3D dosimetry for brachytherapy and heterogeneities

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Abstract. Clinical applications of 3D dosimeters have been considered since the 1950s, when uses of gel dosimeters were first explored [1,2]. Since then, however, the number of investigators has increased, and the body of knowledge regarding gel dosimetry has grown considerably [3-5]. 3D dosimetry is essential for a number of clinical applications, including the complete characterisation of brachytherapy sources, and the evaluation of dose distributions around and within inhomogeneous tissues. However, each of these conditions presents several challenges which must be overcome to allow the accurate and practical determination of dose distributions.

1. Introduction
The dose distributions around brachytherapy sources and within and around heterogeneities present unusual difficulties for dose characterization. Brachytherapy sources are intended to be implanted directly in, or immediately adjacent to diseased tissue, and as a result, the most significant part of the dose distribution occurs within a few millimeters to a few centimeters from the source. In this region, the dose gradient is extremely steep and a measurement device must have both high spatial precision and low sensitivity to rapidly changing doses. In addition, a dosimeter that will be used to measure the dose distribution out to several centimeters from the source must have a dynamic range that accommodates several orders of magnitude change in dose rate.

The determination of doses within and around heterogeneous media is likewise extremely difficult [6,7]. Early measurements focused on measurements behind heterogeneities, because measurements within the heterogeneities were confounded by the effective density of the measurement device [8]. Most radiation detectors are non-unit density. Many diode detectors, for example, are largely silicon with an atomic number of 14, and have the potential to perturb the dose distribution. Inserting an air-filled ionization chamber into a low-density medium introduces electronic disequilibrium. Consequently, the perturbing effects of the dosimeter must be separated from the measurement and accounted for.

2. Brachytherapy sources
During the last several decades, low-dose rate interstitial brachytherapy using 125I and 103Pd encapsulated sources has become a popular treatment for prostate cancer, resulting in an increased demand for these sources. In response, manufacturers developed 131Cs sources and several new models of 125I and 103Pd sources. The dose distributions of these low energy emitters are very sensitive to internal geometry, encapsulation, self-absorption and filtration, all of which vary with each
manufacturer. The American Association of Physicists in Medicine (AAPM) Radiation Therapy Committee has recommended that the dosimetric characteristics of each new source be evaluated by at least two independent investigators [9]. The AAPM Task Group 43 protocol for interstitial brachytherapy sources details the dosimetric parameters needed to calculate the doses from these sources [10].

However, it is not only low-energy sources that present dosimetric problems. Even high-energy sources of $^{137}$Cs and $^{192}$Ir generate dose distributions with steep dose gradients. Measurements around high-dose rate $^{192}$Ir sources are further complicated by the high dose rates and perhaps even by the heat generated by the source.

There is currently no dosimeter in routine use that provides the desired characteristics to measure these dose distributions. Most experimentalists to date have used thermoluminescent dosimetry (TLD) in water-equivalent plastics, which add uncertainty due to their energy dependence, although a technique to use TLDs in water was described recently [11]. However, the use of TLDs to measure sufficient data to fully characterize a brachytherapy source is a very tedious process. Radiographic film allows the collection of a plane of data in a single exposure, but similarly presents issues of energy dependence, dose rate dependence, and dose linearity.

An ideal dosimeter would be tissue equivalent, have high spatial resolution in three dimensions, and be independent of energy and dose rate. In an effort to find a dosimeter that meets these requirements, the use of gels and polyurethane blocks as dosimeters has been investigated. These dosimeters often are nearly tissue equivalent at the energies emitted by the sources, provide high spatial resolution, and allow visualization of three dimensional dose distributions. Polymer gels are made up of water, gelatin, and acrylic monomers that polymerize when exposed to radiation. Ionizing radiation produces free radicals within the gel and these in turn lead to the formation of polymer microparticles that remain attached to or entangled with the gelatin [12]. A polyurethane dosimeter called PRESAGE® consists of a polyurethane matrix doped with radiographic leuco dyes that generate a color change, and consequently a change in optical density (OD) upon exposure to ionizing radiation [13]. Both gels and PRESAGE® must be evaluated for their energy dependence, but as they allow the collection of dose information in almost a complete 3D volume, great interest is shown in these dosimeters.

2.1. Measurements around brachytherapy sources

The ability of gels to record and display dose distributions around a high-dose rate (HDR) source was first demonstrated over a decade ago [14-16]. Most radiation-sensitive gels must be protected from oxygen, as atmospheric oxygen scavenges free radicals and halts the polymerization process, reducing the sensitivity of the gel. This complicates the measurement, as a brachytherapy source must be introduced without also introducing oxygen.

Maryanski et al showed the dose distribution around a single catheter into which a high-dose-rate (HDR) remote afterloader source had been positioned [16]. 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. Care was taken to protect the dosimeter from oxygen, through the use of a catheter material impervious to oxygen.

More recently, measurements have been made in close proximity to HDR $^{192}$Ir sources [17-19]. 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 influence the polymerization of acrylamide monomer gels [20,21].
Efforts also have been made to characterize low-dose rate (LDR) sources such as prostate seeds [22-24], eye plaques [25], $^{137}$Cs afterloading sources [26,27] and intravascular sources [28,29]. Studies have indicated that the diffusion of monomers (or ferrous and ferric ions in Frickel gels) across steep dose gradients can introduce errors in measurement [30]. 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 must 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 [31]. 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 [32]. Changes in mass attenuation coefficient of polymer gels during irradiation can also introduce errors in the dose distributions measured around low-energy sources.

The PRESAGE® dosimeter also has been used to measure dose distributions around brachytherapy sources. These investigations demonstrated some success, and Wai has shown good agreement between measurement and Monte Carlo calculations [33,34].

2.2. Intravascular brachytherapy sources

That gel dosimetry is able to visualize steep dose gradients in 3D makes it very attractive for acquiring the dose distributions of both LDR and HDR brachytherapy sources. In addition, gel dosimeters integrate the dose delivery over time so that the total dose distribution from a moving source can be measured. Different types of sources have been evaluated, as indicated above, including sources designed for intravascular therapy.

Although polymer gel dosimetry has been perceived as a suitable dosimeter for dose verification in brachytherapy, some additional difficulties may be encountered that are related to the insertion of applicators or catheters into the polymer gel. Significant dose deviations near the catheter wall have been noted [35-37]. In polymer gel dosimeters, oxygen diffusion through the wall of the cavity may contaminate the gel and result in an underestimation of absorbed dose [38]. De Deene et al performed a quantitative analysis of the inhibition of gel response by oxygen passing through the catheter wall of a HDR $^{192}$Ir source [17]. Other sources of dose deviations related to the high dose gradients encountered with HDR point sources and measured with MRI have also been established [39].

3. Measurements Around Heterogeneities

Concern about the influence of heterogeneities has figured prominently in dosimetry applications for many decades [6]. However, the measurement of radiation dose distributions within and around heterogeneities is notoriously difficult. Most detectors themselves are non-tissue equivalent, and their introduction into non-unit density media complicates the measurement. For example, the effects of the introduction of the air cavity of an ionization chamber into a water phantom can be handled with carefully-determined replacement factors ($P_{rep}$). Correction for the introduction of the same chamber into a non-water equivalent medium is not as easily managed, and the changes in scattering and absorption become difficult.

As a result, efforts to measure the effects of heterogeneities have largely been limited to determinations of the transmission of radiation through the heterogeneous region, and the effects on the dose at a point or in a plane behind the heterogeneity [7].

One of the attractive features of gel dosimeters is that they are very nearly water-equivalent, particularly at megavoltage photon beam energies. However, several investigators have attempted to use gel dosimetry 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 [40-42]. More recently, measurements have been made behind or adjacent to cavities filled with air or with lung-equivalent plastic. Watanabe assembled a geometric phantom with a rectangular cavity into which air or other
material could be placed [43]. A conventional gel dosimeter filled the surrounding spaces. In contrast, Vergote et al constructed an anthropomorphic thorax phantom into which they placed air cavities to simulate the lungs [40]. In both cases, the dosimetry system was a conventional polymer gel of unit density.

Attempts to construct 3D dosimeters consisting of non-unit density media were described as early as 1998, including reports by several investigators who constructed low-density versions of Fricke gels [44-46]. For these applications, Fricke gels were beaten vigorously to introduce air, or were mixed with expanded polystyrene beads to reduce density. After irradiation, the gels were examined using MR imaging. Subsequently, and to avoid the disadvantages of Fricke gels, Haraldsson et al constructed a normoxic polymer gel with a reduced density by adding expanded polystyrene spheres to a conventional mixture of polymer gel [47]. The antioxidant THP was added in various concentrations. They found the dose versus 1/T2 response to be essentially linear in the dose range of 2 to 8 Gy for all investigated THP concentrations. Good agreement between measured and Monte Carlo calculated data was obtained, in test tubes as well as in a larger 3D phantom. Similar data were shown by Borges who introduced expanded polystyrene beads into a version of MAGIC gel, in a glass vessel [48].

Construction of a low-density gel by adding expanded polystyrene spheres requires a uniform distribution of the gel and the polystyrene, and the buoyancy of the polystyrene spheres makes this difficult. However, Borges et al have shown that dense packing of truly spherical polystyrene beads results in a good approximation of lung density [48]. While these measurements showed promise, there were several potential sources of error. First, the introduction of air, or air-containing polystyrene eliminated the possibility of evaluating the measured dose distribution by optical scanning, and MR imaging must be used. The presence of air also may lead to partial volume imaging effects that could introduce errors into the measurement.

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