means of absorption — the orange gel turns purple on exposure; see [18] for a typical spectrum — polymer gels attenuate light by scattering and hence change from clear to opaque white. Several polymer gel formulations have been studied or developed specifically for optical CT [106, 107] and a variety of different applications have been pursued [35, 36, 39, 48, 49, 80, 89, 108-111]. A disadvantage of some polymer gels is their sensitivity to atmospheric oxygen.

Without doubt, one of the most exciting developments during recent years has been the introduction [112] of PRESAGE™, a novel transparent plastic dosimeter. The material is rigid and easily machinable. It is insensitive to oxygen; moreover, as it is not a gel, PRESAGE™ does not dry out. Hence, one can remove the container and, with it, two optical interaces. This makes modelling of the optical path through the dosimeter [18] much simpler. PRESAGE™ is stable during a prolonged irradiation period, and has good post-irradiation storage properties. It has a linear response at low energies and over an exceptionally wide range of doses [113]. Not surprisingly, the last four years have seen a flurry of interest in this dosimeter [16, 17, 42, 44, 47, 50-52, 59, 92, 93, 114-123].

6. Future perspectives and conclusions
Optical CT in 3-D radiation dosimetry has come a long way over the last 12 years. As pointed out in the introduction, the competing readout systems, MRI and X-ray CT, were already mature imaging modalities when first applied to dosimetry. Nevertheless, the capability of optical CT instruments is now approaching and, in areas such as image resolution and acquisition speed, exceeding that of MRI. Naturally, there is less experience with the technique and a number of the most promising dosimeter
materials are only a few years old. Their formulations are still changing, with key properties not yet measured. Thus, it would probably be fair to say that the state-of-the-art in optical dosimetry is where MRI was 3–5 years ago. We still have to undertake much more painstaking verification work, investigation of anomalous results and the pursuit of absolute dosimetry, topics which have characterised the last two decades of work with MRI and continue to do so.

Future work needs to address one issue above all. The optical CT workflow needs to be optimized. Imaging needs to become more routine and less of a specialised “research” activity. The process needs to be “less messy”, no longer requiring users to spending time worrying about issues such as refractive index matching of their samples. There needs to be better quality assurance built into the scanners. File formats need to be standardised. Finally, with many different designs of scanner available, there need to be more test objects and widely agreed standards for how to compare systems.

Two years ago, I wrote the following confident statement: Optical CT scanners cost between one and two orders of magnitude less than an MRI system and this should ensure that the community of researchers will grow rapidly during the next five years. This hope has certainly been fulfilled by this year’s crop of DOSGEL abstracts. At DOSGEL 2008, there were 21 abstracts with an optical theme, presented by 7 different groups. This year, the programme features 27 such submissions, from 15 groups, with an increased diversity of applications.

I ended the previous review with the words: It is possibly premature to sound the death knell of MRI, but we are catching up fast! A recent, preliminary analysis of the published scientific literature for the period 2006 – 2008 yielded 78 full papers on gel dosimetry. Of these 34 used MRI as the readout modality and 34 used optical methods. Other methods, including X-ray CT and ultrasound, accounted for the other 10.

Optical CT has now arrived!

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