On the accuracy and precision of gel dosimetry

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1. Introduction

Gel dosimeters are radiation sensitive chemical dosimeters that aim at displaying the absorbed dose in three dimensions (3D) and integrated over time. In that perspective are gel dosimeters unique. This makes it also difficult to evaluate the gel dosimeter in terms of accuracy as there is no comparable 3D dosimeter that can be used as a “gold standard” for the evaluation of the gel dosimeter. Accuracy can be defined as the degree of conformity of a measured or calculated quantity to its actual (true) value. Precision can be defined as the ability of a measurement to be consistently reproduced. Accuracy and precision can be defined statistically in terms of the deviation of the mean value from the reference value and the standard deviation of the mean value of many measurements respectively (figure 1). In the case of gel dosimetry, accuracy and precision can be evaluated in terms of dose and space. In the end result of a gel dosimetry experiment (and any radiation treatment), the spatial and dosimetric dimensions are interwoven. In the measured spatial dose distribution (the result of a 3D gel dosimetry

![Figure 1](image_url)

**Figure 1.** Accuracy and precision shown schematically in terms of probabilities of occurrence. Precision can be characterised in terms of the standard deviation of many measurements (experiments). The accuracy is characterized by the deviation of the mean of many measurements from the reference value.

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experiment), it is theoretically impossible to extract both dosimetric and spatial errors. To encompass both spatial and dosimetric performance in one parameter, Low et al have introduced the gamma-index [1]. Other dose distribution comparisons are performed on the basis of maximum allowed dose differences [2].

2. Error sources

Various sources can lead to a loss in both accuracy and precision at different levels of the measurement process (see table 1). To evaluate the performance of a 3D gel dosimeter, the accuracy and precision can be evaluated at different stages of the gel dosimetry experiment (figure 2).

The prescribed dose distribution as obtained from the radiation treatment planning (D_{presc}) is presented to the radiation unit (Linac, gammaknife, afterloader, …) and the radiation is delivered to the gel dosimeter resulting in a delivered dose distribution in the gel dosimeter (D_{deliv}). In the gel dosimeter a dose-dependent polymerization reaction takes place which is results in a spatial distribution of polymer (Polym). The gel dosimeter is read out by use of a non-invasive imaging technique (MRI, Optical CT, X-ray CT, ultrasound,…). The result is a parametric map (Ω) that reflects the absorbed dose distribution. The parametric map displaying a physical quantity (R1, R2, MT, optical density, CT numbers, speed of sound, …) is converted to a dose map by use of a calibration curve (Ω(D)) that is obtained by passing through all of the above-mentioned stages but starting of with a known dose or dose distribution (D_{presc} = D_{cal}).

When a gel dosimeter is subjected many times to a radiation treatment, there will be stochastic deviations in both radiation output (ε(D)) and spatial errors (ε(x,y,z)) that may result in both spatial and dosimetric deviations in the resulting dose distribution. The magnitude of the positioning deviations will depend on the robustness of the positioning operation. Sophisticated methods to accurately position the gel dosimeters apply stereotactic frames and fiducial markers [3]. Note that any positioning error during the irradiation of calibration phantoms may also give rise to dosimetric errors in the final dose distribution. It has been shown that in some gel dosimeters the dose response depends on the temperature of the gel dosimeter during irradiation and thus temperature variations may introduce dose variations (ε(T)).

The response of the gel dosimeter is susceptible to variations in different parameters during fabrication of the gel. Differences in the temperature treatment during (ε(T_{fabr})) and after (ε(T_{storage})) fabrication of the gel dosimeter as well as variations in the concentration of the chemical components (ε(chemical concentrations)) may result in differences in the dose response. Most of these deviations

![Diagram of Gel Dosimetry Process](image)

**Figure 2.** Gel dosimetry is performed in different stages. At each stage errors can add, leading to a decrease in the overall precision and accuracy.
are compensated by using calibration phantoms that are constructed from the same batch of gel. However, it may be difficult to keep the temperature history after fabrication ($\varepsilon(T_{\text{storage}})$) similar for both the gel dosimeter phantom and the calibration vials because of the differences in phantom size.

Upon irradiation, a complex set of radiation-induced chemical reactions take place. On a molecular level, these reactions are probabilistic in nature ($\varepsilon(\text{radiochemical})$). In most gel dosimetry applications the voxel-size is several orders of magnitude bigger than the molecular size. As a result, this intrinsic radiochemical noise contribution can be easily neglected.

After irradiation, the gel dosimeter is scanned. During scanning, detector (thermal) noise ($\varepsilon(\text{thermal})$) will add to the measurements. The processing (fitting) of acquired data, may have a big influence on the amplification of the noise figure. For example, in quantitative R2-NMR scanning it is found that a least-square fit will amplify the thermal noise in the base images to a larger extent than a chi-square based minimization. Imaging artefacts may result in systematic errors. Imaging artefacts can result in dosimetric errors and in geometrical distortions.

**Table 1.** Influencing factors for precision and accuracy in 3D gel dosimetry under conditions of good practice. The components that make up the dose resolution are marked in italic font.

|             | PRECISION                      | ACCURACY                      |
|-------------|--------------------------------|-------------------------------|
|             | dosimetric | spatial | dosimetric | spatial |
| chemical    | Variation in chemical conc. (weighting) and manufacturing temperature | • Spatial variations in manufacturing temperature | • Discrepancies between calibration vials and phantom |
|             | Dose sensitivity of the gel dosimeter | | • Chemical stability |
|             | | | • Spatial integrity |
| radiation   | • Stochastic variations in the delivered dose | • Variations in the phantom position | • Positioning error of the calibration phantom |
|             | • Variations in the temperature during radiation | • Spatial temperature variations in combination with temp. sensitive dose response | • Dose-rate dependent response |
|             | • Reproducibility of calib. phantom-positioning | | • Energy-dependent response |
|             | • Radiochemical noise | | • Temperature dependence |
|             | | | • Tissue equivalence |
|             | | | • Recipient wall effects |
| scanning    | • Stochastic noise | • Voxel size / shape (resolution) | • Voxel shape (bandwidth) |
|             | | | • Imaging artefacts |
|             | | | • Temperature during scanning |
|             | | | • Imaging artefacts |
To convert the measured physical quantity (R2, MT, OD, etc.) to dose a calibration is performed. The calibration phantom has been subjected to the same pathway as the 3D gel dosimetry phantom. As a result, the calibration dose-response curve will also be susceptible to the same error sources. After calibration, the calibration error (e(calibration)) will add to the overall dose error.

3. Precision

Theoretically, the overall precision of the gel dosimeter would be measured by performing a reproducibility study of several 3D gel dosimetry experiments. Such a study would also include inter-operator variability. Due to the infinite number of different radiation treatment configurations, such a study is unfeasible.

The overall dosimetric precision is governed by variations in the several operations that take place in the dosimetry experiment. The first step in a gel dosimetry experiment is weighing the chemicals. Stochastic variations in the weighting will result in variations in the measured dose-related value (R2, MT, OD) as the dose-response is determined by the chemical composition. It is found that other manufacturing conditions may also have an influence on the dose-response such as the temperature during fabrication. Stochastic variations in the controlled temperature will therefore also lead to variations in the measured dose-related value. Also during irradiation there are different sources of stochastic variable contributions that determine the overall dosimetric precision such as variations in the dose delivery, variations in the temperature during irradiation and stochastic variations in the positioning of the calibration phantoms. Any form of scanning the gel dosimeter will introduce thermal detector noise. The noise contribution is determined by some scan parameters. Often, the scan parameters can be optimized in order to achieve an optimal figure of precision.

The concept of dose-resolution was introduced to evaluate the intrinsic dosimetric precision in terms of dose sensitivity and scanning signal-to-noise [4]. The dose resolution, written as \( \Delta D \), is defined as the minimal detectable dose difference within a given level of confidence, \( p \). The dose resolution is related to the standard deviation on dose \( \sigma_D \) by the equation

\[
D_p = k_p \sqrt{2} \sigma_D
\]

For a 95\% confidence level the dose resolution becomes \( D_\Delta = 2.77 \sigma_D \).

In most radiation dosimetry experiments, gel dosimeters are used in a relative manner in the sense that the dosimeter is exposed to the same treatment as the patient but with a different total radiation dose. The total dose delivered to the dosimeter is scaled to cover the active dose range of the dosimeter. In this context, it is preferable to use the concept of dose resolution relative to the operating dose range, here defined as relative dose resolution \( \Delta \% D \):

\[
D_{\Delta \%D} = \frac{D_p}{(D_{max} - D_{min})} = \sqrt{2} \cdot k_p \left( \frac{\sigma_D}{D_{max} - D_{min}} \right)
\]

If the dose maps are derived from quantitative NMR-R2 maps, it can be shown that the relative dose resolution \( D_{\Delta \%D} \) is equal to the relative R2 resolution \( R2_{\Delta \%D} \) which is defined in a similar way:

\[
D_{\Delta \%D} = \sqrt{2} \cdot k_p \left( \frac{\sigma_D}{D_{max} - D_{min}} \right) = \sqrt{2} \cdot k_p \left( \frac{\sigma_{R2}}{R2_{max} - R2_{min}} \right) = R2_{\Delta \%D}
\]

It should be noted that dose-resolution does not include stochastic variations in chemical concentrations, in dose delivery or in the calibration procedure. For that reason, dose-resolution can be considered as an intrinsic lower limit of dosimetric precision.
It should be emphasized that the dose-resolution is not only related to the type of gel dosimeter but also to the scanning protocol \cite{5, 6}. In some publications, the concept of dose resolution has been used as the criteria to compare different types of gel dosimeters. This may be misleading as these studies report on dose resolutions obtained with sub-optimal scanning parameters. The concept of dose resolution however is very practical to optimize the intrinsic NMR sequence in terms of intrinsic precision \cite{7}. In optimizing the NMR sequence, it is also important to take into account the number of slices that are required for the 3D dosimetry application as the optimization is also dependent on the number of slices.

Depending on the scanning facilities three different optimization strategies can be followed:

1. **Only a single spin-echo sequence is available:** It can be shown that if one has only a single spin-echo sequence a two-points method is preferred. In the two-points method, two differently T2-weighted base images are acquired and the R2 is derived algebraically from the signal intensity in the two images according to the equation:

\[
R2 = \frac{\ln S_1 - \ln S_2}{TE_2 - TE_1}
\] (4)

with \(S_i\) and \(TE_i\) the signal intensity and the echo time in image \(i\) respectively.

The relative dose resolution is then given by:

\[
D_{\Delta r}^\beta \approx \frac{\sqrt{2k_T}}{SNR_1 \Delta TE (R_{2\text{max}} - R_{2\text{min}})} \left\{ \frac{e^{2R2 \Delta TE}}{(1 - \psi_N)} + \frac{1}{\psi_N} \right\} \frac{1}{N_1 + N_2}
\] (5)

\(SNR_1\) is the signal-to-noise ratio in the first base image which is the image recorded with echo time \(TE_1\). The echo time spacing \(\Delta TE = TE_2 - TE_1\).

Equation 5 also applies to the case where more acquisitions are taken in each image. The acquisition fraction is \(\psi_N = N_1 / (N_1 + N_2)\) with \(N_1\) the number of acquisitions (averages) of the first image (with \(TE_1\)) and \(N_2\) the number of acquisitions (averages) of the second image (with \(TE_2\)).

An optimal echo time spacing and acquisition fraction can be derived for a gel with R2 values in a range \([R_{2\text{min}}, R_{2\text{max}}]\). In publication \cite{7}, table 1 and 2 give the optimal echo time spacing and the optimal acquisition fraction for the two-points method.

2. **A multiple spin-echo sequence is available with a fixed number of echoes:** To acquire quantitative R2 images, a multiple spin-echo sequence is highly preferred above a single spin-echo sequence because several differently T2-weighted images are acquired within the same measurement time increasing the overall signal-to-noise ratio in the R2-images. The quantitative R2-image is then obtained by fitting a mono-exponential decay-function to the pixel intensities of corresponding pixels. On many scanners, a multiple spin-echo sequence is available on the scanner. Very often, the number of available echoes is ‘hardcoded’ in the sequence. In that case, it is advisable to optimize the echo time interval. The echo time interval is defined as the time interval between the first and the last echo in the multiple spin-echo acquisition window. The optimal echo time interval can be derived from a graph provided in figure 9 of reference \cite{5}. It is also found that the optimization also depends on the fitting algorithm that is used to derive the R2 value. For more than 7 echoes and using a chi-square minimization fitting algorithm, the optimal echo time interval is approximately 2 times the T2 value of the sample. To a good approximation this also applies to the median R2 for a phantom containing a range of R2 values.
3. A multiple spin-echo sequence is available with an arbitrary number of echoes:

This is the most optimal situation from the perspective of optimization of the dosimetric precision when more slices are acquired. The reasoning behind this is based on the fact that within an optimal echo time recording period it is preferable to acquire as many echoes as possible. To optimise this sequence, the echo time spacing is taken as short as possible without introducing any artefacts. Then the number of echoes is optimized. Most often, there is an upper threshold on the available echoes in the multiple spin-echo sequence. When the recommended (optimal) number of echoes exceeds the number of available echoes, the echo time spacing is increased to cover the optimal echo time interval. In publication [7], table 5 gives the optimal number of echoes for a gel with R2 values in a range [R2min, R2max]. The number of echoes is limited by the minimal inter-echo-time-spacing which is determined by machine related characteristics such as maximum gradient strength, sampling rate and SAR considerations. In practice, it is also advisable to check for imaging artifacts (uniformity, dose errors and geometrical distortions) while decreasing the inter-echo-time-spacing.

![Figure 3](image-url)

**Figure 3.** Multi-slice multi-echo sequence showing the optimal sequence parameters. The number of spin-echoes acquired for each slice is optimized.

Then, if more than one slice is acquired, the time interval between the end of the echo-time recording period and the repetition time can be used to scan another slice (figure 3).

As from a certain number of slices, not all slices can be recorded within one repetition time. In that case, the repetition time can be extended to cover exactly the time needed to acquire all slices. The algorithm to determine the optimal echo time looks as provided in figure 4.

The dose-R2 curve is used to calibrate the R2 map. The uncertainty on the dose value $\sigma_{D^*)}$ extracted from the linear dose-R2 plot with equation $R2 = R2_o + \alpha.D$ is given by

$$
\sigma_{D^*} = \frac{\sigma_e}{\alpha} \cdot \sqrt{\frac{(D^* - \overline{D})^2}{N_{cal}} + \frac{1}{N_{cal}}} 
$$

(7)
Figure 4. Flow-chart of the algorithm to determine the optimal number of spin-echoes in a multi-slice multi-echo sequence. The minimum inter-echo-time spacing and the minimum repetition time are imposed by scanner related limitations. The range of R2-values is gel specific (i.e. determined by the dynamic dose range and dose sensitivity of the gel). For the total R2-range, an optimal inter-echo-time can be derived mathematically along the equations provided in reference 5. If the total calculated echo-time recording period for all slices is smaller than the repetition time the echo-time recording period is artificially extended to cover the total repetition time. If the total calculated echo-time recording period for all slices extends the minimum repetition time, it is the repetition time that is increased to the total calculated echo-time recording period.

with $\sigma_c$ the standard deviation on R2 in the calibration points [5]. This value is derived from the standard deviation in a region of interest of the calibration vials $\sigma_{ROI}$. If $N_{ROI}$ is the number of points in the region of interest, the standard deviation on the calibration point is given by $\sigma_c = \sigma_{ROI} / \sqrt{N_{ROI}}$.

$D^*$ is the estimated dose, $\overline{D}$ is the mean dose of all dose values in the calibration plot ($\overline{D} = \frac{1}{N_{cal}} \sum_{i=1}^{N_{cal}} D_i$) with $D_i$ the dose in the $i^{th}$ calibration point and $N_{cal}$ the number of calibration points.

4. Accuracy

The problem in evaluating the final accuracy of the dose maps obtained with gel dosimetry is that there is no such thing as a “golden standard” to compare with. The most reasonable strategy is to compare doses obtained with gel dosimetry with doses obtained by the “most reliable” dosimetry techniques that apply to a certain spatial dimension. As such, dose profiles of a single field (photons...
and electrons) can be compared with dose profiles obtained with an ionization chamber or diamond detector [8-9]. In two dimensions, gel dosimetry can be compared with film dosimetry [8,10-11]. Dose distributions obtained with gel dosimetry have been compared with the outcome of treatment plans [3,10,12-17]. The verification of the treatment plan can be seen as the most important application of gel dosimetry in radiotherapy quality assurance so far.

Factors that have an influence on the accuracy are listed in table 1. These factors can be classified in two categories: (1) Dosimetric factors cause deviations between the measured dose and the described dose. (2) Spatial deviations cause deviations in the spatial distribution of the delivered dose. At the stage of the chemistry of the gel dosimeter, inaccuracies arise from differences in dose-response between calibration phantoms and the dose verification phantom (1a), from chemical instabilities (1b) and from the loss of spatial integrity (1c). At the stage of the radiation delivery other factors may have an influence on the inaccuracy such as a positioning error of the calibration phantom (1d), a dose rate dependent response (1e), an energy dependent response (1f), a temperature dependent response (1g), tissue non-equivalence (1h) and recipient wall-effects (1i). During scanning, dose inaccuracies originate from the imaging voxel shape (1j), dose-related imaging artifacts (1k) and a scanning temperature dependent response (1l). Spatial inaccuracies are attributed to a volume change of the gel dosimeter at the chemical level (2a), a phantom positioning error (2b) during radiation and imaging artifacts (2c) during scanning. Controlling all these factors may largely enhance the accuracy of the gel dosimetry experiment.

1. Sources of dosimetric inaccuracy

1a. Differences in dose response between calibration phantoms and the dose verification phantom

Several groups have observed dose deviations between calibration vials and larger phantoms originating from the same batch of gel. Different potential causes were hypothesized for this phenomenon. In a Monte Carlo study by Michael et al [18] it was shown that the effect of backscatter was negligible. To our experience, the deviations between small calibration vials and larger phantoms are specific to the kind of gel formulation. It was also found that the deviations were not very much reproducible. These findings point in the direction of a physico-chemical factor during fabrication such as a temperature history (cooling rate) of the gel dosimeters after fabrication or traces of oxygen that adhere to the wall of the recipients. Note that the cooling rate of gel dosimeters is different for small phantoms as compared to large phantoms. Typically, in test tubes (13 mm diameter), polymer gel is solid within 10 minutes when stored in a fridge while it may take up to several hours for polymer gel to become solid in large phantoms. Please also note the abstract by Dumas et al elsewhere in this issue.

1b. Chemical instabilities

Two kinds of chemical instabilities have been observed in polymer gel dosimeters [19, 20]. One affects the slope of the dose–R\textsuperscript{2} plot and is related to post-irradiation polymerization of the comonomer/polymer aggregates. It is observed that post-irradiation polymerization only lasts 12 hours after irradiation. The other instability affects the intercept of the dose–R\textsuperscript{2} plot, lasts for up to 30 days and is related to the gelation process of gelatin. The chemical instability depends on the polymer gel composition [20]. In order to minimize systematic errors it is advisable to scan the gel dosimeter phantom together with the corresponding calibration vials.

1c. Loss of spatial integrity

Loss of spatial integrity may be a consequence of the diffusion of the active components or oxygen in the gel dosimeter. An extreme case is the Fricke gel in which the diffusion of ferric and ferrous ions results in a blurring of the measured dose distribution over time [21 - 23]. But also in polymer gel dosimeters can the diffusion of monomers at very high dose gradients
during irradiation [24] and in the first hours after irradiation [20] have an influence on the measured dose distribution.

Figure 5 illustrates the mechanism that is responsible for overshoots in the measured dose distribution near steep dose gradients and high dose regions. Unreacted monomer diffusing from an unirradiated region may react with large polymer radicals in the irradiated region in the first hours after irradiation. A quantitative model is presented that links the loss of spatial integrity with the monomer diffusion and post-irradiation polymerization [25].

Figure 5. Model that describes the loss of spatial integrity in a polymer gel due to diffusion of ‘fresh’ monomer from an unirradiated region to a monomer-depleted region in the presence of large polymer radicals (see chemical instability).

1d. Positioning error of the calibration phantom
A positioning error of the calibration phantom may lead to an error in the presumed absorbed dose in the calibration phantom. This will result in an erroneous calibration curve and will eventually result in a systematic dose error in the dose distribution. The presumed dose to the calibration phantom is most often the result of an independent measurement with a standard QA dosimeter (ionization chamber). Any error in the independent measurement will also contribute to the overall error in the calibration plot.

1e. Dose rate dependent dose-response
It is seen that in certain polymer gel dosimeters the dose-response is also dose rate dependent. This may be the result from competing radiation-induced chemical reactions. This effect is more pronounced in a normoxic THP-based methacrylic acid (MAc) gel dosimeter than in PAG gel dosimeters [26]. This effect should not be underestimated as it may lead to a depth dependent dose-response. A dose-rate dependency may be detrimental for a dosimeter. Nevertheless, the dose-rate dependence has not been documented for many gel dosimeter systems.

1f. Energy dependent dose-response
Only a slight energy-dependence has been found in a few polymer gel dosimeters [26]. Although this may not have such a strong influence on the dose distribution obtained with external radiotherapy, it is a concern for brachy therapy applications. In many papers that report on brachy therapy verifications with gel dosimetry, a dose-response calibration plot is obtained with high-energy photons. A difference in dose-response for low-energy ionizing radiation as compared to the high-energy ionization will lead to a systematic dose error in the measured dose distribution.
1g. Temperature dependent dose-response during radiation
Recently, it was found that in some polymer gel dosimeters the temperature during irradiation has an influence on the dose-response [27]. This is likely due to a temperature dependent change in the diffusivity of the monomers in the gel matrix and a change in the chemical reaction kinetics.

1h. Tissue non-equivalence
For high-energy photon irradiation, most polymer gel dosimeters can be considered as soft tissue equivalent. However, at low energies, some deviations may occur.

1i. Recipient wall-effects
To avoid permeation of oxygen through the recipients wall, Barex® or glass are often used as phantom materials. It should be noted that some glass may contain heavy metals. These specific glass materials may result in a stronger attenuation of the incident beam and may also result in beam hardening. Some caution is therefore advised in selecting cast materials.

1j. Imaging voxel shape
The dose-related parametric map is constituted of voxels (3D for pixels). The signal that is plotted in these voxels originates from a set of measurements. In most measurements, there are also outer-voxel signal contributions. The voxel shape is also defined through the point-spread-function (PSF). The convolution of the point-spread-function with the theoretical (block shaped) voxel gives the voxel-shape. In NMR measurements, the voxel shape is determined by the receiver bandwidth and the slice profile which is determined by the radiofrequency pulse envelope [27]. The imaging voxel shape is very important if it comes to high-resolution applications such as with point sources in brachytherapy and high-LET irradiation.

1k. Dosimetric imaging artefacts
Imaging artefacts may deteriorate the dose value in each voxel. Dosimetric imaging artefacts can be machine-related or object-related. Machine-related artefacts originate from imperfections in the scanning device while object-related artefacts originate from the dosimeter itself.

The most-important machine-related NMR artefacts are attributed to eddy currents, stimulated echoes, B1 field inhomogeneity, imperfect slice profiles and standing waves. These machine-related artefacts may dependent on the dosimeter shape or morphology and make it difficult to make general statements on the accuracy of the dosimeter. A larger phantom or a phantom with sharp edges may perform differently than a smaller cylindrical or spherical

![Figure 6. Schematic representation of an ideal voxel-shape without any outer-voxel contribution (left) and an actual voxel with spread of signal to neighbouring voxels.](image)
shaped phantom. Standing-waves can severely deteriorate the dose distribution in dosimeters with specific shapes and spatial dimensions but may be almost completely absent if the dosimeter phantom has a slightly different shape.

Object-related NMR artefacts are mainly attributed to a temperature drift during scanning or molecular self-diffusion. For a more detailed overview of different NMR imaging artefacts and compensation strategies we refer to previous communications [27] and the review paper by Lepage further in these proceedings. In optical imaging, dosimetric artefacts are related to reflection and absorption by the recipient walls, off-axis positioning of the recipient, variation of the laser output and photo-detector and light-scattering by both impurities in the matching fluid, container and by the polymer [28, 29].

II. Scanning temperature dependent response
It has been shown that the NMR spin-spin relaxation rate (R2) of polymer gel dosimeters is temperature dependent. Temperature coefficients of the dose-R2 response have been published for PAG gel dosimeters [30, 31], for acrylic acid based gel dosimeters [32] and for methacrylic acid based normoxic gel dosimeters [26]. While it can be expected that the temperature dependence of the optical properties of 3D dosimeters and other NMR properties such as magnetization transfer is not as high as for R2, no temperature coefficients have been communicated until now.

2. Sources of spatial inaccuracy

2a. Volumetric changes of the gel dosimeter
Upon gelling, the polymer gel dosimeter may contract which may result in a physical geometrical deformation. However, until now, no significant contraction has been reported that would lead to measurable deformations of the dose distribution. Upon irradiation, some additional contraction may be expected as an increase in density has been reported for different polymer gel dosimeters [33, 34].

2b. Phantom positioning error
Any deviation in the position of the phantom with respect to the presumed position will lead to both a dose error and a geometrical error in the dose distribution. The positioning error can be minimized by using special registration devices such as stereotactic frames [35, 36] and an infrared light-emitting diode optical tracking system [37].

2c. Imaging artifacts that cause spatial deformations
Imaging artifacts may also deform the image space. In NMR imaging, geometrical distortions originate from deviations in the magnetic field distribution of the scanner during excitation or signal acquisition. These deviations in the magnetic field may arise from machine related sources such as eddy currents, main magnetic field inhomogeneities and gradient non-linearity and from phantom related sources such as susceptibility differences and chemical shifts [27]. Different strategies have been proposed to compensate for these artifacts (see ref. [27] and references herein).

5. Comparisons with other dosimeters
As mentioned before, a comparison with another dosimeter is not equivalent to a robust verification of the accuracy of the gel dosimeter as there is no absolute guarantee that the reference dosimeter can be regarded as a ‘golden standard’. However, it is probably the best way to get an ‘idea’ about the accuracy of the gel dosimeter. Different comparisons of a PAG gel dosimeter with other dosimeters are given in table 2.
Table 2. Comparisons of gel dosimetry with other gel dosimeters with different spatial dimensions. Single point measurements (0-D) can be obtained by irradiating test tubes to a ‘known’ dose. The ‘known’ dose is derived by verification with an ionization chamber. The maximum deviation of all test tubes (max{|D_i - D_j|}) is plotted in the third column. Depth-dose profiles and lateral profiles recorded with a diamond detector and an automatic water phantom (1-D) can be compared against profiles extracted from a gel dosimeter. Root-mean-square deviations (RMSD) of the two profiles are calculated. The maximum dose deviations are also plotted. A 2D comparison can be performed by subtracting corresponding dose maps obtained by either film, gel or treatment plan from each other. Structural RMSD and stochastic RMSD values can be obtained. The systematic deviation (Sys. Dev.) is obtained as the average dose difference of the corresponding dose maps. All values for the (2-D)+ table are averages from 20 different slices. The fifth and sixth column display the distance-to-agreement for 50% of the points and for 95% of the points respectively.

| O-D     | 2σD/D | max{|D_i - D_j|} |
|---------|-------|-----------------|
| Intra-batch | 1.6 % | 2.6 %           |
| Inter-batch | 15 %  | 25 %            |

| 1-D     | RMSD  | max{|D_i - D_j|} |
|---------|-------|-----------------|
| Depth-dose | 1.8 % | 2.5 %           |
| Lateral | 2 %   | 8 %             |

| (2-D)+  | RMSD_{struct} | RMSD_{stoch} | Sys. Dev. | d (N > 50%) | d (N > 95%) |
|---------|---------------|---------------|-----------|-------------|-------------|
| Film – Gel | 2.3 %     | 1 %           | - 0.6 %   | < 1.5 mm    | < 4 mm      |
| Plan – Gel | 3.6 %     | 1 %           | 2.5 %     | < 2 mm      | < 5 mm      |

With respect to the dosimetric accuracy, several test tubes (0-D) can be irradiated to a certain dose and compared with the dose recorded by an ionization chamber. However, it should be noted that a calibration curve is also obtained using the same measurement set-up. As a result, the dose accuracy may be overestimated. The standard deviation as plotted in table 2 (for 0-D) is a measure for the reproducibility or precision rather than for the accuracy. Note that the inter-batch reproducibility is very poor. This measurement was obtained by evaluating the dose-response curves of measurements that were obtained spanning an entire year. In this time span, different batches of gelatine were used. 1-D profiles can be measured using a diamond detector with the smallest dimension in the scanning direction [8, 9]. For more details on these measurements the reader is referred to [8]. In order to perform a comparison in more dimensions the gel-measured dose distribution can be compared with a dose distribution as recorded by a stack of films [10] or with the calculated dose distribution from the computer plan [10, 12, 14]. The results presented in table 2 are from a study as reported in detail in [10].

It should be noted that single values as displayed in table 2 only provide some average. Dose difference or gamma maps provide more insight in the deviations between different dosimeters.

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