Clinical Applications of Gel Dosimeters

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1. Gel dosimetry
Clinical applications of gel dosimetry have been considered since the 1950s [1,2]. During the last two decades, however, the number of investigators has increased, and the body of knowledge regarding gel dosimetry has expanded considerably [3,4]. Gel dosimetry is still considered by some to be a research project, and its introduction into clinical use is virtually at a standstill. However, the interest in, and potential of, gel dosimetry for clinical use is highlighted by the level of participation in three successful international workshops held to date on this subject [5-7] with a fourth workshop scheduled in 2006. This paper will review the characteristics of gel dosimetry that make it desirable for clinical use, the postulated and demonstrated applications of gel dosimetry, and some complications, setbacks, and failures that have contributed to the slow introduction into routine clinical use.

1.1. Fricke Gels
NMR-based gel dosimetry was introduced by Gore who recognized that the ferrous sulfate Fricke dosimeter [8,9] could be examined with magnetic resonance rather than spectrophotometry [3]. The Fricke dosimeter is based on the radiation-induced and dose dependent transformation of ferrous (Fe²⁺) ions into ferric (Fe³⁺) ions. These two ions have different electron paramagnetic spin states and different ionic radii [8,9]. Gore et. al. realized that the NMR spin-lattice and spin-spin relaxation rates (1/T₁ and 1/T₂, respectively) of the water protons in the Fricke dosimeter are dependent upon the amount of ferric ion present in the solution and that, because changes in these parameters produce the contrast of MR images, radiation induced changes in the solution should be visible by MRI [3]. Soon afterwards, other researchers began investigating the use of Fricke solutions incorporated into gel matrices (Fricke gels) to provide spatial stability of the dosimeter [10-13]. The most common matrices investigated were gelatin, agarose, and sephadex. Each of these systems had its advantages and limitations, but agarose was probably used more than any other detector system. While agarose dosimeters are more sensitive to dose than gelatin-based systems, they are more difficult to manufacture because they must be bubbled with oxygen to ensure a uniform dose response.

Fricke gel dosimeters have a number of advantages; principal among them is the well-described understanding of the radiation chemistry of this system. In addition, the basic and NMR processes leading to the dosimetry response are well understood [14,15]. Fricke gel dosimeters are tissue equivalent over a large range of photon energies. Like other gel dosimeters, they are prepared in a
liquid form so that phantoms containing heterogeneities or conforming to anthropomorphic geometries can be constructed.

However, there are number of significant problems associated with the use of Fricke gels for radiation dosimetry. The dosimeters require high doses, on the order of 10-40 Gy, for the radiation-induced changes to be observed by MRI. The ferric ions produced by absorption of radiation diffuse readily through the gel or agarose matrix, leading to a decrease in signal intensity, and a loss of spatial information [13-17]. Imaging must be performed within about two hours of irradiation to avoid serious degradation of the dosimetric detail [18]. The diffusion has been reduced by replacing the gelatin matrix with a poly vinyl alcohol (PVA) matrix, which is less porous to the ferric ions [19]. Other investigators have developed further methods to delay diffusion, although imaging must still be performed quite soon after irradiation [20]. Some improvement in the diffusion of ions can be achieved by cooling the gel, but this is rarely practical in a clinical setting. Consequently, Fricke gel dosimetry has seen only limited clinical use.

Several improvements have been reported recently. For example, a Fricke gel dosimeter manufactured using a polyvinyl alcohol (PVA) cryogel technology has been described. PVA is a common water-soluble polymer which can be cross-linked into its cryogel form by simply freezing and thawing. The cryogel is a rubber like material which holds its shape even at elevated temperatures. Preliminary reports of the PVA Fricke gel dosimeter indicate that its $(1/T_1)$ response has been found to be linear from 0-10 Gy, and the ion diffusion constant was found to be only 0.2 to 0.5 that of traditional preparations in gelatin or agarose [21,22]. Representative ion diffusion constants have been described [15].

Some preliminary work using Fricke gel dosimetry in anthropomorphic phantoms has been reported [23]. Several different gel compositions were investigated, including a lung equivalent gel that was developed with a density of 0.4 g/cm$^3$. This allows measurements of dose within the heterogeneity itself. However, diffusion of ions continues to be a problem with this dosimetry system [24-26].

### 1.2. Polymer Gels

Gels that replaced the Fricke solution with acrylic monomers were introduced in 1992 [27-29]. Early work was conducted using a poly-acrylamide gel based on the radiation-induced polymerization and cross-linking of bis and acrylamide. The formation of acrylic polymer chains largely resolved the problem of diffusion exhibited by Fricke gels, as the long polymer chains were too large to diffuse rapidly. The reciprocal of $T_2$, or $R_2$, the relaxation rate, was found to vary proportionally with dose, and MR imaging of polymer gels was shown to yield quantitative dose distributions [28].

A variety of polymerizing gels have been developed, many of which are based on acrylamide or acrylic acid, and are referred to as polyacrylamide gels (PAG) [30,31]. One of these, the BANG™ polymer gel system is a proprietary PAG dosimeter made of a mixture of acrylic monomers in a tissue-equivalent gel(MGS Research, Inc., Guilford, CT). Early BANG gels were made from acrylic acid monomers and methylene-bis-acrylamide crosslinker. More recently, the BANG3® dosimeter was introduced, which contains methacrylic acid monomer (see Table 1, from reference 31). Other proprietary response modifiers were added to adjust the dose range and sensitivity. Dissolved oxygen inhibits free radical polymerization reactions and is removed from the mixture by passing nitrogen through it while the gel remains above the gelling temperature, prior to sealing the vessel. Consequently, vessels of glass or other material not permeable to oxygen must be used for irradiating and imaging the gels.
Table 1. Composition of BANG3® Polymer Gel Dosimeter

| Component                  | Percentage |
|----------------------------|------------|
| 6% methacrylic acid        |            |
| 1% sodium hydroxide        |            |
| 5% gelatin                 |            |
| 88% water                  |            |

The gelling agent in the BANG dosimeter is gelatin, which is used because the transverse NMR relaxation rate of water ($R_2 = 1/T_2$) in a gelatin gel is nearly an order of magnitude lower than that in agarose gels. Therefore the background $R_2$ in the gel is substantially reduced, which improves its dynamic range.

However, the polymer gels continued to show another disadvantage; their response was inhibited by the presence of oxygen. This effect was addressed through the recent introduction of a class of polymer gel dosimeters containing oxygen scavengers [32,33]. Several variations of these normoxic gel dosimeters (so called because they can be prepared under normoxic conditions) have been characterized [34].

1.3. Radiochromic Polymer Dosimeter
A new class of three-dimensional dosimeter (PRESAGE™, Heuris Pharma, LLC, Skillman, NJ) has recently emerged, that consists of an optically clear polyurethane matrix containing a leuco dye. The dye exhibits a radiochromic response when exposed to ionizing radiation, changing color in proportion to the radiation dose. The response of the dosimeter is evaluated through optical computed tomographic (OCT) imaging, and will not be discussed further here. Early results indicate that the optical response of the dosimeter is linear to within 1%. The dosimeter exhibits only a small dependence on dose rate, and is stable over a period of several days following irradiation. However, the dosimeter material is in general more dense than tissue, with CT numbers ranging from 100 to 470. A temperature dependence of the optical response has been observed, indicating that careful temperature control during readout is required [35]. Both the spatial and dose resolution (when OCT is used) are comparable to polymer gels [36].

1.4. MR Imaging of Polymer Gels
Irradiation of the polymer gels induces polymerization and cross-linking of the acrylic monomers. As polymer microparticles are formed, they reduce the NMR relaxation times of neighboring water protons. Magnetic resonance imaging can be used to measure dose distributions in the gel [28,29,37]. Water proton NMR transverse relaxation time $T_2$ can be determined from multiple spin-echo images. Images can be acquired using the Hahn spin-echo pulse sequence: $90^\circ - \tau - 180^\circ - \tau - \text{acquire}$ for four or more different values of $\tau$. Typical pulse sequence parameters are TR = 2 seconds, TE = 11, 200, 400, and 600 ms. A field of view of 24 cm and a matrix of 128 x 256 can be used, with one acquisition and a 3 mm slice thickness.

More recently, it has been shown that spin-echo sequences other than the Hahn sequence described above can be used for gel imaging. Improved dose resolution can be achieved through the use of multiple spin-echo pulse sequences. [38,39] Optimization of the imaging sequence is necessary, especially with regard to the number of echoes measured. The use of multi-planar imaging can reduce imaging times but can also lead to interference between image planes.

Once MR images have been obtained, they are most conveniently transferred via network to a computer for which a data analysis and display program has been written. One example of such a program has been described previously [29]. The program calculates $R_2$ maps on the basis of multiple TE images, using a monoexponential non-linear least squares fit based on the Levenberg-Marquardt algorithm [40]. The program also creates a dose-to-$R_2$ calibration function by fitting a polynomial to a
set of dose and $R_2$ data points, obtained from gels irradiated in test tubes to known doses. This function can then be applied to any other $R_2$ map so that a dose map can be computed and displayed.

Figure 1 shows values of transverse relaxation rates ($R_2$) for the gels as a function of dose. The pooled data show that the dose response was highly reproducible over a wide range of doses. The dose response is well fitted by a straight line [41].

![Figure 1](image.png)

**Figure 1.** The dose dependence of the transverse relaxation rate ($R_2$) as a function of dose. Data from several experiments are shown indicating reproducibility over a wide range of doses. Reproduced with permission from reference 41.

Additional experiments have shown that the response of the BANG gel can be adjusted by varying the concentration of crosslinker used per total amount of comonomer [42]. Similar data have been shown more recently for several different polymer gel mixtures [43].

The temperature of imaging has a large effect on both the gel sensitivity and its dynamic range [42,44]. Sensitivity is seen to reach a maximum at about 50% crosslinker but sensitivity at all concentrations increases as the temperature at the time of imaging is reduced. The maximum $R_2$ achievable, and therefore the dynamic range of the gel, is dependent upon the temperature at the time of imaging. While $R_2^{\text{max}}$ increases with crosslinking, the dependence is enhanced by cooling the gel during NMR measurement.

For a number of gel compositions presently being evaluated, the fundamental chemistry and physics of response are well understood. Several gel compositions have been characterized in great detail [29,34,37,45,46]. In polymer gels, for example, it is understood that the interaction of radiation with water produces free radicals, which trigger the cross linking of monomers into polymer chains [28,47]. The polymer chains bind water protons tightly causing a change in their paramagnetic properties that is detectable by magnetic resonance imaging [37,48]. The relationship between dose and relaxation rate can be influenced by several additional factors, including accuracy of the calibration curve [49] and the aging characteristics of the gel [45,50,51].

The quality of the imaging process is affected by the homogeneity of the $B_1$ field [52] and the presence of eddy currents [53]. Some additional complications due to the distortion of MR imaging systems have been identified [54].
1.5. Optical Scanning of Polymer Gels
The equipment and imaging procedures associated with optical computed tomography (OCT) of polymer gels are addressed elsewhere.

1.6. X Ray CT Scanning of Polymer Gels
The formation of polymer chains increases the physical density of the gel, and the resulting change in attenuation coefficient can be measured by measurements of x-ray transmission such as by computed tomography [55-60]. While the change in density is small, it has been shown to vary proportionally with dose [58-61]. This change in density leads to a small change in CT number when irradiated gels are examined with CT. Recent data show that this change can be as much as 0.2 kg/m\(^3\) per gray [62]. An image of tubes of gel irradiated to different doses appears in Figure 2. Methods for improving the quality of x-ray imaging have been developed, and include the acquisition of multiple images, background subtraction, and filtering [61-62].

![Figure 2. CT image of several vials of polymer gel irradiated to different doses. Reproduced with permission from reference 63.](image)

1.7. Ultrasound Imaging of Polymer Gels
Polymerization leads to changes in elasticity of the medium, and the corresponding changes in ultrasound absorption can be exploited [64-67]. Ultrasound has been used to evaluate changes in density and elastic constant of a number of materials. Several different ultrasonic parameters can be measured and these can be used to characterize materials. The most commonly measured parameters are attenuation and reflection coefficients, and the speed of propagation. A pulse-echo technique using one probe or a transmission technique using two probes is used to measure these parameters. These parameters can be related to structural properties of the sample, including bulk density, elastic constants as well as sample inhomogeneities.

1.8. Vibrational Spectroscopic Imaging of Polymer Gels
Finally, vibrational spectroscopy can be used to demonstrate the conversion of monomers to polymer chains [68-72]. FT-Raman vibration spectroscopy of polymer gel dosimeters has been investigated as a means by which the fundamental structure and properties of the dosimeters might be better understood. Raman spectroscopy has also been used to investigate the track structures of proton beams in polymer gel dosimeters [73]. This study illustrated the difficulty in using polymer gel dosimeters to extract quantitative dose maps when exposed to proton radiation. Further studies will be required to
determine whether Raman microscopy can be used routinely in the evaluation of polymer gel dosimeters.

2. Characteristics of Gel Dosimeters
Gel dosimeters have a number of characteristics that make them attractive for radiation dosimetry [74]. Oldham has shown that gels compare favorably with other common detectors in most characteristics, including their relative accuracy, volumetric nature, inherent three-dimensionality, high resolution and lack of energy dependence over much of the important energy range [75]. Methods for characterizing the response of gels have been found, and in particular, a technique for characterizing the dose resolution has been described [38,76,77].

However, today gels are still time-consuming and relatively expensive. Several dosimetric aspects have not yet been realized, including the absolute accuracy of measurement, and the ability to render a 3-dimensional dose distribution as opposed to multiple planes of data, although progress is being made rapidly on both aspects. In addition, the issues of cost and time required are being addressed. The availability of optical CT scanning and other imaging techniques are likely to drive down the cost of gel analysis, and improve the penetration of this modality into the clinic. At the same time, newer optical CT scanners equipped with more powerful computers are faster and can perform comprehensive imaging of gels in the time previously required for a single slice.

3. Applications of Gel Dosimetry
Potential applications of gel dosimetry have been summarized on several occasions in the recent past [46,74,78-80] although the field is developing rapidly. Today it is considered by many that gel dosimetry has useful characteristics that can facilitate radiation therapy dosimetry, especially in situations that are not handled well by conventional dosimeters. These characteristics include the ability to measure complex three-dimensional dose distributions; to integrate dose accurately without dependence on dose rate, at least over a fairly wide range; tissue-equivalence; high spatial resolution; and lack of energy dependence over most of the kilovoltage and megavoltage range. With most gels, data are stored permanently, making gels suitable for performance of dosimetry at remote locations [81]. They also are relatively safe to manufacture and handle, although some components such as acrylamide are toxic and must be handled with appropriate protection until mixed.

Demonstrated applications of gel dosimetry to date include basic dosimetry (depth dose, penumbra, wedge profiles) in photon, electron, neutron and proton beams; dose distributions from imaging procedures; conformal therapy, stereotactic radiosurgery, and intensity-modulated radiation therapy (IMRT); dose distributions around brachytherapy sources (low- and high-dose rate, and intravascular sources); internal dosimetry (iodine-131 doses); and evaluation of tissue heterogeneities. The advances made recently in these areas will be discussed.

3.1. Basic Dosimetry
Gel dosimeters have the capability to record and display the dose distribution throughout a three-dimensional volume. This ability affords advantages over conventional dosimeters, even for basic dosimetry parameters such as percent depth dose in photon and electron beams [2,37,82]. Gel dosimetry has been shown to be useful to validate simple multiple-field arrangements [83] and more complex anatomical situations including tangential breast treatment [84,85], conformal therapy [86] and scalp treatment with electron beams [87]. Dynamic functions, such as a programmable wedge filter are difficult to measure with ionization chambers or diodes, and film is often used to provide data in a single plane. Gels have proven useful for capturing the dose distributions from programmable wedge filters, and allow distributions in multiple planes to be demonstrated from a single exposure [88].
3.2. Dose from Imaging Procedures

More recently, the use of gels to demonstrate dose distributions from imaging procedures has been explored [89,90]. In a novel experiment, a high-sensitivity gel was used to determine the dose from CT imaging. The benefit of this measurement is that the dose distribution throughout a patient volume can be estimated without requiring the use of numerous point dosimeters (e.g., TLD) and without averaging the dose along a line or throughout a volume (e.g., a pencil ionization chamber). These benefits may be most apparent in evaluating the dose distribution from helical CT scanners.

4. Evaluation of conformal dose distributions

4.1. Stereotactic radiosurgery

Gels have been used to demonstrate the dose distributions from stereotactic treatments both from dedicated multisource cobalt units and from linear accelerators [91-98]. A clear benefit of gel dosimeters is that they can display a dose distribution, especially a highly conformal one as is produced by stereotactic radiosurgery techniques, so that it can be appreciated qualitatively in three dimensions without need of imaging systems or processing (see Figure 3, reference 95).

In one series of measurements, gels were prepared in glass flasks chosen for their size and shape, which was comparable to that of a human head. Additional polymer gel material from the same batch was prepared in glass test tubes, for irradiation to selected doses, to generate a dose-response curve. The gels were prepared in Guilford, CT, and were shipped to Lexington, Kentucky for irradiation and analysis.

A gel prepared in a 16 cm diameter flask was fitted with a radiosurgical head frame (Leksell, Elekta Corporation, Atlanta, GA.), as shown in Figure 4. This flask was also equipped with a glass rod extending to near the center of the flask, to be used as a target. MR images were obtained and were transferred to a Gammaknife treatment planning computer (Gammaplan; Elekta Corporation), where a complicated dose distribution was planned using multiple target points. Once the plan was completed, the coordinates of the individual target points were determined, and the gel was moved to the Gammaknife irradiation unit. Treatments were delivered to each of the target points, in accordance with the treatment plan. A dose of 10 Gy was delivered to the 100% isodose line.

![Figure 3. A BANG® gel irradiated with a highly-conformal dose distribution produced by a Gammaknife treatment unit. The distribution can be appreciated qualitatively without the need of imaging systems or processing. Photograph by the author. See also reference 95.](image-url)
Figure 4. Photograph of a glass flask filled with the BANG® Polymer Gel dosimeter. A glass rod was inserted into the gel to provide a target around which to localize the dose distribution. The flask was fitted with a Leksell stereotactic head frame. The gel is shown as it appeared following irradiation. Photograph by the author.

Figure 5. \( R_2 \) maps obtained from the irradiated gel (a) Distribution in the axial plane. (b) Distribution in the sagittal plane. Reproduced with permission from reference 95.
Figure 6. Composite figures showing both the treatment plan prepared using a Gammaplan treatment planning computer (drawn in black, labeled in percent of maximum dose) and isodose curves measured by the technique described in the text (drawn in gray, labeled in Gy). (a) The distribution in the axial plane containing the 8 isocenters. (b) The distribution in a perpendicular sagittal plane. Reproduced with permission from reference 95.

Dosimetric imaging of the flask and test tubes containing gel was performed between twenty-four and thirty-six hours after irradiation. The flask was placed in the head coil of the imager and the test tubes irradiated for calibration purposes were placed around the flask. The images were transferred via network to a Macintosh computer, and the “DoseMap” program was used to compute the maps of transverse relaxation rate ($R_2$).

A dose-response calibration curve was obtained as described earlier. Images of the gel-filled test tubes were obtained, and $R_2$ determined as a function of dose.

The calibration curve was then applied to $R_2$ maps of the flask irradiated with the Gammaknife unit. The result yielded an image of the dose distribution, as shown in Figures 5 a and b. As all scans were performed with the head ring and localizer box in place, the coordinates of the image plane could be determined. These image planes were located 1 mm from each of the corresponding treatment plans shown in Figures 5 a and b. Finally, isodose curves were drawn (by the DoseMap program) by interpolating within the measured dose distribution.

The measured dose distributions were compared with the treatment plans prepared prior to irradiation by superimposing the two data sets. The superimposed data are shown in Figures 6 a and b. The calculated and measured dose distributions were registered by aligning the point representing the tip of the glass rod.

4.2. Evaluation of repeat-fixation stereotactic radiotherapy
In recent years, fractionated stereotactic radiation therapy has been seen as a desirable method of delivering high-dose radiation therapy to malignancies of the brain. Techniques developed for immobilizing the patient have also been applied more recently to intensity-modulated radiation therapy, in which conformal dose distributions are delivered through multiple fractions to one or more target volumes. In both techniques, reproducible positioning of the patient is critical, to ensure that the target volume receives the intended dose, and normal tissues are spared to the extent determined by
treatment planning techniques. The BANG® gel dosimeter has been used in a fractionated regimen to demonstrate the reproducibility of multiple setups under stereotactic position methods (Meeks, 1999).

4.3. Intensity-Modulated Radiation Therapy (IMRT)

Gels dosimeters have proven themselves to be valuable for evaluating and confirming IMRT dose distributions [83,99-106]. Most investigations have been conducted in simple geometric phantoms (Figure 7), but others have employed anthropomorphic phantoms in arrangements that allowed direct comparison with measurements using other techniques such as film and TLD [100,102,103].

Beach developed a gel insert for an existing anthropomorphic phantom that had been developed with film and TLD dosimeters [107]. The phantom design revision included converting the existing imaging/dosimetry insert from a block-style design to a cylindrical design (Figure 8). This insert contained embedded structures that simulated a primary and secondary target volume as well as an organ at risk (OAR). An additional insert was then constructed to house the polymer gel dosimeter. This insert was specially designed using Barex® plastic. Both the imaging insert and the gel insert had an image registration system incorporated into their construction.

X-ray CT images were obtained of the phantom with the imaging insert in place, and an IMRT treatment plan was developed. The phantom was then taken to the linear accelerator, the imaging insert was replaced with the gel insert, and the IMRT treatment was delivered.

Repeated measurements showed that a polymer gel imaged with optical CT was reproducible to within 1% [108]. Repeated OCT imaging was shown to be consistent to within 1%. However, the results also showed that the techniques used to calibrate the gel (irradiation of a similar gel container with small-diameter beams delivering doses spanning the expected range) did not provide absolute dose measurements offering better agreement than 10% with the calculated data.

**Figure 7.** A cylindrical flask containing a normoxic gel shortly after irradiation with an IMRT treatment. The dose distribution is clearly visible, demonstrating the change in optical density with dose. Reproduced with permission from reference 104.
Duthoy compared the dose distribution measured with gels to the calculated distribution, for complex intensity-modulated arc therapy (IMAT) treatments in the abdomen [109]. Vergote also examined IMAT with gels and observed a reproducible difference between calculations and measurements in low-dose regions near steep dose gradients; a phenomenon also observed by Cadman et. al. and attributed to the failure of treatment planning systems to model the transmission of radiation through the rounded ends of multileaf collimator leaves [106,110].

4.4. Brachytherapy
Determining dose distributions and confirming the results of planning for brachytherapy treatment is historically difficult. No suitable methods of dosimetry have existed in the past to enable

Figure 8. An anthropomorphic head-and-neck phantom developed by the Radiological Physics Center [107] showing the modifications made to accommodate a gel dosimeter.

Figure 9. (a) A calculated dose distribution for an IMRT treatment, shown in a gray-scale format. (b) The measured dose distribution obtained from optical CT of a polymer gel, following irradiation with the treatment plan shown in (b). Reproduced with permission from reference 102.
measurement and display of these three-dimensional and complex distributions. Measurements around single sources have been possible only in a point-by-point fashion, such as with small ionization chambers or with thermoluminescence dosimeters (TLDs), [111] or in planar fashion with film [112]. These methods are quite unsatisfactory for anything other than distributions around single sources, or very simple source arrangements. In contrast, the BANG polymer gel dosimetry system has the capability to measure and display complex dose distributions from complicated source arrangements. It is necessary to immerse the applicator containing the sources into the gel, or arrange for its introduction into a catheter already placed in the gel.

The ability of gels to record and display dose distributions around a high-dose rate (HDR) source was first demonstrated over a decade ago [37,113,114]. Maryanski et al showed the dose distribution around a single catheter into which a high-dose-rate (HDR) remote afterloader source had been positioned [115]. The HDR unit was programmed to dwell the source at several locations in the catheter, to deliver an elliptical dose distribution. After irradiation, the gel was imaged with MR, and a map of the dose distribution was computed. The map is shown in Figure 10, where the color intensity is proportional to dose. Isodose lines, determined from the dose map data, are superimposed on the intensity map. Points at which the dose was computed by the treatment planning system also are shown. Excellent agreement between the position of the calculated dose points and the corresponding measured isodose lines indicates the agreement between doses measured by the gel and computed by the treatment planning system.

More recently, measurements have been made in close proximity to HDR 192Ir sources [116-118].

**Figure 10.** Use of the BANG gel to measure the dose distribution around an HDR source. The source was positioned in a catheter implanted in a BANG polymer gel. The figure illustrates a comparison between the dose distribution determined from a MRI image of the gel and the calculated dose distribution. (From reference 115).
These measurements have shown that complications occur when measurements are made in the steep dose gradients close to an HDR source. Polymerization of the gel causes an increase in the gel density and a corresponding decrease in the volume filled by the gel. The change in density causes shrinkage of the gel in the vicinity of the source, distorting the resulting measured distribution. Changes to the composition of the gel to increase the concentration of gelatin can mitigate the amount and effects of the density changes. Furthermore, there are suggestions that the high dose rates found near brachytherapy sources, particularly those of HDR afterloaders, can introduce temperature gradients which influences the polymerization of acrylamide monomer gels [34,42,119,120].

Efforts also have been made to characterize low-dose rate (LDR) sources such as prostate seeds [121-123], eye plaques [124], $^{137}$Cs afterloading sources [125,126] and intravascular sources [127,128]. Studies have indicated that the diffusion of monomers (or ferrous and ferric ions in Fricke gels) across steep dose gradients can introduce errors in measurement [37,129]. As the use of gels to measure dose distributions from LDR sources requires long exposure times, diffusion of monomers or ions could introduce significant errors, and gels exhibiting high diffusion rates should be avoided for these measurements.

A further problem with gel dosimetry for LDR brachytherapy has been demonstrated by recent studies indicating energy dependence at lower energies. Data show that a polymer gel dosimeter under-responds to radiation in the 20 keV – 60 keV range [130]. Others have shown differences in gel response from one formulation to another, and suggest that the MAGAT gel is most water-equivalent over a wide range of energies [131]. Changes in mass attenuation coefficient of polymer gels during irradiation can also introduce errors in the dose distributions measured around low-energy sources.

4.5. Internal Dosimetry

Gel dosimetry has shown promise in the determination of dose distributions from administrations of unsealed radioactive sources. [132] The authors embedded a vial of iodine-131 into a flask of polymer gel and observed a distance-dependent change in the T2 signal. They also mixed iodine-131 into the gel and demonstrated a change in T2 signal that was dependent on distance from the concentration of activity. No more-recent investigations have been located.

4.6. Measurement of Neutron Dose Distributions

Some developments have been reported in characterizing fast and epithermal neutron beams with gel dosimetry [133-135]. Thin layers of Fricke-xylene orange gels have been irradiated in phantoms composed of insensitive gel. Adding $^{10}$B or other nuclides with large cross-sections has increased the sensitivity of the gel dosimeter to neutrons. This technique has been used to determine the profiles of neutron beams used for boron neutron capture therapy. Some benefits of the use of gel dosimetry are the tissue-equivalence of the dosimeter to these energies, and the ability to separate the components of dose.

4.7. Measurement of Particle Dose Distributions

Several investigators have demonstrated the use of polymer gel dosimeters to record the dose distributions produced by proton beams [41,73,136-138]. However, several authors have noticed disagreements between measurements with gels and conventional dosimeters such as diodes in the peak region of the distribution. Gustavsson has suggested that the response of gels, as they are based on the formation of free radicals, is dependent on the LET of the radiation [137,138]. As the LET of the beam increases in the peak region, the local ionization density increases. As the distance between the radicals formed in the gel decreases, the likelihood of recombination of radicals increases. A decrease in the production of radicals with increasing LET has been described previously [139]. Consequently, significant differences appear between depth dose measurements with gels and those with detectors such as diodes (see Figure 11, reference 138).

Jirasek et. al. performed track energy-deposition calculations and raman spectroscopy and reported agreement between these techniques and gel measurements [73]. Their conclusion also was that the
high density of delta-ray interactions close to the track of a proton resulted in high doses being
delivered to the gel. These doses saturated the response of the gel by consuming the available
monomer. This effect was greater near the end of the proton range, consistent with the results of other
authors.

Figure 11. The variation in LET as a function of depth for a monoenergetic proton beam (dashed
curve, left-hand scale) and the measured relative sensitivity for the gel dosimeter (full curve, right-
hand scale). Also shown is the depth dose curve for the proton beam (dotted curve), normalized to
100% at the Bragg peak. Reproduced with permission from reference 138.

Gels have been used also to demonstrate the dose distribution produced by $^{12}$C ions [140]. Similar
effects associated with decreased radical formation at high LET were observed in the carbon beam.

4.8. Evaluation of Tissue Heterogeneities
A valuable feature of gel dosimeters is that they are very nearly tissue-equivalent, particularly at
photon beam energies above about 100 kV. Previous investigations have shown that the BANG™ gel,
the MAGIC and MAGAS normoxic gels, as well as gels based on Fricke or vinyl solutions have
electron densities within 1% of soft tissue, and effective atomic numbers in the range of 7.4 [130].
However, several investigators have attempted to measure the effects of non-unit density tissues on
external beam dose distributions. Early measurements were performed to estimate the dose
distribution behind high atomic number heterogeneities, to simulate the presence of bone [141-143]
More recently, measurements have been made behind or adjacent to cavities filled with air or with
lung-equivalent plastic [105,144]. To attempt a measurement in lung-equivalent gel, Olberg produced
a foam of gel with the approximate density of lung tissue [145]. Other investigators have evaluated
the promise of gel dosimeters to simulate lung tissue, by introducing polystyrene foam beads into a gel
mixture [146,147]. While these measurements showed promise, there were several potential sources
of error. First, the introduction of air, or air-containing polystyrene beads introduced the possibility of
oxygen contamination. Purging the polystyrene beads with nitrogen, or using nitrogen rather than air to foam the gel addressed this problem. The introduction of air or polystyrene eliminated the possibility of evaluating the measured dose distribution by optical scanning, and MR imaging must be used. The presence of air may lead to partial volume imaging effects that could introduce errors into the measurement.

5. Complications to be Addressed
As was suggested earlier in this paper, there are a number of complications associated with gel dosimetry that remain to be addressed, and that are inhibiting the routine use of gels in the clinic. Some of these are listed below, with short descriptions of the causes of the problems, and possibilities for correcting them.

5.1. Imaging Artifacts
This paper has discussed several methods of generating images of dose distributions using gels. Principal methods are magnetic resonance imaging (MRI), optical computed tomography (OCT), and x-ray computed tomography (CT). Each of these imaging methods is prone to imaging artifacts, although the type of artifact and its causes are different with the different modalities. In MRI, for example, susceptibility artifacts can result from variations in the conductivity of the volume being imaged, and interference is likely when multiplanar imaging of adjacent planes is attempted. The presence of air or low-density structures can lead to partial volume effects or susceptibility artifacts.

In OCT, any structure that blocks the light beam is likely to cause a streak artifact, similar to those produced by high densities in x-ray CT images. In addition, the refraction of the light at interfaces between the gel and other materials can cause ring artifacts or distortion of the image. The artifacts found in OCT images have been described [75]. An example of the artifact caused by high optical densities is shown in Figure 12.

![Figure 12](image)

**Figure 12.** An optical CT scan of a normoxic gel irradiated with a low dose rate $^{125}\text{I}$ brachytherapy source. The high optical densities close to the source completely attenuate the laser, resulting in a streak artifact.

When x-ray CT is used, artifacts can result from the low signal to noise ratio that occurs because of the very small density differences present in the gel. These artifacts have been investigated in some detail previously [57].

5.2. Temperature Dependence
The existence of a dependence on temperature during irradiation of polymer gels was not recognized immediately, but it has since been shown that this dependence exists. Furthermore, the temperature dependence can be more pronounced for some polymer gel formulations than others. The
polymerization that occurs as a result of irradiation of the gel is exothermic, and consequently can lead to a temperature rise that influences further polymerization of gel in response to continuing exposure. In extreme cases, this temperature rise can exceed several degrees C [148].

5.3. Oxygen Sensitivity
The sensitivity of polymer gels to oxygen has been discussed extensively, and several investigators have responded by developing gels that contain oxygen scavengers, such as the MAGIC gel [33]. The oxygen scavenger removes oxygen present in the gel at the time of manufacture, even if this is done in normoxic conditions. It can remove additional small amounts of oxygen, but ultimately will be overwhelmed if exposure to normal atmosphere is ongoing. While this problem has been addressed, it still creates minor inconvenience that might limit the successful introduction of gels into routine clinical use. The characteristics of several normoxic gel dosimeters have been investigated in detail [39,149].

5.4. Tissue Equivalence and Energy Dependence
Gels, both Fricke and polymer types, compete well when compared to other dosimeters in terms of their tissue equivalence and energy dependence. In comparison to thermoluminescence dosimeters (TLD), radiographic film, and even ionization chambers, for example, gels are considerably less energy dependent and are much more tissue equivalent [150]. However, under extreme conditions of photon energies below 60 keV, and LET values greater than about 2.5 keV/micron, gels show a dependence that remains to be fully characterized [130].

5.5. Simulation of Non-Unit Densities
The benefits of gels discussed in the previous paragraph lead to the inability of gels to easily simulate non-unit density tissues. To date, limited efforts have been described to create low-density gel mixtures, to simulate lung tissue. No attempts have been described to date to create high-density mixtures and are unlikely to be with today’s emphasis on the use of gelatin-based formulations.

5.6. Diffusion of Monomer in Steep Gradients
The diffusion of monomer, and the shrinkage of gel proportional to the creation of long polymer chains, can be addressed through the development of better gel mixtures. Avoiding small monomers such as acrylamide can reduce the rate of diffusion in regions of steep dose gradient, such as the penumbra of radiation beams [151]. Employing different concentrations of gelatin might reduce or eliminate problems associated with shrinkage of gels in high-dose regions. Some normoxic gels may demonstrate decreased diffusion in regions of steep dose gradient [39,147].

6. Summary and Conclusions
The importance of quality assurance in radiation therapy is well documented. High quality, safe, and effective radiation therapy is dependent upon the proper operation of equipment, the accuracy of alignment devices, and the dependability of dosimetry procedures. The accurate delivery of the prescribed dose depends on procedures involving dosimetry systems. Properly functioning dosimeters are necessary to assure that the equipment is properly calibrated and that treatment planning procedures are conducted correctly.

A wide variety of dosimetry systems are available to medical physicists today. Choosing the correct dosimetry system for any given application requires an understanding of the operation of the device and its appropriateness for the intended circumstances. This presentation has reviewed a number of dosimetry systems presently available, their design and operation, and some of the uses for which they are valuable.

Gel dosimetry offers the promise of accurate and convenient dosimetry under a variety of circumstances. In most of the examples discussed above, gel dosimeters offer a number of advantages over conventional dosimeters. Chief among these is the ability to measure a complex dose distribution
throughout a volume with a single radiation exposure. Additional advantages include tissue-equivalence, high spatial accuracy, good dose precision, and reasonable convenience.

However, gel dosimetry continues to experience little acceptance in the clinic, largely because some aspects of promise have not been achieved, and because of a perceived lack of convenience. Members of the radiation physics community are apparently not convinced that the benefits of gels sufficiently outweigh conventional dosimeters such as film and TLD. It is incumbent on those of us working with gels to encourage more widespread use, by taking every opportunity to display the results of measurements with gels.

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