Absolute dosimetry with polymer gels—a TLD reference system

P Mann1,2,3,4, A Schwahofer1,2,4 and C P Karger1,2

1 Department of Medical Physics in Radiation Therapy (E040), German Cancer Research Center (DKFZ), Im Neuenheimer Feld 280, Heidelberg, Germany
2 National Center for Radiation Research in Oncology (NCRO), Heidelberg Institute for Radiation Oncology (HIRO), Im Neuenheimer Feld 280, Heidelberg, Germany
3 Author to whom any correspondence should be addressed.
4 These authors contributed equally to this study.

E-mail: p.mann@dkfz.de

Keywords: radiation dosimeters, absolute dosimetry, 3D dosimetry, polymer gel, thermoluminescence dosimetry, TLD

Abstract

Background and purpose. Absolute dosimetry in 3D with polymer gels (PG) is generally complicated and usually requires a second independent measurement with conventional detectors. This is why, PG are often used only for relative dosimetry. To overcome this drawback, we combine PG with a 1D thermoluminescence (TL) detector within the same measurement. The TL detector information is then used as additional information for calibration of the gel.

Materials and methods. The PAGAT dosimetry gel was used in combination with TLD600 (LiF:Mg,Ti). TL detectors were attached on the surface of the PG container placed inside a cylindrical phantom. To test the usability of this setup, two irradiation geometries were carried out: (a) homogeneous target coverage and (b) small-field irradiation. PG was evaluated with magnetic resonance imaging (MRI) and the TL detectors with a Harshaw 5500 hot gas reader.

Results. PG dosimetry alone showed deviations of up to 4% as compared to calculations. Including additionally the dose information of the TL detectors for PG calibration, this deviation was decreased to less than 1% for both irradiation geometries. This is also reflected by the very high $\gamma_{3D}$-passing rates of $>96\%$ (3%/3 mm) and $>93\%$ (2%/2 mm), respectively.

Conclusion. This study presents a novel method combining 3D PG and TL dose measurements for the purpose of absolute 3D dose measurements that can also be applied in complex anthropomorphic phantoms using only a single measurement. The method was validated for two different irradiation geometries including a homogeneous large field as well as a small field irradiation with sharp dose gradients.

Abbreviations

PG Polymer gel
TL Thermoluminescence
CC Calibration containers
MGC Measured gel container
TTP Time temperature profiles
PAGAT PolyAcrylamide gelatin gel fabricated at atmospheric conditions
rp Renormalization point

1. Background and purpose

Polymer gel (PG) dosimetry has shown to be a valuable tool in radiation therapy (RT) to verify dose in three dimensions (3D) (Baldock et al 2010). Besides optical readout systems (Abtahi et al 2014), magnetic resonance imaging (MRI) is widely used in PG dosimetry to acquire a 3D dose distribution (Venning et al 2005). Although absolute dosimetry with MRI can be performed with an uncertainty of 2.58% (Vandecasteele and De Deene 2013a), the effort is large as both calibration and irradiated volume has to be equally treated throughout the
entire process of (i) gel production, (ii) irradiation and (iii) evaluation (Vandecasteele and De Deene 2013a). While this can be realized for (i) and (iii) by a dedicated workflow (Vandecasteele and De Deene 2013a, Mann 2017), the temperature during irradiation is not always easy to control. The ability to keep the irradiated sample and the calibration containers (CC) at equal temperature strongly depends on the phantom geometry. This holds especially if complex anthropomorphic phantoms are used, where neither the PG container is completely surrounded by water nor a waterflow unit can be utilized (Biederer and Heller 2003, Niebuhr et al 2016) and therefore temperature stability cannot be guaranteed. By this temperature difference, uncertainties are induced which may result in dose deviations of up to 4% (Vandecasteele and De Deene 2013b). Therefore, PG is mostly used for relative dosimetry for a large variety of different phantom settings (Sandilos et al 2004, Venning et al 2005, Ceberg et al 2010) and irradiation techniques including intensity-modulated radiotherapy (IMRT) (Sandilos et al 2004) or measurements in presence of respiratory motion (Ceberg et al 2008). To overcome the drawback of relative dosimetry, the measured dose distribution is usually renormalized to the dose at a reference point calculated by the treatment planning system (TPS) (Ballock et al 2010, De Deene and Vandecasteele 2013). This well-established method shows good results if calculations do not differ significantly from the delivered dose. However, this is not necessarily the case when new motion management techniques are being tested. Examples are (i) gated treatments including residual target motion (Berbeco et al 2005), (ii) gated treatments where the correlation between external surrogate signal and irradiated volume are not necessarily constant (Korremann 2012) or (iii) in general where poor motion management can result in inaccurate dose delivery. To perform absolute PG measurements, we combine the PG with a point dosimeter within the same measurement, which allows a renormalization of the dose distributions based on this point dose information.

For the point dose measurements, we use thermoluminescence dosimetry (TLD) because of the small dimension of the solid crystals. Thermoluminescence (TL) detectors have a long history (McKinlay 1981, Horowitz and Moscovitch 1986, Ferguson et al 1997, Horowitz and Moscovitch 2013) in absolute dose measurements in RT. If TLD is carried out properly, a very high accuracy of about 0.5% can be obtained by calibrating each individual TL detector (Feist 1992). A highly accurate calibration procedure is the most important basis for the use of absolute TLD. This was first established by Feist (1992) and adopted by Schwahofe et al (2016, 2017) in a clinically feasible way. To determine the absolute TL dose, standard protocols are available (ISO 28057 2014).

The purpose of this first study is to demonstrate that absolute dosimetry with PG can be performed accurately when combined with TLD reference measurements. To benchmark this new method against the TPS-based renormalization procedure, only experiments in well-defined static phantom geometries were performed.

2. Materials and methods

All 3D conformal radiation therapy (3DCRT) treatment plans were calculated with the Raystation 6 (RaySearch Laboratories, Stockholm, Sweden) TPS based on computer tomography (CT) scans with a Somatom Flash (Siemens Healthineers, Germany). The TPS uses a collapsed cone dose calculation algorithm with a dose grid resolution of $1 \times 1 \times 1 \text{mm}^3$. Further information about dose calculation accuracy is described elsewhere (Fogliata et al 2007). All irradiations in this study were performed with 6MV photons from a linear accelerator (LINAC, Artiste, Siemens Healthineers, Germany) with a dose rate of 300 MU min$^{-1}$.

The LINAC was commissioned with carefully selected detectors respecting the purpose of measurement (beam profiles, depth dose curves and output factors). The smallest commissioned field size was $1 \times 1 \text{cm}^2$. Therefore the standard conditions of dosimetry for all field sizes were fulfilled. Deviations between measured and simulated depth dose distributions beyond the maximum were $\pm 0.5\%$ (field size $1 \times 1 \text{cm}^2$) and $\pm 0.2\%$ (all other field sizes). All relative data sets are related to the calibration in order to calculate absolute dose to water. The LINAC is calibrated by a Farmer ionization chamber (IC) (TM 30013, PTW Freiburg, Germany) at 5 cm depth using a source-surface distance (SSD) of 95 cm. At the calibration point, 100 MU correspond to 1 Gy and the stability of the LINAC output is checked on a daily base to be within $\pm 1\%$.

2.1. Dosimeter preparation

PG dosimetry was performed using the in-house produced PAGAT (PolyAcrylamide Gelatin gel fabricated at Atmospheric conditions) following the protocol described in De Deene et al (2006). The PG was stored in BAREX containers (BAREX™, VELOX GmbH, Hamburg, Germany) as this material is known to have no significant interactions with the PG itself and especially shows a very low oxygen permeability (Vergote et al 2004). The containers exhibit a volume of 25 ml (length = 45 mm, diameter = 26.8 mm). To acquire a calibration curve, a set of 8 CC were filled with PG from the same batch in addition to the measured gel container (MGC). To avoid volume effects, the same BAREX containers were used for calibration and measurement.

PG containers were stored 24h at 4 °C prior to irradiation and were covered by aluminum foil to prevent light-induced polymerization. A detailed description of the production and storage procedure can be found elsewhere (Mann et al 2017a).
For TLD, 50 rods of TLD600 (LiF:Mg,Ti, Harshaw, Thermofisher scientific, USA) were used in one batch. One batch is always treated with the same thermal processes. The thermal tempering and annealing was carried out with the TLD oven TLD-O (PTW Freiburg, Germany).

The calibration of the TL detectors is carried out by irradiating the TL detectors repeatedly with the same dose, Feist (1992) observed a systematic shift from one to the next calibration irradiation. This indicates that the response of all TL detectors of the same batch decreases or increases on average by about the same amount. Thus, one calibration irradiation is defined as reference measurement and the results of all subsequent calibrations (number of calibrations is \( n = 1 \ldots m \)) are specified in terms of deviations to this reference measurement. The signal shift of all 50 TL detectors of one measurement is averaged (\( \bar{S} \)) and the signal of the individual detectors \( S_{corr} \) is corrected by this averaged shift. Finally, the individual calibration factor for each TL detector \( i \) can be calculated by:

\[
N^{i}_{m, Q_b} = \frac{1}{N} \sum_{n=1}^{m} \left( \frac{D^{IC}_{m, Q_b}}{S^{corr}} \right)
\]

2.2. Irradiation experiments

The PG was removed from the fridge 4 h prior to irradiation to allow thermal adaptation to room temperature at the LINAC. To acquire a calibration curve, CC were irradiated with known doses between 0 to 7 Gy. In the next step, TL detectors were individually inserted into waterproofed PMMA capsules and were attached equidistantly to the surface of the PG container. The combination of gel container and TL capsules was then inserted into a water filled cylindrical phantom (Mann et al 2017b) (figure 1). The phantom was positioned at the LINAC by means of the room lasers with the center of the PG container at the isocenter.

Two different irradiation geometries were used to verify the validity of the combined use of TLD and PG:

(a) A homogeneous irradiation of the PG container and the TL detectors was carried out with two opposing beams (90°, 270°), a field size of 10 × 10 cm² and a dose of 4 Gy. For this scenario, a total of four TL detectors were attached to the surface of the BAREX™ container at 0°, 90°, 180° and 270°.

(b) A small field irradiation resulting in steep dose gradients within the PG was carried out with three equally-spaced beams (0°, 120° and 240°), a field size of 1 × 1 cm² and a maximum dose of 5.17 Gy at the center of the container. For this scenario, a total of six TL detectors were attached to the container (0°, 60°, 120°, 180°, 240°, 300°) by means of a 3D printed holder (VeroClear material, Stratasys, MN, USA) shown in (figure 1). The TL detector-positions were selected to measure both entrance- and exit-dose of each beam. As the TL detector provides only a single point dose, this irradiation was repeated for better statistics.

2.3. Dosimeter evaluation

PGs were imaged 48 h after irradiation on a 3 T Biograph mMR (Siemens Healthineers, Erlangen, Germany) inside a water-flow phantom to ensure constant temperature of 20.0 °C ± 0.1 °C during image acquisition. The PG \( T_2 \) relaxation time was measured using a multi spin-echo CPMG sequence with the following image parameters: TR = 6000 ms, TE = 27.5 – 880 ms, resolution = 1 × 1 × 1 mm³, number of slices = 1; 32 (CC; MGC), BW = 130 Hz/pixel and a total acquisition time of \( T_{sat} = 12 : 48; 76 : 48 \) min (CC; MGC). The calibration curve was calculated using a Matlab (Mathworks Inc., Natick)-based PG evaluation tool (Mann 2017) by fitting the mean relaxation rate \( R_2 = 1/T_2 \) within a circular region of interest (ROI) against the corresponding dose values by using a mono-exponential fit function (Vandecasteele and De Deene 2013c):

\[
R_2 (D) = R_{2,sat} - \Delta R_2 \cdot e^{-\alpha \cdot D}.
\]

In equation (2), \( R_{2,sat} \), \( \Delta R_2 \) and \( \alpha \) are the fit parameters describing the asymptotic relaxation rate at high doses, the difference between this value and the y-intersection of the curve, and slope-related parameter, respectively (Vandecasteele and De Deene 2013c). This procedure allows converting any measured \( R_2 \)-value of the MGC into a corresponding dose value.

For the MGC, \( R_2 \)-maps were calculated from the MR-images. For the purpose of noise reduction, an edge conserving total variation (TV) filter (Rudin et al 1992) was applied. It has been shown previously that this filter has no significant impact on the dose distribution (Mann et al 2017a). Treatment plan and MR data were...
co-registered by means of a point-based 3rd order B-Spline interpolation algorithm using three uniquely defined
dots indicated by external markers (Beekly Medical, Bristol, USA) on the MGC that were visible in both CT and
MR images. This was done with the image processing platform MITK (Nolden et al 2013).

TLD read-out was performed with a hot gas reader (Harshaw 5500, Thermofisher scientific, USA) capable
of automatic sequential read-out of 50 TL detectors. The setting for the time temperature profiles (TTP) was
10 °C s⁻¹ up to 350 °C. The signals of all TL detectors are gathered by a photomultiplier and are stored in an
ASCII file. The ASCII file of the TL read-out was evaluated by an in-house developed software ‘TLD Analyzer’
(Schwahofer 2016). This tool allows calculating the dose of each TL detector within a few seconds.

2.4. Comparison of measured and calculated dose
Prior to the comparison of TPS and PG dose, it is an established procedure in gel dosimetry to renormalize
the PG calibration curve at a suitable point to a known dose value (Gustavsson et al 2003, Ceberg et al 2008,
Vandecasteele and De Deene 2013c). This dose value may be obtained either (i) from the TPS or (ii) by an
independent IC measurement performed by repeating the initial experiment. Using the obtained dose value
$D_1$ and the corresponding $R_2$-value $R_{2,1}$, the calibration curve of the PG is renormalized by maintaining the
intersection with the $y$-axis ($D = 0$ Gy) and the fitting parameter $\alpha$. The new fit parameters $R_{2,\text{sat}}'$ and $\Delta R_2'$ are
then calculated as follows (Mann et al 2017a):

$$\Delta R_2' = \frac{R_{2,0} - R_{2,1}}{e^{\alpha D_1} - 1} \quad (3)$$

$$R_{2,\text{sat}}' = \Delta R_2' + R_{2,0}. \quad (4)$$

Here, $R_{2,0}$ is the intersection of the original calibration curve with the $y$-axis defined by

$$R_{2,0} = R_{2,\text{sat}} - \Delta R_2. \quad (5)$$

Using this renormalized calibration curve, the PG dose distribution is then calculated from the PG $T_2$-map. A
major drawback of method (i) is that the planned dose is assumed to be correct which is not necessarily the case.
With this respect, method (ii) is advantageous as it uses an independently measured dose. For this, however, a
second measurement under identical experimental conditions is required.

To overcome these problems, a simultaneous PG and TLD measurement is proposed (method (iii)). As the
TL detectors are located outside while the renormalization point is located inside the gel, the ratio of the TPS
dose at the renormalization point (rT) $D_{\text{TPS}}^{rT}$, and the TL position ($D_{\text{TL}}^{\text{rT}}$), is multiplied by the measured TL dose ($D_{\text{TL}}$),
to calculate the dose at the renormalization point ($D_{\text{TPS}}$):

$$D_{\text{rT}} = D_{\text{TL}} \frac{D_{\text{TPS}}^{\text{rT}}}{D_{\text{TL}}^{\text{rT}}}. \quad (6)$$

This dose is then used in the same way as in the other methods to renormalize the PG calibration curve.
Practically, $D_{TPS}^{rp}$ was obtained by contouring the TL detectors in the TPS and taking the average dose. For irradiation geometry (a), the average of all four TL detectors was used, whereas geometry (b) was based on the average of the TL detectors that were located at the beam entrance side. For irradiation geometry (a), the renormalization point is located within the homogeneously irradiated region of the MGC, whereas for geometry (b), the maximum of the small field is chosen as renormalization point. A schematic summary of the renormalization procedures using method (i)–(iii) is shown in figure 2.

Renormalized PG 3D dose distributions using both method (i) and method (iii) are finally compared with the original TPS dose distribution using the 3D $\gamma$-map analysis (Low et al 1999) of the commercial software iSoft (PTW, Freiburg, Germany). Passing criteria of 3%/3 mm and 2%/2 mm where dose deviations refer to the local dose, were used excluding doses below 10% of the maximum dose.

3. Results

Table 1 compares the absolute doses measured by the TL detectors with those of the original treatment plan. The deviations were below 0.02% for scenario (a). For scenario (b), a single TLD showed a deviation of 3.1% while all other deviations remained below 1.3%.

Comparison of measured and calculated doses prior and after applying both renormalization methods (i) and (iii) are displayed in table 2.

Prior to renormalization, PG measurements show deviations of up to 4% for both irradiation geometries, which corresponds well with values reported in the literature, if a temperature offset during irradiation between CC and MGC is present (Vandecasteele and De Deene 2013b). By performing either the TPS- or TLD-based renormalization procedure, this deviation was decreased to less than 1%. This is also reflected by the very high $\gamma_{3D}$-passing rates of $>96\%$ (3%/3 mm) and $>93\%$ (2%/2 mm), respectively. Figure 3 displays 2D dose profiles along the sagittal plane. For the irradiation geometry (a), the PG shows a uniform dose throughout the entire volume. For scenario (b), the sharp dose gradients were well reproduced and after renormalization, the measured full width half maximum was 9 mm as compared to the TPS-value of 9.9 mm.

4. Discussion

Performing absolute dose measurements with PG is generally difficult and uncertainties of around 5% are reported (De Deene 2010, Vandecasteele and De Deene 2013d, De Deene and Jirasek 2015). Therefore, the PG-based dose distribution is often renormalized either (i) to the planned dose in a homogeneous region or (ii) to the dose determined by an independent measurement using e.g. an IC in the same phantom (Vandecasteele and De Deene 2013c). While (i) provides only relative dose measurements, (ii) still reveals an independent absolute measurement of the 3D-dose distribution. However, this requires a separate measurement under identical conditions, as the IC cannot be inserted into the MGC without inactivating the PG. In general, it is desirable to obtain the absolute 3D dose distribution in a single measurement.

With the proposed workflow, we successfully combined for the first time a 1D TLD-based detector system and a 3D PG dose measurement. This combination allows performing a PG renormalization based on a single measurement. For this, the TL-measurements proved to be highly accurate with an uncertainty of only 0.62%. The measured TL-dose is then used to determine the dose at the renormalization point using only the information of the relative TPS dose distribution. After renormalization, a very good agreement between PG and TPS in terms of absolute dose was obtained as shown by the high $\gamma_{3D}$ passing rates (table 2). Furthermore, the results of this study are well-comparable to the TPS-based renormalization method (i). This is due to the fact that the experiments were performed in well-defined static phantom geometries, hence deviations between measured and calculated doses were not expected. This, however, may not be the case in presence of motion or other disturbing factors, if they are not explicitly taken into account in the dose calculation. This is an advantage of normalization method (iii) over method (i).

A limitation of this method, however, is that the TL detectors are not located within the PG. Thus, the determination of the dose at the renormalization point relies on the correctness of the TPS dose ratio $D_{TPS}^{rp}/D_{TL}^{rp}$ for the selected renormalization point. The adequate choice of this point, in turn, depends on the irradiation setup and can generally be divided into two cases: (a) homogeneous irradiation of the entire ROI consisting of the MGC and TL detector, and (b) small-field irradiations where sharp dose gradients are located within the PG volume. For (a) the renormalization point in the MGC should be located close to the TL detector used for the calculation of $D_{TPS}^{rp}$. For (b) this point should be located at the dose maximum or homogeneous regions, but not within steep dose gradients as the renormalization method is very sensitive to positioning errors in the latter case. Hence, a highly accurate positioning of the phantom is required. However, these positioning errors are likely to be smaller than errors caused by a repeated phantom setup in the case of renormalization by an independent IC measurement.
Furthermore, it is important to notice that both MGC and CC are of same size and therefore no volume effects are expected. However, if the MGC is significantly larger than the CC, this could result in an inhomogeneous dose response (De Deene and Jirasek 2015) and as in the case of renormalization method (i) and (ii), also the TLD-based renormalization method (iii) is not expected to compensate for this. Therefore, we suggest that the proposed method is only used when MGC and CC do not differ in size.

The combination PG and TLD together with the proposed renormalization method provides the possibility to use PG in more complex phantom settings, e.g. anthropomorphic (Miyamoto et al 2011, Niebuhr et al 2016) or dynamic phantoms (Biederer and Heller 2003, Court et al 2010, Ehrbar et al 2016) where exact repetitions of the experiment for renormalization purpose is usually not possible. Moreover, this method could be especially useful for experiments testing new motion management techniques. In these cases, the TPS-based normalization of the PG (i) could be inaccurate due to blurred dose distributions. This would not be the case for the proposed method as the delivered dose is measured directly by the attached TL detectors.

The proposed measurement method is well-suited for end-to-end testing when new treatment techniques are implemented. Moreover, the response of PG and TL detectors during irradiation have been shown to be only marginally influenced by external magnetic fields (Lee et al 2017, Green et al 2018). The described method may

Table 1. Comparison of the averaged TL detector and TPS doses for the two irradiation geometries. Results are shown for the homogeneous (a) and the small-field irradiation (b) compared with the doses calculated by the TPS. ∆ exhibits the relative difference between measured and calculated doses. The calibration dependent uncertainty of all TL measurements in this study is ±0.62%.

(a) Homogeneous irradiation

| Gantry angle | 0  | 90 | 180 | 270 | Mean |
|--------------|----|----|-----|-----|------|
| $D_{TPS}$ [Gy] | 4.01 | 4.02 | 4.00 | 4.00 | 4.01 |
| $D_{TL}$ ($n = 1$) [Gy] | 4.01 | 3.99 | 4.01 | 4.03 | 4.01 |
| ∆ (1 SD) [%] | < 0.01 | 0.02 | 0.01 | 0.02 | < 0.01 |

(b) Small-field irradiation

| Gantry angle | 0  | 60 | 120 | 180 | 240 | 300 |
|--------------|----|----|-----|-----|-----|-----|
| $D_{TPS}$ [Gy] | 1.83 | 1.47 | 1.94 | 1.47 | 1.94 | 1.47 |
| $D_{TL}$ ($n = 1$) [Gy] | 1.85 | 1.46 | 1.93 | 1.52 | 1.92 | 1.48 |
| ∆ (1 SD) [%] | 0.93 | −0.36 | −0.40 | 3.08 | −1.31 | 0.73 |

Furthermore, it is important to notice that both MGC and CC are of same size and therefore no volume effects are expected. However, if the MGC is significantly larger than the CC, this could result in an inhomogeneous dose response (De Deene and Jirasek 2015) and as in the case of renormalization method (i) and (ii), also the TLD-based renormalization method (iii) is not expected to compensate for this. Therefore, we suggest that the proposed method is only used when MGC and CC do not differ in size.

The combination PG and TLD together with the proposed renormalization method provides the possibility to use PG in more complex phantom settings, e.g. anthropomorphic (Miyamoto et al 2011, Niebuhr et al 2016) or dynamic phantoms (Biederer and Heller 2003, Court et al 2010, Ehrbar et al 2016) where exact repetitions of the experiment for renormalization purpose is usually not possible. Moreover, this method could be especially useful for experiments testing new motion management techniques. In these cases, the TPS-based normalization of the PG (i) could be inaccurate due to blurred dose distributions. This would not be the case for the proposed method as the delivered dose is measured directly by the attached TL detectors.

The proposed measurement method is well-suited for end-to-end testing when new treatment techniques are implemented. Moreover, the response of PG and TL detectors during irradiation have been shown to be only marginally influenced by external magnetic fields (Lee et al 2017, Green et al 2018). The described method may
therefore be used in dynamic anthropomorphic phantoms to perform end-to-end test of adaptive treatments at MR-Linac devices. This would allow benchmarking the absolute 3D dose distributions with high accuracy.

5. Conclusion

This study presented a novel method combining 3D PG and TL dose measurements for the purpose of absolute 3D dose measurement that can also be applied in complex anthropomorphic phantoms. The method used TL detectors to renormalize the PG calibration curve and shows high accuracy of absolute dosimetry with PGs. The method was validated for two different irradiation geometries including a homogeneous large field as well as a small field irradiation with sharp dose gradients.
Conflict of Interest

One of the authors, Dr Mann is shareholder of the company HQ-Imaging.

Acknowledgments

This work has received funding from the EMPIR programme co-financed by the Participating States and from the European Union’s Horizon 2020 research and innovation programme under Grant No. 15HLT08. Additionally, the authors would like to thank Professor Dr K Kopka for providing the lab space for the production of the polymer gel.

References

Abtahi S M, Aghamiri S M R and Khalafi H 2014 Optical and MRI investigations of an optimized acrylamide-based polymer gel dosimeter J Radioanal. Nucl. Chem. 300 287–301

Ballock C et al 2010 Polymer gel dosimetry Phys. Med. Biol. 55 R1–63

Berbeco R, Nishoka S, Shirato H, Chen G and Jiang S 2005 WE-C-J-6C-01: residual motion of lung tumors in gated radiotherapy with external respiratory surrogates Med. Phys. 32 2124–4

Biederer J and Heller M 2003 Artificial thorax for MR imaging studies in porcine heart-lung preparations Radiology 226 250–5

Ceberg S et al 2010 Tumor-tracking radiotherapy of moving targets; verification using 3D polymer gel, 2D ion-chamber array and biplanar diode array J. Phys.: Conf. Ser. 250 012051

Ceberg S, Karlsson A, Gustavsson H, Wittgreen L. and Back S A J 2008 Verification of dynamic radiotherapy: the potential for 3D dosimetry under respiratory-like motion using polymer gel Phys. Med. Biol. 53 N387–96

Court L E et al 2010 Use of a realistic breathing lung phantom to evaluate dose delivery errors Med. Phys. 37 5830

De Deene Y 2010 How to scan polymer gels with MRI J. Phys.: Conf. Ser. 250 012015

De Deene Y and Jirasek A 2015 Uncertainty in 3D gel dosimetry Radiology

De Deene Y and Vandecasteele J 2013 On the reliability of 3D gel dosimetry J. Phys.: Conf. Ser. 444 012015

De Deene Y, Vergote K, Claey S Cand De Wanger C 2006 The fundamental radiation properties of normoxic polymer gel dosimeters: a comparison between a methacrylic acid based gel and acrylamide based gels Phys. Med. Biol. 51 653–73

Ehrbar S et al 2016 Respiratory motion management in stereotactic body radiation therapy for lung cancer—a dosimetric comparison in an anthropomorphic lung phantom (LuCa) Radiother. Oncol. 121 328–34

Feist H 1992 Entwicklung der Thermolumineszenzdosimetrie mit LiF zu einer präzisionsmethode für absolute energiedosisbestimmungen in der strahlentherapie mit photonen- und elektronenstrahlung mit hoher energie Habilitation Thesis (Munich: Medical Faculty Ludwig-Maximilians—University Munich)

Ferguson H M, Lambert G D and Harrison R M 1997 Automated TLD system for tumor dose estimation from exit dose measurements in external beam radiotherapy Int. J. Radiat. Oncol. Biol. Phys. 38 899–905

Fogliata A et al 2007 On the dosimetric behaviour of photon dose calculation algorithms in the presence of simple geometric heterogeneities: comparison with Monte Carlo calculations Phys. Med. Biol. 52 1363–85

Green O L et al 2018 First clinical implementation of real-time, real anatomy tracking and radiation beam control Med. Phys. 45 3728–40

Gustavsson H et al 2003 MAGIC-type polymer gel for three-dimensional dosimetry: intensity-modulated radiation therapy verification Med. Phys. 30 1264–71

Horowitz Y and Moscovitch M 1986 LiF-TLD in the microgray dose range via computerised glow curve deconvolution and background smoothing Radiat. Prot. Dosim. 17 337–42

Horowitz Y and Moscovitch M 2013 Highlights and pitfalls of 20 years of application of computerised glow curve analysis to thermoluminescence research and dosimetry Radiat. Prot. Dosim. 153 1–22

ISO 28057 2014 Dosimetry with Solid Thermoluminescence Detectors for Photon and Electron Radiations in Radiotherapy (Geneva, Switzerland: International Organization for Standardization) (https://www.iso.org/standard/44482.html)

Korremann S 2012 Motion in radiotherapy: photon therapy Phys. Med. Biol. 57 R161–91

Lee H J, Roed Y, Venkataraman S, Carroll M and Ibbott G S 2017 Investigation of magnetic field effects on the dose–response of 3D dosimeters for magnetic resonance—image guided radiation therapy applications Radiat. Oncol. 125 426–32

Low D A et al 1999 Evaluation of polymer gels and MRI as a 3D dosimeter for intensity-modulated radiation therapy Med. Phys. 26 1542–51

Mann P 2017 Development and implementation of 3D dosimetric end-to-end tests in adaptive radiation therapy of moving tumors PhD Thesis (Heidelberg: Combined Faculties for the Natural Sciences and for Mathematics Ruperto-Carola University of Heidelberg)

Mann P et al 2017a 3D dosimetric validation of motion compensation concepts in radiotherapy using an anthropomorphic dynamic lung phantom Phys. Med. Biol. 62 573–95

Mann P et al 2017b Validation of 4D dose calculation using an independent motion monitoring by the calypso tracking system and 3D polymer gel dosimeter J. Phys.: Conf. Ser. 847 012040

McKinlay A F 1981 Medical Physics Handbooks vol 5 (Bristol: Adam Hilger Ltd) pp 1–59

Miyamoto N et al 2011 Optimization of fluoroscopy parameters using pattern matching prediction in the real-time 3D tumor-tracking radiotherapy system Phys. Med. Biol. 56 4803–13

Niebuhr N et al 2016 Technical note: radiological properties of tissue surrogates used in a multimodality deformable pelvic phantom for MR-guided radiotherapy Med. Phys. 43 908–16

Nolden M et al 2013 The medical imaging interaction toolkit: challenges and advances Int. J. Comput. Assist. Radiol. Surg. 8 607–20

Radin L I, Osher S and Fatemi E 1992 Nonlinear total variation based noise removal algorithms Physica D 60 259–68

Sandilos P et al 2004 Dose verification in clinical imrt prostate incidents Int. J. Radiat. Oncol. Biol. Phys. 59 1540–7

This work has received funding from the EMPIR programme co-financed by the Participating States and from the European Union’s Horizon 2020 research and innovation programme under Grant No. 15HLT08. Additionally, the authors would like to thank Professor Dr K Kopka for providing the lab space for the production of the polymer gel.
Schwahofer A 2016 Establishing a process for the use of thermoluminescence dosimetry (TLD) in radiotherapy—an approach concerning the entire range of diagnostic and therapeutic photon energies PhD Thesis (Heidelberg: Medical Faculty of Ruprecht-Karls-University Heidelberg)

Schwahofer A, Feist H, Georg H, Häring P and Schlegel W 2017 Experimental determination of the photon-energy dependent dose-to-water response of TLD600 and TLD700 (LiF:Mg, Ti) thermoluminescence detectors Z. Med. Phys. 27 13–20

Vandecasteele J and De Deene Y 2013a On the validity of 3D polymer gel dosimetry: III. MRI-related error sources Phys. Med. Biol. 58 63–85

Vandecasteele J and De Deene Y 2013b On the validity of 3D polymer gel dosimetry: II. Physico-chemical effects Phys. Med. Biol. 58 43–61

Vandecasteele J and De Deene Y 2013c On the validity of 3D polymer gel dosimetry: I. Reproducibility study Phys. Med. Biol. 58 19–42

Vandecasteele J and De Deene Y 2013d Evaluation of radiochromic gel dosimetry and polymer gel dosimetry in a clinical dose verification Phys. Med. Biol. 58 6241–62

Venning J, Hill B, Brindha S, Healy B J and Baldock C 2005 Investigation of the PAGAT polymer gel dosimeter using magnetic resonance imaging Phys. Med. Biol. 50 3875–88

Vergote K et al 2004 Validation and application of polymer gel dosimetry for the dose verification of an intensity-modulated arc therapy (IMAT) treatment Phys. Med. Biol. 49 287–305