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

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1. Introduction
It is now a decade since the first optical computed tomography (CT) images of 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 and it is therefore timely to produce a thorough review of the field. Whilst it should be recognised that, in the limited space of an extended conference abstract, it is not possible to discuss fully all the contributions to the subject, a comprehensive reference list has nevertheless been assembled and the authors hope that this will provide a valuable resource for the 3-D dosimetry community.

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 elementary 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. A separate article [3] will discuss in more detail the class of optical-CT scanner based on a translating laser beam, whilst, here, we will describe the particular considerations relating to ultra-rapid, true-3D scanners based on charge-coupled device (CCD) or complementary metal-oxide semiconductor (CMOS) imaging detectors. 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, we will speculate briefly on the future of the technique.

2. Historical 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 [4] and have, historically, used UV spectroscopy as the readout method. Two factors contributed to the relatively late appearance of optical CT on the scene. 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 advanced to the extent of having the required performance at an appropriate cost. As soon as the equipment became available, the technique was “discovered”.

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 discovery, Winfree presented [5] what appears to be the first CCD-based optical-CT scanner, in an elegant experiment to investigate self-organising chemical structures. In 2002, Sharpe [6] 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 introduced with the name optical transillumination tomography in studies of tissue-engineered blood vessels [7-10].
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 [11], optical coherence tomography (OCT) [12] and diffuse optical tomography (DOT) [13] are techniques that are all commonly used to image biological tissues. Laser scanned confocal microscopy is capable of producing high-resolution (1 µm in-plane and 5 µm 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. For a review of other 3-D microscopy techniques, see Steltzer [14]. 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 [15-17] has much in common with schlieren and shadowgraphy methods [18], 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

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 $\mu$, the light intensity, as measured by a detector placed at depth $d$ is given by

$$l(d) = I_0 \exp(-\mu d),$$

(1)

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

$$l(N\Delta y) = I_0 \exp(-\mu_1 \Delta y) \cdot \exp(-\mu_2 \Delta y) \cdot \exp(-\mu_3 \Delta y) \cdots = I_0 \exp\left\{- \sum_{i=1}^{N} \mu_i \Delta y \right\}. \quad (2)$$

Figure 1. (a) Graphical representation of Beer’s law; (b) Schematic diagram of a laser optical-CT scanner: a laser is projected through the sample and the attenuated beam is detected by a photodiode; the source and detector move in tandem along a gantry; (c) The sample rotates, allowing projection data to be acquired at different angles.

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 [19]. 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 the limit $\Delta y \to 0$, $N \to \infty$, this becomes

$$l = l_0 \exp \left( -\int_{\text{across sample}} \mu(y) \, dy \right).$$

(3)

As shown in Figure 1(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 $x$:

$$l(x) = l_0 \exp \left( -\int \mu(x, y) \, dy \right),$$

(4)

where $(x, y)$ is now the 2-D distribution of optical attenuation coefficient. This is related to the optical density (OD) by the relation $\text{OD} = \ln 10$. 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.)

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 $-\log_T T$, 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.

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**Figure 2.** Relationship between real space and Radon space. The presentation of data in Radon space in 2-D image form is often called a sinogram. Note how the square feature in the object leads to a sinogram track with substructure and this may be interpreted by saying that the square is made up of many points, each producing a sinogram tracks of their own, which are overlaid.
As the sample rotates by a positive angle $\phi$ in Figure 1(c), the function describing the attenuation coefficient changes to $\mu_\phi(x, y)$ by the following transformation of co-ordinate system:

$$\mu_\phi(x, y) = \mu(x_\phi, y_\phi) \quad \text{where } \begin{pmatrix} x_\phi \\ y_\phi \end{pmatrix} = R_\phi \begin{pmatrix} x \\ y \end{pmatrix},$$

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

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

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

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

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

We define the Radon transform as

$$\lambda_\phi(x) = \int_{\text{sample}} \mu(x_\phi, y_\phi) \, dy = -\ln \left( \frac{I_\phi(x)}{I_0} \right).$$  

**Figure 3.** Illustration of the relationship between acquired profiles and sinogram tracks in Radon space.
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. 2, Radon space has two dimensions, \(x\) and \(\phi\), and, at the general point \((x, \phi)\), we “store” the result of the projection \(\lambda \phi (x)\). 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.

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 \(\lambda \phi (x) = 0\), except the one coinciding with the point \((x_0, y_0)\) has been rotated. Thus, all points in the Radon space corresponding to the single-point object are zero, except along the track

\[
x = x_0 \cos \phi - y_0 \sin \phi = R \cos (\phi + \phi_0)
\]

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.

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 \(\lambda \phi (x)\), 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 3, the value of the Radon transform \(\lambda \phi (x)\) 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/).

\(^3\) 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.
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. This is explained further in [20]. Filtered back-projection is not the only method of image reconstruction. A family of methods based on the algebraic reconstruction technique (ART) are also available. These are potentially more flexible, allowing one to cater for various deficiencies in the raw projection data, but they are more time-consuming to run. See Ch. 7 of [20] for a general description and [7] for an application in optical CT.

4. Optical CT in 3-D radiation dosimetry

The first-generation scanner geometry illustrated in Fig. 1(b) was the one employed by Gore et al. [1]. Fig. 5(a) gives a more detailed schematic of their apparatus, which developed into what is now marketed by MGS Inc. as OCTOPUS, 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 [21] and, in the years that have followed, a number of other groups have published work of increasing sophistication using first-generation scanners [22-42].

Excellent reviews by Jordan [43-45], 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 [46], reflection, dynamic range, wavelength selection, wall corrections [47], 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. [33, 36].
A major disadvantage of first-generation systems is their relatively slow scanning speed. In order to obtain a profile, the laser beam and photodiode must be stepped in parallel across the sample. Until very recently, when van Doorn’s group [48] (closely followed by Conklin et al. [49]) demonstrated the feasibility of a rotating mirror approach, this was done by mechanical translation of a mirror assembly and took of order 2 s per profile, leading to an imaging time per slice of order 12 minutes [50]. To obtain further slices, the sample was raised or lowered 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.

A faster alternative is the use of scanners based on CCD or CMOS area detectors. These 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 [43]. Each 2-D projection gives the required data 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, in particular the rate at which the data may be transferred out of the camera to the host computer.

Three classes of CCD scanner have so far been presented in the gel dosimetry literature. Our group has developed scanners based on the parallel-beam geometry [15-17, 51-55], illustrated in Fig. 5(c), whilst the cone-beam geometry (Fig. 5(b)), first introduced by Wolodzko et al. [56], was pursued further by Jordan et al. [57, 58] and 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 [59-63]. In these Proceedings [64], Oldham et al. have introduced a dual-purpose CCD-based scanner, which is able both to image radiation dosimeters at a small scale and to perform 3-D micro-imaging along the lines of Sharpe [6, 65].

Not enough data are present in the literature yet to conclude which of the geometries is superior. However, several comparisons may be drawn. (i) The cone-beam design is generally cheaper and is easily scaleable [58]. It has the advantage of being able to scan larger objects without recourse to expensive optical components. However, whilst large optics are expensive, the pair of bespoke lenses for our 30 cm FOV parallel-beam scanner still cost only around £5000, which is a relatively small fraction of the current list price of the Modus 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 [16] 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.

Figure 6. (a) Illustration of the principle behind the generation of sclieren effects; (b) an extreme example of sclieren and other effects that degrade the quality of projection images. The dark region in the centre is a brachytherapy irradiation.
5. Particular considerations for CCD/CMOS-based scanners

(i) Schlieren are refractive index variations in the nominally homogeneous dosimeter and matching liquid. As light rays pass through these, they are deviated. 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. Fig. 6(a) illustrates this phenomenon schematically, whilst Fig. 6(b) is an extreme example that we observed. Such variations in refractive index result typically from incomplete mixing of dosimeter or matching liquid components, or local temperature gradients, which lead to convection currents. If these inhomogeneities are present as the dosimeter 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. The effect on the final image is difficult to predict a priori, and manifests itself as a structured, but complex artefact.

As shown in Fig. 6(a), the length of the optical path through the scanner is long enough such that a deviation of only a small fraction of a degree (and hence only a minor change in refractive index) is necessary to create the schlieren patterns. Cone-beam scanners should be less susceptible to this problem than parallel-beam ones and the phenomenon is generally not observed with laser measurements, since the photodiode detector is often relatively large (5–10 mm diameter) and a small beam deviation is well tolerated [1].

(ii) Ring artefacts are generated when a feature not associated with the dosimeter sample is present in all projections. A typical cause might be a bubble or scratch on the wall of the tank containing the matching liquid. Such artefacts have been studied both in the context of laser scanning [36] and CCD scanning [16]. 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 (e.g., the “wobbling” technique introduced in [16]) to overcome this. 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 [66] to ameliorate any residual image artefacts.

(iii) Calibration: Some CCD’s have a non-linear response and calibration is necessary to ensure that the correct ratio $I / I_0$ is calculated in Eq. (8). 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. It is also necessary to subtract the value of the CCD “dark current” from pixel values in both the light-field and dosimeter images. See abstract [54] for further details.

Figure 7. Illustration of some of the effects than can occur when using CCD detection. In each case the profile across the centre of the reconstruction is displayed, with inset images corresponding to the original object (left) and the reconstructed image (right): (a) simulation of the effect of failing to correct for a dark-current signal of 1.5% of dynamic range of CCD, with noise level 0.1% and 12-bit detection; (b) same situation with 8-bit detection. In each case, the object simulated has high attenuation such that the integrated optical density along the longest path is 2.
PRESAGE polymer gels is their sensitivity to atmospheric oxygen. Formulations have been studied or developed specifically for optical CT [79, 80]. A disadvantage of some polymer gels is that they attenuate light by scattering, which might be less than ideal. However, their potential in optical CT studies was obvious from early on [2]. Whilst contrast in FXG is unacceptable for medical applications, the dose pattern within a few hours of irradiation is stable during a prolonged irradiation period, and has good post-irradiation storage properties. It 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 [16] 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. Not surprisingly, the capability of optical CT instruments is now approaching these limits.

(iv) 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., \( I / I_0 = 0.01 \)). A crude simulation of the type of effects that may result is shown in Fig. 7. The ORCA BT-1024G has a a 14-bit ADC and thus allow us to overcome these problems.

(v) Image signal-to-noise ratio and accuracy: When describing the performance of an imaging CCD, manufacturers often quote the full-well capacity (i.e., how many electrons a pixel on the chip can hold) and the 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 [67], Healey and Kondepudy describe the roles of shot noise (i.e., photon counting statistics), dark current, readout noise and “scene variation” (i.e., genuine random changes in the variable being imaged) in determining the SNR performance of a machine vision system.

(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, but this is in many cases time consuming and runs the risk of images being corrupted by blooming.

6. 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 xylenol-orange system (e.g., [68]). 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 xylenol-orange gels (FXG) [51, 69-78]. 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.

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 [16] 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 [79, 80]. 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 [81] of PRESAGE™, a novel transparent plastic dosimeter. The material is rigid and easily machineable. 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 [16] 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. Not surprisingly, the last two years have seen a flurry of interest in this dosimeter [39, 52, 53, 55, 82-87].

7. Future perspectives and conclusions
Optical CT in 3-D radiation dosimetry has come a long way over the last decade, particularly given the rather small number of groups worldwide that have been working in this area. As pointed out in the introduction, the competing readout systems, MRI and (to a lesser extent) X-ray CT, were already mature imaging modalities when first applied to dosimetry. Nevertheless, the capability of optical CT instruments is now approaching these and, in areas such as image resolution and acquisition speed, it already exceeds 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 all the painstaking verification work, investigation of anomalous results and the pursuit of absolute dosimetry, which have characterised the last two decades of work with MRI. With this in mind, the quality assurance work of Guo et al. [88] and Bosi et al. [59] presented in these proceedings is welcome.

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. It is possibly premature to sound the death knell of MRI, but we are catching up fast!

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