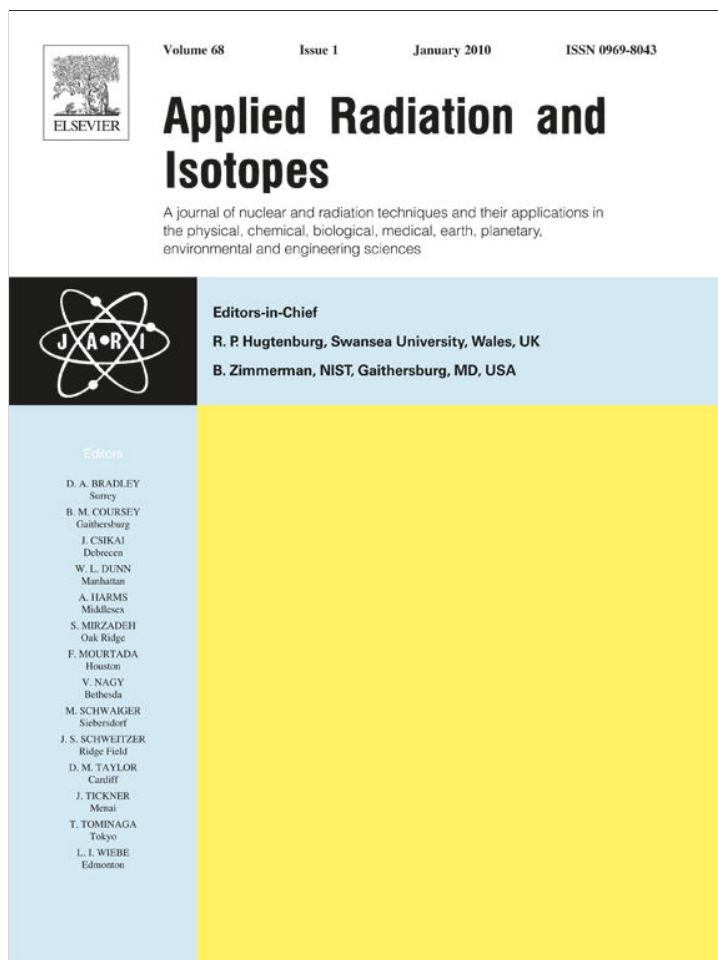




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Characteristics of a new polymer gel for high-dose gradient dosimetry using a micro optical CT scanner

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ABSTRACT

The properties of a new polymer gel with two sensitivities, made specifically for high-dose-gradient dosimetry, were investigated. The measurements were performed at NIST using a 1 cm × 1 cm calibrated ⁶⁰Co field, and a 1 cm active diameter ⁹⁰Sr/⁹⁰Y beta particle source. A high-resolution laser CT scanner was used to quantify the response. The results show that the high-sensitivity gel responds linearly to the absorbed dose for doses from 0.5 up to 15 Gy, while the low-sensitivity one is linear up to 225 Gy. For both radiation types, the gel response remains stable in time up to a month after the irradiation. The response of the gel was found to have no dose rate dependence for dose rates ranging from 3.7 to 15 mGy/s. Within the measurement uncertainty, the gel response is more sensitive for beta particles than high energy photons.

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1. Introduction

Nowadays, low-energy photons and beta particles are being used for radiotherapy treatment purposes (ICRU 72, 2004; Chiu-Tsao et al., 2007; Soares et al., 2001; Massillon-JL et al., 2009). As is well known, the absorbed dose in a condensed medium for photon and beta particle seed sources depends strongly on the source to measurement distance at close distances (< 1 cm). This can be understood, for example, for the simple case of a point source, where the variation in the distance of the dose rate is a result of the inverse square law, buildup and attenuation in the medium. At very close distances, the inverse square dependence dominates the other effects, and can result in large dose rate

gradients. Due to these steep dose gradients, accurate absorbed dose distribution measurements at close distances from both photon and beta sources are very difficult. In addition, commercially available detectors place serious constraints on measurements due to their finite sizes; most are not amenable for measurements of three-dimensional (3D) dose distributions around complex geometries with the required high spatial resolution necessary for radiotherapy dosimetry.

In the past few years, a muscle-equivalent polymer gel dosimeter, commonly known as BANG has been used for measuring absorbed dose distributions delivered during radiotherapy treatment in 3D with high spatial resolution and with the necessary accuracy (Maryanski et al., 1994, 1996a,b). The acronym was based on first letters of bis, acrylamide, nitrogen and gelatin. The dose response of BANG-type gels is based on radiation-induced polymerization and cross-linking of acrylic, acrylate or vinyl monomers in a gel matrix. As the polymer microparticles precipitate from the liquid phase of the gel, their concentration is proportional to the absorbed dose, and so is the optical

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attenuation coefficient in the gel. The polymer is partly grafted, crosslinked, and entangled with the gelatin polypeptide chain, which prevents its diffusion away from the site of the initial ionization event. Therefore, the spatial distribution of the optical attenuation coefficient in the irradiated gel represents the dose distribution. Consequently, optical tomographic imaging of irradiated BANG gels can be applied to 3D dosimetry (Gore et al., 1996, 2001). The gel studied in the present work is based on the BANG3 formulation, developed in 1998, which uses methacrylic acid as the sole monomer instead of acrylamide (linear monomer) and bis (crosslinking monomer) (Maryanski, 1999). In the presented formulation, a proprietary high-viscosity compound is also added with the purpose of reducing the mobility of the growing polymer chains. It is believed that this could reduce the probability of premature chain termination that normally occurs in less viscous media under high ionization density conditions.

A special small-format, high-resolution laser computer tomography (CT) scanner has been developed (Maryanski and Ranade, 2001) in parallel, to measure the dose distribution in gels exposed to low-energy photon and beta particle radiation fields. Contrary to magnetic resonance imaging (MRI), where the dose distribution is represented in the resultant spatial distribution of water proton relaxation rates, the optical readout relies on the optical attenuation coefficient (or optical density per unit length) which depends on the degree that the polymer micro-particles scatter light, and is proportional to absorbed dose. Therefore, the optical CT image of the gel represents the dose distribution within the gel. Gel dosimetry is attractive for low energy photons due to its water equivalence (ICRU 72, 2004).

The objective of this work is to study the properties of this new gel under different irradiation conditions in order to determine its suitability for high-dose gradient dosimetry. In particular, we have evaluated the linearity of the new polymer gel's response as a function of absorbed dose, reproducibility, the long-term stability after irradiations and the influence of temperature during the irradiation and readout process. Since low-energy photon brachytherapy seeds and beta particle sources have very high-dose rate gradients at short distances, the polymerization effect on the accumulated-dose response as well as the dose rate dependence were also investigated. As a prerequisite to use the new CT scanner, the calibration of a small reference radiation field (1 cm × 1 cm) in terms of absorbed dose to water was required, and the result of that calibration is also presented in this paper.

2. Materials and methods

2.1. The gel

The muscle-equivalent gel¹ used in this study² is specifically made for high-ionization-density radiation and composed of 12% gelatin (C₁₇H₃₂N₅O₆), 12% proprietary high-viscosity-component (C_xH_yO_z), 6% methacrylic acid monomer (C₃H₄O₂)_n and 70% water (H₂O). The density of the gel was measured and found to be (1.03 ± 0.03) g/cm³. In this gel, the methacrylic acid which is soluble in water, acts as the active ingredient that undergoes free-radical polymerization during the irradiation process. The preparation processes were carried out by the manufacturer and at NIST using a commercially available kit form of gel with the same chemical composition. We also investigated the gel at two

Table 1
Composition of the gel.

Components	Weight/weight fractions	
	High sensitivity (HS)	Low sensitivity (LS)
C	0.1419	0.1418
H	0.1008	0.1008
N	0.0209	0.02089
O	0.7364	0.7361
Cu	3.20×10^{-7}	–
S	1.60×10^{-7}	1.6×10^{-4}
Fe	–	2.8×10^{-4}

different sensitivities (Table 1): high-sensitivity (HS), that has a linear response at low dose levels and is useful for low-energy photons where the half-life of the radionuclide is very short and dose rates are relatively low, and low-sensitivity (LS), that is linear from 7.6 up to 225 Gy, which is useful for medical beta particle dosimetry where dose rates are relatively high.

2.2. BANG gel preparation at NIST

About 48 h before preparing the gel, 10 empty PMMA cylinders and caps (nine with 3 mm PMMA caps and one with a 10 mm cap in order to compare with gels prepared by the manufacturer in that geometry) were cleaned and placed under nitrogen atmosphere in a glove box positioned in a fume hood. At preparation time, a 2 L BANGkit gel was placed in a hot water bath at approximately 50 °C for 2 h; the gel kit was removed from the bath and flipped over every 20–30 min in order to mix the gel while it was melting. Meanwhile, a 1 mol/L stock solution of CuSO₄ (catalyst) was prepared, followed by a solution of L-ascorbic acid (oxygen scavenger) made by dissolving 0.704 g of ascorbic acid powder in 40 mL of de-ionized water at 50 °C. Once the gel was fully melted, it was poured gently into a 2 L beaker on a hot-plate set to maintain a temperature of 50 °C. A syringe was used to add the top 20 mL of the saturated ascorbic acid solution to the gel. Then, 10 mL of the 1 mol/L CuSO₄ solution was added slowly. The gel was gently stirred for the next 5 min. To fill the cylinders, the gel was first poured into an 800 mL beaker and subsequently added to each container. Once a container was filled, the lid was placed on top, waxed plastic wrap was pressed around the lid for sealing, and adhesive tape was wrapped firmly around the circumference of the lid to ensure that it was air tight. These cylinders were placed back in the original air tight bag in which the kit came, and were immersed in a nitrogen atmosphere at approximately 23 °C. Aluminum foil was used to cover the cylinders, preventing photo-polymerization of the monomers, and the gel was given 72 h to harden and cure.

2.3. Readout process using an optical computer tomographic scanner

Fig. 1a displays a top view of the laser CT scanner³ used in this work. The schematic depicted in Fig. 1b explains its operating principles. After each slice scan, the linear stage to which the rotating gel cylinder is mounted can be moved vertically up or down to acquire a new slice whose thickness can be as small as 50 μm (z-dimension). During the readout process, the light from a 5 mW, 635 nm polarized laser diode passes through a built-in focusing lens on a rotating mirror whose surface is placed at the focal spot of the gel-filled PMMA cylinder of 76.2 mm outside diameter and 3.175 mm wall thickness. The cylinder itself acts as a

¹ BANG3-Pro-1 gel (MGS Research Inc., Madison, CT)

² In this paper certain manufacturers and products are referred to by name. These identifications are for informational purposes only and do not imply that these are the best or only suppliers or products available, nor do they imply endorsement by the National Institute of Standards and Technology.

³ DRYOCTOPUS CT scanner (MGS Research Inc., Madison, CT).

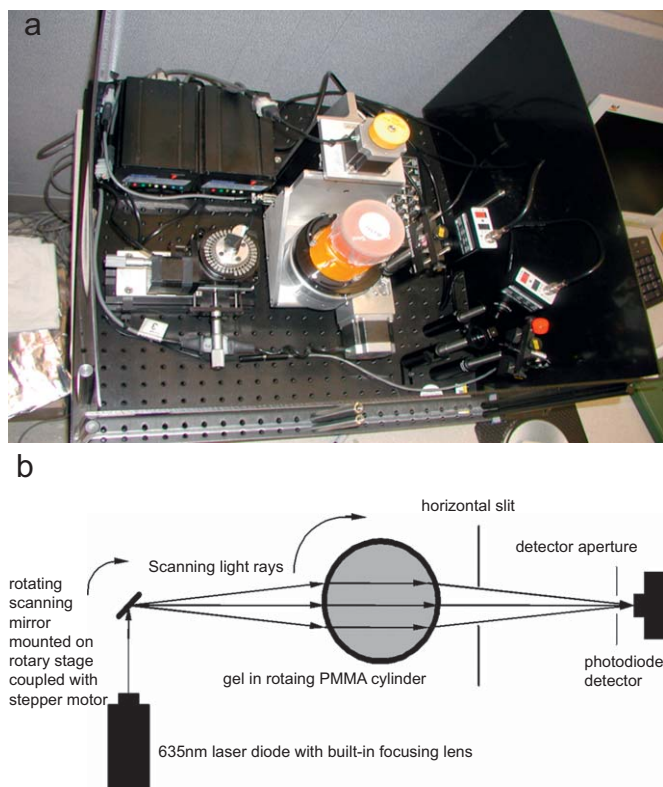


Fig. 1. (a) Laser CT scanner with an acrylic cylinder filled with gel and (b) schematic of the laser CT scanner used in this work.

lens in this scanning system. Therefore, the paraxial rays inside the gel are all parallel to each other, and all the transmitted rays converge at a focal length, $f=r/(n-1)$, from the exit surface of the cylinder, where r and n are the radius and refractive index of the cylinder, respectively. In order to avoid cylindrical aberration, the scan is limited to one third of the cylinder's diameter, which means that the scanner has a field of view of 25 mm. At this field of view, the scanner has the maximum spatial resolution of 100 μm which is determined by the laser spot size. The transmitted light is collected by a 1 cm^2 silicon photodiode detector to produce the projection data, as the cylinder rotates in 0.45° increments between projections. The number of projections depends on the pixel size selected. For example, the number of projections for a pixel size of 100 μm is 800 for a complete scan of 360°. The total time to obtain such an image is approximately 7 min. In this work the scans were made only over 180°. The photodiode signal is sampled at constant frequency using a single channel of a USB 16-bit data acquisition module to form individual projection data as the laser beam scans the gel. Each pixel is defined as an arithmetic average of 100 samples. These projections are then fed into a filtered back-projection image reconstruction program "recon" written in MATLAB (The MathWorks, Natick, MA). The program forms the sinogram for each slice, normalizes it by a reference sinogram, then uses a MATLAB built-in inverse radon transform function "iradon" to reconstruct a stack of two-dimensional maps of the optical attenuation coefficient, expressed as optical density per cm (OD/cm). The final data analysis, including OD/cm-to-dose calibration, is performed using the public-domain ImageJ software developed at the National Institute of Health (ImageJ, 2009). ImageJ provides multiple functionalities such as global density calibration, image arithmetic (useful for correcting baseline optical density drifts in the gel), profile plotting, generating orthogonal reslices from the original stack or z-projections, and many others.

2.4. Irradiation processes

The gels were calibrated at NIST using a special small field-size ^{60}Co gamma ray beam of 1 cm $\times$ 1 cm and a calibrated $^{90}\text{Sr}/^{90}\text{Y}$ beta particle source. For the calibrations, the gel was sealed in polymethyl methacrylate (PMMA) cylinders of 76.2 mm outside diameter, with 3.175 mm walls and a height of 50 mm. For the ^{60}Co beam irradiations the cylinders were covered with 10 mm PMMA caps, while 3 mm PMMA caps were used for the $^{90}\text{Sr}/^{90}\text{Y}$ beta source irradiations.

2.4.1. Calibration of a small ^{60}Co field in terms of absorbed dose to water

As explained in Section 2.3, the scan is limited to one third (25 mm) of the translucent cylinder diameter which means that the size of the radiation field used to calibrate the gel must be always smaller than this dimension. The dosimetry of such a narrow photon beam presents a challenge due to the lack of charged-particle-equilibrium in the lateral dimensions for most of the detectors that can be used for such conditions. Consequently there are no reference conditions for calibration purposes. Normally, the ^{60}Co gamma-ray beam at NIST is calibrated with the water calorimeter in terms of absorbed-dose to water at 5 cm depth in a reference radiation field size of 15.4 cm $\times$ 15.4 cm using a source to surface distance (SSD) equal to 95 cm (Minniti et al., 2007). To transfer this calibration from the reference radiation field size to a smaller one, three different detectors were used: a 0.125 cm^3 small volume micro-ionization chamber,⁴ thermoluminescent dosimeters⁵ and radiochromic film.⁶ Before these measurements, all detectors were calibrated in terms of absorbed dose to water in the reference ^{60}Co gamma-ray field at NIST.

The-ionization chamber was placed at 1 cm depth in a PMMA phantom of 30 cm $\times$ 30 cm $\times$ 14 cm. The measurements were carried out using two different SSDs equal to 40.8 and 95 cm, with two different field sizes: the reference field size of 15.4 cm $\times$ 15.4 cm, and the radiation field size of interest for our gel dosimeter calibration of 1 cm $\times$ 1 cm. To validate the dose-rate obtained in the PMMA phantom, the ion chamber measurements were compared to the dose-rate measured directly in a water phantom with the reference field size and multiplied by a scaling factor of 1.2087 which is taken from the depth dose profile in water to convert the dose from 5 cm depth in water to 1 cm depth in PMMA at 95 cm SSD in the large field. A difference of about 0.1% was observed between the two measurements.

The thermoluminescent dosimeters (TLDs) as well as the radiochromic film were calibrated using the same PMMA phantom mentioned above at 1 cm depth in PMMA, at 95 cm SSD in the reference ^{60}Co field size, using the dose rate measured with the ionization chamber under these conditions. The dose levels ranged from 25 mGy up to 20 Gy for the TLDs and from 3 to 50 Gy for the radiochromic film. The calibration curves are shown in Figs. 2a and b for both detectors. To measure the dose-rate at the non reference geometries, the dosimeters were irradiated at 1 cm depth in a PMMA cylinder phantom of 14.5 cm height and 7.5 cm diameter using a SSD of 40.8 cm for both field sizes.

For the TLDs, the exposure time was such that the delivered absorbed dose was not greater than 1 Gy to avoid the onset of the supralinearity of the dosimetric peak for TLD-100 (Massillon-JL et al., 2006). An average of three dosimeters was used for each dose value for the calibration curve and four for the dose-rate

⁴ PTW type N31011 (PTW, Freiburg, Germany).

⁵ TLD-100 Harshaw/Bicron chips 3.1 mm $\times$ 3.1 mm $\times$ 0.89 mm (Thermo Scientific, Solon, OH).

⁶ Gafchromic type MD-55-2 (International Specialty Products, Wayne, NJ).

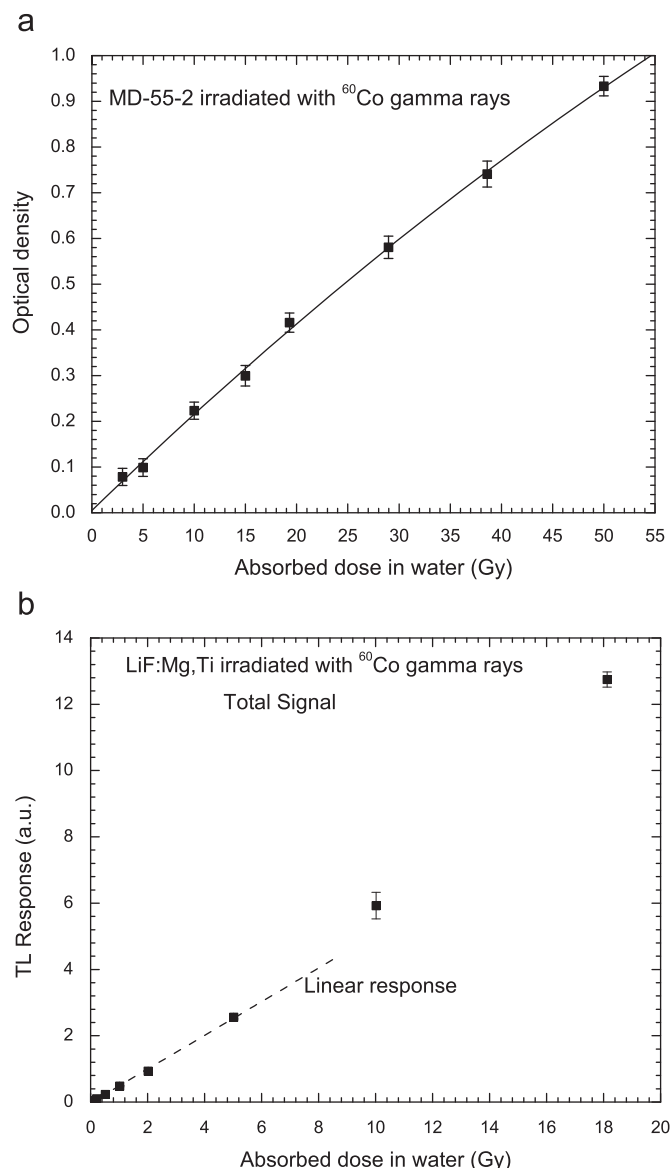


Fig. 2. (a) Calibration of radiochromic film with ^{60}Co gamma rays. The error bars indicate the measurement uncertainty at one standard deviation and (b) calibration curve for LiF:Mg,Ti (TLD-100) after exposure to ^{60}Co gamma rays. The error bars indicate the measurement uncertainty at one standard deviation.

measurements. The annealing and readout procedure were carried out according to Massillon-JL et al. (2006).

For the dose rate measurement using radiochromic film, the delivered dose was about 30 Gy. The readout process was performed approximately 48 h after the irradiation using a 633 nm laser densitometer scanner (LKB Pharmacia model 2222-020, no longer manufactured) whose spot size is about 100 μm in diameter. The step size used during the scan was 40- μm increments in both dimensions.

2.4.2. The gel response to ^{60}Co

Eight phantoms of each gel that was prepared by the manufacturer were used. The HS gels were irradiated to doses from 0.5 Gy up to 30 Gy, while the LS gel to doses from 8 to 300 Gy in the small calibrated ^{60}Co gamma field of 1 cm $\times$ 1 cm at 40.8 cm SSD. After the irradiation, the gels were stored in a refrigerator and scanned at different times (referred to as time post irradiation) to evaluate the stability of the gel over time. That is, for how long the changes that occurred in the gel due to the irradiation (gel

response) remain constant over time. The readout process was performed with a spatial resolution of 100 μm in the plane just behind the 10 mm PMMA cap at a depth of 1 mm in the gel.

2.4.3. The gel response to $^{90}\text{Sr}/^{90}\text{Y}$ beta particles

The measurements were performed using a $^{90}\text{Sr}/^{90}\text{Y}$ beta ophthalmic applicator at NIST that was calibrated in terms of the radiation quantity absorbed dose to water. This source has an active diameter of 9 mm and the reference dose rate is measured and therefore well known at a depth of 1 mm in water. To determine the dose rate at other depths in water, the depth dose curve for this source, that has been previously reported elsewhere (ICRU 72, 2004; Soares et al., 2001), was used. In order to calibrate the response of the gel (optical density per unit length) in terms of the absorbed dose to water delivered by the source, a total of eight phantoms were exposed to individual doses D_j^{ind} where j represents each one of the eight gels used. The doses D_j^{ind} delivered ranged from 1 to 30 Gy. That is, each phantom was exposed to a single dose value. This data allowed constructing a calibration curve (gel response vs. absorbed dose to water) that will be addressed later in the discussion section. For these irradiations all phantoms had a 3 mm PMMA cap, and contained a HS gel that was prepared at NIST. A special holder was built to position the beta source reproducibly relative to the gel phantom.

Alternatively, we investigated a second approach for building the calibration curve for the gel (gel response vs. dose). This consisted in exposing a single gel to various small fractions of the total desired dose referred to as *accumulated doses* D_i^{acc} , where the index i indicates each one of the accumulated doses delivered. This second fractionated dose method has the advantage of being able to use only one gel instead of various gels as described above in the first approach. The fractionated dose method consists of the following; first, the gel phantom is exposed to a single dose D_1^{acc} by placing the $^{90}\text{Sr}/^{90}\text{Y}$ beta source on top of the phantom for an interval of time required to deliver the specified dose value D_1^{acc} . To perform the irradiation, the beta source was placed inside the holder made of PMMA and centered on top of the 3 mm PMMA slab of the gel phantom. This setup allows reproducing exactly the position of the beta source during different irradiations. After the first irradiation is completed, the gel is scanned and the gel response is recorded for this given delivered dose D_1^{acc} . After that, the source is placed a second time on top of the gel phantom at exactly the same position relative to the gel phantom. A second dose D_2^{acc} is delivered to the gel. The total accumulated dose delivered to the gel is therefore $D_1^{\text{acc}} + D_2^{\text{acc}}$. Once again, after this second irradiation is completed, the gel is scanned and the gel response is recorded. This process is repeated several times for various small accumulated doses (or fractional doses) D_i^{acc} . For our particular case we exposed the gel to a total of 10 fractional doses. That is, the index i had values between 1 and 10 and each accumulated dose D_i^{acc} had values between 1 and 30 Gy.

The validation of the fractional dose method is an important result that provides and validates this method to overcome the challenge of measuring dose distributions at distances close to the source. As mentioned in the introduction, the accurate measurement of the absorbed dose at close distances to low-energy photon and beta sources is very difficult and consequently the dose at these distances is currently unknown. Thus, instead of exposing the gel only once to a single individual total dose, which could result in saturation effects at distances very close to the source, we can use the fractionated dose method. That is, we exposed a single gel to small fractions of the desired dose value to ensure that the response of the gel is always within the linear response region. This approach was reported recently elsewhere

(Massillon-JL et al., 2009) for characterizing the 3D dose distribution of a $^{90}\text{Sr}/^{90}\text{Y}$ intravascular brachytherapy seed source.

To evaluate the effect of the temperature on the irradiation and readout process, the individual gels were divided into two groups: the first group was irradiated at room temperature (22 °C), stored in the refrigerator overnight and scanned the next day, while the second group was stored in the refrigerator overnight prior to irradiation, irradiated cold (4 °C) (the time exposure was such as the gel could remain cold) and scanned at room temperature by waiting for about 4–5 h. The scanning process, which takes approximately 4 min, was carried out with the same spatial resolution mentioned above, at 1 mm depth in the gel behind the 3 mm PMMA cap.

2.5. Dose rate dependence

The dose rate dependence was evaluated by scanning the gels irradiated with the $^{90}\text{Sr}/^{90}\text{Y}$ beta source at different depths in the gel. The depths were chosen such that the dose rates ranged from 5 to 15 mGy/s as determined from the NIST reference dose rate calibration and the published depth dose rate curve. Independently, the LS gel was exposed in the ^{60}Co field of 1 cm $\times$ 1 cm at two SSDs equal to 54.9 and 75.2 cm, different from the calibration reference SSD. The dose rate at these distances was measured by exposing calibrated TLDs at the same position using the cylindrical acrylic phantom. Due to the divergence of the beam and the size limitation of the scanner, larger distances were not used.

3. Results

3.1. Calibration of a small ^{60}Co field in terms of absorbed dose to water

Table 2 presents the dose-rate obtained with different dosimeters at various distances and field sizes. For the large field, the dose rate measured with the ionization chamber agrees well, within 0.2%, with the radiochromic film and within 1.8% with the TLDs, while for the small field a difference of up to 12% is observed. Although this chamber has a small active volume, there is a fraction of the chamber body that is not fully exposed to the small beam size. This results in a lack of lateral electronic equilibrium and is consistent with the difference of 12% observed between the chamber and the other two passive dosimeters. Due to the recent increase of radiotherapy techniques in the last years that rely on the use of small radiation fields, this level of agreement between the different types of dosimeters must be taken into account in the future development of protocols for small reference fields (Alfonso et al., 2008). We have also

evaluated the homogeneity of the beam at 40.8 cm SSD in the small field. Fig. 3 shows the dose profile obtained with the radiochromic film exposed to 30 Gy. A variation of 1.9% is observed across an area of 8 mm $\times$ 8 mm.

3.2. Gel response to ^{60}Co gamma rays

Fig. 4a displays the gel sealed with a 10 mm PMMA cap in a PMMA cylinder before and after irradiation where the small ^{60}Co radiation field of 1 cm $\times$ 1 cm can be observed. The beam was incident from the top of the phantom. The square shaped region in the central of the gel shown on the right hand side figure is the change in the gel produced by the exposure to the square gamma-ray beam. The square region results in a change of the optical density of the gel material. As mentioned previously, the optical density is proportional to the dose delivered to the gel. That is, the higher the dose delivered to the gel, the higher is the optical density in the exposed region. As will be discussed in the sections below, there is a range of doses delivered for which the response of the gel is linear. Therefore, a conversion factor can be determined to express the optical density per unit length (OD/cm) in terms of dose units (Grays). By the same token, the

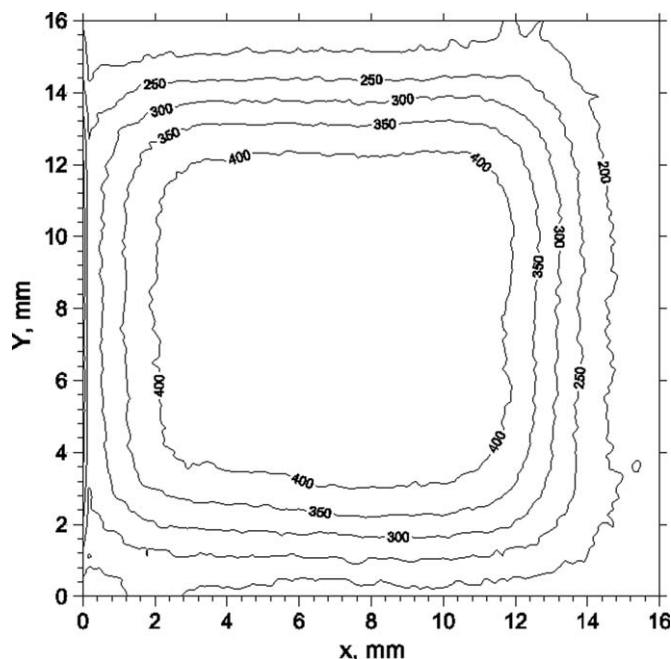


Fig. 3. Beam profile measured at a SSD of 40.8 cm using the radiochromic film. The collimator opening of 1 cm $\times$ 1 cm was located at a SSD of 36.8 cm.

Table 2
Dose rate measured using different detectors.

SSD (cm)	Dose rate (Gy s ⁻¹) ^a					
	Field size=1 $\times$ 1 cm ²			Field size=10 $\times$ 10 cm ²		
	RCF	TLD	IC	RCF	TLD	IV
40.8	1.56(6) $\times$ 10 ⁻²	1.51(7) $\times$ 10 ⁻²	1.40(1) $\times$ 10 ⁻²	1.87(1) $\times$ 10 ⁻²	1.84(1) $\times$ 10 ⁻²	1.87(5) $\times$ 10 ⁻²
54.9	–	7.08(9) $\times$ 10 ⁻³	–	–	–	–
75.2	–	3.65(7) $\times$ 10 ⁻³	–	–	–	–
95	2.64(3) $\times$ 10 ⁻³	–	–	–	–	–

The relative combined uncertainties on the dose rate measurement are 1%, 3% and 4% for the ionization chamber (IC), radiochromic film (RCF) and the TLDs, respectively.

^a The digits shown in parenthesis do not represent the accuracy of the measurement. Nevertheless, it is included to prevent rounding errors when these values are used in calculations.

measured 2D optical density distributions like the one shown in Fig. 4b can be converted to 2D dose distributions by applying the corresponding conversion factor (OD/cm per Gy). The image on the left side of Fig. 4b shows the reconstructed 2D distribution image behind the 10 mm PMMA cap at a depth of 1 mm in the gel for a gel exposed to 10 Gy. The image on the right shows the quantitative result, i.e., the isodose curve in the plane perpendicular to the beam axis. The uniformity across an area of $8\text{ mm} \times 8\text{ mm}$ from Fig. 4b is 1.5%. As described in the previous section, the variation in the beam uniformity observed using the radiochromic film over the same size area was 1.9%. The measured variations using both film and the gel are mainly the result of variations in the imaging material and not the beam uniformity itself since similar measurements reported elsewhere (Minniti et al., 2006) using ion chambers report a beam variation of 0.1%.

To evaluate the time required for this gel to be scanned after irradiation, the time between irradiation and readout was varied. Fig. 4c and d present two images and their associated dose profiles

for the same gel exposed to 30 Gy and scanned at two different times post-irradiation. Clearly two effects can be observed in Fig. 4d that depend on the time post irradiation. An enhancement appears in the edge of the field when the gel was scanned 648 h after the irradiation was completed. This effect is not seen in Fig. 4c, i.e., for the case in which the gel was scanned immediately after the irradiation was completed. The second effect is the change in the value of the intensity value around the central region of the radiation field. In Fig. 4c the intensity around the central region of the field was 0.7 OD/cm. This initial value represents the response of the gel due to the irradiation. However, as shown in Fig. 4d the gel itself changes over time and as a result the original value of the response does not remain constant for long periods of time. Instead, the value of the original response (created by exposure to the radiation) drops to a value of 0.6 OD/cm after 648 h as shown in Fig. 4d. The latter effect, i.e., the decrease of the dose intensity in the central region of an irradiated gel phantom was observed for doses equal to or $> 15\text{ Gy}$ as shown

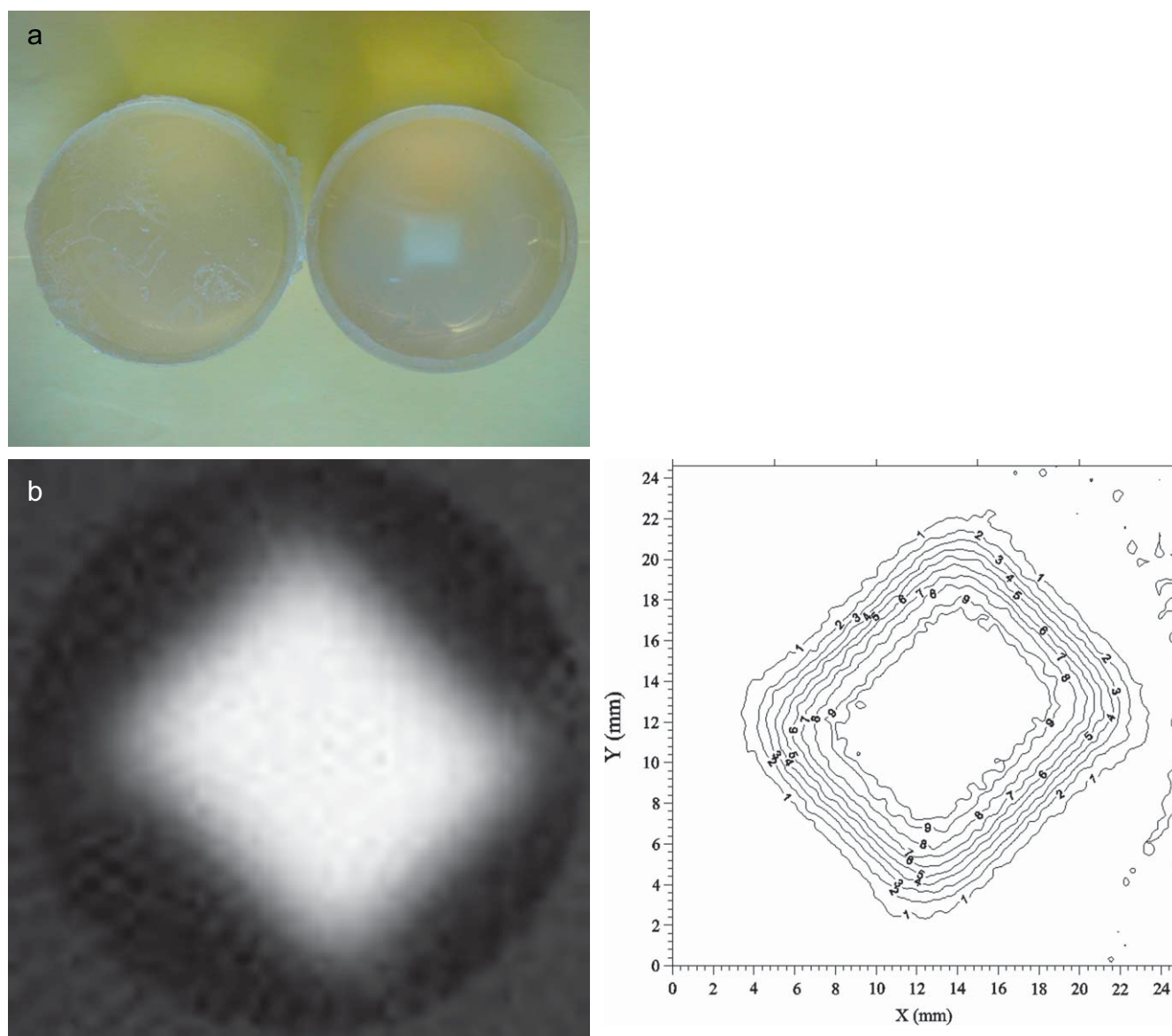


Fig. 4. (a) Image showing two PMMA cylinders filled with polymer gel before and post irradiation, (b) single-slice optical CT reconstruction from the gel irradiated at 10 Gy, scanned at 1 mm depth behind 10 mm PMMA cap, and its 2D dose distribution, (c) 2D dose reconstruction and profile measured at 1 mm depth in gel behind 10 mm PMMA cap, for 30 Gy and scanned $< 1\text{ h}$ after irradiation. The line joining the points is drawn to guide the reader's eyes, and (d) 2D dose reconstruction and profile measured at 1 mm depth in gel behind 10 mm PMMA cap, for 30 Gy and scanned $< 1\text{ month}$ after irradiation. The line joining the points is drawn to guide the reader's eyes.

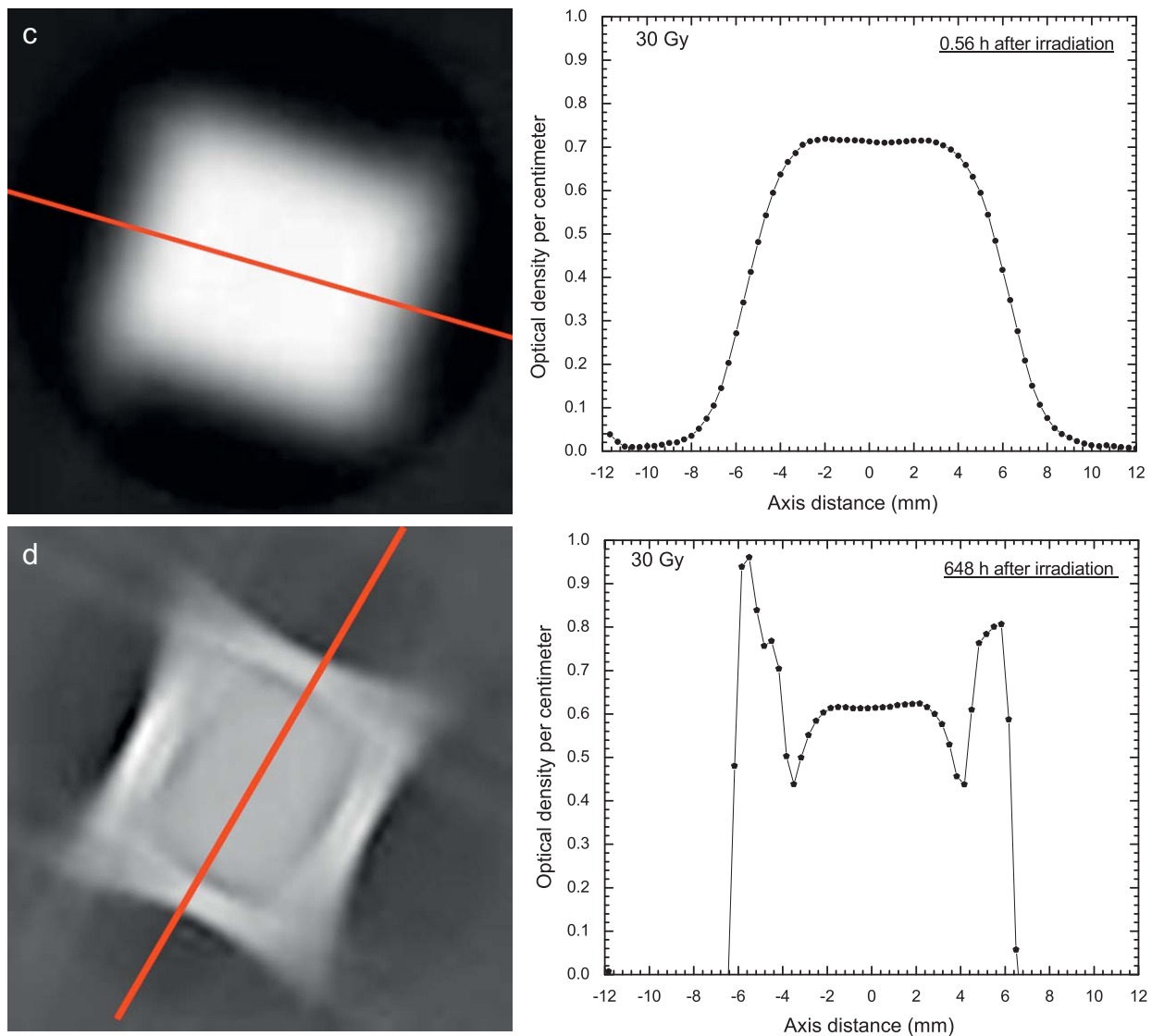


Fig. 4. (Continued)

in Fig. 5a and b. Furthermore, the fading or decrease of the initial dose response is larger for larger doses. So, for example, the fading of the original response is larger for a dose of 30 than for 20 Gy. The edge enhancement shown in Fig. 4d, is becomes larger also for higher doses. Furthermore, the onset of this effect is faster for larger doses. For example for the gels exposed to doses of 15 Gy and lower, the edge enhancement was observed for the first time after one month.

The response for the low-sensitivity gel after exposure to ^{60}Co gamma rays is shown in Fig. 6. A linear fit is included in the figure and the data points follow this linear trend, within uncertainty bars, up to 225 Gy.

3.3. Gel response to $^{90}\text{Sr}/^{90}\text{Y}$ beta particles

Fig. 7 shows two gels that were exposed to the $^{90}\text{Sr}/^{90}\text{Y}$ beta source. As discussed in Section 2.4.3, the HS gel is sealed in the PMMA phantom with a 3 mm PMMA cap. One of the gels was exposed for 195 s while the other one was exposed for a longer period of time of 2443 s. As mentioned in Section 2.4.3, the dose rate at any depth is well known from the calibration of the source. A clear circular spot can be seen on the gel located on the right

hand side of Fig. 7. This is indicative of the change in the optical density due to the irradiation. However a very faint spot can barely be seen on the left side consistent with the fact that the gel has been exposed for a much shorter period of time. The circular spot has a diameter of 10 mm which corresponds to the active diameter of the $^{90}\text{Sr}/^{90}\text{Y}$ beta source used. Fig. 8 presents the 2D dose distribution with its associated isodose curve produced by the beta particles penetrating the 3 mm PMMA cap and 1.5 mm of the gel; the delivered absorbed dose to water at this depth is 6 Gy. The dose at this depth was determined from the known reference absorbed dose rate to water as described in Section 2.4.3, and by scaling the PMMA and gel thicknesses to water depth using their respective mass densities. For this source, the response of the gel was determined by taking the average intensity of the 2D distribution across a circular area of 4 mm diameter

As discussed in Section 2.4.3, a total of eight gel phantoms were exposed using individual doses while a single gel phantom was exposed to several accumulated doses. The response for each exposure to the beta particle source as a function of the absorbed dose delivered to the gel is shown in Fig. 9. Following the notation introduced earlier, the response for each one of the accumulated doses D_1^{Acc} , $(D_1^{\text{Acc}} + D_2^{\text{Acc}})$, $(D_1^{\text{Acc}} + D_2^{\text{Acc}} + D_3^{\text{Acc}})$, etc..., delivered to the single gel using the fractional dose method, are shown in Fig. 9 by

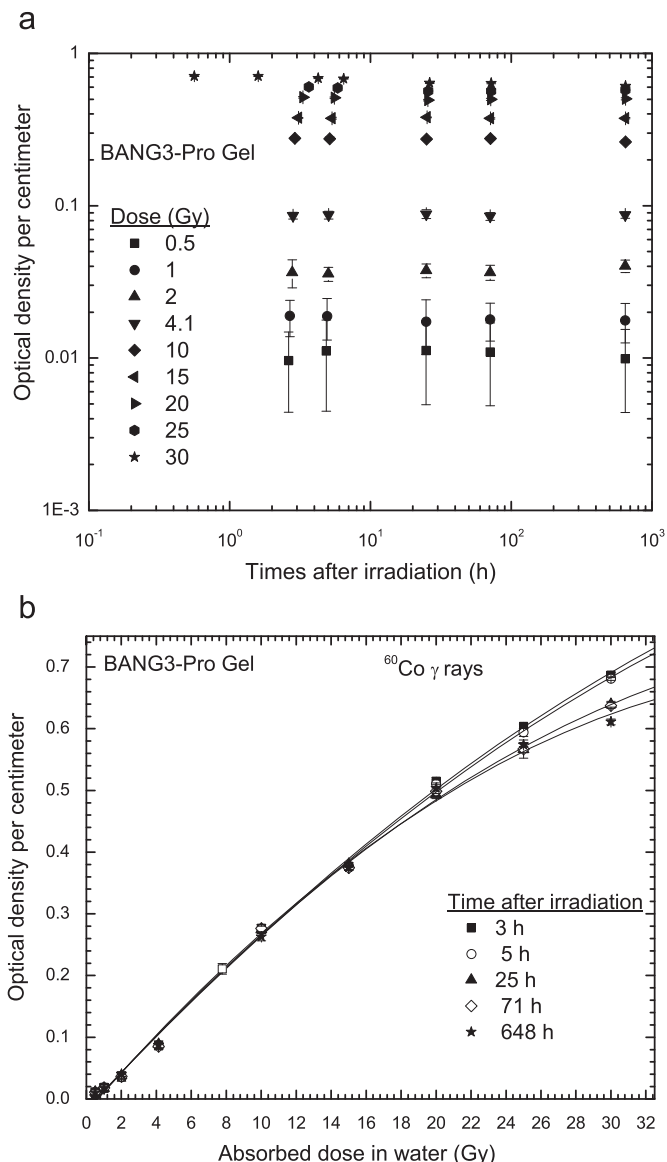


Fig. 5. (a) Time effect between irradiation and reading process. The error bars indicate the measurement uncertainty at one standard deviation and (b) response of the high-sensitivity gel to ^{60}Co gamma rays. The lines represent a polynomial fit.

empty square symbols. On the other hand, the gel responses due to the individual doses delivered D_j^{ind} to each one of the eight phantoms used are shown in Fig. 9 with solid circles. As seen in Fig. 9, the responses obtained using both methods, i.e. the individual dose method and the fractional dose method, fall on the same curve within the uncertainty of the measurement indicated by the horizontal bars on each data point. This means that there are no differences between the two methods, validating in this way the delivery of small accumulated doses to a single gel.

Fig. 10 displays the effect of the temperature on the polymerization processes, and shows that the difference in the response of the gel irradiated cold and at room temperature is negligible. To evaluate the dependence of the gel response on the type of radiation, we compare the response obtained with ^{60}Co gamma rays versus $^{90}\text{Sr}/^{90}\text{Y}$ beta particles. Fig. 11 presents this result. In spite of the uncertainty in the beta particle source calibration (ICRU 72, 2004; Soares et al., 2001), an appreciable difference is observed between the two measurements.

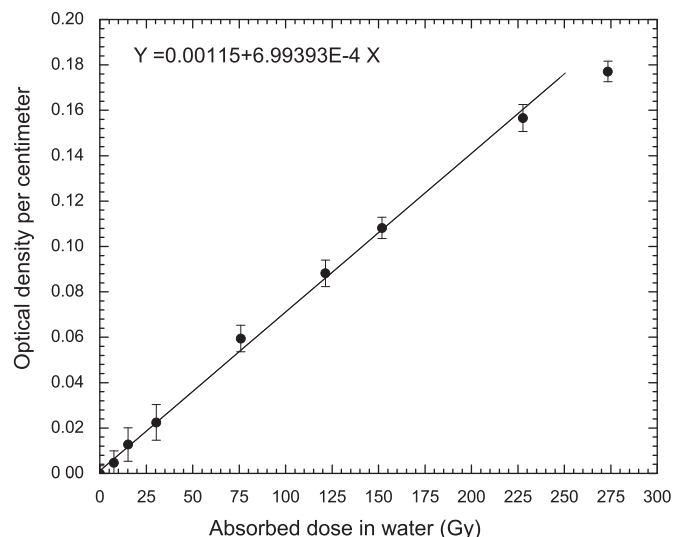


Fig. 6. Response of the low-sensitivity gel to ^{60}Co gamma rays. The error bars indicate the measurement uncertainty at one standard deviation.

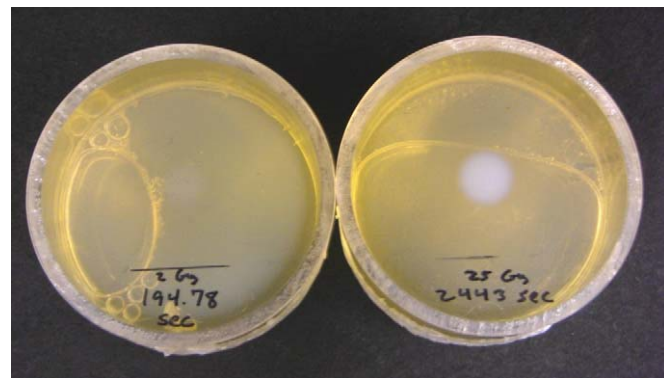


Fig. 7. Imaging showing two cylinders filled with gel after exposure to $^{90}\text{Sr}/^{90}\text{Y}$ beta applicator at two different exposure times.

3.4. Dose rate dependence

Figs. 12a and b present the response of the gel as a function of the dose for different dose rates using ^{60}Co gamma rays and beta particles, respectively. For both photons and electrons, the response of the gel does not show any dependence on the dose rate in the studied interval.

4. Discussion

For this study we focused on the properties of the new BANG gel evaluated by a micro optical CT scanner, in order to determine how effective this dosimeter is for measuring high dose gradient of radiation. Firstly, we calibrated a small radiation field in terms of absorbed dose to water using a small volume ionization chamber, TLDs and radiochromic film. We found that radiochromic film as well as TLDs can be used to calibrate small radiation fields in term of absorbed dose rate to water within an uncertainty of 3–4%, while the 0.125 cm^3 volume ionization chamber used is too large to measure the absorbed dose rate in small ($1\text{ cm} \times 1\text{ cm}$) radiation field. As mentioned earlier in Section 3.1, this result is relevant to ongoing efforts in the medical community (Alfonso et al., 2008) to develop protocols for small field dosimetry.

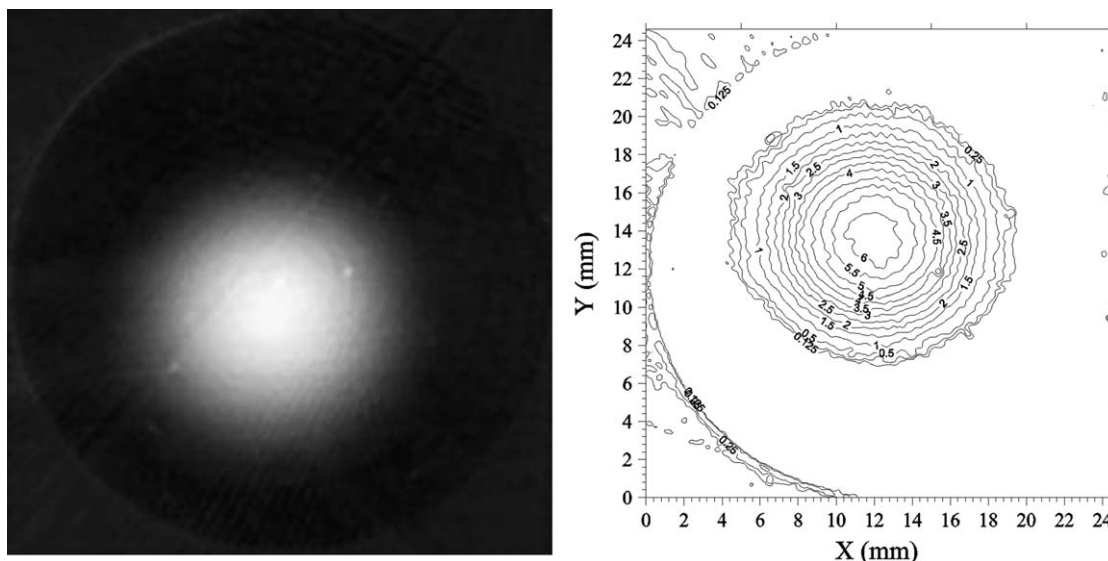


Fig. 8. 2D dose distribution from $^{90}\text{Sr}/^{90}\text{Y}$ beta particles, measured at 1 mm depth in the gel behind 3 mm PMMA cap.

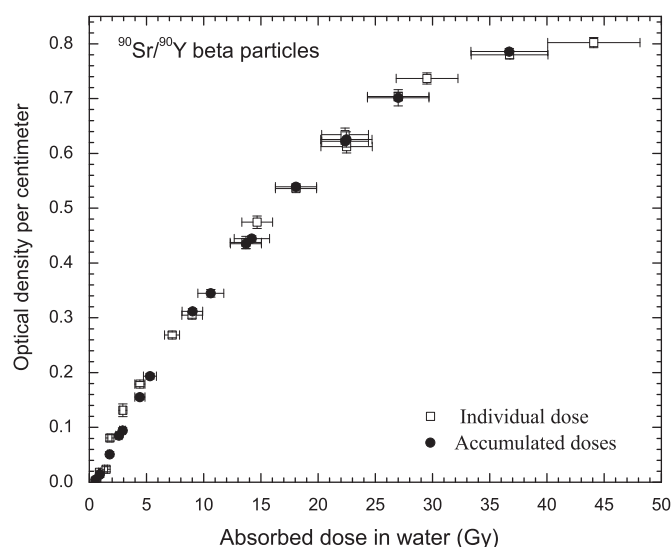


Fig. 9. Response of gel to $^{90}\text{Sr}/^{90}\text{Y}$ beta particles: individual exposure versus accumulated doses. The error bars indicate the measurement uncertainty and dosimetry uncertainty at one standard deviation.

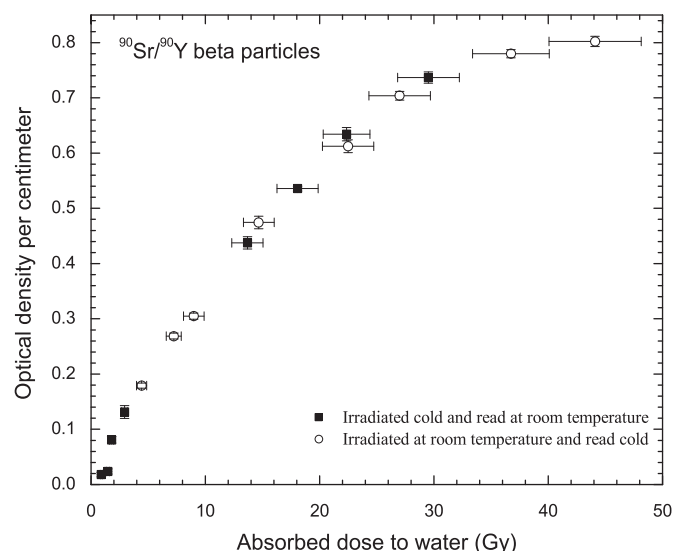


Fig. 10. $^{90}\text{Sr}/^{90}\text{Y}$ beta particles induced-response for the gel irradiated and read at different temperatures. The error bars indicate the measurement uncertainty and dosimetry uncertainty at one standard deviation.

The evaluation of the gel response (radiation-induced polymerization process) as a function of post-irradiation time for ^{60}Co gamma rays (Figs. 5a and b) shows that the gel is stable and consequently, its dose–response curve is reproducible for doses lower than 15 Gy. For doses larger than 15 Gy, two effects were observed which were described in the previous section in relation to Figs. 4c, d, 5a and b. One of these two effects is the enhancement of the edge of the radiation field. This effect is observed after a given period of time after the irradiation was completed. The higher the dose the earlier the onset of this effect occurs. For example for a dose of 20 Gy delivered to the gel, the edge enhancement appears a little above 24 h the irradiation was completed, and for a dose of 30 Gy the effect appears earlier. Curiously, the edge enhancement was observed for previous versions of BANG gel exposed to 6 MV X-ray beams at doses equal to or larger than 10 Gy using the MRI readout technique (Maryanski et al., 1994). It was argued that the edge enhancement could be attributed to a diffusion process of monomers from the

low to the high-dose region which are depleted by the polymerization reaction (Maryanski et al., 1996a,b). Therefore, this diffusion can react with long-lived macroradicals possibly created by the radiation acting on the gelatin or the polymer chains. Contrary to the enhancement at the edge, a diminution of about 14% is observed in the apparent response in the center of the radiation field (0.7 OD/cm in Fig. 4c versus 0.6 OD/cm in Fig. 4d). A similar behavior was observed for beta particle irradiations. In this work, the response of the gel was obtained by taking the average intensity of the 2D distributions over an area centered on the center of the field.

From a practical point of view, this feature of the gel is a very important limiting factor that must be taken into account when the gel is to be used for dosimetry where a high dose rate gradient is present. In the case of low-energy X-ray brachytherapy seeds or medical beta particle sources, where the dose rate varies drastically at very short distances from the source, the response can begin to diminish even during the irradiation and scanning

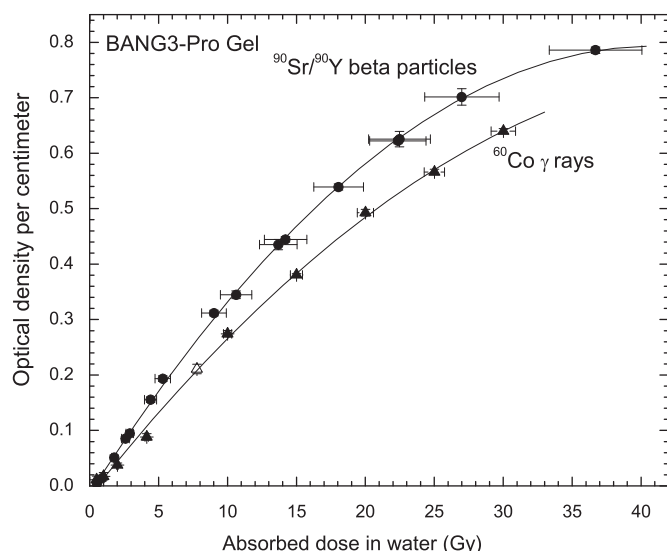


Fig. 11. Induced-response of the gel after exposure to ^{60}Co gamma rays and $^{90}\text{Sr}/^{90}\text{Y}$ beta particles. The lines represent a polynomial fit. The error bars indicate the measurement uncertainty and dosimetry uncertainty at one standard deviation.

processes, since at very high resolution an average of 4 min is needed to acquire a single slice. Thus, during the irradiation, if exposure time is not such that the response of the gel can be considered linear, the gel could be less efficient as the ionization density increases, which could not only be associated with the saturation effect but perhaps also with an intrinsic property of the gel.

As Fig. 9 shows, the response of the gel irradiated with a single dose is statistically similar to that one irradiated with fractionated doses. This fact is as an important issue since the problem of under-response of the gel can be solved by irradiating the gel with small accumulated doses in a high dose rate gradient radiation field, as long as the gel can be positioned reproducibly between irradiations.

We have evaluated the temperature effect on the response of the gel when it is irradiated at room temperature and scanned cold and vice versa (Fig. 10). We found that the temperature at the time of irradiation does not affect the response of the gel, and neither does the temperature at the time of scanning process. The latter differs from a result reported before (Maryanski et al., 1994) for a previous formulation of BANG gel, where a strong dependence on the temperature during the readout process through MRI was observed. This difference can be attributed to diverse physical phenomena forming the basis of each reading systems. The water proton NMR relaxation rates, which are calculated from series of MRI images and which are calibrated to dose in BANG-type polymer gels, are known to be temperaturesensitive. However, the mechanism of temperature dependence (or rather lack thereof) of Mie-scattering of light on sub-micron-sized poly (methacrylic acid) particles surrounded by a mixed solvent and grafted to gelatin in the gel is less obvious as the refractive indices (which ultimately determine light-scattering) of all components are likely to be somewhat temperature-dependent. It may be hypothesized that they all undergo very similar trend with varying temperature, therefore the light scattering efficiency coefficients, which depend upon the relative, not absolute refractive indices, exhibit little change (Van de Hulst, 1981).

With respect to the radiation quality, Fig. 11 shows that the response of the gel is quite linear up to 15 Gy, independently of the

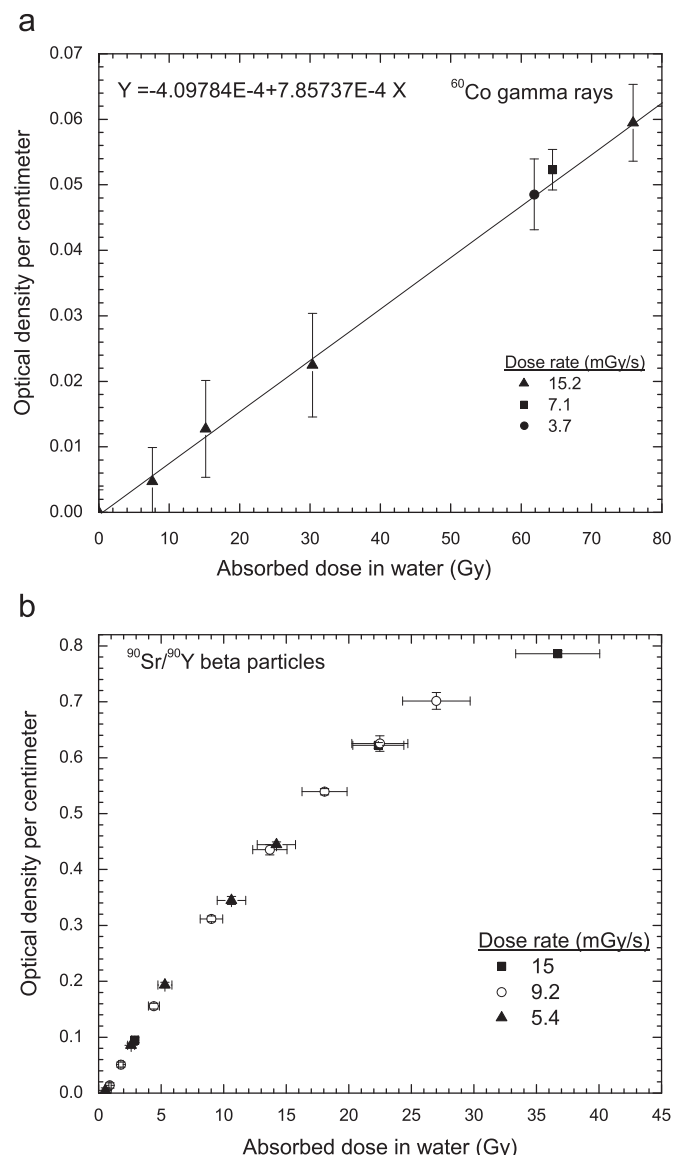


Fig. 12. (a) ^{60}Co gamma rays induced-response; dose-rate dependence. The line presents a linear fit. The error bars indicate the measurement uncertainty at one standard deviation and (b) response to $^{90}\text{Sr}/^{90}\text{Y}$ beta particles; dose-rate dependence. The error bars indicate the measurement uncertainty and dosimetry uncertainty at one standard deviation.

radiation type. However, there is a difference between beta particles and gamma rays for doses higher than 2 Gy. As it can be observed in Fig. 11, this polymer gel appears to be more sensitive to $^{90}\text{Sr}/^{90}\text{Y}$ beta particles than ^{60}Co photons, although this may not be significant given the uncertainty in the source calibration (ICRU 72, 2004; Soares et al., 2001). It has been reported that for another version of BANG gel, the sensitivity decreases as the mean energy electron increases (Novotny et al., 2001).

In addition, the response of the gel for both ^{60}Co gamma rays as well as $^{90}\text{Sr}/^{90}\text{Y}$ beta particles displayed in Figs. 12a and b, respectively, indicates that there is no dose rate dependence within the dose rate interval studied, because all the results fall on the same curve. This fact agrees well with results reported elsewhere for different version of BANG gel where dose rate dependence was not observed (Novotny et al., 2001; Maryanski et al., 1996a,b).

5. Conclusion

We have studied the properties of a new polymer gel under different radiation conditions using a micro optical CT scanner. We found that the gel response is linear from 0.5 Gy up to 15 Gy. It is independent of the dose rate for dose rates lower or equal to 15 mGy/s. Furthermore, it is reproducible over post-irradiation times of up to one month. On the other hand, for doses larger than 15 Gy, the response of the gel due to the irradiation does not remain stable over time. The value drops slowly as a function of the time post-irradiation, i.e., after a given period of time after which the irradiation was completed. The decrease in the value of the response is larger for higher post irradiation times. Therefore these effects should be taken into account if the gel is used in high dose rate radiation fields. For these cases, the gels should be scanned as soon as possible after the exposures are completed to minimize these effects. In this work we proposed to address this issue by performing multiple irradiations using low fraction doses. In addition, contrary to magnetic resonance imaging, with optical CT, the readout process is not affected at all by the temperature of the gel. A higher sensitivity is observed for beta particles than for ^{60}Co gamma rays.

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