Preliminary evaluation of optical CT scanning versus MRI for nPAG gel dosimetry: the Ghent experience

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Abstract The aim of this study was to evaluate fast laser-scanning optical CT versus MRI for an nPAG gel dosimeter in terms of accuracy and precision. Three small cylindrical volumetric gel phantoms were fabricated and irradiated with photon beams. The gel dosimeters were scanned with an MR scanner and an in house developed laser scanning optical CT scanner. A comparison between MRI and optical CT scanning was performed based on the reconstructed images. Preliminary results show a fair correspondence in the MRI acquired and optical CT acquired dose maps. Still, ringing artifacts contaminate the reconstructed optical CT images. These may be related to sub-pixel misalignments between the blank projection and the acquired transmission projection of the gel phantom. Another artifact may be caused by refraction near the edges of the field. Further optimisation of our optical CT scanner is required to obtain the same accuracy as with MRI. To make a comparison between the two imaging modalities in terms of precision, the intrinsic dose precision on readout (IPD) was calculated which is independent of spatial resolution and acquisition time. It is shown that optical CT has a better intrinsic dose precision.

1. Introduction

In the last decade radiotherapy has focused on shaping the radiation dose distribution according to the shape of the tumour, what is called conformal radiotherapy (IMRT, IMAT, radiosurgery, tomotherapy). As a consequence there are steep dose gradients in three dimensions that pose a serious dosimetric challenge. Several dosimetric techniques are able to record the dose in one or two dimensions. Gel dosimetry is the only technique able to measure the absorbed dose distribution of a complete treatment in three dimensions and integrated over time. During the last years, gel dosimetry has evolved into a more reliable technique. This is due to extensive research into the chemical and physical characteristics of the gel system. Gel dosimeters have been used to validate conformal radiotherapy treatments [1]. A strong advantage of gel dosimetry above other techniques is that the gel dosimeter follows the complete treatment pathway as the patient during his or her treatment.
MRI is the most extensively studied imaging modality for gel dosimetry. Still only a few radiotherapy centres use this valuable dosimeter as an overall control for complex radiotherapy treatment verification. This can be partially attributed to the fabrication procedure, which requires a chemical laboratory. The greatest difficulty to implement gel dosimetry in the first place is the limited access to MR scanners. Also the expertise to get good quantitative images is not always readily available.

As an attempt to make gel dosimetry more accessible, optical computed tomography (optical CT) was developed and being researched for 12 years now. Different generations of optical CT scanners were proposed in the past [2].

In this study, an optical CT scanner was constructed on site which is able to measure 3D polymer gels. The aim was to evaluate the performance in terms of accuracy and precision of the novel optical CT scanning versus MRI for an nPAG gel dosimeter.

2. Materials and methods

2.1. Gel fabrication

The gel consists of a hydrogel matrix of 7% (w/w) gelatin from porcine skin (bloom 300, Sigma-Aldrich, Bornem, Belgium) in which (3% (w/w) N,N'-methylene-bis-acrylamide (Bis) and 3% (w/w) acrylamide (Aam) (both monomers were obtained from Sigma-Aldrich and are of electrophoresis grade) are dissolved. An antioxidant (tetrakis hydroxymethyl phosphonium chloride) is added to the mixture in a concentration of 3mM. The monomers and gelatin were dissolved into equal amounts of the total water volume (87% (w/w)). The monomer solution was heated to 60 °C until everything was dissolved. The gelatin was dissolved in water at room temperature and allowed to swell. Next the solution was heated up to 45 °C, at which the gelatin dissolves. Subsequently, both solutions were cooled down to 30 °C before mixing. After mixing, the antioxidant was added and the solution was thoroughly stirred for 30 s. The gel was transferred into three cylindrical transparent volumetric phantoms and sixteen test tubes that served as calibration vials.

2.2. Gel irradiation

The irradiation was performed using a clinical linear accelerator (SL-18, Elekta, Stockholm, Sweden). Each calibration vial (R: 0.8 cm, h: 10 cm, glass thickness: 0.3 cm) was irradiated separately in a cubic water tank (22x22x22 cm³) with 6 MV photons at a reference depth of 5 cm and with a source-to-surface distance (SSD) of 95 cm. The field size was 10x10 cm².

![Figure 1](image-url) Photographic overview of the three cylindrical phantoms. A radiation beam with a fluence of 1500 MU was given to phantom A. Phantom B was irradiated with two centered square beams of 3x3 cm² and 1x1 cm² respectively, both with a fluence of 1000 MU. Phantom C was irradiated along 6 beam directions. The fluences are shown for each incident beam. The last photograph is a side view of phantom C. The black dotted line represents the field shapes.
The dose rate for all volumetric phantoms was 4 Gy min\(^{-1}\). The axis of the calibration vials was perpendicular to the axis of the beam. Using this method, irradiation doses of 0 Gy to 30 Gy with increment of 2 Gy were delivered in the calibration vials.

The three small cylindrical volumetric gel phantoms (R: 2.6 cm, h: 6.7 cm) (figure 1) were irradiated with 6MV photon beams at an SSD of 100 cm. The field size used to irradiate phantom A was 2x2 cm\(^2\). A radiation beam with a fluence of 1500 MU was given. Phantom B was irradiated with two centered square beams of 3x3 cm\(^2\) and 1x1 cm\(^2\) respectively, both with a fluence of 1000 MU. The axis of both phantoms A and B was parallel to the axis of the beam. For phantom C, the field size was 1.5x1.5 cm\(^2\). Six beam directions were chosen (0°, 60°, 120°, 180°, 240°, 300°) with fluences of respectively 100 MU, 200 MU, 400 MU, 100 MU, 200 MU, 400 MU. The axis of phantoms C was perpendicular to the axis of the beam.

2.3. MR measurements and data processing

Irradiated gels together with the calibration vials were scanned in a Siemens SP 1.5 T whole body scanner using a standard transmit/receive RF head coil. A multiple spin-echo sequence (32 equidistant spin echoes) with a CPMG encoding scheme was used. Imaging parameters were TE = 40-1280 ms, TR = 5000 ms, FOV = 150 mm and MS = 256x256.

In house Matlab (The MathWorks, Natick, USA) code was employed to calculate R2 images from the set of base images. A calibration curve, extracted from the calibration vials, was used to convert the R2 maps of the volumetric gel phantoms to dose maps.

2.4. Optical CT scanner

2.4.1. General overview

An optical CT scanner (figure 2) was constructed in house (OPTOSCAN). We use a 2.0 mW, 632.8 nm HeNe-laser (JDS Uniphase Model 1122p, CA,USA) with a beam diameter of 0.63 mm (1/e\(^2\) points). Before reaching a galvanomirror, the laser beam is guided through two pinhole collimators to sharpen the laser beam size. The galvanomirror reflects the laser beam to a plano-convex lens (Melles-Griot, Albuquerque, USA) which is positioned at focus distance (450mm) from the galvanomirror.

![Figure 2. Left: Schematic basic design of the in house optical laser scanner. The laser beam is pointed at a rotating galvanomirror. Before reaching this galvanomirror the laser beam is guided through two pinhole collimators. The galvanomirror reflects the laser beam towards a plano-convex lens. The laser beam leaving the plano-convex lens travels in parallel lines through the gel dosimeter until it passes another plano-convex lens which guides the laser beam towards the photodetector. Right: Photograph of the OPTOSCAN optical CT scanner.](image)

The laser beam leaving the plano-convex lens travels in parallel lines through the gel dosimeter until it passes another plano-convex lens which guides the laser beam towards the photodetector. The
A photodetector is a large area balanced photoreceiver (Thorlabs, model 2307, Munich, Germany). In between the lenses a small fluid bath (14.2x7.5x 5.5 cm³) is placed. The refractive index matching procedure is performed automatically based on a dry and wet off axis refraction measurement and using an automated pump system with a refractive index matching fluid (water-glycerol solution). Matlab (The MathWorks) code was developed in house to control the optical scanner, construct sinograms and optical density images.

2.4.2. Optical CT scanning protocol
One hundred projections of 128 mm were recorded per slice with a phantom rotation increment of 1.8°. 128 spatial increments were acquired per projection yielding a profile resolution of 1mm. The same region was scanned both optically and with MRI.

2.4.3. Post processing
The measured projections per rotational increment were aligned in the sinogram using an in house Matlab (The MathWorks) script before reconstructing the optical density map. The optical density was calculated using equation 1

\[
OD' = \log_{10} \left( \frac{SB - SD}{S(x, \Theta, z) - SD} \right)
\]  

were SB is the mean of a blank projection, SD is the dark current of the photodetector and \( S(x, \Theta, z) \) represents the signal at position \( x \), angle \( \Theta \) and vertical slice position \( z \).

2.4.4. Comparison between MRI and optical CT
The optical CT measured OD images were scaled to the MRI measured dose maps using a bilinear interpolation. The optical density maps were normalized to the MRI measured dose maps using a conversion factor that is derived from the maximum dose regions in the three volumetric phantoms. In order to make a fair comparison of precision between both imaging modalities, the intrinsic dosimetric precision on readout (equation 2) was calculated, which is independent of spatial resolution and acquisition time.

\[
IF_D = \sqrt{\frac{1}{V} \cdot \frac{1}{\Delta V} \cdot \frac{1}{\Delta \text{meas}}}
\]  

\( V \) is the voxel volume, \( \Delta \text{meas} \) is the total acquisition time and \( D_{\text{rel}}^\% \) is the relative dose resolution (equation 3)

\[
D_{\text{rel}}^\% = \sqrt{2 \cdot k_p \cdot \frac{\sigma_D}{D_{\text{max}} - D_{\text{min}}}}
\]

with \( k_p \) the coverage factor for a 95% confidence interval and \( \sigma_D \) the standard deviation on dose. \( D_{\text{max}} \) and \( D_{\text{min}} \) are the maximum and minimum doses in the image.

To align the resulting images of MRI and optical measurements a rotation was carried out using a bilinear interpolation. A subtraction image was calculated between MRI and optical CT data for every phantom. Cross-profiles through both MRI and optical CT derived dose maps were averaged in the perpendicular direction over 5mm wide.

3. Results

Figure 3 shows the MRI and optical CT derived dose maps. The conversion factor derived from the three phantoms did not differ by more than 1.3%. The OD images were normalized to dose using the average of the three conversion factors.
Figure 3. MRI acquired (top row) and optical CT acquired (second row) dose maps for the tree volumetric phantoms (from left to right: phantoms A, B and C). The bottom row shows the corresponding difference maps. Colorbar is expresses dose in Gy.

For further comparison, cross-profiles were also taken along the center of the irradiated fields for phantom A and phantom B (figure 4).

Figure 4. Lateral cross-profiles through phantoms A (left) and B (right) in both orthogonal directions.

The acquisition time for MRI data was 255 min and for optical CT was 20 min. Acquisition and performance parameters are listed in table 1.

| Parameter | MRI                   | Optical CT               |
|-----------|-----------------------|--------------------------|
| Voxel size (mm³) | 1.72 (0.586 x 0.586 x 5.00) | 1.27 (1.00 x 1.00 x 1.27) |
| Minimum R2 (T⁻¹)/ minimum OD (-) | 1.18 | -5.92 10⁻³ |
| Maximum R2 (T⁻¹)/ maximum OD (-) | 2.38 | 1.43 10⁻² |
| Dₜ₉₅% (%) | 5.96 10⁻³ | 1.09 10⁻⁴ |
| Dₜₔ₁₈% (%) | 1.38 | 1.49 |
| IP₀ (mm³⁻¹/²) | 3.42 10¹ | 1.83 |

Table 1. Acquisition and performance parameters for MRI and optical CT measurements.
4. Discussion

To investigate the overall performance of an in house built fast-laser scanning optical CT scanner, a comparison with a ‘gold standard’ is required. In this study, we have used an nPAG gel dosimeter and MRI to obtain an ‘idea’ of the accuracy and precision of the optical CT scan technique. A fair quantitative correspondence was found between MRI measured dose maps and optical CT measured dose maps. However, from figure 3 and figure 4 it can be seen that some imaging artifacts in the optical CT images still compromise the accuracy.

A dose hotspot is clearly visible inside the square field of phantom A. This may be related to some residual ringing artifacts. The field lines are less sharply delineated in the optical CT images which can also be seen in the subtraction image (figure 3) were the biggest deviation occurs near the edges of the irradiated field. We also observed refraction of the laser beam when grazing the edge of the field. This is probably due to a change in refractive index between the two structures (polymerized and non polymerized gel). This artifact is more pronounced at higher doses which was the case for phantom B.

Phantom C shows the potential of scanning more complex dose distributions. The optical CT data has a high degree of alignment with the MRI data within the irradiated field. Still further work is needed to develop an adequate calibration procedure for optical CT. This could be done using calibration films or with calibration phantoms.

For the intrinsic dose precision on readout (IP$_D$), it is shown that optical CT scanning has a five times better intrinsic dose precision than MRI.

5. Conclusions

A fast laser scanning optical CT scanner was constructed and optimized at the Ghent university. To make an evaluation of the overall accuracy and precision of the optical CT scanner, a comparison was needed with a ‘gold standard’. Therefore an nPAG gel dosimeter was used to make the comparison between MRI and optical CT. Preliminary results show a fair correspondence in the derived dose maps for MRI and optical CT. Still, ringing artifacts contaminate the reconstructed optical CT images. These may be related to sub-pixel misalignments between the blank projection and the acquired transmission projection of the gel phantom. Another artifact may be caused by refraction near the edges of the field. Further optimisation of our optical CT scanner is required to obtain the same accuracy as with MRI. To make a comparison between the two imaging modalities in terms of precision, the intrinsic dose precision on readout (IP$_D$) was calculated. It is shown that optical CT has a better intrinsic dose precision.

References

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