FlexyDos3D: a deformable anthropomorphic 3D radiation dosimeter: radiation properties

To cite this article: Y De Deene et al 2015 Phys. Med. Biol. 60 1543

View the article online for updates and enhancements.
FlexyDos3D: a deformable anthropomorphic 3D radiation dosimeter: radiation properties

Y De Deene1,2, P S Skyt2,3, R Hil2,4 and J T Booth2,5

1 Department of Engineering, Faculty of Science and Engineering, Macquarie University, Sydney NSW 2109, Australia
2 Institute of Medical Physics, Faculty of Science, University of Sydney, Sydney, Australia
3 Department of Medical Physics, Aarhus University Hospital, Nørrebrogade 44, 8000 Aarhus Denmark
4 Department of Radiation Oncology, Chris O’Brien Lifehouse, Sydney, Australia
5 Northern Sydney Cancer Centre, Royal North Shore Hospital St Leonards, Sydney, Australia

E-mail: Yves.DeDeene@mq.edu.au

Received 26 August 2014, revised 10 December 2014
Accepted for publication 17 December 2014
Published 23 January 2015

Abstract

Three dimensional radiation dosimetry has received growing interest with the implementation of highly conformal radiotherapy treatments. The radiotherapy community faces new challenges with the commissioning of image guided and image gated radiotherapy treatments (IGRT) and deformable image registration software.

A new three dimensional anthropomorphically shaped flexible dosimeter, further called ‘FlexyDos3D’, has been constructed and a new fast optical scanning method has been implemented that enables scanning of irregular shaped dosimeters. The FlexyDos3D phantom can be actuated and deformed during the actual treatment. FlexyDos3D offers the additional advantage that it is easy to fabricate, is non-toxic and can be molded in an arbitrary shape with high geometrical precision.

The dosimeter formulation has been optimized in terms of dose sensitivity. The influence of the casting material and oxygen concentration has also been investigated. The radiophysical properties of this new dosimeter are discussed including stability, spatial integrity, temperature dependence of the dosimeter during radiation, readout and storage, dose rate dependence and tissue equivalence.

* The first authors Y De Deene and P S Skyt made an equivalent contribution to the experimental work presented in this paper.
Keywords: gel dosimetry, 3D radiation dosimetry, deformable dosimeter

(Some figures may appear in colour only in the online journal)

1. Introduction

Three dimensional dose integrating dosimeters are chemical dosimeters in which the radiation sensitive chemical agent is held by a matrix such as a hydrogel or a synthetic polymer. Ideally the dosimeter phantom can be shaped directly in an anthropomorphic shape or be inserted into an anthropomorphic phantom holder. Upon irradiation, a radiation induced chemical reaction occurs within the dosimeter phantom that results in a change in physical properties. Over the last two decades, much research has been conducted towards both the development of different radiation sensitive three dimensional dosimeters and adequate scanning techniques. Two different classes of dosimeters have been developed:

(a) Polymer gel dosimeters: in these dosimeters, a radiation induced polymerization reaction occurs. Polymer gel dosimeters consist of vinyl monomers and a hydrogel. The degree of radiation induced polymerization is dose dependent and can be read out by use of magnetic resonance imaging (MRI), x-ray CT or ultrasonic scanning (Ballock et al 2010). Although some groups have scanned polymer gel dosimeters with optical CT, it is important to note that Tyndall scattering of polymer gel dosimeters compromises the accuracy of the dose distribution (Bosi et al 2009).

(b) Radiochromic dosimeters: these dosimeters contain a leucodye as radiation sensitive chemical (Adamovics and Maryanski 2003, Jordan and Avvakumov 2009) or a redox indicator in combination with a reductants such as Fe2+ in a Fricke solution (Olding et al 2010). The matrix can be an optically transparent hydrogel or a polymer.

Polymer gel dosimeters offer a wide range of applications in the 3D verification of complex radiation dose distributions such as in intensity modulated radiotherapy (IMRT) (De Deene et al 2000) and dynamic radiotherapy treatment verification (IMAT) (Vergote et al 2004). In order to acquire the 3D dose distribution with polymer gel dosimeters and MRI readout with a clinically acceptable uncertainty, adequate artifact compensation methods need to be implemented (De Deene and De Wagter 2001, De Deene and Vandecasteele 2013, Vandecasteele and De Deene 2013a, 2013b, 2013c).

With the introduction of new delivery techniques that are able to gate or track the beam with organ motion, there is a growing need for motion actuated and deformable 3D dosimeters. A deformable polyacrylamide gel dosimeter named DEFGEL, has been proposed recently (Yeo et al 2012). The deformable polymer gel dosimeters are read out with cone beam optical CT. Although these deformable polymer gel dosimeters can be useful in demonstrating the effect of deformation during treatment, their use as accurate dosimeters can be problematic, because of the effect of light scattering on the transmission images (Bosi et al 2009).

Different staining dosimeters have been developed including polyurethane based dosimeters, PRESAGETM (Adamovics and Maryanski 2004) and micelle gel dosimeters (Jordan and Avvakumov 2009, Vandecasteele and De Deene 2013d). A deformable dosimeter based on PRESAGE has been recently suggested (Juang et al 2013). Although promising results have been obtained with the new dosimeter named ‘PRESAGE-DEF’, the fabrication of PRESAGETM dosimeters requires a specialized chemical laboratory facility.

In this study, we developed a novel deformable dosimeter based on a transparent silicone elastomer which is relatively easy to fabricate, is non-toxic and of which the mechanical properties can be easily tuned (Gutierrez and Groisman 2011).
2. Materials and methods

2.1. Dosimeter fabrication

The dosimeter consists of an optically transparent silicone elastomer, Sylgard® 184, a corresponding curing agent, chloroform as a source of radiation induced radicals and the leucodye, leucomalachite green (LMG). The concentration of chloroform and leucomalachite green was varied in order to optimize the sensitivity of the dosimeter. We will further refer to the dosimeter by the name ‘FlexyDos3D’.

2.1.1. Cuvette samples and cylindrical finger phantoms. To remove gas bubbles from the Flexydos3D phantoms, samples are placed in a desicator and vacuum pumped. It is well known for other dosimeters that the oxygen concentration in the phantom may also have an effect on the dose response. To perform a calibration of the dosimeter phantom with calibration vials, we propose the use of plastic cuvettes that are filled with the same batch of dosimeter elastomer and which can be easily removed prior to scanning. In this study of radiation properties, most experiments were performed on samples that were poured in PMMA cuvettes. However, it is well known that pure chloroform is able to dissolve PMMA which could potentially lead to deviations in dose response. In order to investigate the influence of vacuum pumping, the dependence on oxygen concentration and the PMMA cast on the dose response, four series of 4 ml glass tubes with an inner diameter of 8 mm were filled with the FlexyDos3D elastomer.

Four different batches of cylindrical finger phantoms were fabricated using the standard FlexyDos3D recipe but following different procedures:

• **Series 1:** Standard procedure (control samples); samples are stored under normal atmospheric conditions in air.
• **Series 2:** Containing PMMA flakes (8% (w/w)); samples are stored under normal atmospheric conditions in air. The PMMA flakes sank to the bottom of the tubes.
• **Series 3:** Degassed in a desiccator by use of vacuum
• **Series 4:** Storred in a reaction vessel filled with Nitrogen

All phantoms were cured for one hour in a hot air oven at 65°C Celsius and allowed to cool down to room temperature for another 2 h before exposure. The FlexyDos3D was then removed from the glass test tubes by breaking the glass.

2.1.2. Volumetric cylindrical phantoms. The silicone solution was also poured in cylindrical teflon phantoms with a diameter of 5 cm and a length of 10 cm to investigate the spatial integrity of the dosimeters. The phantoms were cured by heating up to 65 °C for two hours. The phantoms were exposed to radiation the subsequent day.

2.1.3. Deformable phantom. By way of illustration, a cylindrical deformable dosimeter phantom was also constructed. To further increase the elasticity of the dosimeter the ratio of resin and curing agent of the elastomer was changed. The deformable phantom dosimeter was composed of 90% (w/w) Sylgard® 184 resin, 5% curing agent, 5% Chloroform and 0.03% LMG. Halving the curing agent concentration reduced the elasticity modulus with a factor of approximately 3, as measured from the strain of a cylindrical finger phantom that was elongated under the influence of a weight.
2.2. Irradiation

Dosimeter phantoms were irradiated with high-energetic photon beams using a clinical Varian 21EX linear accelerator (Varian Medical Systems, Palo Alto, USA) with a beam energy of 6 MV unless otherwise stated. The dosimeter was also found to be sensitive to UVC light. To facilitate the optimization of the chemical procedure of the dosimeter, a 9 W germicidal lamp (TUV PL-S 9 W, Philips) was used to expose dosimeters to UVC light for various time periods. According to the manufacturer specifications, the predominant wavelength of the lamp was 255 nm. The lamp was suspended in a polystyrene box that was coated with aluminium foil on the inside to homogenize the light flux. In the investigation of the sensitivity of the dosimeter to curing temperature, cuvettes were exposed to the UVC light at a distance of 10 cm from the lamp while all sides except for one were shielded with black paper. The cuvettes were then read out with the UVC exposed surface perpendicular to the optical light path. As a result, the measured optical density corresponds with the integrated absorbance across the UVC depth dose profile. Although this method doesn’t yield absolute dose values, it was found a practical and fast method in performing measurements immediately after exposure. This method was also used in a preliminary study of dose sensitivity for different chemical compositions (not reported in this paper). Once a preliminary idea on sensitivity was obtained, the dosimeter was further analyzed by irradiation with photon beams from a clinical linear accelerator.

Cylindrical finger phantoms were also exposed radially to UVC light by rotating the sample at a rate of 500 rotations per minute. This further homogenized the exposure to UVC light in the The distance between the cylindrical finger phantoms and the light source amounted to 15 cm. Total exposure times ranged from 15 s to 3 min.

A similar dosimeter recipe optimization study as reported in section 3.1.2 was performed with the UVC light source (not reported in this paper) instead of with high-energetic photon beams from a linear accelerator. The same optimum in concentration was found for both exposure modalities and the trends in response with concentration variations were also similar. Although this is a good indication of comparable reaction kinetics, caution is required to draw sound quantitative conclusions from measurements obtained with a UV light source. In this work, dose rate dependence, temperature dependence during readout, temporal and spatial stability were all studied for high-energetic photon beams. Important to note is also that the dose distribution in a small cuvette is not uniform in the direction of the beam with UV light. However, when the samples are positioned in the spectrometer with the observation lightbeam in the same direction as the exposure beam, the attenuation is integrated along the radiation beam path which allows comparisons between different samples.

The deformable phantom was deformed by squeezing the phantom by use of eight screws that were connected by a suspension PMMA cylinder as illustrated in figure 1. A gap was made in the suspension cylinder to allow optical scanning during deformation without obstructing the light transmission. The dosimeter phantom was irradiated while it was in the deformed state.

The tissue equivalence of the FlexyDos dosimeter was determined by calculation of the mass energy absorption coefficients and compared to those of the PRESAGE dosimeter and water (Berger et al 1998, Hill et al 2014).

2.3. Optical readout

To create an optical parametric map that is correlated with radiation dose without the need of a blank scan, we introduced a double wavelength scanning technique. With this approach, two sets of transmission images are recorded with light of two different colors and the difference
Figure 1. Results from a deformation experiment on a cylindrical phantom: (a) photographs of the deformed cylindrical phantom which has been irradiated with a square photon beam of 2 cm × 2 cm in the longitudinal direction. (b) The dose distribution in 10 axial slices of a cylindrical phantom that was irradiated in the non-deformed state. (c) The dose distribution in the deformed phantom under deformation and (d) in the released state demonstrating the biggest deformation of the field at the contact points of the screws.
optical density ($\Delta$OD = OD(red)–OD(blue)) is calculated. The $\Delta$OD transmission scans are then cone beam reconstructed to create a set of $\Delta$OD CT images.

2.3.1. Optical spectroscopy and absorption measurements. Spectroscopic measurement on the square cuvettes were performed with a high-resolution spectrometer (USB4000, Ocean Optics) equipped with a cuvette holder and a manual benchtop spectrometer. The measurement uncertainty in $\Delta$OD was first assessed by use of cuvettes containing various amounts of food dye solution and was found to be less than 0.02 cm$^{-1}$ in a $\Delta$OD range of 0–1.2 cm$^{-1}$ measured as the maximum deviation between different samples with the same dye concentration. This value of uncertainty also accounts for possible variations in plastic cuvettes. All optical absorption spectra are calibrated against a water sample. The spectrum of the optical cone beam CT scanner has also been measured by use of the optical spectrometer. It was found that the spectrum of each of the light sources could be fit by use of a skew-normal distribution (figure 2(b)) described by the function:

$$g(\lambda) = A \cdot (1 + \text{erf}(B(\lambda - C))) \cdot \exp(-D(\lambda - C))$$

(1)

where $A$, $B$, $C$ and $D$ are the fitting parameters.

2.3.2. Optical imaging. Optical imaging was performed with a modified dual wavelength cone beam optical scanner based on the Modus Vista™ scanner (Modus Medical Devices Inc, London, Canada). In the dual wavelength scanner, the light source is replaced by a diffuse light source containing tricolor LEDs (red, green and blue). Transmission scans were recorded at 512 different angles with equidistant angular increments with both the blue light source and red light source. The pixel resolution in the transmission scans was 0.2 mm × 0.2 mm. CT images were constructed using in house developed conebeam reconstruction software in the Matlab (The Mathworks) environment and using multi-threading. The dual wavelength conebeam CT scanner has been optimized and benchmarked which will be discussed in future work. It is shown that the optical density measured with the optical spectrometer matches very well with the optical density of the dual wavelength optical CT scanner if the spectral distribution of the flat-bed diffuse light sources is taken into account. This enables comparison between the values obtained with the three optical instruments and absolute calibration of the dosimeters.

$\Delta$OD images were also acquired of several cylindrical finger phantoms that were suspended together by use of a dedicated silicone holder. Depth-$\Delta$OD profiles were acquired by angular averaging of axial $\Delta$OD images (figure 5(e)).

The deformable dosimeter phantom was scanned in the deformed state and after it was released and returned to its original cylindrical shape (figure 1).

3. Results

3.1. Spectral response and dose sensitivity

3.1.1. Spectral response. The optical absorption spectra of cuvette samples exposed to UVC light for different times is shown in figure 2. From the optical absorption spectra it is clear that preferential absorption occurs at a wavelength of 627 nm. Minimum light absorption occurs at 480 nm and for this wavelength no significant change in absorption is observed upon irradiation.

To account for the spectral irradiance profile of the light source of the scanner ($g(\lambda)$), the optical density measured with the optical scanner (OD$_{\text{scan}}$) was derived by integrating the spectral response (OD$_s$) (figure 2(a)) with the irradiance profile of the corresponding light source (figure 2(b)).
The fitting coefficients of $g(\lambda)$ (equation (1)) for the different light sources are listed in table 1.

The OD-dose response for monochrome light (OD$_L$) at the peak wavelength of the scanner light source is plotted as circular symbols in figure 2(c), while the integrated scanner-related OD-dose response is plotted as square symbols in figure 2(c). To convert optical density differences measured with the spectrometer to optical density differences acquired with the optical scanner, a conversion factor (CF) can be derived. Figure 2(d) shows that the conversion factor is independent of optical density. Hence, a single conversion factor (CF) can be applied to

$$\text{OD}_{\text{scan}} = \int_0^\infty \text{OD}_L \cdot g(\lambda) \, d\lambda$$

(2)
convert optical density differences obtained with the spectrometer at distinct frequencies to optical density differences obtained with the scanner at two colors:

\[
CF = \frac{\text{OD}_{\text{scan (red)}} - \text{OD}_{\text{scan (blue)}}}{\overline{\text{OD}}_{627\text{nm}} - \overline{\text{OD}}_{480\text{nm}}}
\]

(3)

### 3.1.2. Dose sensitivity and dosimeter recipe optimization.

The \(\Delta\text{OD}\)-dose sensitivity (slope) and offset were determined for different concentrations of CHCl\(_3\) and LMG. Phase plots of the slope and offset are displayed in figure 3. A significant change in slope and offset is observed over time as a result of auto-oxidation and fading when the samples are kept at room temperature for several days. Fading, manifested as a decrease in slope, appears to be more pronounced than auto-oxidation (manifested as an increase in \(\Delta\text{OD}\) offset). For LMG concentrations higher than 0.5% (w/w) the dosimeter became turbid. For a dosimeter scanned 3 h after irradiation (figures 3(a) and (b)), an optimal composition was found for a LMG concentration of 0.025% and a chloroform concentration of 5%. The dose sensitivity (slope) does not change significantly for higher concentrations of LMG but the \(\Delta\text{OD}\) offset increases significantly. A second maximum in slope could be seen for samples at high chloroform concentrations but these samples suffered from gas bubbles and became less rigid.

### 3.1.3. Influence of treatment temperature on the dose response.

Curing of FlexyDos3D can be accelerated by increasing the temperature after fabrication. To observe the effect of this temperature treatment upon the dose response and auto-oxidation, several samples were placed in the oven after fabrication and removed from the oven after different time intervals and measurements were performed after the samples cooled down to room temperature. We chose a curing temperature of 65° because most plastic casts can tolerate this elevated temperature. All samples were also rescanned after 3 d to observe the long term stability. From figures 4(a) and (b), it can be clearly seen that increasing the timespan that samples are cured at elevated temperatures, also increases the background optical density. The sensitivity of the dosimeters decreases slightly with increasing curing times at elevated temperature (figure 4(c)) but figure 4(d) suggests that the curing was incomplete when samples are heated for less than 2 h.

### 3.1.4. Influence of cast material and sample treatment.

The influence of the PMMA cast, oxygen concentration and degassing by use of vacuum on the dose-response was assessed by exposing rotating cylindrical finger phantoms to UVC light. Reconstructed \(\Delta\text{OD}\) CT images are shown in figures 5(a) and (c). Radial \(\Delta\text{OD}\) profiles through samples exposed to UVC light for different times are shown in figures 3(b) and (d). To make an easy comparison between the different samples, the \(\Delta\text{OD}\) profiles are integrated in the radial direction and the integral is plotted as a function of the exposure time \(t_{\text{exp}}\) (figure 5(e)). From this figure it is clear that exposing the dosimeters to vacuum to get rid of gas bubbles does not have a significant effect. Also PMMA does not seem to affect the \(\Delta\text{OD}\)-dose response curve. But storing the samples in a nitrogen atmosphere causes a decrease in sensitivity.

---

**Table 1.** Fitting coefficients (equation (1)) of the scanner light sources.

<table>
<thead>
<tr>
<th>Light source</th>
<th>A</th>
<th>B</th>
<th>C</th>
<th>D (nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Blue</td>
<td>0.030</td>
<td>0.082</td>
<td>0.072</td>
<td>470.2</td>
</tr>
<tr>
<td>Green</td>
<td>0.024</td>
<td>0.058</td>
<td>0.068</td>
<td>530.9</td>
</tr>
<tr>
<td>Red</td>
<td>0.000</td>
<td>0.079</td>
<td>0.425</td>
<td>664.3</td>
</tr>
</tbody>
</table>
3.2. Stability

3.2.1. Auto-oxidation during curing. Immediately after mixing all the chemicals, a small change in color is visible. In order to observe the change in optical density with time after mixing, optical density measurements were performed at two wavelengths immediately after
The change in optical density in a cuvette at 400 nm and 630 nm after mixing is shown in figure 6. It can be seen that the absorption at 630 nm increases rapidly in the first minutes after mixing and then increases slowly until saturation after approximately 1 h. To rule out that the auto-oxidation was caused by exposure to light of the spectrometer, additional measurements were done on a cured sample demonstrating that no observable changes in optical density were caused by the spectrometer light. The auto-oxidation was more pronounced when the Flexydos3D was cured at elevated temperatures.

3.2.2. Temporal stability after exposure. The optical density change after exposure is shown in figure 7. It is found that a small change in optical density in the order of 4% occurs immediately after exposure which reaches a steady state after 3 h. The spectrum does not change significantly but the decrease in ΔOD amounts to 1.3% per hour in a time period of 3 h till 1 d after irradiation and at room temperature.
Figure 5. ΔOD cross sections through cylindrical finger phantoms (a, c); labels in the figure indicate samples cured in air (A), samples containing PMMA (P), samples cured in vacuum (V) and samples that were cured in a nitrogen atmosphere (N). Graphs (b, d) are radial ΔOD profiles through each of the cylindrical finger phantoms obtained by angular averaging as illustrated in (e). The integrated average ΔOD is plotted against exposure time (f). A linear fit indicates a similar response in all samples except for the samples that have been cured in nitrogen.
A series of irradiated FlexyDos3D samples were stored in the fridge (6 °C) and compared against a series that was stored at room temperature (22 °C). Although significant changes in $\Delta OD$-dose were found for the samples that were stored at room temperature, the changes were only moderate for the samples stored in the fridge (figure 7(e)). However, a small increase in $\Delta OD$ in the low dose region is visible.

### 3.3. Dose rate dependence

The dose rate dependence of the FlexyDos3D dosimeter is shown in figure 8 for dose rates between 200 cGy min$^{-1}$ and 400 cGy min$^{-1}$. Remarkable is that the dose rate dependence seems to be more pronounced at higher dose levels and that the dose response is non-linear. The non-linearity in dose response increases with decreasing dose rate. For dose values between 0–15 Gy, the dose rate dependence of the dose sensitivity amounts to 12%, while in the dose range [15 Gy, 40 Gy] the dose sensitivity varies to up to 32% for dose rates between 200 cGy min$^{-1}$ and 600 cGy min$^{-1}$.

### 3.4. Temperature dependence during radiation

At high doses [20 Gy, 40 Gy] there is a significant dependence of the $\Delta OD$-dose response (figure 9) while the effect is less pronounced in the low dose region ([0 Gy, 20 Gy]). The change in $\Delta OD$-dose sensitivity amounts to 0.8% per degree Celsius in the dose range [0 Gy, 20 Gy] and 1.8% per degree Celsius in the dose range [20 Gy, 40 Gy].

### 3.5. Temperature dependence during scanning

The change in optical density difference for FlexyDos3D samples irradiated to 50 Gy (figure 10(a)) and 20 Gy (figure 10(b)) readout at different temperatures shows a small decrease in sensitivity with increasing readout temperature. The temperature dependence of the optical density of FlexyDos3D (nominal composition) is in the order of 0.3% per degree Celsius (figure 10).
Figure 7. Change in ΔOD after exposure of a cuvette with UV light for 5 min (a)–(b). Change in absorption spectrum over time in a time span of 3 h up to 1 d after irradiation of a cuvette with high energetic x-rays (6 MV; 40 Gy) and kept at room temperature (c). Change in optical density at the wavelengths 480 nm (blue dots) and 627 nm (red dots) and ΔOD (black dots) (d). A fit (dashed line) is applied as a guide to the eye. The uncertainty in OD at each data point is 0.02 cm⁻¹. Samples stored in the fridge at 6 °C are found to be far more stable than when the samples are kept at room temperature after irradiation (e).
3.6. Tissue equivalence and beam quality

The basic composition of the dosimeter is polydimethylsiloxane and 5% (w/w) chloroform. The relative electron density of the dosimeter as compared to the electron density of water is in the order of 1.0038 (for a dimethylsiloxane chain length of 10). For higher chain lengths the electron density converges to that of water. The mass attenuation coefficient for water,
PRESAGE™ and FlexyDos3D are shown in figure 11(a). Within the energy range of 100 kV to 20 MV, the dosimeter is highly tissue equivalent but at energies lower than 100 keV, the mass attenuation coefficient in both PRESAGE™ and FlexyDos3D is significantly higher and

Figure 9. ΔOD-dose response curves obtained after irradiating cuvettes at different temperatures (from bottom to top: 6 °C, 11 °C, 20 °C, 30 °C, 40 °C) (a). Dashed lines are monoexponential fits to the data points. The change in slope as a function of irradiation temperature (b) obtained from a linear fit to the ΔOD-dose response curves in the dose interval [0 Gy, 20 Gy] (black circular symbols) and in the dose interval [20 Gy, 40 Gy] (red square symbols).
consistent with previous published data (Hill et al 2010). Calculated interaction probabilities for water, PRESAGE™ and FlexyDos3D are plotted in figure 11(b). The photoelectric effect dominates in water up to 26 keV, while for PRESAGE™ and FlexyDos3D the photoelectric effect dominates up to 34 keV and 43 keV respectively. The difference in dose sensitivity for FlexyDos3D dosimeters irradiated with photon beams of 6 MV and photon beams of 18 MV amounted to less than 5% which is within the measurement uncertainty for these measurements.
3.7 Spatial integrity

No measurable effect of leucodye diffusion was observed over a time span of 4 d for dosimeters that were exposed to a 2 cm × 2 cm field with the 6 MV photon beam (figure 12 a–c).

3.8 Deformation

Figure 1 illustrates the feasibility of the dosimeter to study deformation. Transverse ΔOD images of a non-deformed phantom exposed to a square photon beam are shown in figure 1(b) by way of comparison. Transverse ΔOD images of a deformed phantom exposed to a square photon beam but scanned while still being deformed are shown in figure 1(c). Transverse ΔOD images of the same deformed phantom after being released and returned to its original cylindrical shape are shown in figure 1(c). The deformation is more pronounced in slices that were located close to the screws.
4. Discussion

The spectral absorption response demonstrated an absorption peak with a maximum at a wavelength of 627 nm and was independent of radiation dose within the clinical dose range ([0 Gy, 50 Gy]). However, at extremely high doses obtained by long exposures with UV-light we found a decrease in absorption and shift in peak wavelength towards the red (data not shown). Total decolourisation was obtained in a sample that was exposed to UV light for several days. The peak absorption wavelength at 627 nm for FlexyDos3D is slightly lower than the peak absorption wavelength of 635 nm observed in LMG based micelle gel dosimeters (Vandecasteele et al 2011) and the peak absorption wavelength reported for PRESAGE™ for which values between 628 nm and 633 nm have been reported (Adamovics and Maryanski 2004, Guo et al 2006, Yates et al 2011). The specific mechanism behind the difference in absorption spectrum with respect to micelle gel dosimeters is unknown but is not surprising as the leucodye is in a significantly different chemical environment. Minimum absorption occurs at 480 nm while a slight increase in absorption is also observed at lower wavelengths. For dual-wavelength readout, optimum wavelengths are therefore 627 nm for the absorption scans and 480 nm for the reference scans. Imaging artifacts as a result of impurities and refractive index mismatch are then compensated while the sensitivity in terms of optical density difference is optimized. Diffuse light sources have a broad spectral distribution which are skewed normal distributed in wavelength. If the ΔOD maps are calibrated by using a set of calibration

Figure 12. Longitudinal and axial sections through ΔOD maps recorded at various time intervals (a) demonstrate a good spatial integrity. From left to right, post irradiation times are 3 h, 3 h 30, 4 h 30 and 6 h. Corresponding profiles are shown in (b). A 2D plot of axial profiles is also shown for images recorded at times 3 h, 3 h 30, 4 h 30, 6 h and 4 d after irradiation (c). The phantoms and calibration vials were stored in the fridge between 6 h and 4 d post-irradiation.
cuvettes of which the absorption is measured at a particular wavelength, a conversion factor needs to be applied that takes into account the spectral distribution of the diffuse light sources. An optimal composition in the LMG/chloroform phase diagram was found for 0.025% (w/w) LMG and 0.5% (w/w) chloroform. The elasticity of the FlexyDos3D dosimeter can be tuned by varying the ratio of resin and curing agent. It is possible that the optimum in LMG and chloroform will also shift by adding the ratio resin/curing agent concentration as a third variable in the chemical phase diagram. The investigation of the effect of the resin/curing ratio on the sensitivity and offset and on the radiophysical properties will be the subject of future studies.

The curing temperature has an effect on the $\Delta$OD offset and the $\Delta$OD-dose sensitivity (slope). Complete curing was achieved after 2 h at 65 °C. Vacuum pulling to remove gas bubbles from the dosimeter during curing did not have a significant effect on the dose response but keeping the dosimeter in anoxic conditions upon curing caused a significant decrease in $\Delta$OD-dose response. This indicates that oxygen also plays a role in the radiation induced oxidation. The PMMA cast material of the cuvettes did not appear to influence the $\Delta$OD-dose response significantly.

The $\Delta$OD offset is established in a time period of approximately 1 h after mixing all ingredients. A post-irradiation reaction appears to occur in a timespan of approximately 100 min when the dosimeter is irradiated and kept at room temperature. On a longer time scale in the order of days, the irradiated dosimeter fades when kept at room temperature. However this fading effect is found to be inhibited by storing the dosimeter at lower temperatures after irradiation. While also the auto-oxidation (unirradiated dosimeter) is slowed down by cooling down the dosimeter, the effect of cooling down on the auto-oxidation is not as pronounced as on the $\Delta$OD-dose sensitivity.

A significant dose rate dependence in the order of 12% in the low dose region [0 Gy, 20 Gy] and 32% in the high dose region [20 Gy, 40 Gy] has been observed for the FlexyDos3D dosimeter. These values of dose rate dependence are in the same order of magnitude as what was previously found for micelle gel dosimeters (Vandecasteele et al 2011) but the dose response and dose rate dependence in micelle gel dosimeters was found linear in the dose interval [0 Gy, 40 Gy]. A low dose rate dependence has been reported for electron beams in PRESAGETM (Guo et al 2006) but only limited data is available on the dose rate dependence of PRESAGETM for photon beams. The dose rate dependence of the FlexyDos3D dosimeter was only studied for the nominal composition. While a similar dose rate dependence in micelle gel dosimeters did not impact significantly on the performance of a clinical IMRT dose verification (Vandecasteele and De Deene 2013d), future research will be needed to reduce the dose rate dependence of the $\Delta$OD-dose response of the FlexyDos3D dosimeter through extensive chemical analysis and optimization of the dosimeter formulation.

No significant energy dependence is observed for the FlexyDos3D dosimeter for photon beam qualities of 6 MV and 18 MV which is similar to other leucodye based dosimeters (Guo et al 2006, Vandecasteele et al 2011) and polymer gel dosimeters (De Deene et al 2006). This result is not unexpected as the photon interactions for both energies are dominated by the Compton effect.

A measurable but not detrimental dependence of the $\Delta$OD-dose response on the temperature during irradiation was found. The temperature dependence of 0.8% per degree Celsius, implies that the temperature distribution within the phantom should be uniform to at least 2 degrees Celsius in order to keep the dose uncertainty well within 2%. Also calibration vials need to be irradiated at a temperature that matches the temperature of the dosimeter phantom. The sensitivity of the dose response to the temperature during irradiation is significantly more pronounced in PRESAGETM (4.6% ·(°C)$^{-1}$) (Guo et al 2006) and micelle gel dosimeters (2% ·(° C)$^{-1}$) (Vandecasteele et al 2011) than in FlexyDos3D (0.8% ·(° C)$^{-1}$).
The dosimeter $\Delta$OD-dose response of the FlexyDos3D dosimeter is also slightly dependent on the temperature during scanning ($0.3\% \cdot (^{\circ}C)^{-1}$). As the dosimeter is normally dispersed in a matching fluid which is equilibrated at room temperature, temperature variations are expected to be small.

In the energy range of [100 kV, 10 MV], the FlexyDos3D dosimeter is tissue equivalent. Slightly higher mass attenuation coefficients are found outside this energy interval which is similar to PRESAGE™ (Brown et al 2008) but more pronounced at lower photon energies. The larger mass attenuation coefficient in the lower photon energy range for the FlexyDos3D dosimeter as compared to PRESAGE™ is attributed to the more significant contribution of photoelectric interactions at the silicon atoms. The spatial integrity is preserved for several days if the dosimeters are stored in a fridge.

A considerable advantage of the FlexyDos3D dosimeter is the ability to modify its' elasticity over a broad range, so that it can mimic deformable organs. The silicone matrix remains strong so that the dosimeter doesn't crack upon large deformations. A pincushion deformation experiments showed that the $\Delta$OD response is not significantly altered under compression.

Future research will be performed to investigate the behavior of the dosimeter under various clinical deformation conditions.

5. Conclusions

A new flexible 3D dosimeter is proposed that can be cast in an anthropomorphic shape and which is easy to fabricate and non-toxic. The new dosimeter can be readout optically using a dual wavelength optical conebeam CT scanner after two hours post-radiation. Compared to other leucodye based 3D dosimeters, the FlexyDos3D dosimeter demonstrates favourable dosimetric properties such as a low dependence of the OD-dose response on temperature during radiation and scanning, tissue equivalence for clinical radiotherapy beams and a good spatial integrity. However, the dosimeter phantoms should be stored in a Fridge after radiation to avoid fading of the dose response. The dose rate dependence of the FlexyDos3D dosimeter is in the same order of magnitude as that of micelle gel dosimeters. A further reduction of the dose rate dependence would improve the accuracy of the dosimeter. The dose response of the FlexyDos3D dosimeter with nominal composition is mono-exponential in the dose range of [0 Gy, 50 Gy] and the dose rate dependence increases with higher doses. The dosimeter has great potential as a dosimetric tool for dose verification in deformed tissue structures as the elasticity can be easily tuned by changing the fraction of resin to curing agent.

Acknowledgments

The authors thank T Gorjiara, E Colvill and A Quinn for their assistance with the irradiation of the samples.

References
