How to perform dosimetry with Optical CT

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Abstract. Both polymer gels and PRESAGE radiochromic solid dosimeter, in conjunction with optical CT scanning system, have been employed to measure 3-D dose distribution. The 3-D dose maps obtained from these systems can provide a useful tool for dose verification on complex treatments such as IMRT, radiosurgery, and RapidArc. These complex treatments present high dose gradient regions in the boundaries between the target and the surrounding critical organs. Dose accuracy in these areas can be critical, and may affect the treatment. There is a pressing need for a dosimeter that allows for accurate determination of 3-D dose distribution with high spatial resolution. In this paper, performance of polymer gels and PRESAGE dosimeter with optical CT scanning is reviewed and evaluated in terms of their sensitivity calibration, irradiation, optimization of scanning procedures, precision, and accuracy. Clinical applications of optical-CT dosimetry are presented.

1. Introduction
It has long been recognized that 3-D dosimetric information is important in clinical situations where high dose gradients exist such as IMRT and stereotactic radiosurgery. Both polymer gels and PRESAGE radiochromic plastic dosimeter, in conjunction with optical CT scanner, have been employed to record the radiation dose distribution in three-dimensions [1-3]. When irradiated, Polymer gels become opaque through polymerization and PRESAGE presents colour change. Their optical densities increase in response to absorbed dose. The 3-D dose information can be reconstructed based on the readout of optical computerized tomography (CT) of polymer gels and PRESAGE. In 1996, Gore et al [1] demonstrated the potential of this optical-CT imaging technique for polymer gel dosimeters, although the concept of this technique can be traced further back. The history and principles of optical CT for scanning 3-D radiation dosimeters were reviewed in details previously by Doran [4-5]. The fundamental science for polymer gels has recently been thoroughly reviewed by Baldock et al. [6]. This article will discuss the practical aspects of optical scanners with polymer gels and PRESAGE in implementing a 3-D dose distribution of clinical applications.

2. Optical-CT systems
The collected data for the optical-CT technique are optical projections acquired either by scanning a laser across the dosimeter sample and detected by a photodiode detector, or by passing a broad light field through the sample and collected by CCD or CMOS area detectors. The principles of optical computed tomography are similar to that for the x-ray CT. Detailed information and the mathematical procedure of filtered back-projection used in the 3-D reconstruction of dose image can be found in the reference [5]. Besides introduction of various optical-CT systems, the main objective of this article is
to review and discuss the operation and procedures as to how to perform clinical 3-D dosimetry with optical CT systems.

2.1. First-generation optical-CT scanner

The concept of original optical-CT scanner, designed by Gore et al, is illustrated in Figure 1(a). We describe here the basic running principle and the configuration of a commercial product, OCTOPUS 5x (MGS Research, Inc.), of this invention. Briefly, a single He-Ne laser beam (633-nm wavelength) is guided by a stepper motor and a series of mirrors and lenses to scan across a rectangular water tank containing the gel/PRESAGE sample repeatedly with a fixed field of view. The light transmitted through the tank is collected by a photodiode during each cycle of translational motions as single projection data. After each full translational scan, the sample mounted on a turntable is rotated by a small angular increment and followed by the linear scan. In earlier version of OCTOPUS scanner, the turntable rotates 360 degrees to collect projection data for each planar image. OCTOPUS x5 has improved imaging time to approximate 8-9 min per slice by reducing the rotation to 180 degrees, since each scan is duplicated in 360-degree scan. In addition, the manufacturer discontinued the use of the reference diode, as shown in Figure 1(a), for higher sampling rate. The laser drift and output variation are taken into account in the driver software. The tank is otherwise filled with a mixture of water, glycerol, and a blue dye.

First-generation optical-CT systems have been demonstrated to be reproducible and accurate, and can be employed, in conjunction with polymer gels and PRESAGE radiochromic plastic dosimeter, to measure 3-D dose distribution with good quality [7-11]. They are also most efficient at minimizing the artefacts caused by scattered light. However, the first-generation systems have not been routinely used in the clinical application, partially because of its slow scanning speed.

Fig. 1. Schematic diagram of the different types of optical CT scanners (diagram reproduced from reference [6] ) : (a) first-generation laser system (Gore et al, reference [1]); (b) fast laser scanner (Krstajic et al, reference [14] ); (c) cone-beam CCD scanner (Wolodzko et al, reference [15]); (d) parallel-beam CCD scanner (Krstajic et al, reference [17])
2.2. Fast optical-CT scanners

In the design of fast optical-CT scanner, several methods for increasing the speed of the scanning laser beam have been proposed. A number of groups have introduced the approach of a rotating mirror, in which the laser was reflected by the rotating mirror instead of the mechanical translation [12-13]. Another approach is the use of a pair of galvanometer-controlled mirrors proposed by Krstajic and Doran [14], as shown in Figure 1(b). A faster alternative is the use of a broad beam with CCD or CMOS area detectors. A cone-beam configuration was proposed by Wolodzko et al [15] (Figure 1(c)). A parallel-beam configuration was proposed by Doran et al [16], and telecentric optics was added to minimize sensitivity to scattered light in a refined design [17]. Similar to x-ray CT, the presence of scattered light is a well known source of artifact in the 3-D dose reconstruction using optical CT scanning of polymer gels. Both Xu et al [18] and Islam et al [19] reported a cross-shaped artifact that appeared in the optical-CT dose images, especially in a high-dose irradiation where the signal-to-noise ratio is relatively small. It should be noted that artifact becomes noticeable when scattering is a significant portion of the signal. Successful 3-D dose verifications of complex IMRT have been reported using first-generation optical-CT systems [7-8, 10]. While scatter correction is still the main obstacle for broad-beam with CCD systems to become a successful 3-D dosimeter, the first-generation optical-CT scanner can be used as a “gold standard” [18].

3. 3-D radiation dosimetry by optical CT scanning of polymer gels/ PRESAGE

3-D dose distributions of various radiotherapy techniques have been obtained by optical-CT scanning of polymer gels on the basis of light-scattering [7-8]; as well as optical scanning of PRESAGE dosimeter based on light-absorbance [9-10]. In this paper, the complete process of 3-D dose determination using the optical CT approach, with either polymer gel or PRESAGE, will be reviewed.

3.1. Polymer gels/PRESAGE sensitivity calibration

The characterization of both polymer gels and PRESAGE radiochromic solid dosimeter have been well studied, including diffusion effect, temperature effect, dose rate dependence and energy dependence [3, 19-21]. It was concluded that, for both polymer gel and PRESAGE dosimeters, neither photon energy nor dose rate has an effect on the dose response for all the megavoltage photon and electron beams tested. One important step in the determination of dose distribution using optical-CT based polymer gels/PRESAGE dosimeter is the sensitivity calibration. To achieve this goal, three calibration methods have recently been used to study the optical dose response of BANG®3 polymer gels and compare their performances [20]. As shown in Fig.1, the methods include a) measurements across a series of small glass-vials dosimeters irradiated to graded doses; b) optical-CT scanning of a single Barex cylindrical gel phantom irradiated to graded doses with small photon fields; c) percent depth dose comparison between optical-CT scans and beam data for a single 16 MeV electron field.

(a)  
(b)  
(c)

Fig. 2. Optical dose response calibration using three different methods described above: (a) glass vials of BANG polymer gels irradiated to graded doses; and reconstructed images from irradiations of (b) four 4 cm x 4 cm photon fields; c) four 6 cm x 6 cm electron fields. Reproduced from reference [20].
In that study, the optical response of the BANG®3 gels was found to be linear and energy-independent within the uncertainty of the experimental method used. The dose response curves obtained from the six Barex gel phantoms agree within 2.2%, indicating a good reproducibility of the gel dose response within phantoms of the same geometry. The dose sensitivity of the BANG®3 gels obtained from the small glass vial experiment was substantially different from that of the large Barex phantoms, owing possibly to the difference in temperature inside the two types of phantoms during gel formation and irradiation [21], and possible oxygen contamination of the glass vial walls. In addition, it was observed that the readouts were sensitive to the positioning of the laser path relative to the glass vials. The maximum difference among the 6 readings taken for one glass vial was found to be 15%. The ratio between the standard deviation and the mean value for the group of 6 readings taken from a glass vial could be as much as 6%. It was suggested that cautions need to be taken with the dose sensitivity calibration method using small glass vials irradiated to graded doses in optical-CT of BANG®3 gels. The large variation in laser path through a glass vial may be reduced if an optical cuvette with flat surface is used instead [3]. Based on the energy independence and the dose response reproducibility established for the BANG®3-Barex gel phantoms, the PDD matching method using electron depth dose curve from ion chamber measurement as a benchmark is demonstrated to be a convenient and relatively accurate way for dose response calibration in optical-CT scanning of BANG®3 gel dosimeters. A 16 MeV electron field irradiation with a field size of 6 cm or larger is appropriate for this purpose. The calibration method using multiple small-field irradiations is in general more labor intensive and less accurate than the PDD matching approach in that it can’t provide optical density values for all desired doses.

PRESAGE™ is a solid 3-D dosimetry material consisting of an optically clear polyurethane matrix, containing a leuco dye that exhibits a radiochromic response when exposed to ionizing radiation. It has the advantages of insensitivity to oxygen, radiation induced absorption contrast rather than scattering contrast, and a solid plastic texture that can be molded to a variety of shapes and sizes without an external container. Guo et al [3] have investigated the basic characteristics of PRESAGE dosimeter, including optical response linearity, dose rate dependence, reproducibility, stability, special changes in absorption, and temperature effect. In that study, sensitivity calibration of the PRESAGE was performed with PRESAGE cuvettes, as shown in Fig. 3(A).

![Fig. 3. (A) Pictures of various cuvettes, (a) nonirradiated cuvette, (b) irradiated cuvette by 6 MeV electrons. (c) irradiated cuvette by 6 MV photons. (d) A PRESAGE™ column without an external container, irradiated by 16 MeV electrons. (B) Picture of the linear optical scanner. A He-Ne laser is the light source. Reproduced from reference [3].]
The linear optical scanner used to monitor optical density variations along the long axis of the PRESAGE samples is shown in Fig. 3(B). The optical response sensitivity was determined from the plot of normalized depth OD changes along the long PRESAGE columns for a 16 MeV electron and 6 MV photon irradiations plotted against the calculated PDD. Good agreement between the normalized depth OD change and the PDD was observed. The OD change versus the absorbed dose along the long PRESAGE column, irradiated by the 16 MeV electrons, was used to determine the sensitivity from a linear fit. The linear fit showed a high linearity between OD change and absorbed dose, with RMSE < 1%. However, “over response” at the low dose region, for both 6 MeV and 16 MeV electrons, was observed (at tails of the PDD curves). This over response phenomenon was also observed at the dose region below 2 Gy in the linear fit of dose range spanning from 1 Gy to 12 Gy. While the reason for this is not fully clear, it is always prudent to perform the sensitivity calibration covering only the full dose range that is used in the specific clinical application (e.g. 0 to 3 Gy for a fractionated external beam treatment; 2 to 20 Gy for a stereotactic radiosurgery).

At the present time, there is not enough data to show if polymer gels and/or PRESAGE dosimeter can be consistently used for absolute dosimetry. More studies are needed in terms of the uniformity, stability, and consistency of polymer gels/PRESAGE; as well as the optimization of sensitivity calibration, dosimeter irradiation, and optical scanning procedures.

3.2. Irradiation of dosimeters
In the optical CT scanning of gel/PRESAGE dosimeters, the net optical density varies in a much wider range than it does in the 2-D scan of the same plan, because of the longer and varying path lengths of the laser beam through the dosimeter. The usable dynamic range of a scanner is limited by several factors, including the signal-to-noise ratio and the electronic and mechanic noises in the scanner. Therefore, gel/PRESAGE sensitivity and the dose delivered should be optimised such that the optical density increments along all possible paths during scanning of an irradiated dosimeter never exceed a certain maximum. As such, readout uncertainty can be minimized and reconstruction artefacts can be avoided. On the other hand, the sensitivity of the dosimeter should be high enough to maximize the dynamic rage of the scanning system, while a high signal-to-noise ratio and thus high dosimetric accuracy can be maintained. Xu et al developed a method for determining the minimum transmitted signal along all scan paths through a gel of known dose response irradiated with a planned dose distribution [23]. This calculation allows either the dose to be set for the optimal transmission range of the dosimeter or to prepare a dosimeter with an optimal sensitivity for the delivered dose. In that study, optical CT experiments were performed to determine the range of transmitted signal that would allow a reconstruction accuracy of less than 4%. It was concluded that, given the optical CT scanning system (OCTOPUS, MGS Research, Inc), optical densities from 0.5 to 2.5 met the 4% criteria. The optimal dose delivery for each dosimeter may vary depending on the sensitivity and the dosimeter size. It is suggested that, for clinical dose verification such as IMRT, the dosimeter sensitivity be adjusted to its optimum such that verification can be performed at clinical dose range.

Temperature effects on both polymer gels and PRESAGE have been reported [3, 20]. The polymer gels/PRESAGE can have a different optical response to irradiation at a different temperature. The temperature sensitivity of the PRESAGE dosimeter around room temperature is about 1.7% OD variation per degree. Uniform temperature throughout the dosimeter volumes during calibration and phantom irradiation is essential in order to achieve an accurate dose measurement. Special cautions should be taken when the phantom size is significantly different from the samples (e.g. cuvettes) used for sensitivity calibration. In general, it is adequate to keep the dosimeters under the irradiation environment for 24 hrs before the experiment.

3.3. Scanning process
The majority of optical-CT dosimetry of polymer gels/PRESAGE, especially those related to clinical dose verification, was obtained from the first-generation laser scanning systems. The most significant
sources of artifacts in optical CT scanning of polymer gels/PRESAGE result from reflection and refraction of light at the surface of the dosimeter. One key step to minimize the artifact in dose reconstruction is to match the refractive index of the tank liquid (water, glycerol, and a blue dye) with the refractive index of the polymer gels. The addition of a dye to the matching fluid is to match the light attenuation between the gel and the liquid. This allows for maximum use of the dynamic range of the light detection system. Depending on the differences in refractive index between the matching fluid, the gels, and gel container wall, an edge artifact was observed in the outer 1-2 cm of the gel container [7].

Optical CT scanning with refractive index matching liquids currently involves open systems for placement of the gel samples. Jordan has a thorough discussion of matching liquids that are used in current scanners [25]. Experimenters should consider: toxicity, safety (flammable), vapor pressure, viscosity, chemical stability, color stability, disposal and corrosive nature of solutions when choosing suitable materials for matching liquid. The matching procedure can be lengthy and time-consuming. Vandecasteele and De Deene have developed an automated procedure [26], based on a dry and wet off-axis refraction measurement of the gel dosimeter used. Based on the determined refractive index of the gel and the fluid, the required glycerol concentration can be determined from all the experimental data. An automated pump and mixing system was developed for concentration adjustment. For PRESAGE/optical CT scanner, the matching fluid currently used is a mixture of octyl salicylate and methoxy octyl cinnamate [10]. With a better refractive index matching of the fluid, the use of a more attenuating fluid, and an improved PRESAGE formulation, the edge artifact can be reduced to 3mm in the outer region of the dosimeter.

The only commercially available first-generation optical CT laser scanner is the OCTOPUSTM, manufactured by MGS Research Inc. In the OCTOPUSTM series of optical CT scanners, the data acquisition process is controlled by a windows executable program written with TESTPOINT (CEC Inc., Billenca, MA). The scanning field of view, the pixel size and the number of pixels per projection are defined by the users. The data acquisition rate and the velocities of the 3 stepper motors are optimized by the control program. After the data acquisition process is completed, the obtained laser signals from the data acquisition board are imported into an image reconstruction program written in MATLAB. The image reconstruction program reconstructs the optical density distributions slice by slice using the filtered back projection method and a set of filters. The total time for obtaining a full set of 3D images of a dosimeter, ranging from 4 to 8 hours, depends on the dimension of the dosimeter, pixel size, and the version of the scanner.

The reconstructed 3D optical density map from the image reconstruction program represents the three dimensional relative dose distribution within the dosimeter scanned. Sensitivity calibration procedures can be performed to obtain the absolute dose distribution delivered to the dosimeter. The viewing and the preliminary analysis of the reconstructed 3D relative or absolute dose distributions can be done using most of the standardized imaging software, ImageJ (http://rsb.info.nih.gov/ij), for example. Detailed analysis of the reconstructed images from the gel measurement, including the comparison of the 3D dose distribution with those from the planning system or other measurement methods, can be done using in-house developed computer software. In addition, evaluation tools for 3D dose comparison can be incorporated in the computer program.

3.4. Artifacts from light scattering

The dose response of optical-CT based polymer gel dosimeter arises from light scattering at the radiation-induced polymer micro-particles in the gel. Similar to the scattered radiation in x-ray CT, the scattered light can be a potential source of artifacts in the optical CT dose reconstruction, as reported by Xu et al and Islam et al [7, 27]. As shown in Fig. 4, a cross-shaped artifact that appeared in optical-CT reconstructed dose images of a high-dose square-field irradiation. The distortions of the images in Figs. 4(b) and 4(c) can be attributed primarily to the significant contribution from scattered photons at high ODI. In polymer gels, optical attenuation is caused by the Rayleigh/Mie scattering of light on polymer particles. Therefore, a collection of some of the scattered photons at the detector’s aperture is inevitable but normally does not cause any detectable error in the reconstructed image. With
increasing optical density increments, the number of scattered photons produced increases whereas the intensity of the transmitted primary beam decreases (low signal-to-noise ratio). The fraction of the scattered light in the signals collected will increase and eventually become a significant source of error. The cross artifact was confirmed to result from light scattering in optical-CT Monte Carlo simulations by Oldham et al [28]. The impact caused by light scattering can be minimized to less than 4% uncertainty, if the optical density is between 0.5 and 2.5 as described in section 3.2 above.

PRESAGE dosimeter exhibits optical contrast through light absorbance rather than light scattering, and has an advantage for the optical CT scanning technique since the scattering artifact displayed above is not an issue.

Fig. 4 Reconstructed dose images for 3 axial slices at depth 6, 4, and 3 cm, from a 6 cm x 6 cm single 6 MV field with 100 cm SSD. The maximum optical density increments for these planes (left to right) are 2.48, 2.8 and 3.1, respectively. Reproduced from reference [7].

4. Dose verification in clinical applications

Both polymer gels and PRESAGE dosimeter, in conjunction with the optical CT system OCTOPUS, have been employed to verify complex 3D dose distribution, including IMRT [8,10] and stereotactic radiosurgery [29]. Fig. 5 shows comparisons of IMRT dose distributions in three planes, obtained from gel (blue), EDR2 film (green), and the Helios IMRT treatment planning system (red). The results show good agreement among all three techniques at selected planes. PRESAGE/optical-CT scanner was also applied to verify a complex IMRT dose distribution [10]. A highly modulated IMRT plan was delivered to a cylindrical PRESAGE dosimeter, and the dose distribution was readout using a commercial optical CT scanner OCTOPUS. As shown in Fig. 6, comparisons were made with GAFCHROMIC EBT film measurements, and the calculated dose distribution from ECLIPSE treatment planning system. Excellent agreements were observed.

Fig. 5. Comparison of dose distributions at (a) axial central plane (40%, 60%, 100%, and 115%), (b) coronal central plane (40%, 60%, 100%, and 115%), and (c) sagittal central plane (30%, 70%, and 100%). Dose distributions were obtained from treatment planning calculation (red), polymer gel measurement (blue), and EDR2 measurements (green).
Radiosurgery, either cranial or extracranial, is a procedure that precisely delivers large radiation doses to tumors and other relevant anatomical targets in a single fraction or in a small number of fractions (typically up to five). Therefore, the success of stereotactic radiosurgery requires precise targeting and rapid dose falloff in surrounding normal tissues. We have employed both polymer gel and PRESAGE dosimeters to measure the 3-D dose distribution of an IMRT radiosurgery plan for a 2.5 cc brain tumor. Figure 7 shows dose distribution comparison at the central axial plane between measured (polymer gels and PRESAGE dosimeter) and calculated dose distributions. Both measured and calculated dose distributions are in reasonable agreement. However, the isodose lines from both gel and PRESAGE measurements show more variation than those from the planning calculation. These discrepancies may be partly attributed to the fact that the calculation grid for the planning system is 2.5 mm yet the resolution of measurements is 1 mm, and partly attributed to the impurities and air bubbles in the phantoms. It appears that the accuracy and precision of the 3-D dosimetry technique, with optical CT scanning of polymer gel/PRESAGE, requires further improvement for detailed 3-D dose verification.
Figure 7. Comparison of the isodose lines (110, 100, 90, 80, 50, and 30 percent) from the Eclipse planning system (green lines) and measured dose distributions (Gel and PRESAGE) for the central axial slice. Gel: red lines; PRESAGE: black lines.

5. Conclusion
With the increasing use of complex treatments such as IMRT, stereotactic radiosurgery, and RapidArc, there is an urgent need for a dosimeter that allows for accurate 3-D dose distribution measurement. Both polymer gels and PRESAGE solid dosimeter, with an optical CT scanning system, have been used to verify complex 3D dose distributions from IMRT and radiosurgery plans. The results demonstrate that an accurate 3D dose distribution with high resolution can be implemented. Future development should be focused on improvement of the scanning speed, as well as the consistency and uniformity of the dosimeters.

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