The use of the BANG-3 polymer gel to quantify the three-dimensional dose distribution of IMRT

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Recommended Citation
Bruce, Paul Anthony, "The use of the BANG-3 polymer gel to quantify the three-dimensional dose distribution of IMRT" (2003). LSU Master's Theses. 2336.
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THE USE OF BANG-3™ POLYMER GEL TO QUANTIFY THE THREE-DIMENSIONAL DOSE DISTRIBUTION OF IMRT

A Thesis

Submitted to the Graduate Faculty of the Louisiana State University and Agricultural and Mechanical College in partial fulfillment of the requirements for the degree of Master of Science in Medical Physics and Health Physics in The Department of Physics & Astronomy

by
Paul Bruce
B.S., Louisiana State University, 2000
May 2003
Acknowledgements

A special thanks goes out to Dr. Marek Maryanski and MGS Research, Inc. for the vast amount of advice given to me for assistance in my study. The guidance and encouragement from the MGS group allowed me to better understand the procedures involved with my investigation, which helped to obtain more accurate data in my results. Also, I thank Dr. Clive Baldock from the Queensland Institute of Technology for providing me with numerous articles related to gel dosimetry. With his help, I obtained a better understanding of this dosimetry technique, which gave me insight on the popular topics being investigated with gel dosimetry.

This polymer gel study turned out to be a difficult task that required a vast amount of support and opinions from others. Dr. Kenneth Matthews provided insight on procedures to follow and knowledge of the imaging aspects of the study. Also, with Dr. Matthews’ help, I learned IDL and used this program to analyze my data. Assistance from the Mary Bird Perkins Cancer Center staff is also greatly appreciated. Pat Summers, Eddie Singleton and Chad Dunn assisted me with dosimetry aspects of my study. Eddie helped to plan the irradiation to the gel, while Pat and Chad answered my questions dealing with the ADAC treatment planning software. CT scans and MRI scans were performed by the CT tech Doug Naden and MRI tech Guy St. Amant, respectively. Our Lady of the Lake Regional Medical Center provided access to the MRI scanner used in this project. The MIS director, Ronnie Meadors, provided technical support involved with the transfer of data to the IDL workstation. Finally, support involved with any physics-related questions such as ion chamber usage, film analysis, beam quality and geometries and machine operation were answered by the Physics staff, Dr. Thomas Kirby, Dr. William Bice, Dr. Oscar Hidalgo,
Angela Stam, Daniel Neck and Kara Ferachi. I thank each and every one of you, as well as the entire Perkins’ staff.

Finally, none of the above would have been possible without the loving support of my family and friends. I thank each and every one of you for your support throughout the years as I obtained my bachelor’s and master’s degrees.
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Abstract

The sophistication of radiation therapy delivery techniques at Mary Bird Perkins Cancer Center (MBPCC) creates the need for an advanced dosimetric system that can quantify and verify the dose distributions in three-dimensions. Current dosimetric systems perform this dose analysis in only one or two dimensions. This paper evaluates the application of BANG-3™ polymer gel dosimetry to quantify the 3-D dose distribution of Intensity-Modulated Radiation Therapy (IMRT) using a “step and shoot” approach. The gel was irradiated by 10 MV photons at a dose rate of 400 MU/min. Relaxation rate maps were computed from proton density and T2-weighted magnetic resonance images acquired with a GE Horizon 1.5T scanner; scans were performed 5 days and 2 months post-irradiation. The dose distribution within the gel was compared to the dose distribution calculated by the Pinnacle³ planning system. Three techniques were used for analysis: image subtraction, dose-volume analysis and contour analysis. Also, a dose correction factor was used to attempt to correct for excess dose delivered to the gel as the gels were erroneously placed in the treatment room two days prior to irradiation. Corrected 5-day post-irradiation dose maps show reasonable agreement with the Pinnacle³ plan. The absolute measurement error was +/-50 cGy; however, the relative errors were large compared to the total dose of 2 Gy delivered to the gel. Delivering a larger total dose should reduce the relative error to a reasonable magnitude. Exposure to light and other environmental factors caused substantial additional polymerization with time. The results of this project indicate that polymer gel dosimetry could be a useful routine 3D dosimetric technique at MBPCC. However, utilizing a commercial scanning service may simplify use of the gels.
Chapter 1
Gel Dosimetry

1.1 Radiation-sensitive Gels

Advancements in radiation therapy delivery techniques have presented cancer facilities with an increased need to obtain dosimeters that can measure three-dimensional dose distributions with good precision, accuracy and spatial resolution. Current dosimeters (e.g., ionization chambers, thermoluminescent dosimeters, silicon diodes and film) are effective for dose measurements at single points, in two-dimensional planes and sometimes in three-dimensions, but none offers the capability of routine three-dimensional (3D) dose visualization. Each dosimetry technique has limitations that restrict its usefulness in evaluating the 3D-oriented radiotherapy modalities, such as brachytherapy, intensity-modulated radiotherapy (IMRT), stereotactic radiosurgery (SRS) and charged-particle beams. An example of these limitations is the finite size of ionization chambers, creating a difficulty in resolving areas of high dose gradient. A second example is the difficulty of performing a complete 3D dose measurement with film (McJury 2000). Moreover, conventional dosimeters are not capable both of continuously measuring 3D absorbed dose distributions in phantoms of arbitrary geometries and of integrating the dose (Novotny 2001).

Radiation-sensitive gels provide an imaging technique for measuring the 3D dose distributions inherent in the new treatment modalities (McJury 2000). Absorbed dose causes a chemical change within the gel that can be visualized (measured). In gel dosimetry, the phantom itself forms the detector, which is a characteristic not shared with any other dosimetric system. The measuring system, typically a magnetic resonance scanner, visualizes the complete 3D dose distribution recorded in the gel (Gum 2002). Along with the 3D
measurement capabilities, the tissue-equivalence of the gel makes this method a promising dosimetric tool (Novotny 2001).

1.1.1 Polymer Gels and Fricke Gels

In 1984, Gore, et al., first suggested the potential of gel dosimetry in combination with magnetic resonance imaging (MRI) as a system that can measure dose distributions in 3D. The spin-lattice relaxation rate, \( R_1 = 1/T_1 \), was shown to have a linear correlation with the amount of ferric (\( Fe^{3+} \)) ions produced by irradiation of a ferrous (\( Fe^{2+} \)) solution (Haraldsson 2000). As the ferrous ions are oxidized to ferric ions, a change in the net paramagnetic moment results, altering the spin-lattice relaxation of the water molecules near the coordination shells of the ions (De Deene 2000). Ferric ions produce paramagnetic enhancement of the water-proton relaxation rates, which can then be measured with MRI (Maryanski 1994). The above process is known as Fricke gel dosimetry.

Maryanski, et al., proposed a new type of gel dosimetry system in 1993. In this system, acrylic monomers within a gel matrix polymerize upon irradiation with an observed relation between the degree of polymerization and the absorbed dose received by the gel (McJury 2000). Upon irradiation of the gel, dissociation of water molecules occurs, whereby free \( HO^- \) and \( H^+ \) radicals are formed. These free radicals attack the double bonds of the co-monomers whereby the co-monomer radicals then attack other co-monomers and form a polymer chain (De Deene 2000). The spin-spin relaxation rate, \( R_2 = 1/T_2 \), of neighboring water protons increases in proportion to the absorbed dose (Haraldsson 2000). Thus, MRI can be used to visualize the absorbed dose distribution. This process is known as polymer gel dosimetry.
The MRI-polymerization method eliminates some problems encountered with the Fricke-infused gels, mainly the diffusion of ferric ions, which occurs within minutes after irradiation (Novotny 2001). The dose distribution is altered over time due to ion diffusion; thus, there are limits to the use of the Fricke-infused gel (Haraldsson 2000). The diffusion coefficient of $Fe^{3+}$ and $Fe^{2+}$ in 1% agarose gel at a pH of 1.1 is $2.7 \pm 0.3 \times 10^{-6}$ cm$^2$s$^{-1}$ and $3.3 \pm 0.5 \times 10^{-6}$ cm$^2$s$^{-1}$, respectively (Balcom 1995). Pederson, et al., determined that the addition of xylenol orange to a gelatin gel reduces the ferric ion diffusion coefficient from $4.03 \times 10^{-6}$ to $2.25 \times 10^{-6}$ cm$^2$s$^{-1}$. Polymer gels are stable over a period of days to weeks and can be imaged at any time during this period, resulting in greater accuracy primarily in the regions of high dose gradient. In addition, polymer gels show insignificant radiofrequency (RF) attenuation due to their lack of ionic species. Fricke gels, on the other hand, strongly attenuate RF fields, producing significant variations of the RF pulse flip angle and a decreasing signal-to-noise ratio (SNR). Both the variations of the RF pulse flip angle and the decreasing SNR impair the accuracy of dose measurement based on MRI relaxation data (Maryanski 1994). A greater sensitivity is obtained with the polymer gel as compared to the Fricke gel. Drawbacks to the polymer system include a complicated mixing procedure and the toxicity of the chemicals (Haraldsson 2000). Also, polymer gels cannot be used with oxygen-containing materials (e.g. lung-tissue substitutes) as oxygen inhibits polymerization. Unlike the Fricke gel, the polymer gel is not suitable as an active inhomogeneous dosimeter (Gum 2002). Based on MRI, gel dosimetry has been shown to provide verification of calculated dose distributions in soft tissue-equivalent homogeneous phantoms; however, comparable results for a lung-tissue equivalent phantom have only been achieved with a porous Fricke gel dosimeter (Oldberg 2000).
BANG\textsuperscript{TM} gel (MGS Research, Inc.), a type of polymer gel, is comprised of $N,N'$-methylene-bis-acrylamide (BIS), acrylamide, nitrogen and gelatin. Due to its broad stability and sensitivity, this gel offers a potential for the verification of radiotherapy dose distributions. Radiation-induced polymerization is utilized by BANG\textsuperscript{TM}-gel dosimetry (Oldham 1998). These gels are useful for high-energy photon dosimetry as the physical and electron densities are within 3\% and 2\% of water, respectively, and within 1\% and 3\% of muscle (Low 1999). Like x-ray film, BANG\textsuperscript{TM} gels contain a batch-to-batch variability of the dose-response characteristics. This variability is due to the presence of gelatin, a large naturally occurring polymer. A non-reproducible complex final structure, causing this batch-to-batch variability, has been reported (McJury 2000). For this reason, it is recommended that all gels in a given dosimetric session come from the same batch.

1.1.2 Polymerization and Dose Map Calculation

The polymer gels usually consist of an acrylamide monomer along with a cross-linking agent such as BIS uniformly dispersed in an aqueous gel (Maryanski 1994, McJury 2000, Novotny 2001). Approximately 90\% of the gel is composed of water (Maryanski 1994). Irradiation-induced polymerization occurs via free radical production during water radiolysis. The spatial dose-response characteristics of the gel originate from the polymerization (i.e., cross-linking of regions within the gel). The amount of polymerization depends on the quantity of free radicals generated by the absorbed radiation. The polymer structure and concentration determine the relaxation characteristics of the gel’s water protons. NMR relaxation rates of neighboring water protons are increased by the formation of cross-linked polymers in the irradiated regions of the gel (Novotny 2001). The transverse relaxation rates, $R_2=1/T_2$, can be measured, allowing the dose maps to be calculated. In the range of 1-10
Gy, R2 of the water protons in the polymerized regions, measured with MRI, is linearly related to absorbed dose (McJury 2000). Proton density and T2-weighted MR images of an irradiated gel can be converted to dose yielding a 3D dose map (Oldham 1998).

Unfortunately, some uncertainties in dose determination do occur. These uncertainties are influenced by a number of factors: accuracy of the calibration curve, aging dynamics of the polymer gel, B1-field inhomogeneity, and eddy currents (Baldock 2001). Eddy currents occur in the cryogenic and metal casings of the MR system when magnetic field gradients are switched; they result in spatial and time-dependent changes in the magnetic field of the magnet (De Deene 2000). Oldham, et al., proposed a method for calculating the uncertainty of gel dosimeters and it was concluded that the lowest uncertainty obtainable with a standard polymerization gel is about 3 % at 8 Gy and 7% at 2 Gy (Baldock 1999).

1.1.3 Gel Preparation and Specifications

The change in R2 with dose (i.e., sensitivity) and polymer gel saturation limit are critically dependent upon gel preparation conditions. Such conditions include oxygen and light exposure, weight fractions of gelatin, monomer and cross-linker and temperature during imaging. The effects of the monomer/cross-linker density are discussed in Chapter 2.2.3.

Oxygen, a free radical scavenger, inhibits gel polymerization. De-ionized distilled water is degassed with an inert gas (e.g., pure N2) to remove any dissolved oxygen. A gelling agent is added to this de-ionized water. Fogging of the gel can result from the bubbling of nitrogen gas through the gel mixture, which will lead to an increased background R2 value. Oxygen exposure is minimized by manufacturing the gel in either a sealed reaction flask or in a sealed glove box with a nitrogen environment. Oxygen-impermeable vessels are usually used to store the gel after it has been fabricated; glass or BAREX (BP Chemicals Inc.,
London, UK) are common container materials (McJury 2000). Glass walls are usually thicker than the dose build-up region for external beams. BAREX is the preferred phantom vessel when studying dose build-up regions and dose distributions near the entrance surface of the phantom. Furthermore, BAREX may be molded into any shape (Haraldsson 2000).

Light, like oxygen, can reduce the overall sensitivity of the gel as it can initiate photopolymerization. Shielding must be provided once the monomer, cross-linker, and gelatin solution have been mixed (McJury 2000). The polymer gel should be stored in a dark area and its exposure to natural and artificial light should be minimized during use.

Irradiation of the gel dosimeter should take place soon after manufacturing due to the potential for oxygen contamination (McJury 2000). Post-manufacture gels show a continuous change in $R_2$, resulting from both free radical impurities in the components of unirradiated gels and the increasingly attenuated dynamics of the free radical reactions in irradiated gels (McJury 1999). The optimum time to image the irradiated gel is upon reaching a steady state with all polymerization reactions completed and maximum sensitivity having been reached. Imaging prior to this point would result in a loss of sensitivity (McJury 1999). In the case of the BANG-3™ polymer gel, polymerization reactions require 30 minutes to complete; in addition, up to two days may be required to achieve equilibrium in the scanning room. To minimize the effects of spontaneous polymerization, the gel manufacturer recommends gel scanning within one week post-irradiation.

The sensitivity of the gel dosimeter is inversely proportional to the temperature; a reduction in temperature during imaging will lead to larger gel sensitivity. Proton correlation times lengthen and proton exchange rates increase as the polymer chain motion becomes slower with a decrease in temperature. In a similar fashion, the gelatin relaxation rates also
increase with decreasing temperature. With this in mind, prior to MRI, a gel must be allowed to equilibrate to a uniform temperature. The temperature of the experimental gel and the calibration gels must also be the same.

A greater visibility of the polymerized region is observed the gelatin as opposed to agarose, both of which are gelling agents. Gelatin produces clearer gels than agarose. Furthermore, the transverse relaxation rates of gelatin systems are approximately one order of magnitude lower than those of the agarose gel. Sensitivity is therefore higher in the gelatin system compared to the agarose system as the baseline $R_2$ is decreased. Finally, a gel consisting of 5% gelatin by weight is best because an increase in gelatin concentration causes a decrease in both $R_2$ and overall gel sensitivity (McJury 2000).

1.2 MRI Pulse Sequence Techniques

The spin-echo pulse sequence is a common acquisition method for MRI images. Spin-echo methods include the single echo (Hahn spin echo) sequence and the multiple spin-echo among other techniques (Baustert 2000). With the single echo technique, only a single echo is collected by each acquisition, but the acquisition is repeated with different echo times. The multiple spin-echo technique acquires a train of spin echoes, with different trains acquired with different echo times (McJury 2000). To achieve optimal results in determining $T_2$, the multiple-spin-echo pulse sequence parameters can be varied. These parameters include echo spacing (ES), number of echoes, and repetition time (TR) (Baldock 2001). Baustert, et al., provided insight into an efficient method of measuring $T_2$. Unfortunately, multiple-spin-echo sequences available on clinical MRI scanners are generally not optimized for determining the range of $T_2$ values encountered in polymer gel dosimetry (Baldock 2001). Non-optimal multiple-spin-echo acquisitions can result in errors in the $R_2$ values owing to
imperfect refocusing of the 180° pulses, which can lead to a standing wave effect or RF attenuation in large aqueous samples (McJury 2000). Thus, the single echo technique is preferred for polymer gel analysis.

1.3 **Intensity-Modulated Radiation Therapy (IMRT) and Polymer Gel Dosimetry**

With the advent of IMRT techniques, the validation of dose distributions produced by planning systems is required from the radiation physicist. Intricately shaped conformal dose distributions, particularly with concavities and steep dose gradients between the boundaries of the planning target volume (PTV) and the organ-at-risk (OAR) in all three dimensions, are characteristic of IMRT (Oldham 1998). Through IMRT, a conformal irradiation is achieved by delivering beams composed of different beam segments to the patient (De Deene 2000).

With complex (e.g., concave-shaped) PTVs, IMRT is the major hope for achieving the maximum protection of the OAR with the best possible tumor control probability (TCP) without an increase of normal tissue complication probability (NTCP) (Gum 2002).

Treatment planning for IMRT must have high spatial accuracy to ensure adequate coverage of the PTV and sparing of the OARs in the presence of the highly conformal distributions. According to Oldham, *et al.*, there is much research effort aimed at measuring IMRT distributions in 3D and comparing these to the planned distributions.

For successful clinical implementation of IMRT, a dosimetric verification process must be established to ensure that the delivered doses are consistent with the calculated doses for each patient. There are generally three types of measurements that can be used to provide dose verification (Zhu 2002):

1) an absolute measurement using calibrated detectors such as ion chambers

2) relative 2-D dose distribution measurements using radiographic or radiochromic film; relative 3-D measurements with polymer gels
3) the use of charge-coupled device (CCD) imaging, electronic portal imaging devices (EPID) or radiographic films to verify the projected fluence pattern perpendicular to the incident beam

Clinical IMRT distributions are commonly verified in a two-step process. Positional accuracy of the high-dose regions and gradients is confirmed by film measurement followed by a quantitative diode or ion chamber measurement to verify the absolute dose at the reference point in the distribution. Gel dosimetry provides potentially both results with a single measurement, making it attractive for the verification of IMRT (Oldham 1998).

1.4 Project Objective

The objective of this study is to evaluate the application of BANG-3™ gel dosimetry to quantify the 3D dose distribution of IMRT with a “step and shoot” approach delivered by the Varian Clinac 21EX. This dose distribution will then be compared to the predicted static tomotherapy dose distribution calculated by the Pinnacle³ planning system. In this plan, the target volume will be the prostate, while the organs at risk will include the bladder, rectum and femoral heads. This comparison will provide the staff at Mary Bird Perkins Cancer Center (MBPCC) some insight into the effectiveness of using the BANG-3™ gel as a verification tool for IMRT treatments. Also, this study will provide MBPCC’s staff a recipe for the future use of this dosimetric system and some understanding into the feasibility and limitations of its use.

Calibration of the dosimeter is done at the time of use for each gel batch. To calibrate, a range of known doses is applied to a portion of the gel batch. The calibration is done in a water-filled calibration phantom; with the use of MRI, a plot of $R_2$ (s$^{-1}$) versus known dose (Gy) can be obtained (McJury 2000).
The MRI scanner produces the 3D gel image in the form of $T_2$–weighted and proton-density-weighted images that are combined to compute the relaxation rate ($R_2$). A conversion of the image to absorbed dose must occur to allow verification of radiotherapy dose distributions (Oldham 1998). Dose distributions are imaged in multiple planes with the gel dosimetry system. Dose maps obtained from the polymer gel are compared to the Pinnacle$^3$ dose plan pixel by pixel using a commercial data-processing package. Comparisons are done by image subtraction, by direct comparison of isodose curves and by dose volume analysis. The desired accuracy is about 5% relative to the maximum dose deliverable to the gel (approximately 10 Gy).

1.5 Conclusion

Polymer gel dosimetry for measuring 3D dose distributions is promising but may not yet be clinically practical. Numerous questions related to the methods and cost remain. Gel phantoms for IMRT dose mapping currently cost $350 each and $250 each for SRS dose mapping. Additional materials (e.g., calibration vessels, MRI system access and fiducial markers for reproducible positioning of the phantom) are necessary to conduct research and plan verifications. The gels can be self-manufactured, but along with material costs, safety is a concern because the components of polymer gels are extremely toxic.

Fortunately, research is underway to better understand this dosimetry system and interest in the development and application of the polymer gel dosimetry technique is growing. In 1999, the first international workshop dedicated to gel dosimetry (DOSGEL’99) took place, followed two years later by the second international workshop (DOSGEL’01). Due to this increased research and recent interest, the polymer gels may be realistically available for widespread clinical use within a couple of years. Ideally, greater demand for the
gel should lower the cost. With the advent of IMRT and SRS, the increased use of high-dose-rate sources at MBPCC and the potential benefits of polymer gel dosimetry, an evaluation of the effectiveness of the BANG-3™ gel system for IMRT verification should be beneficial to MBPCC.
Chapter 2

Literature Review

2.1 Historical Overview

2.1.1 Introduction

Radiation-sensitive gels were first developed in the 1950s when Day and Stein investigated a color change upon irradiation of a gel containing Folin’s phenol. In 1957, Andrews, *et al.*, made measurements of photon and electron depth doses using agar gels. Later studies were done which utilized Fricke solutions and gels, and in 1958, Hoecker and Watkins studied an alternative method. This method was based on radiation-induced polymerization in monomer and polymer solutions (McJury 2000).

Several studies investigated the use of Fricke gel dosimeters. Gum, *et al.*, determined that an inhomogeneous anthropomorphic Fricke gel phantom and 3D magnetic resonance imaging are valuable tools for verification of IMRT treatment plans. The phantom was that of the human thorax including the lungs and spine, where the thorax and spinal cord consisted of undiluted Fricke gel and the lungs were filled with low-density Fricke gel. An average deviation of less than 5% between measurement and planning was found in regions containing the standard Fricke gel, whereas a higher deviation was found in regions consisting of the low-density gel. The greatest error was due to decreased SNR in the magnetic resonance measurement for the low-density gel. Oldberg, *et al.*, demonstrated that Fe$^{2+}$-infused low-density gel was suitable for measurements of absorbed dose distributions in phantoms containing lung tissue compartments. In this study, a low-density dosimeter gel and a conventional ferrous sulphate gel were filled in separate compartments of a Perspex container.
Chu, et al., conducted a study on two new Fricke dosimeter gel systems with low Fe\(^{3+}\) diffusion. These two systems, polyvinyl alcohol-Fricke hydrogel and cryogel, have been developed for 3D radiation dosimetry purposes. Because of its optical transparence, the hydrogel can be imaged by either optical or MRI detection methods. On the other hand, the cryogel is opaque, so the internal Fe\(^{3+}\) concentration can only be measured by MRI.

### 2.1.2 Polymer Gel Dosimetry

The BANANA gel (composed of BIS, acrylamide, nitrous oxide, and agarose) was the first polymer gel. In this type of gel, acrylamide and BIS were added to a simple agarose gel. Future work involved the use of polyacrylamide gels (PAG) and BANG\(^{TM}\) gels. BANG\(^{TM}\) gels were produced to provide the gel solution with a greater sensitivity. Nitrous oxide and nitrogen are added to the gel mixture to displace oxygen from the gel during manufacture (McJury 2000). BANG-1\(^{TM}\) and BANG-2\(^{TM}\) gels are current models of BANG\(^{TM}\) gels. The former is made using acrylamide in powder form, while the latter replaces acrylamide with acrylic acid and NaOH to buffer the pH. Gel response is improved with acrylic acid compared to acrylamide, allowing larger relaxation-rate changes per unit dose. Acrylamide is a neurotoxin that can lead to nervous system disorders. Safe handling of acrylamide is essential and appropriate care should be taken when disposing of the gel. (McJury 2000).

BANG-3\(^{TM}\) gels have recently been developed. This type of gel has strong optical and MR responses (Oldham 2001). With the BANG-3\(^{TM}\) gel, acrylic acid is replaced with methacrylic acid. Table 2.1 summarizes the gel types and compositions. Of the BANG\(^{TM}\) polymer gels, BANG-3\(^{TM}\) polymer gel reportedly has the highest MR sensitivity upon photon irradiation (Ramm 2000).
Table 2.1. Polymer Gel Types and Compositions

<table>
<thead>
<tr>
<th>Type</th>
<th>Composition</th>
</tr>
</thead>
<tbody>
<tr>
<td>BANANA</td>
<td>BIS, acrylamide, nitrous oxide and agarose</td>
</tr>
<tr>
<td>BANG-1\textsuperscript{TM}</td>
<td>BIS, acrylamide, nitrogen and gelatin</td>
</tr>
<tr>
<td>BANG-2\textsuperscript{TM}</td>
<td>BIS, acrylic acid, sodium hydroxide, nitrogen and gelatin</td>
</tr>
<tr>
<td>BANG-3\textsuperscript{TM}</td>
<td>BIS, methacrylic acid, sodium hydroxide, nitrogen and gelatin</td>
</tr>
</tbody>
</table>

2.2 Gel Response Studies

2.2.1 Introduction

To use the polymer gel dosimeter in a clinical setting, the response of the dosimeter to different physical conditions must be known. Such physical conditions include: polymer-gel composition, temperature during irradiation and NMR evaluation, type and energy of radiation, dose rate dependence, time from irradiation to NMR evaluation (Chapter 1.1.3) and magnetic field strength. These different conditions can potentially affect polymer gel dosimeter response, thereby significantly influencing the measured results (Novotny 2001).

2.2.2 Temperature Effects

Temperature effects have been evaluated on polymer gels. The dose response of BANG\textsuperscript{TM} reportedly shows little dependence on the temperature during irradiation, but does depend upon the temperature during imaging (Maryanski 1994). A temperature rise of 1-3 °C is not uncommon during imaging due to RF power absorption by the gel dosimeter. A temperature rise of 3 °C results in a dose underestimation of 10% over the whole dose map (De Deene 2001). Salomons, et al., reported that the temperature during irradiation is a very important parameter requiring control and that temperature change effects on gel performance are important issues that require further study. They argue that polymerization reaction rates within the dosimeter are affected by temperature changes and that the size and shape of the
gel dosimeters along with their environment during and after irradiation are factors that should be controlled to ensure reproducibility.

### 2.2.3 Weight Fraction Variation

A study involving the variation of both the absolute and relative weight fractions of monomer and cross-linking agents was performed by McJury, et al., in 2000. They reported that expanding the total monomer content of the gel from 3% to 6% increases the sensitivity and the dose saturation point of the gel dosimeter. A limit to the concentration of co-monomers is set by the low solubility of BIS, as well as by crystallization during storage of high co-monomer content gels. Also, McJury reported that peak sensitivity is obtained for equal amounts of cross-linker and monomer content, while the dose saturation point continues to rise for increasing cross-linker fraction. With an increase in the cross-linker fraction, the monomer-to-polymer conversion per unit dose decreases; there is a lower reactivity of the BIS cross-linker to the acrylamide. However, there is increased rigidity of the polymerized gel as the cross-linker density increases. The greater rigidity allows efficient spin diffusion, decreasing the motional averaging due to dipolar interactions. This, in turn, causes the $R_2$ of the gel to rise (McJury 2000).

### 2.2.4 Energy and Dose Rate Dependence

To date, only two studies have been reported that evaluate energy and dose rate dependence for the BANG-2™ polymer gel dosimeter (Maryanski, et al. and Novotny, et al.) Maryanski, et al., compared diode measurements in water to central axis percentage depth dose measured in a gel dosimeter for 6 MV X-rays and 15 MeV electron beams. They concluded that there is no energy or dose rate dependence of the gel dosimeter response in the range of 2-15 MeV and dose rates in the range of 20-400 cGy·min\(^{-1}\).
Novotny, et al., studied dosimeter sensitivity dependence on photon energy using 4, 16, and 18 MV X-rays from the Varian Clinac 600C and 2100C. For electron beams, 9, 12, 16, and 20 MeV beams were produced by a Varian Clinac 2100C, while the same machine was used to measure the dose rate dependence with repetition rates of 80, 160, 240, 320, and 400 MU min⁻¹. This study concluded that polymer-gel dosimeter sensitivity decreases with increasing photon or electron energy; however, no clear trend was observed for sensitivity dependence on dose rate. Several studies measuring the energy and dose rate dependence of the BANG-1™ polymer gels have been reported (Novotny 2001), but no studies of these effects on the BANG-3™ polymer gel have been published. Polymerization is independent of dose rate in the BANG-1™ gel (Ramm 2000), and the response is independent of energy and type of radiation (Novotny 2001).

2.2.5 Other Response Effects: Magnetic Field Strength, pH Studies and Gel Aging

In a study performed by Haraldsson, et al., both the slope of the dose-response curve and intercept were found to decrease with increasing magnetic field strength. During a study in 1993, an opposite relationship was found by Maryanski, et al. Differences between the results can be attributed to different gel types and different temperatures during evaluation (Haraldsson 2000). Other investigations of the polymer gel dosimeters include pH studies (Gochberg 1998) and studies on the effects of gel aging on image quality (McJury 1999). McJury, et al., report a continuous change in R₂ for post-manufacture gel (irradiated and unirradiated). For the unirradiated gel, this change in relaxivity is due to the initiation of polymerization by the presence of free radical impurities, whereas for the irradiated gel, the relaxivity change can be attributed to the increasingly attenuated dynamics of the free radical reactions as the viscosity of the gel increases with increasing polymer density.
2.3 Imaging of Polymer Gels

2.3.1 Magnetic Resonance Imaging

Two methods for imaging the spatially localized polymerization are MRI, as discussed in Chapter 1.1.2, and optical scanning methods (McJury 2000). In an MRI scanner, relaxation time constants (T₁ and T₂) are measured by applying a radiofrequency (RF) pulse to excite the magnetization of the spin system followed by sampling during the return to equilibrium (McJury 2000). With the Fricke gel dosimeter, maps of the T₁ relaxation parameter can be obtained (Oldham 2001). Using the polymer gel dosimeter, T₂ can be measured by fitting the collected data from at least two points on the transverse relaxation curve (McJury 2000).

2.3.2 Optical Imaging

Much research effort is being conducted into developing alternative methods to image the dose distribution recorded in gel dosimeters (Oldham 2001). New techniques are being investigated to find imaging techniques with less inherent noise than MRI and to eliminate reliance on MR technology with its associated issues of limited access and high scanning cost (Oldham 2001). Polymer gels can be easily visualized upon irradiation due to their opalescent appearance as the radiation dose increases. Qualitative testing can thus be done by visual inspection or through the use of an optical densitometer (Maryanski 1994). Oldham, et al., performed a study on the effectiveness of gel dosimetry and optical-CT (computerized tomography) scanning as a verification method for complex radiosurgery deliveries and by extension IMRT deliveries. As the laser was stepped in increments of 1 mm across the flask, horizontal line scans were taken. A total of 100 projections were acquired which corresponded to 180 degrees of projection data. A RTAP (Resolution-Time-Accuracy-Precision) criteria of ≤ 1mm³ resolution, less than 1 hour imaging time, within 3 % accuracy
and less than 1 % precision was applied. The paper demonstrated that the optical-CT scanning method for BANG-3™ gels yielded maps of sufficient accuracy, resolution, and precision and approached a major goal of dosimetry.

McJury, et al., performed an investigation that utilized optical scanning as an imaging technique. Due to the increase in opacity of the gel upon irradiation, they determined that it is possible to use the optical scanning technique to generate a two-dimensional dose distribution. Light attenuation was found to be related to polymer density and thus absorbed dose, with the understanding that light within a BANG-1™ gel is scattered rather than absorbed by the polymer particles. McJury, et al., noted that the attenuation coefficient for 500 nm wavelength light increased by about 0.7 mm\(^{-1}\) with a dose increase of 0 Gy to 5 Gy using a gel sample irradiated with a series of uniform dose regions. Moreover, the attenuation was found to be directly related to absorbed dose over the dose range from 0-10 Gy. It was concluded that from the preliminary optical scanning studies the optical scanning technique could either replace or at least supplement the NMR imaging method of dose measurement (McJury 2000).

### 2.3.3 X-ray Computed Tomography

Several studies have been conducted using X-ray computed tomography (CT) as a means to image the polymer gel (Audet 2002, Hilts 2000, Trapp 2001). In Audet’s study, two polyacrylamide gels (PAGs) of different compositions were irradiated with a four-arc stereotactic treatment to maximum doses of 15 Gy and 8 Gy, respectively. Image averaging and background subtraction of the CT images were used to improve the image quality. Conversion of the PAG images to relative dose maps used a CT number-dose calibration curve to allow for comparison to the planned isodose distribution. Registration of the PAG
images to the planning CT images was reported to show good agreement. Investigation of more complex dose distributions and improvement of dose resolution are future studies of this group (Audet 2002).

Hilts, et al., also studied the use of X-ray CT as a method to analyze polymer gels. The experiment involved used two 1.5 L cylindrical polymer gel phantoms composed of 3% acrylamide, 3% BIS and 5% gelatin by weight. One phantom was irradiated with four intersecting 10 MV photon beams whereas the second phantom was irradiated with 10 sets of parallel-opposed radiosurgery fields. In addition to developing an imaging protocol to optimize the CT images of polymer gels, the group investigated the relationship of polymer gel CT number to dose (N_{CT}-dose response), determined the reproducibility of the response and compared CT to MRI to analyze the polymer gels. An optimized imaging protocol is necessary due to the fact that CT imaging is sensitive to radiation induced density changes that occur within the polymer gel. The group concluded that the N_{CT}-dose response is not only reproducible and linear up to 8-10 Gy, but is also relatively insensitive to the gel temperature during imaging. They note a downfall of this imaging method is the poor dose resolution of approximately 50 cGy for a slice thickness of 10 mm; MRI has a dose resolution of about 20 cGy for a slice thickness of 5 mm, but is very sensitive to gel temperature during imaging. The group concluded that CT imaging technique has three major disadvantages compared to MRI: lower dose sensitivity, greater dose uncertainty and the post irradiation wait prior to imaging (waited one week before CT imaging to prevent the gel from responding to dose from the CT scanner). Despite these downfalls, the CT polymer dosimetry technique appeared to be accurate and capable of localizing high dose gradients (Hilts 2000).
Trapp, et al., investigated the effects of the composition of the polymer gel dosimeters on the CT-dose sensitivity. This group studied several gel compositions, comparing acrylamide to BIS comonomers and comparing agarose to gelatin gelling agents. An increase in comonomer concentration was found to increase the CT-dose sensitivity. Also, the use of agarose instead of gelatin increased the sensitivity. Dose resolution was found to be optimal for a polymer gel dosimeter composition of 5% gelatin, 3% acrylamide, 3% BIS and 89 % water was reported (Trapp 2001).

### 2.3.4 Fourier-Transform Raman Spectroscopy

Fourier transform (FT)-Raman spectroscopy studies have been performed on polyacrylamide gels (Baldock 1998, Jirasek 2001). Baldock, et al., investigated cross-linking changes during the copolymerization of PAGs in the spectral range of 200-3500 cm⁻¹. The vibrational bands of the single binding modes of acrylamide and bis-acrylamide were found to decrease in amplitude with increasing absorbed radiation dose. They concluded that FT-Raman spectroscopy has potential for ionizing radiation dosimetry using PAGs (Baldock 1998).

The study by Jirasek, et al., utilized FT-Raman spectroscopy to characterize the rates of monomer and crosslinker consumption and polymer formation. The consumption rates of monomer and crosslinker were characterized in the spectra as a function of absorbed dose and were found to be monoexponential up to 13 Gy. The rates of consumption are different for the two molecules, with a higher rate of consumption found for BIS compared to acrylamide. Polymer formation, up to a dose of 13 Gy, was monoexponential and correlated well with the consumption rate of acrylamide. Like the previous study, these results indicate
that Raman spectroscopy provides a direct and useful tool for characterizing an irradiated PAG.

2.3.5 Ultrasound Imaging

One study involving the use of ultrasound to evaluate a polymer gel has been reported. Mather, et al., investigated this method for evaluating radiotherapy 3D polymer gel dosimeters. The method involves the use of ultrasound to evaluate the important structural changes that occur following dosimeter irradiation. Acoustic speed of propagation, attenuation and transmitted intensity were measured as a function of absorbed radiation dose. Each parameter showed a strong variation with absorbed dose continuing beyond absorbed doses of 15 Gy. Also, a larger dynamic range in the dose response curves was found with the ultrasonic measurements as compared with MRI dose response data previously published. The group concluded that the ultrasound technique shows great potential for the evaluation of polymer gel dosimeters (Mather 2002).

2.4 Radiotherapy Applications for Polymer Gels

2.4.1 Intensity-Modulated Radiation Therapy (IMRT)

Research effort is aimed at measuring IMRT distributions in 3D and comparing the measured to the planned distributions (Oldham 1998). In a study performed by Oldham, et al., it was concluded that gel-measured distributions compared well with plans from a Peacock planning system using an in-house dose algorithm. Two irradiation experiments were conducted with a homemade BANG-gel dosimeter. The two distributions were that a parallel-opposed irradiation and a nine-field “static tomotherapy” intensity-modulated irradiation delivered with a Nomos MIMiC. They concluded that the gel dosimeter agreed well with the predicted dose distributions and the steep penumbral fall-off of the dose was reproduced very accurately. At medium to high doses (50-90% isodose lines), the predicted
distribution agreed well with the gel-measured distribution. Differences of up to 10% were found at lower doses (30% isodose line). For doses greater than 50%, isodoses were in general agreement to 1 mm with a maximum discrepancy of 3 mm.

### 2.4.2 Stereotactic Radiosurgery and Radiotherapy

Investigations of polymer gel dosimetry for stereotactic radiosurgery include comparison of a planned and measured 3D stereotactic dose volume (Audet 2002), image distortion in MRI-based polymer gel dosimetry of gamma knife stereotactic radiosurgery systems (Watanabe 2002) and reproducibility of polyacrylamide gel (PAG) dosimetry applied to stereotactic conformal radiotherapy (Cosgrove 2000). The study performed by Audet’s group was previously discussed (Chapter 2.3.3), whereby the high dose region produced by the stereotactic treatment was accurately localized by the CT gel technique.

Watanabe, et al., investigated the effects of collimator size and MR image distortion on polymer gel dosimetry of gamma knife stereotactic radiosurgery systems. Both CT scans and magnetic resonance images of the BANG™ polymer gel were acquired and directional frequency encoding and receiver bandwidth artifacts in the images were studied. An image shift of the measured dose distributions of two pixels was observed in the frequency encoding direction for both MRI and CT. The results were found to be reproducible and independent of collimator size and the observed shifts were said to be caused by MR image distortion due to magnetic susceptibility effects.

Cosgrove, et al., tested the reproducibility of polyacrylamide gel dosimeters for a three-field coplanar arrangement and a four-field, conformal, non-coplanar plan using precision-cast lead alloy shielding blocks. The three-field plan used linac jaws for field shaping. Reproducible relative dose distributions were encountered with a standard deviation
on the mean areas enclosed by the 50% isodose lines, measured in three orthogonal planes of 6.4% and 4.1% for the coplanar and non-coplanar plans, respectively. The group also found consistency between the measured and planned distributions. Isodose lines agreed generally to within a few millimeters; however, the measured absolute doses were, on average, 23.5% higher than those planned.

2.4.3 Carbon Ion Therapy

Ramm, et al., studied the application of three-dimensional BANG™ gel dosimetry to carbon ion therapy. The BANG™ polymer gel dosimeters were visualized by MRI. Their objective was to examine saturation effects for densely-ionizing (high linear energy transfer (LET)) radiation. The gels were irradiated with monoenergetic $^{12}$C ions at different beam energies. Results demonstrated that BANG™ gel dosimetry with MRI could be used for conformal heavy-ion radiotherapy. Measurements of target volume contours and dose gradients can be obtained with high spatial resolution. Furthermore, the influence of LET on saturation effects and the importance of the microscopic dose pattern for high-LET radiation were demonstrated (Ramm 2000).

2.4.4 Brachytherapy

Farajollahi, et al., investigated the use of polymer gel dosimetry in low dose rate brachytherapy. The response of gel was found to be reproducible and linear up to 10 Gy. Measurements of the absorbed dose distributions for a straight applicator containing 36 $^{137}$Cs sources measured with the gel and TLDs were compared to the calculated values. A good agreement for relative measurements was found. Additionally, measurements for a complex gynecological insert were compared with the isodose curves from a Helax TMS planning
system. Within areas unaffected by oxygen diffusion, an agreement within 0.5 mm was found for the 50-100% isodose levels (Farajollahi 1999).

McJury, et al., investigated the 3D dosimetry of a high-dose-rate clinical $^{192}$Ir source to avoid errors imminent in the dosimetry of a brachytherapy source. Experimental dosimetry measurements can be inaccurate for several reasons: poor spatial resolution for ion chambers, lack of 3D information using film and errors due to distortion of the radiation flux by the dosimeter itself. The group compared experimental measurements of the dose versus radial distance from the center of the source to calculations from a Nucletron planning system (NPS). In the selected planes, a good agreement was found between the planning system and the gel measurement. The group concluded that polymer gel dosimetry shows promise for brachytherapy applications and can offer complete 3-D dose information, good spatial resolution and small measurement errors. However, measurements close to the source were difficult to acquire due to limiting properties of the polymer gel (i.e., saturation of gel) composition at around 10 Gy and oxygen diffusion at surface of catheter (McJury 1999).

2.4.5 Boron Neutron Capture Therapy

Two groups are studying the use of polymer gel dosimetry in boron neutron capture therapy (BNCT), a tumor specific treatment modality, (Farajollahi 2000 and Wojnecki 2001). Farajollahi, et al., exposed polymer gels with and without 60 ppm of $^{10}$B to an epithermal neutron beam. The gels were irradiated in vials in pairs in a water phantom for 5 hours each. The variation of relaxation rates of the gels with depth were measured and compared to MCNP Monte Carlo calculations. Boron enhanced absorbed dose in the gel measurements. The group concluded that polymer gels could measure the enhancement of absorbed dose due to boron for epithermal or thermal neutrons (Farajollahi 2000).
Wojnecki, et al., performed a computational study on the use of polyacrylamide gel and A-150 plastic as substitutes for brain tissue in BNCT. Using a phantom material that closely resembles brain tissue is a necessity to precisely evaluate the dosimetric performance of epithermal neutron beams designed for BNCT of brain tumors. The study compared the performances of polyacrylamide gel and A-150 plastic to standard phantom materials such as water and polymethyl-methacrylate (PMMA) as substitutes for brain tissue. The group concluded that the polyacrylamide gel is promising for the use with BNCT as it provides a good simulation of the radiation dose components in brain tissue. The polyacrylamide gel can be used as both a phantom material and a dosimeter. Also, A-150 was found to provide a better simulation of the radiation transport in the brain tissue than PMMA and thus should be used as a solid phantom material. It was determined that water also is a suitable phantom material for BNCT. However, further research is required into calibration of the MR-derived dose images (Wojnecki 2001).
Chapter 3

Materials and Methods

A summary of the necessary equipment, their functions and the measurements and analysis to be carried out in this project are discussed in this section. The ultimate goal of this project is to provide guidance to MBPCC in using polymer gels to verify 3D IMRT treatment plans from the Pinnacle\(^3\) planning system.

3.1 Equipment Overview

3.1.1 Introduction

The BANG-3\(^{TM}\) polymer gels require a complicated mixing procedure and some of the chemicals within the gel (e.g., acrylamide, BIS) are toxic, thus this project utilizes pre-manufactured gels (MGS Research, Inc.). Other equipment which is used include: the Pinnacle\(^3\) treatment planning system, the Varian Clinac 21EX, an MRI scanner, an ionization chamber, an electrometer, the RIT system, XV film and solid water (Table 3.1).

3.1.2 In-house Equipment

Several pieces of equipment necessary for this project are available at MBPCC. The Pinnacle\(^3\) treatment planning system is the system currently used at MBPCC. Conventional therapy plans have been done with this system for several years, but recently the workload has risen tremendously due to the advent of IMRT therapies. Currently, IMRT plans are generated by the Pinnacle\(^3\) treatment planning system for prostate cancers, head and neck cancers, lung cancers and brain cancers. Eventually, the Pinnacle\(^3\) system will be used to plan IMRT treatments for breast cancers. The cubical water phantom, ionization chamber, electrometer, RIT system, XV film and solid water are all easily accessible at MBPCC. The MRI scanner is available at Our Lady of the Lake Regional Medical Center (OLOLRMC).
Table 3.1 Equipment utilized in study, quantity and supplier of materials and total cost.

<table>
<thead>
<tr>
<th>Equipment</th>
<th>Quantity</th>
<th>Supplied By</th>
<th>Cost ($)</th>
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<td>OLOL Regional Medical Center</td>
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</table>

3.1.3 BANG-3™ Polymer Gel

The BANG-3™ polymer gel used in this experiment has a maximum measurable dose of 10 Gy. Technical specifications are listed in Table 3.2. The maximum measurable dose is set by the amount of reducing agents added to the gel (Maryanski presentation). Gaseous oxygen contamination strongly affects the dose response of BANG™ gels. The gels are therefore made under anoxic conditions and must be hermetically sealed at all times. Ultraviolet light and blue light induce photo-polymerization of monomers in the BANG™ gels. Exposure to daylight and fluorescent light should be minimized (MGS Research Inc.)
Table 3.2 Technical specifications for the BANG3™-polymer gel. Adapted from MGS Research, Inc. website.

<table>
<thead>
<tr>
<th>Technical Specifications</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum Dose:</td>
</tr>
<tr>
<td>Maximum Dose Rate:</td>
</tr>
<tr>
<td>Accuracy:</td>
</tr>
<tr>
<td>Specific gravity:</td>
</tr>
<tr>
<td>Atomic composition (weight fractions):</td>
</tr>
<tr>
<td>Dose response is independent of photon or electron energy. Spatial resolution is limited by MRI voxel size.</td>
</tr>
</tbody>
</table>

website. Irradiation should occur no more than two weeks from shipment. MGS Research, Inc., the manufacturer of BANG-3™ polymer gel, asserts that the dose response of the gel is independent of photon or electron energy. No other data analyzing the relation of the dose response to the photon or electron energy are available. Likewise, no data have been reported on the dependence of dose-response of the BANG-3™ polymer gel on dose rate. Several studies are available on the BANG-1™ and BANG-2™ polymer gels; almost all of them conclude that the dose-response is independent of dose rate.

### 3.1.4 IMRT at MBPCC

At MBPCC, treatment using IMRT is currently done with two Varian Clinac 21EX systems, one in Baton Rouge and the other in Covington. Both accelerators are equipped with multi-leaf collimators (MLCs) to achieve a conformal dose distribution. Photon energies of 4 and 10 MV are available for treatments with the Baton Rouge machine, while the accelerator in Covington provides photon energies of 6 and 18 MV. IMRT treatments are achieved with MLCs because the computer-driven multileaf collimators modulate the incident radiation fluence as a function of gantry angle and off-axis distance (Low 1998). In particular, a “step and shoot” approach is used by MBPCC, whereby the dose distribution is delivered with the MLCs immobile during each irradiation. IMRT was designed in an effort to maximize the
dose to the planning target volume (PTV) while sparing the organ(s)-at-risk. MBPCC uses photon beams to execute the IMRT treatments. Dose rates commonly used for these treatments are $250 \frac{MU}{min}$ for the 4 MV photon beams and $500 \frac{MU}{min}$ for the 10 MV photon beams.

### 3.1.5 Gel Calibration

For calibration, a quantity of the gel is contained in borosilicate glass vials (VIAL) with dimensions of 25 mm OD, 95 mm length, 1.3 mm thick wall, and a 1 mm thick bottom. The vials are positioned parallel to the radiation beam in a cubic water phantom with the water filled to at least 10 cm above the vial surface (see Figure 3.1). Each vial receives a different total dose to obtain a calibration curve, which relates the dependence of the transverse relaxation rate, $1/T_2$, to absorbed dose.

### 3.1.6 Dose Mapping

For purposes of dose mapping of the IMRT treatment field, BANG-3™ polymer gel fills a BAREX plastic cylinder (CGEL-S) with phantom dimensions of approximately 2L volume, 17 cm OD, and 10 cm height. The CGEL-S is irradiated with a simulated IMRT plan obtained from Pinnacle³ using 10 MV photon beams. The treatment plan is detailed in Appendix A. The Varian Clinac 21 EX with a dose rate of $400 \frac{MU}{min}$ is utilized in this study as $4 \frac{Gy}{min}$ is the maximum allowable dose rate with this gel (Table 3.2).

### 3.1.7 Imaging of the Polymer Gel

MRI is used at OLOLRMC as a means of obtaining high quality images of the internal anatomy of the human body. The method is based on principles of nuclear magnetic...
resonance to obtain microscopic chemical and physical molecular information (Hornak website, Webb 1988). A radio-frequency pulse to excite the magnetization of the spin system is applied followed by sampling during return to equilibrium. The sampling allows for the relaxation times ($T_1$ and $T_2$) to be measured (McJury 2000). Currently, the MRI scanner at OLOLRMC is utilized by MBPCC during MRI/CT fusion for stereotactic radiosurgery (SRS). With polymer gel dosimetry, the MRI scanner can also be used to obtain images of the 3-D dose distribution.

3.1.8 Experimental Procedure

A 3D distribution of absorbed dose within the polymer gel is fixed upon irradiation. Upon receiving the gels, they are stored in a room with temperature between 18-23 °C, away from sunlight and fluorescent light. A few days after gel acquisition, an initial MRI scan is taken of the CGEL-S phantom and gel vials providing background measurements of the gel.
CT scans of the BANG-3\textsuperscript{TM} gel phantom are then obtained to develop a plan on the ADAC treatment planning system. Five fiducial markers are placed on the CGEL-S phantom to facilitate repositioning between irradiations and imaging (e.g. shifting and rotation). In addition, crosshairs are marked on the outside of the CGEL-S. CT scans deliver less than 5 cGy of dose, which is less than 0.5\% of the maximum deliverable dose (10 Gy) (Maryanski email) and 2.5\% of the actual 2 Gy delivered. After the CT scan, a second MRI scan is taken of the CGEL-S phantom to measure any dose from the CT scan or photo-polymerization. This scan is also used for background subtraction of the polymer gel within the phantom.

The phantom is mounted on the treatment couch and irradiated according to the treatment plan. This treated volume simulates a prostate treated with a seven-field IMRT plan. MRI scans of the irradiated BANG-3\textsuperscript{TM} are then taken. Calibration gels are also irradiated and scanned by the MRI scanner. The $R_2$ (relaxation rate) maps calculated from the images are used with the calibration data to acquire quantitative dose maps. Spatial visualization of the treatment volume is obtained from $T_2$-weighted images of the irradiated BANG-3\textsuperscript{TM} gel.

A fourth imaging session some time after irradiation is used to test the stability of the polymer gel. These images are taken approximately two months after the gel irradiation and provide some insight as to how the polymerized regions change over time. These results provide a timetable into when the polymer gels should be imaged and whether or not imaging sessions at a later time would be accurate or beneficial.

The dose determined from the polymer gel is compared to an IMRT QA procedure using solid water and XV film for independent verification. Also, an ionization chamber is used to measure the absolute dose at isocenter. All three procedures are compared with the
planned dose distributions from the Pinnacle³ system whereby the three techniques are evaluated.

Finally, all data analysis involving the polymer gel dosimeter is done via the IDL computer software. IDL provides a wide range of procedures, including aligning data with one another so that it can be accurately compared. This analysis software also has excellent graphing capabilities that provide the experimenter with a means to display figures.

3.2 Dose Verification

3.2.1 Introduction

A comparison is made between the calculated dose from the Pinnacle³ system and the dose within the polymer gel, as measured by MRI. To further verify the gel dosimetry results, additional measurements are made with conventional dosimeters. An ionization chamber (correction factor = 5.362 x 10⁷ Gy/C) and electrometer are used to measure the dose within the cubic water phantom where the vials are irradiated. XV film, solid water and the RIT system are used to determine the 2D distribution delivered by the beams at a source-to-axis distance (SAD) of 100 cm. Also, the ionization chamber and electrometer are used to measure dose at isocenter within the solid water.

3.2.2 Calibration Vials

For calibration purposes, eight vials are placed on the bottom of the water tank at the delivered to each vial in order to obtain a calibration curve for the 4 MV and 10 MV photon beams. These doses ranged from 1 Gy to 9 Gy. To ensure that these doses are properly delivered to the vials, an ionization chamber is used to determine the dose at \( d_{\text{max}} \), the location used to obtain the calibration curves.
As the cubic water phantom contains 1 cm of acrylic at the bottom, a correction factor is computed to determine the effects of the acrylic on the beam energy. To obtain a correction factor, three measurements with the ionization chamber (PTW-30006) are taken with electrometer A with the chamber positioned 10.36 cm below the water surface and the gantry rotated to 180°. The three measurements are then averaged. Three additional measurements are taken and averaged at a gantry angle of 0° with the ion chamber positioned 10.36 cm above the bottom of the tank. This depth was chosen due to an inability of the ion chamber to be placed at a lower depth in the tank. The source-to-surface distance (SSD) is 100 cm for both measurements and 100 MU are delivered. A bias voltage of –300V is set on the electrometer. The correction factor is obtained by dividing the measurement with the acrylic by the measurement without the acrylic and is expected to be close to unity as the effective density of acrylic relative to water is 1.15 (Khan 1994).

The TG-51 calibration protocol is followed to obtain dose values at the CAX. The ion chamber is placed at a depth of 10 cm below the surface of the water and an SSD of 100 cm is again used. Three readings each are taken with at electrometer bias voltages of +300 V, -300 V and -150 V with 100 monitor units (MU) delivered. For each bias voltage, the three measurements are averaged together. The final measurements are designated as $M_A$, $M_B$ and $M_C$ for the bias voltages of +300 V, -300 V and –150 V, respectively. A fourth measurement, $M_D$, is determined by taking a cumulative reading from the electrometer at a bias voltage of -300 V from four exposures of 25 MU each. Units for measurement is Coulombs (C) or readings (rdg).
The dose at 10 cm is determined by the following equation, equation 3.1:

\[ D_{10cm} = M_B \cdot P_{TPC} \cdot P_{ion} \cdot P_{pol} \cdot ECF \cdot N_{w}^{60} \cdot K_Q \cdot \frac{\mu}{\rho_{water}} \cdot P_{TPC} \]

where \( P_{TPC} \) is the temperature/pressure correction factor, where \( P_{TPC} = \frac{273.15 + T(°C) \cdot 760.0}{295.15 \cdot P(mmHg)} \); this corrects the charge or measured current to the standard environmental conditions at which the equipment was calibrated. \( P_{ion} \), the recombination correction factor, takes into account the incomplete collection of charge from an ion chamber and is determined by dividing \( M_B \) by \( M_C \). \( P_{pol} \) is the polarity correction factor, which accounts for any polarity effect on the response of the ion chamber and is computed as \( \frac{M_A + M_B}{2 \cdot M_B} \). \( ECF \), or \( P_{elec} \), is the electrometer correction factor. This factor is 1.00 if the electrometer and ion chamber are calibrated together; however, for a separate calibration of these two instruments, the calibration factor corrects the electrometer reading to true Coulombs (Almond 1999). These correction factors, along with \( M_B \), correspond to the fully corrected charge reading from an ion chamber (Almond 1999).

\( N_{w}^{60} \) is the absorbed dose to water calibration factor for the ionization chamber used in this study, for a \(^{60}Co \) beam under reference conditions. The quality conversion factor, \( K_Q \), accounts for changes in the absorbed-dose to water calibration factor between the beam quality of interest and the beam quality for which the absorbed-dose calibration factor applies (commonly \(^{60}Co \) ) and is chamber dependent (Almond 1999). \( \left( \frac{\mu}{\rho_{water}} \right)_{tissue} \) is the ratio of the photon mass attenuation coefficients of tissue to water. The final dose at 10 cm (\( DF_{10cm} \)) is computed by multiplying \( D_{10cm} \) by the acrylic correction factor. Verification of the dose rate
at \( d_{\text{max}} (DR_{d_{\text{max}}}) \) is computed from the dose at 10 cm as
\[
DR_{d_{\text{max}}} = \frac{DF_{10\text{cm}}}{DDF_{10}} (MU + ee),
\]
where \( DDF_{10} \) is the depth dose fraction at 10 cm determined from machine data book, and
\( MU + ee \) is the addition of the monitor units delivered plus the end effect. This equation will be referred to as equation 3.2. The end effect is calculated as
\[
\frac{100 \cdot (M_D - M_B)}{4 \cdot M_B - M_D}.
\]
\( DR_{d_{\text{max}}} \) should be 1.0 cGy/MU as calibrated by MBPCC. By multiplying \( DR_{d_{\text{max}}} \) by the monitor units delivered (MU) to the vial, the dose (cGy) at \( d_{\text{max}} \) is determined.

### 3.2.3 CGEL-S Polymer Gel Vessel

Presently, the Pinnacle\(^3\) treatment plan and the actual dose distribution delivered by the treatment are being compared by a quality assurance (QA) procedure using XV film and the RIT system. Approximately 8-13 films per patient are taken on the treatment couch with the films sandwiched between two 5-cm slabs of solid water. One film each is taken for a parallel calibration, a perpendicular calibration, a composite, and for each gantry angle at which the treatment is administered. The films are processed and then analyzed by the RIT system whereby beam profiles and isodose curves are obtained. This process is used to verify that the calculated dose from Pinnacle\(^3\) agree with the dose delivered by the beams.

The film is positioned between the two 5-cm solid water slabs for the perpendicular calibration film. An SSD of 95 cm is set to the top of the slab (film at 100 SAD), with a gantry and collimator angle of 180\(^o\) and a field size of 5 x 5 cm\(^2\). Four exposures to monitor units of 20, 40, 60 and 80 MU are made at different positions on the film. An H &D curve is obtained for this calibration film using the RIT system. This curve shows the relationship of the optical density on the film to the applied dose.
Seven perpendicular films are then exposed with a similar setup to that of the perpendicular calibration film. The field size (9 cm x 6 cm), energy (10 MV), SSD (95 cm), gantry angle (180°), collimator (180°) and couch (180°) are the same for each film exposed. For these films, the number of monitor units delivered is determined by an in-house spreadsheet. The input to the spreadsheet is calculated dose rate (cGy/MU) for each field, as determined by the Pinnacle³ planar dose computation and monitor units are the output. The “MU delivered” is the number of MU required to achieve 30 cGy on the film at the point of calculation, an arbitrary point chosen by the experimenter. To determine this point in the RIT system, the jaw setting in the x and y direction is added to the position determined from the Pinnacle³ system. The actual monitor units delivered are rounded from the calculated monitor units (Table 3.3).

XV film is positioned between two slabs of 5-cm solid water to obtain a parallel calibration film. This time, however, the solid water is placed on its side with its largest surface area facing the gantry and the end of the couch. An SSD of 100 cm is set to the top of the phantom with the gantry and collimator angles and the field size remaining the same for the perpendicular calibration film. The correct position of the setup is determined by aligning the lateral lasers along the crevices between the slabs and the sagittal laser 15 cm from each side of the phantom laterally. The dose at $d_{\text{max}}$ should be about the maximum dose expected in the axial film; in this case, 80 monitor units are delivered to the film. An H&D curve is obtained for the parallel calibration film as in the case for the perpendicular calibration film.

A parallel composite film is obtained with the solid water in the same position as that during the parallel calibration film; however, unlike the parallel calibration film, an SSD of 85 cm is set to the top of the phantom. This SSD places the isocenter in the middle of the
Table 3.3 Monitor Units for perpendicular films. Monitor units delivered are calculated from a spreadsheet with the input for the calculated dose rate (cGy/MU) from ADAC. The output is monitor units (MU). Actual monitor units delivered is rounded from the calculated monitor units.

<table>
<thead>
<tr>
<th>Field Name</th>
<th>Calculated dose rate (cGy/MU) from ADAC</th>
<th>Monitor units calculated from spreadsheet</th>
<th>Monitor units delivered</th>
</tr>
</thead>
<tbody>
<tr>
<td>RPO 295</td>
<td>0.884065</td>
<td>33.9</td>
<td>34</td>
</tr>
<tr>
<td>Rt Lat 265</td>
<td>0.463454</td>
<td>64.7</td>
<td>65</td>
</tr>
<tr>
<td>RAO 225</td>
<td>0.924311</td>
<td>32.5</td>
<td>33</td>
</tr>
<tr>
<td>AP 180</td>
<td>0.511203</td>
<td>58.7</td>
<td>59</td>
</tr>
<tr>
<td>LAO 135</td>
<td>0.802424</td>
<td>37.4</td>
<td>37</td>
</tr>
<tr>
<td>Lt Lat 90</td>
<td>0.404187</td>
<td>74.2</td>
<td>74</td>
</tr>
<tr>
<td>LPO 65</td>
<td>0.905329</td>
<td>33.1</td>
<td>33</td>
</tr>
</tbody>
</table>

phantom. Laser alignment is used for correct positioning as in the case of the parallel calibration film. The field size (9 cm x 6 cm) and energy (10 MV) is the same for each beam; however, the gantry, collimator, and couch were rotated as indicated by IMPAC, a treatment verification software system. The monitor units are calculated by the Pinnacle$^3$ system with a prescription dose of 40 cGy at the isocenter (Table 3.4).

For all films, a pin prick is made in the upper left corner (facing the gantry) on the film’s envelope. The number of holes pricked is different for each film, allowing the films to be identified (Table 3.5). Planar doses for the seven fields and the composite are obtained from ADAC. This information is transferred to the RIT scanning system. All ten films (7 planar dose, one composite and two calibration films) are digitized with the pin pricks positioned at the bottom left of the film. The isodose curves and absolute dose profiles for each beam and composite are obtained from the RIT software system, along with the absolute dose at the point of interest. A scaling factor (MU delivered) is multiplied by the relative dose rate at the point of interest (cGy/MU) from the Pinnacle$^3$ system (Table 3.3) to obtain an absolute dose measurement. This factor is the actual number of monitor units to be delivered.
Table 3.4 Parallel Composite film. The table summarizes gantry angle, collimator angle, couch angle and monitor units delivered for each field. Monitor units delivered is computed from ADAC with a prescription of 40 cGy at the isocenter.

<table>
<thead>
<tr>
<th>Field Name</th>
<th>Gantry Angle (degrees)</th>
<th>Collimator angle (degrees)</th>
<th>Couch angle (degrees)</th>
<th>Monitor units, computed from ADAC</th>
</tr>
</thead>
<tbody>
<tr>
<td>RPO 295</td>
<td>295</td>
<td>180</td>
<td>180</td>
<td>7</td>
</tr>
<tr>
<td>Rt. Lat. 265</td>
<td>265</td>
<td>180</td>
<td>180</td>
<td>12</td>
</tr>
<tr>
<td>RAO 225</td>
<td>225</td>
<td>180</td>
<td>180</td>
<td>9</td>
</tr>
<tr>
<td>AP 180</td>
<td>180</td>
<td>180</td>
<td>180</td>
<td>20</td>
</tr>
<tr>
<td>LAO 135</td>
<td>135</td>
<td>180</td>
<td>180</td>
<td>11</td>
</tr>
<tr>
<td>Lt. Lat. 90</td>
<td>90</td>
<td>180</td>
<td>180</td>
<td>13</td>
</tr>
<tr>
<td>LPO 65</td>
<td>65</td>
<td>180</td>
<td>180</td>
<td>7</td>
</tr>
</tbody>
</table>

Table 3.5 Film Identification. Holes are pricked into each film to distinguish each film after processing.

<table>
<thead>
<tr>
<th>Film</th>
<th>Number of Holes</th>
</tr>
</thead>
<tbody>
<tr>
<td>Perpendicular Calibration</td>
<td>1</td>
</tr>
<tr>
<td>RPO 295</td>
<td>2</td>
</tr>
<tr>
<td>Rt. Lat. 265</td>
<td>3</td>
</tr>
<tr>
<td>RAO 225</td>
<td>4</td>
</tr>
<tr>
<td>AP 180</td>
<td>5</td>
</tr>
<tr>
<td>LAO 135</td>
<td>6</td>
</tr>
<tr>
<td>Lt. Lat. 90</td>
<td>7</td>
</tr>
<tr>
<td>LPO 65</td>
<td>8</td>
</tr>
<tr>
<td>Parallel calibration</td>
<td>9</td>
</tr>
<tr>
<td>Composite</td>
<td>10</td>
</tr>
</tbody>
</table>

to each film to achieve a dose of 30 cGy on the film at the calculation point. Both a horizontal and vertical line are chosen for the dose profiles.

The measured absolute dose at the point of interest is compared to the calculated dose for the seven perpendicular films. These values are expected to be approximately 30 cGy for each field. The measured dose on the composite film also is compared to the dose calculated for the composite field by the Pinnacle³ system; both should be approximately 40 cGy.

A parallel-plate ionization chamber (N23342-868) is used to confirm the absolute dose at the isocenter for each individual field. Electrometer B is used to measure the current within
the ionization chamber. The setup for electrometer is as follows: Display: 10 V, Thimble: -300 V, Scale: 10^{-8} C. This measurement provides another method of measuring the dose. For the isocentric measurement, the setup is as follows: SSD = 85 cm, chamber parallel to beam and a 5-cm piece solid water surrounding the center on each side. The setup for the individual fields include: SSD = 95 cm, chamber perpendicular to the beam and a 5-cm piece of solid water above the chamber. The setup for the ionization chamber measurement is the same as that of the composite field setup.

3.2.4 Conclusion

The output of the beam with and without the acrylic is expected to be similar since the mass density of acrylic is similar to that of water, 1.18 $\frac{g}{cm^3}$ and 1.00 $\frac{g}{cm^3}$, respectively (Khan 1994). The dose rate at $d_{\text{max}}$ is expected to be 1.0 $\frac{cGy}{MU}$, which is the calibrated dose rate for the 21EX machine; therefore, the doses delivered to the vials are expected to agree with the doses obtained with the ionization chamber. The RIT system and films are expected to confirm that the calculated dose from the Pinnacle$^3$ agrees with the measured dose produced by the beams. The ion chamber and film measurements provide assurance that the planned dose is accurately delivered to the CGEL-S vessel.

3.3 Energy Calibration of Polymer-Gel Dosimeter

3.3.1 Introduction

A calibration curve is required to relate the measured $R_2$ to the absorbed dose. Novotny, et al., reported a linearity with total dose in the range of 0-9 Gy. These calibration measurements check the linearity of the $R_2$ response to dose. Gels used for calibration must
be stored, irradiated, and MR-imaged under identical temperature conditions as the experimental gels.

3.3.2 Procedure

Nine borosilicate glass vials (VIAL in Table 3.1) are purchased from MGS Research, Inc. to obtain calibration curves. Vial 1 is designated as the unirradiated vial. Vials 2-4 are irradiated with 4 MV photons to total doses of 1, 4 and 7 Gy, respectively, at a dose rate of 250 MU/min. Vials 5-9 are irradiated with 10 MV photons. Vials 5 and 6 are irradiated to a total dose of 1 Gy with a dose rate of 400 and 500 MU/min, respectively; Vials 7-9 receive total doses of 4, 7 and 9 Gy, respectively (Table 3.6) at a dose rate of 400 $\frac{MU}{min}$. The cylindrical vials are placed in a cubic water-filled phantom (35 cm x 35 cm x 38 cm) where a photon beam from the Varian Clinac 21EX is administered parallel to the cylindrical axis. The vials are positioned vertically at the bottom of the water tank with approximately 10 cm of water above the vials (Figure 3.1). A gantry angle of $0^\circ$ and a field size of $10 \times 10 \text{ cm}^2$ is used to irradiate the tank.

The Varian Clinac 21EX is calibrated at MBPCC to give a dose of $1 \frac{cGy}{MU}$ at $d_{\text{max}}$ (SSD=100 cm, FS = $10 \times 10 \text{ cm}^2$), where $d_{\text{max}}$ is at a depth of 1.2 cm for 4 MV and 2.5 cm for 10 MV. The radiation beam passes through 1 cm acrylic and 1 mm glass at the bottom of the tube. Because the MRI slices are 2 mm thick, it is expected that the slice containing the maximum dose is slice 1 for the 4 MV photons and slice 7 or 8 for the 10 MV photons. Because of the potential for vial misalignment, the slice at which $d_{\text{max}}$ occurs is not certain. For the background vial, transverse relaxation rates ($R_2$) are computed for slices 6, 7 and 8 for the 10 MV photons and slices 1, 2 and 3 for the 4 MV photons. $R_2$ numbers are adjusted.
Table 3.6 Vial Irradiation. The table gives total dose (Gy), monitor units, dose rate ($\frac{MU}{\text{min}}$) and energy (MV) given to each vial. The vials will be imaged via MRI whereby a calibration curve can be obtained ($R_2$ vs. dose). Vial 1 is left unirradiated to determine the $R_2$ value when no dose is given to the gel within the vials (background reading). Due to experimental error, vial 6 was irradiated to a different dose rate than vials 5 and 7-9.

<table>
<thead>
<tr>
<th>Vial</th>
<th>Total Dose (Gy)</th>
<th>MU delivered</th>
<th>Dose rate ($\frac{MU}{\text{min}}$)</th>
<th>Energy (MV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Unirradiated</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>2</td>
<td>1</td>
<td>100</td>
<td>250</td>
<td>4</td>
</tr>
<tr>
<td>3</td>
<td>4</td>
<td>400</td>
<td>250</td>
<td>4</td>
</tr>
<tr>
<td>4</td>
<td>7</td>
<td>700</td>
<td>250</td>
<td>4</td>
</tr>
<tr>
<td>5</td>
<td>1</td>
<td>100</td>
<td>400</td>
<td>10</td>
</tr>
<tr>
<td>6</td>
<td>1</td>
<td>100</td>
<td>500</td>
<td>10</td>
</tr>
<tr>
<td>7</td>
<td>4</td>
<td>400</td>
<td>400</td>
<td>10</td>
</tr>
<tr>
<td>8</td>
<td>7</td>
<td>700</td>
<td>400</td>
<td>10</td>
</tr>
<tr>
<td>9</td>
<td>9</td>
<td>900</td>
<td>400</td>
<td>10</td>
</tr>
</tbody>
</table>

according to the ratio discussed in Chapter 3.4.2. The smallest $R_2$ value observed in these slices is designated as the $R_2$ value for vial 1. For the irradiated vials, one axial slice in each vial receives the desired dose at $d_{\text{max}}$. Within these vials, $R_2$ is computed for the first axial slices 6,7 and 8 for the 10 MV beam and slices 1,2 and 3 for the 4 MV beam. The largest value is used as the $R_2$ at $d_{\text{max}}$ (Chapter 3.6.6). A graph of $R_2$ vs. absorbed dose is produced (Chapter 4.5), which is the calibration curve. Two calibration curves are computed, one that includes background subtraction (y-intercept=0) and one that does not include background subtraction (non-zero intercept). The slope of the linear portion of the calibration curve gives the gel sensitivity. The storage, irradiation and temperature during MR imaging of the gel vials used for dose calibration and the large experimental gel were kept under identical conditions. All MRI scanning is performed at the same time post-irradiation.
3.4 Application of BANG-3™ Gel Dosimetry for the Verification of an IMRT Treatment Plan

3.4.1 Introduction

Verifying a plan with the BANG-3™ polymer gel would be a tremendous boost to MBPCC in terms of having a modality to image 3D dose distributions of an IMRT plan. Equipment utilized in this study includes one BAREX CGEL-S vessel, one fiducial marker set, the Pinnacle³ treatment planning system, the Varian Clinac 21EX and the MRI scanner. The role of each piece of equipment is briefly discussed.

3.4.2 Procedure

The CGEL-S vessel is filled with BANG-3™ polymer and is used to verify the IMRT plan in three dimensions (Figure 3.2). A BAREX plastic cylinder contains the BANG-3™ polymer gel. A set of fiducial markers (SCM set) is used for alignment purposes. The SCM set contains 4 vinyl suction cups and 5 CT/MRI fiducial marker cups; the vinyl suction cups, intended for stereotactic radiosurgery, are not used. The fiducial markers are placed on the outside of the gel. A central cavity within the fiducial marker cup remains empty during CT scanning but is filled with water for MRI scanning. A syringe is used to fill the cavity with water; care is taken to prevent air bubbles in the cavity.

The CGEL-S phantom is positioned on the CT scanner table. The crosshairs from the alignment lasers are marked on tape placed on the outer surface of the phantom. The marks facilitate repositioning of the phantom for irradiation and imaging. Five fiducial markers are placed at various positions on the phantom’s surface. An outline of the fiducial markers is traced on the phantom just in case the marker falls off of the phantom. A CT scan of the CGEL-S phantom is acquired and sent to the Pinnacle³ treatment planning computer. A simulated IMRT treatment is planned; the plan is a prostate treatment with seven fields.
Figure 3.2 Photograph of the irradiation set-up of the BAREX CGEL-S vessel used in this study. The irradiation consisted of a seven-field prostate technique (see Appendix A). The photo shows the gantry angle of $270^\circ$. One sees on the phantom the repositioning mark and fiducial markers.

(Appendix A). Contours of the right and left femoral heads, prostate, bladder and rectum are drawn and an optimal isodose distribution around these structures is computed. With this prostate plan, dose prescribed to the center of the prostate (isocenter) is 2 Gy, while the dose to the organs-at-risk is kept under 1 Gy.

To irradiate the gel, the treatment plan is first transferred to the IMPAC record and verify system used at MBPCC. This system allows the physicist/dosimetrist a means to monitor the radiation delivered to a particular patient and also provides a mechanism to view the important parameters used during treatment (e.g., couch, gantry and collimator positions).

Two days prior to irradiation, the gels are placed in the 21EX treatment room and stored in a cardboard box. This step allows for thermal equilibrium within the gel but was later determined to be very costly as the gel received excess scattered radiation from patient treatments and leakage radiation from the machine head. These excess doses are measured to
attempt to determine the amount of excess radiation. The scattered radiation measurement will be an approximate value due to the inability to recreate the patient treatments, while the leakage radiation will be somewhat of an absolute value as it will be calculated from the monitor units delivered in those two days. Within the CGEL-S phantom, unirradiated regions are analyzed for scanning sessions 1-4. As the dose is expected to be zero, the $R^2$ values should be approximately the same for all sets of images. In an attempt to correct for any excess radiation, a dose correction factor is applied. The $R^2$ value for the vials for scanning session 1 (unirradiated vials) are compared to vial 1 (unirradiated vial) from scanning session 3. If no contamination is present, these values should also agree. If some contamination is present (i.e., unirradiated vials do not agree) then the $R^2$ values will be adjusted according to the ratio of the $R^2$ values of the two scanning sessions. This ratio is known as the dose correction factor.

For irradiation, the gel is mounted on the treatment couch of the Varian Clinac 21EX with the lasers aligned to the crosshairs traced from the CT scanner lasers. An SSD of 91.3 cm places the prescription point at the center of the gel. However, the lasers alignment indicated an SSD of 92.8 cm. In turn, the couch was lowered by 1.5 cm to give the appropriate SSD whereby the coronal lasers were no longer aligned with the crosshair mark on the tape. An irradiation with 10 MV photon beams is then administered to the phantom. The 21 EX contains MLCs and uses a “step and shoot” approach to perform the IMRT treatment. An MRI scanner located at OLOLRMC is used to image the dose distributions within the BANG-3™ gel.
3.4.3 Conclusion

This part of the experiment focuses on the use of the BANG-3™ polymer gel dosimeter to obtain quantitative dose distributions from an IMRT treatment. Comparison of the dose map obtained from the MRI scanner to that of the IMRT treatment plan of the Pinnacle³ system is described in Chapter 3.6 and Chapter 4.6. Dose distributions are contained within the polymer gel. Dose maps from the MRI scanner are expected to agree with a slight deviation due to the exposure of the gel to scattered radiation in the treatment room.

3.5 Imaging of the Calibration Vials and the CGEL-S Vessel

3.5.1 Introduction

Imaging of the polymer gel provides useful information regarding the amount of polymerization within the gel. Imaging of the BANG-3™ polymer gel is done by either MRI or optical-CT. Though optical-CT may have advantages over MRI, the scans in this study is done with the latter due to availability. Two types of pulse sequences are used for the MRI imaging of a polymer gel: multiple spin-echo sequence and the single echo (Hahn spin-echo sequence); the user is free to decide which sequence to use.

Four separate imaging sessions are conducted in order to monitor the gel at separate times in the study. The first session images the CGEL-S phantom and the nine borosilicate vials (session 1). This information is useful in determining optimal parameters for viewing the gel and can be used for background subtraction of the vials. Upon completion of the CT scan, a second image of the CGEL-S phantom is taken (session 2) in order to perform background subtraction on the gel within the phantom. An image is useful at this time due to the possibility of contamination of the gel due to light, temperature variation and the CT
scanning. Minimal contamination is expected as the gel is stored at room temperature away from light and CT scanning delivers less than 5 cGy dose to the gel. Images of the pre-irradiated gels are later used to compare the $R_2$ values of the unirradiated gel (scanning sessions 1 and 2) to the $R_2$ values in the unirradiated vial (scanning session 3). During session 2, marks are placed on the coils to allow a repositioning of the phantom during the next imaging session.

After irradiation, imaging of the polymerization is necessary to obtain maps of the dose distribution. Both the CGEL-S phantom and borosilicate vials are imaged in session 3. A fourth imaging session of the CGEL-S phantom is done approximately two months post-irradiation (session 4) to determine the stability of the gel and its usefulness at later times.

Certain precautions must be taken during imaging to obtain optimal results from the MRI scanner. These precautions include the temperature dependency of the gel, proper calibration of the scanner and laser misalignment. BANG-3™ gels are temperature dependent and because of this dependence the gels must be equilibrated to room temperature before imaging. MGS Research, Inc. recommends that the gel be allowed to equilibrate to room temperature for two days prior to imaging. Also, it is suggested that both the calibration vials and the gel phantom be kept at similar temperature conditions to provide consistency between the measurements. Information on the scanner calibration and coil calibration were obtained from the MR technician (Guy MRI). From this conversation, it was determined that the scanner was calibrated but the coils were not calibrated. Finally, through discussions with the MR technician, it was determined that preventative maintenance is performed on the alignment laser once per month, assuring us that the laser position has not changed between scanning sessions.
3.5.2 Procedure

Imaging is done using the MRI scanner at OLOLRMC. All gel samples are left in the MRI room two days prior to imaging to allow them to become thermally equilibrated to the room temperature. Recommendations for the parameters of the scan have been given by the MGS Research, Inc. website and are closely followed for MRI scanning (Table 3.7). The actual parameters utilized in this study are shown in Table 3.8. The MRI should not be delayed by more than one week after irradiation. For all sessions, the Hahn spin-echo sequence is used for localization, proton density maps and $T_1$ maps. Vials 1, 2 and 3 are placed on the bottom row, 4, 5 and 6 on the middle row and 7, 8 and 9 on the top row. Localization scans are done to determine the position of the gel within the head coil. Repetition times (TR), slice thickness, pixel size, field of view, and acquisition matrix size are the same for all sessions. Also, all scans took place in the head coil. Within each session, the echo time (TE) is changed from 21.8 ms ($TE_1$) for proton density images, carrying information on coil sensitivity and spatial non-uniformities of the flip angle, to 116 ms ($TE_2$) for a heavily $T_2$-weighted image, carrying information on the dose distribution. Thirty-seven slices are used to image the vials, while seventy slices are used for the imaging of the CGEL-S phantom. An interleaved acquisition is used to eliminate magnetization cross-talk between neighboring slices. Acquisition times for the vials (10 min) are shorter than times for the CGEL-S phantom (17 min) because of the fewer slices. The field of view (FOV) is 22 x 22 cm$^2$ for an acquisition matrix of 192 x 192 pixels. Data is reconstructed into a 256 x 256 matrix upon imaging. Eight acquisitions are averaged for each slice (NEX=8). All image data are transferred to an in-house computer at Louisiana State University and analyzed with IDL software (Research Systems, Inc.). Data is also transferred to a DAT tape for backup.
Table 3.7 MRI parameter recommendation. Parameters recommended by MGS Research, Inc. to obtain optimal results when imaging (MGS website).

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Instructions</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pulse sequence</td>
<td>Hahn spin-echo (single echo); TR at least 1000 ms; First acquisition: TE=20 ms, Second acquisition: TE=100 ms</td>
</tr>
<tr>
<td>Transmitter and receiver gains</td>
<td>Keep constant for both TE acquisitions.</td>
</tr>
<tr>
<td>Multiple slices</td>
<td>Use Interleave mode (i.e. acquire slices #1, 3, 5, 7, etc., then #2, 4, 6, 8, etc.); long TR may be necessary to accommodate many slices to cover the required volume of the gel.</td>
</tr>
<tr>
<td>SNR</td>
<td>Use smallest coils available: head coils or, for smaller gels, extremity coils. Use spacers to position the gel in the center of the coil; use the largest pixel size (i.e. the smallest image matrix and/or the largest field of view) that is acceptable to your specific study; use as much signal averaging as possible, especially for longer TE’s, as they produce weaker signals; consider using longer TR for more magnetization recovery.</td>
</tr>
<tr>
<td>Consistency</td>
<td>use the same parameters for all gels in the experiment for consistency</td>
</tr>
</tbody>
</table>

Table 3.8 MRI Parameters. Coil type, echo time (TE), repetition time (TR), slice thickness, pixel size, field of view and number of acquisitions used for each session. Session 1 was performed 3 days after the gels arrived. Session 2 was performed 3 days after the CT scan. Session 3 was performed 5 days after irradiation, while session 4 was performed approximately two months after irradiation.

<table>
<thead>
<tr>
<th>Image type</th>
<th>coil</th>
<th>TE (ms)</th>
<th>TR (ms)</th>
<th>slice thickness (mm)/spacing (mm)</th>
<th>acquisition matrix size (pixels)</th>
<th>FOV (cm²)</th>
<th>acquisitions (averaging)</th>
</tr>
</thead>
<tbody>
<tr>
<td>proton density</td>
<td>head</td>
<td>21.8</td>
<td>1300</td>
<td>2/2</td>
<td>192x192</td>
<td>22x22</td>
<td>8</td>
</tr>
<tr>
<td>T₂</td>
<td>116</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

To qualitatively examine the location of the treatment volume in the gel phantom, the T₂ images alone are used. Quantitative R₂ maps are computed from the T₂ and proton density image data. The transverse NMR relaxation rate (R₂ = 1/T₂) of the water protons in
the BANG-3\textsuperscript{TM} gel is proportional to the dose ($R_2 = R_0 + kD$). $R_2$ maps are produced by the following equation: $R_2 = \frac{1}{TE_2 - TE_1} \ln\left( \frac{S(TE_1)}{S(TE_2)} \right)$. $S$ is the signal strength (i.e., pixel values) for the images. From the $R_2$ maps and the vial calibration, the quantitative dose maps are obtained.

The use of background subtraction is investigated within the CGEL-S vessel by subtracting the $R_2$ value obtained from scanning session 2 from the values obtained from sessions 3 and 4. The zero-intercept calibration curves (Chapter 3.6.6) are used to obtain the dose maps within the vessel when image subtraction is performed (Chapter 4.5).

3.5.3 Conclusion

An accurate MRI image is important for determining the relation between the transverse relaxation rate and the distributed dose. With this information, dose maps are determined, whereby the IMRT plan is verified. Data that account for background subtraction are expected to be more accurate than the data with no subtraction; subtraction should minimize the effects of imaging artifacts or photopolymerization of the gel. Proper precautions for thermal equilibration are followed because the BANG-3\textsuperscript{TM} polymer gel reportedly is dependent upon temperature during imaging.

3.6 Comparison of Measured to Planned Dose Maps

3.6.1 Introduction

IMRT treatments have recently become common at MBPCC. Through the use of the BANG-3\textsuperscript{TM} polymer gel dosimeter, the 3D-dose distribution from an IMRT treatment is obtained by MRI imaging of the irradiated polymer gel. Sub-micron sized polymer particles are formed within the gel. These polymer particles locally alter the $T_2$ of the gel, producing
3D images of dose distribution (MGS Research website). Comparison of the Pinnacle\textsuperscript{3} treatment plan to the dose map obtained from the polymer gel provides a means of verifying the IMRT plan in three dimensions.

Comparisons are made through the use of the IDL computer program. Dose matrices are obtained for both the Pinnacle\textsuperscript{3} planning system as well as the polymer gel, whereby these individual maps are compared. Dose maps within the polymer gel are determined at two separate times post-irradiation, 5 days and approximately two months. Image subtraction, point by point, is done by subtracting the dose matrix of MRI/polymer gel from the dose matrix of the Pinnacle\textsuperscript{3} planning system. Finally, dose-volume analysis from Pinnacle\textsuperscript{3} and the MRI is compared. The dose within the prostate volume and OAR volumes is computed with IDL and compared with the data obtained from the Pinnacle\textsuperscript{3} system. Global dose volume analysis is also conducted, whereby dose is analyzed plane by plane. Qualitative comparisons are made by analyzing the shape of the contours and color images for the Pinnacle\textsuperscript{3} dose maps and MRI dose maps.

Several data-processing steps are required for generating and comparing the dose matrices. Such procedures include: the elimination of noise fluctuations and background (non-gel related) regions, the alignment of the MR images to CT images as well as alignment of the pre-irradiated MRI images to the images obtained post-irradiation, image rotation and image subtraction. From this point, the $R_2$-dose calibration curve and the dose matrix are obtained along with the comparison of the dose maps from the treatment plan and the polymer gel.
3.6.2 Procedure

Once the gel has been irradiated and scanned, the polymerized region within the gel is ready to be analyzed. When compared to the treatment plan, images of the polymer gel will demonstrate the effectiveness of BANG-3™ polymer gel as a 3-D dosimeter. Expected dose distributions are computed by the Pinnacle\textsuperscript{3} planning system software. IDL, a commercial data processing and analysis software package, is used to determine the measured dose distribution from the polymer gel images. IDL is also used to compare the two sets of data by image subtraction, contour comparisons and dose-volume analysis.

Upon completion of the MRI scans, the gel images are saved on both an optical disk and a DAT tape for later analysis. The images are transferred to the CT scanner from the optical disk, followed by a network file transfer (ftp) of the data to a computer at Louisiana State University (LSU). The CT images of the polymer gel are also transferred to LSU. Dose information from the treatment planning system is transferred manually by a floppy disk to the same computer as that for the MRI and CT data. From this point, IDL codes are developed to perform the data processing and comparisons, as described in the following sections. The IDL code is included in Appendix B.

3.6.3 Noise Fluctuations and Dose Points Outside the CGEL-S

When performing the calculation to determine the $R_2$ values from the echo times and signal intensities, a problem occurs in pixels where $S(TE_2) = 0$ and $\frac{S(TE_1)}{S(TE_2)} < 1$. The latter case results in negative $R_2$ values, which is impossible as it implies a negative relaxation time, while the former situation gives undefined results due to division by zero. To avoid these problems, pixels exhibiting the above characteristics are excluded from the $R_2$ calculation and
assigned values of zero. Because the FOV (22 cm x 22 cm) is larger than the CGEL-S phantom (17 cm diameter), the problem points correspond to background noise fluctuations in the regions outside the phantom and are ignored with no major consequences.

Also, to remove noise fluctuations, a circular mask (Appendix B) is set around the CGEL-S vessel, whereby any points outside of the mask are set to zero and all points inside are set equal to one then analyzed. These masks are developed for the two post-irradiation image sets (scanning sessions 3 and 4) and the pre-irradiation image (scanning session 2). The diameter \( d \) of the region to mask is determined by first viewing the profiles of the gel and its components (profile function) then analyzing which points corresponded to outer surface of the gel. From here, the radius \( r \) of the masked region can be determined \( r = \frac{d}{2} \). These surface points are computed in the x and y coordinates. The central point of the circular region is computed by taking the average of the surface points determined from above.

**3.6.4 Alignment of CT and MRI Images**

With data information being obtained at different time intervals, reproducible gel positioning is a matter that must be addressed. A single CT scan is taken of the gel for planning purposes and four MRI images of the gel are taken over a two-month span. Ideally, the gel will be lined up in the same position for the CT scan and the four MRI scans; however, this perfect alignment is hard to obtain and is not expected. Therefore, steps are taken to allow for correction of small misalignments of the CGEL-S phantom. Alignment corrections of the vial images are not attempted due to the fact that reproducible positioning from scan to scan of all nine vials was unfortunately neglected, thus it would be impossible to accurately align the vials. However, the method used to compute the \( R_2 \) -dose calibration data from the vials does not depend on accurate vial alignment as described in Chapter 3.3.2.
The first step taken to properly align these images is the placement of fiducial markers and crosshair marks on the gel prior to the CT scan. The laser crosshairs are used as reference points for subsequent irradiation and MRI scans. The fiducial markers were previously described in Chapter 3.4.2. During analysis, the location of the fiducial marks between the MRI scans and CT scan are compared to determine the proper rotations, shifts and scaling to align the various image data. Shifts and rotations of the gel images from MRI sessions 3 and 4 are needed for alignment with the CT scans (the MRI scans are shifted and rotated relative to the CT scans). The gel images from MRI scanning session 2 are shifted to align with the data from MRI scanning sessions 3 and 4 (i.e., pre-irradiation scans are shifted to align with the post-irradiation scans).

To determine rotation angles, the location of the central cavities of the fiducial markers and the center of the gel cylinder are calculated for the CT scan and MR images. Three markers are analyzed; two marks appear in slices 23 and 19 in the CT scans (corresponding MRI slices at slice 40 and 47, respectively). The CT scans are cropped in matrix size from 512 pixels x 512 pixels to 409 pixels x 408 pixels then rebinned to 256 pixels x 256 pixels to convert the CT and MRI images to identical image sizes. From here, the deviation of the fiducial marker position is computed by determining the angle between the marker and the adjacent axis. The horizontal axis was the reference axis for marker in MRI slice 40 (CT slice 23) while the vertical axis was the reference axis in MRI slice 47 (CT slice 19). A third slice with the third fiducial marker (MRI slice 13, CT slice 40) is examined for the MRI data from session 3 to check the computed rotation angle. Built-in IDL functions are used to do the actual matrix rotation. Due to an observed slanting of the vials...
and gel vessel in the x and y directions, rotations are also considered for out-of-plane situations, whereby a rotation function “rot function” is developed and used (Appendix B).

### 3.6.5 Image Subtraction

Image subtraction is important for data analysis because it allows for exclusion of polymerization reactions that occurred prior to irradiation. These unwanted reactions occur because of contamination from oxygen, light or dose from the CT scanner. With image subtraction, the measured change in $R_2$ values is the result of irradiation only. Before subtraction of the images can occur, the gel image data must be correctly aligned. In this study, the MRI image sets are rebinned from 256 pixels x 256 pixels to 512 pixels X 512 pixels in order to provide more precision in the shift. The shift magnitude is selected visually from overlays of the shifted images. The shifted pre-irradiation values from session 2 are subtracted from the post-irradiation $R_2$ data (session 3 and 4). The zero-intercept calibration curve (Chapter 3.6.6) is used to determine the dose distribution within the images.

### 3.6.6 $R_2$-Dose Calibration Curve

For the $R_2$-dose calibration, values are obtained for 4 and 10 MV photons with and without image subtraction. Relaxation rates ($R_2$) are determined in a 3 pixel X 3 pixel region of interest with the center of this region positioned at the centers of the vials. The $R_2$ values are determined for slices 1, 2 and 3 for the 4 MV beam and 6, 7 and 8 for the 10 MV beam. The largest average $R_2$ value for the three slices is used in the calibration curve. With this information, a comparison is made between the calibration curves for the pair of photon energies. As the depth of maximum dose is at 2.5 cm for the 10 MV photons and 1.2 cm for the 4 MV photons, one expects to observe this maximum $R_2$ at depths of 1.4 cm and 0.1 cm, respectively (due to 1 cm acrylic for bottom of the tank and 1 mm glass in the vial), from the
bottom of the gel in the vials. The corresponding slices are slice 7 for 1.4 cm and slice 1 for 0.1 cm. For the unirradiated vial, the $R_2$ values are determined by taking the smaller value in slices 6-8 for the 10 MV photons and slices 1-3 for the 4 MV photons.

Calibration curves are generated with Microsoft Excel with delivered doses plotted on the x-axis and $R_2$ values on the y-axis. For both photon energies, a trend line is fit to the graphs produced and a linear equation is determined. For cases where background subtraction is analyzed, a zero-intercept calibration curve is generated as the y-intercept being set to zero.

3.6.7 Dose Maps From Pinnacle$^3$ and Polymer Gel

$R_2$ values are determined for each pixel within the cylindrical gel vessel by the same calculation used to determine the values within the vials. The $R_2$ map within the gel is determined 5 days post-irradiation and approximately two months post-irradiation. Because of the increased dose expected in the images obtained 2 months post-irradiation, an age dose correction factor is used to better depict these images. The correction factor is determined by dividing the maximum dose obtained from the images in scanning session 3 by the maximum dose obtained in scanning session 4. This age correction is discussed in Chapter 4.6.3 and is shown in Figure 4.24.

The MRI scans used a 22 cm X 22 cm FOV and a pixel size of 0.086cm x 0.086 cm. The Pinnacle$^3$ software has the capabilities of determining the dose values for every pixel in a specified region of interest. The field of view and pixel sizes are also set to 22 cm x 22 cm and 0.086 cm x 0.086 cm, respectively. Accordingly, the Pinnacle$^3$ boundary region data is examined at pixel sizes of 0.086 cm x 0.086 cm. Dose values are obtained for every pixel of each slice from the inferior edge of the gel vessel to 10 cm superior to this boundary in increments of 2 mm. Dose information above this boundary is expected to be zero. Data
from the Pinnacle³ is transferred to the same computer as that used for the dose maps from the polymer gel.

### 3.6.8 Dose Error

For a pixel in a gel image with a measured $R_2$ value, the absolute dose error ($\sigma_D$) can be computed with the following equation:

$$\sigma_D = \sqrt{(R_2 \cdot \sigma_\alpha)^2 + (\alpha \cdot \sigma_{R_2})^2 + (\sigma_{\alpha})^2}$$

(Baldock 1999). In this equation, $R_2$ is the measured relaxation rate, $\sigma_\alpha$ is the slope uncertainty, $\alpha$ is the slope, $\sigma_{R_2}$ is the relaxation rate uncertainty and $\sigma_\alpha$ is the uncertainty in the y-intercept. Units are in cGy. $R_2$, $\sigma_{R_2}$ and $\alpha$ are values that can be measured, whereas $\sigma_\alpha$ and $\sigma_{R_2}$ are values that are computed from the least-squares fit of the calibration data. It has been reported that the smallest percent error occurs at higher doses and the most significant uncertainties are due to $\sigma_{R_2}$ for low doses and $\sigma_\alpha$ for high doses (Oldham 1998).

### 3.6.9 Data Comparisons

Once the proper procedures are performed prior to data acquisition and the results have been acquired, comparisons between the Pinnacle³ and polymer gel dose distributions is done. Three techniques used for comparisons are image subtraction, contour shapes and dose volume analysis. All three techniques are performed with an in-house computer program using IDL (Appendix B).

For image subtraction analysis, the average dose difference and standard deviation is computed both globally for each plane (94.4 SPD-104.4 SPD) and locally for points within the regions of interest (e.g. prostate, femoral heads, bladder and rectum). These points are determined from a Pinnacle³ file, where the boundary regions from the Pinnacle³ file around the organ volumes is converted from units of cm to pixels. The data points are examined at a
slice thickness of 2 mm slices; this is the slice thickness used throughout this study.

Programming allows for a determination of which pixels correspond to the organs of interest.

Contour analysis is used for a qualitative comparison between the Pinnacle\(^3\) dose and MRI dose. The planes analyzed are between 98.6 SPD and 101.6 SPD as this region contains the area of interest (i.e., isocenter, critical structures and tumor volume). Doses contoured are 200 cGy, 190 cGy, 180 cGy, 150 cGy and 100 cGy. In some figures, the organs of interest are included at 100 SPD along with the contour levels from both the Pinnacle\(^3\) plan and MRI measurements. MRI measurements are shown for 5 days and two months post-irradiation.

For the same pixel regions used in image subtraction analysis, the total volume within each organ is determined, whereby the total dose within the organ volume is computed (DVH). The total volume of each organ determined is compared to the volume obtained by the Pinnacle\(^3\) plan. This procedure allows for a verification of the volume used in the dose-volume-histograms. The measured data are shown with and without the dose correction factor (Chapter 3.4.2). Also, the total dose plane-by-plane for each contour region (100 cGy, 150 cGy, 180 cGy, 190 cGy and 200 cGy) per volume (cm\(^3\)) from the Pinnacle\(^3\) system and MRI is compared. MRI data is shown both with and without the dose correction factor (Chapter 3.4.2). Finally, the maximum dose measured with MRI and calculated with the Pinnacle\(^3\) for each plane is compared. Again, MRI data is shown with and without the dose correction factor.

**3.6.10 Conclusion**

This section discusses the methods involved with comparing the dose maps obtained from the polymer gel and the Pinnacle\(^3\) planning system. All computer programming is done with the IDL software. Proper corrections must first be applied to ensure compatibility.
between the image sets. Removal of noise fluctuations and background dose points, as well as alignment corrections, are key steps. Also, image subtraction is suggested to eliminate data from polymerization that occurs before irradiation. Methods to obtain dose maps for the polymer gel from the $R_2$-dose calibration curve and the maps from planning using the Pinnacle³ software are explained.

Three procedures are described to compare the data sets, including: image subtraction, contour analysis and dose-volume analysis. For image subtraction, an ideal result would be zero, which implies no difference between the Pinnacle³ plan and the polymer gel dose map. Results chosen to consider the study a success would be a difference in the data sets of 5% of maximum dose that can be delivered to the gel. Contour shapes are expected to agree between the MRI data and CT data. The desired difference in dose analysis results is 5% dose agreement. The Pinnacle³ treatment planning system and the dose map obtained from the polymer gel are expected to agree if the images are properly aligned. Alignment is very important as a deviation of even 1 mm can lead to inaccurate results.
Chapter 4

Results

4.1 Dose Verification

4.1.1 Correction Factor

With the presence of acrylic, the measurements were 1.398, 1.399 and 1.399 with the average of these three numbers at 1.399. When acrylic was removed from the radiation beam, the three measurements were 1.406, 1.405 and 1.405 with the average at 1.405. Dividing the beam average with acrylic by the beam average without the acrylic, a correction of 0.996 was computed.

4.1.2 Central Axis Dose

The measurements, $M_1$, $M_2$ and $M_3$ and their averages/cumulative measurement for the four sets of data taken can be found in Table 4.1. Also, the number of monitor units delivered and the bias voltages set on the electrometers can be found in this table. $\text{PTP}$ was computed to be 1.003. The ECF, determined by the calibration company, was $1.002 \times 10^{-8}$

$$\frac{C}{\text{Rdg}}$$

$\text{P}_{\text{ion}}$ and $\text{P}_{\text{poly}}$ were also calculated and were found to be 1.004 and 1.000, respectively. The water calibration factor ($N_w^{\text{Co-60}}$), obtained from the calibration company was $53.62 \times 10^8$

$$\frac{c\text{Gy}}{\text{C}}$$

$K_Q$ was found to be 0.979 for the PTW-30006 chamber (Almond 1857). Finally, $\left(\frac{\mu}{\rho}\right)_{\text{tissue}}$, was determined to be 0.99 for 10 MV photons. From equation 1 (see Chapter 3.2.2), the dose at 10 cm, $D_{10\text{cm}}$ was computed to be 75.001cGy. The final dose at 10 cm ($DF_{10\text{cm}}$) was computed to be 74.701 cGy.
Table 4.1 Measurements taken using the TG-51 protocol to determine the dose at $d_{\text{max}}$. $M_A$, $M_B$ and $M_C$ are determined by averaging the three reading taken at the respective bias voltages, whereas $M_D$ was determined by taking a cumulative reading of the measurements.

<table>
<thead>
<tr>
<th>Bias Voltage (V)</th>
<th>MU</th>
<th>$M_1$</th>
<th>$M_2$</th>
<th>$M_3$</th>
<th>$M_{AVG}/M_{CVE}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>+300</td>
<td>100</td>
<td>1.432</td>
<td>1.430</td>
<td>1.434</td>
<td>$M_A=1.432$</td>
</tr>
<tr>
<td>-300</td>
<td>100</td>
<td>1.431</td>
<td>1.431</td>
<td>1.432</td>
<td>$M_B=1.431$</td>
</tr>
<tr>
<td>-150</td>
<td>100</td>
<td>1.426</td>
<td>1.426</td>
<td>1.427</td>
<td>$M_C=1.426$</td>
</tr>
<tr>
<td>-300</td>
<td>4X25</td>
<td>.357</td>
<td>.716</td>
<td>1.074</td>
<td>$M_D=1.430$</td>
</tr>
</tbody>
</table>

The $DDF_{10}$ was found in a table at MBPCC and was determined to be 0.7326. This factor is dependent on depth, energy and field size. The number of monitor units delivered was 100 MU and the end effect was computed to be -0.023. From here, the dose rate at $d_{\text{max}}$ ($DR_{d_{\text{max}}}$) was computed from equation 2 to be $1.020 \frac{cGy}{MU}$. Vials 2, 5 and 6 were irradiated with 100 MU, thus the dose at $d_{\text{max}}$ was computed to be 102.0 cGy or 1.02 Gy. A similar multiplication can be done for the 400MU, 700 MU and 900 MU delivered to the remaining vials yielding doses of 4.08 Gy, 7.14Gy and 9.18 Gy at $d_{\text{max}}$.

4.1.3 Planar Dose

Dose verification methods for the seven irradiated fields can be found in Table 4.2. Dose data from the Pinnacle$^3$ is compared to the film and ionization chamber dose data. The jaw setting for each beam is 4.5 cm for x and 3.0 cm for y, while the position determined from Pinnacle$^3$ for the seven beams, RPO 295, Rt. Lat. 265, RAO 225, AP 180, LAO 135, Lt. Lat 90 and LPO 65 was (0,0), (1.80, 0.10), (0.03, 0.05), (-2.11, 0), (-0.76, 0.42), (-2.06, 0.47) and (-0.039, -0.17). The points of interest (see Chapter 3.2.3) for the seven treatment beams in the RIT system were calculated to be (4.5, 3.0), (6.3, 3.1), (4.53, 3.05), (2.39, 3.0), (3.74, 3.42), (2.44, 3.47) and (4.11, 2.83). Leakage from electrometer B was computed to be $0.001 \times 10^{-8}$.
Table 4.2  Dose verification for the seven irradiated beams using film and an ionization chamber. Dose values were measured at 100 SAD.

<table>
<thead>
<tr>
<th>Beam</th>
<th>ADAC dose (cGy)</th>
<th>Film dose (cGy)</th>
<th>Ion chamber dose (cGy)</th>
<th>Percent Difference Film</th>
<th>Percent Difference IC</th>
</tr>
</thead>
<tbody>
<tr>
<td>RPO 295</td>
<td>30.06</td>
<td>33.04</td>
<td>30.48</td>
<td>9.914</td>
<td>1.397</td>
</tr>
<tr>
<td>Rt. Lat. 265</td>
<td>30.12</td>
<td>29.11</td>
<td>28.27</td>
<td>0.702</td>
<td>6.544</td>
</tr>
<tr>
<td>RAO 225</td>
<td>30.50</td>
<td>33.59</td>
<td>30.19</td>
<td>10.130</td>
<td>1.027</td>
</tr>
<tr>
<td>AP 180</td>
<td>30.16</td>
<td>32.44</td>
<td>30.39</td>
<td>7.560</td>
<td>0.763</td>
</tr>
<tr>
<td>LAO 135</td>
<td>29.69</td>
<td>32.25</td>
<td>28.75</td>
<td>8.622</td>
<td>3.270</td>
</tr>
<tr>
<td>Lt. Lat. 90</td>
<td>29.91</td>
<td>31.06</td>
<td>27.21</td>
<td>3.845</td>
<td>9.923</td>
</tr>
<tr>
<td>LPO 65</td>
<td>29.88</td>
<td>32.60</td>
<td>30.19</td>
<td>9.103</td>
<td>1.037</td>
</tr>
<tr>
<td>Composite</td>
<td>39.01</td>
<td>36.88</td>
<td>no measurement</td>
<td>5.775</td>
<td>N/A</td>
</tr>
</tbody>
</table>

C/600s = $1.67 \times 10^{-14}$ A. No reading was taken with the ionization chamber for the composite field. Units for dose are cGy.

Isodose curves at the central axis (100 source-to-axis distance) for the composite film can be found in Figure 4.1. Absolute dose profiles for horizontal and vertical lines through the composite field are shown in Figure 4.2 and Figure 4.3, respectively.

4.2  Pinnacle³ Treatment Planning

4.2.1 Dose Distribution

Figure 4.4 shows a transverse, sagittal and coronal slice of the BANG-3™ polymer gel developed with the Pinnacle³ planning system and imaged with a CT scanner. Within the gel, the tumor structure (prostate) and normal structures (bladder, rectum and right/left femoral heads) are shown. These structures are located between 98.5 SPD and 102.5 SPD. Also, the isodose distributions around these structures can be seen. The dose distributions shown are the 200 cGy, 190 cGy, 180 cGy, 150 cGy, 105 cGy and 80 cGy isodose lines.

4.2.2 Dose Volume Histogram (DVH)

Figure 4.5 shows a dose-volume histogram for the prostate, bladder, rectum, right femoral head and left femoral head. The volume of the organ irradiated is depicted on the y-
axis, while the dose delivered to that particular volume is shown on the x-axis. Dose values are in cGy. The volume axis is displayed as the normalized volume to the 100% volume mark and the dose values are displayed to absolute dose. Also, the DVH calculation is cumulative throughout the entire tissue volume. From this figure, one can see that nearly 100% of the prostate volume (target volume) receives the total prescribed dose (2 Gy). Also, less than 50% of the femoral heads and rectum receive 50% of the prescribed dose (1 Gy) along with 50% of the bladder receiving approximately 1.2 Gy.

4.3 Leakage and Scatter Irradiation

For the two days in which the gel vessel and vials were left in the treatment room prior to irradiation, the total number of monitor units delivered during the treatments was
Figure 4.2 Horizontal dose profile comparison between the calculated dose from the Pinnacle\(^3\) plan and the measured film dose. The dotted line represents the Pinnacle\(^3\) plan and the solid line refers to the film.

Figure 4.3 Vertical dose profile comparison between the calculated dose from the Pinnacle\(^3\) plan and the measured film dose. The dotted line represents the Pinnacle\(^3\) plan and the solid line refers to the film.
Figure 4.4 This figure shows the tumor structure and critical structures within the BANG-3™ polymer gel including the isodose distributions surrounding these structures. The structures are defined as follows: prostate = red, bladder = yellow, rectum = purple, right femoral head = blue and left femoral head = green. The isodose distributions are defined as follows: 200.0 cGy = yellow, 190.0 cGy = dark purple, 180.0 cGy = blue, 150.0 cGy = orange, 105.0 cGy = green and 80.0 cGy = light purple.

determined to be 23,029, which corresponds to a dose of 23,029 cGy \( \left( \frac{cGy}{MU} \right) \) at \( d_{max} \) or 230Gy. The average maximum allowable leakage as reported by the standards organizations (IEC, NCRP and SSRCR) is 0.1% of the total dose delivered. Multiplying this number by the maximum dose delivered gives the maximum leakage received by the gel, 0.230 Gy (23.0
Figure 4.5 Dose-Volume histogram showing the dose delivered to a particular volume of the irradiated structure. DVH lines are as follows: prostate = red, left femoral head = green, right femoral head = blue, bladder = yellow and rectum = purple. The volume is normalized to the 100% volume mark and the dose values are depicted as absolute dose in cGy.

cGy). The scattered radiation received by the gel was assumed to be 2 times the leakage radiation or 46.0 cGy.

4.4 Shifts and Rotations

The horizontal and vertical shifts to match the pre-irradiation images (scanning session 2) of the gel vessel to the images 5 days post-irradiation (scanning session 3) were –5 pixels and 2 pixels, respectively. These same shifts were –3 pixels and 2 pixels when aligning the pre-irradiation images with the images obtained 2 months post-irradiation (scanning session 4). When analyzing the data between the MRI scanning session and CT scanning session, the horizontal and vertical shifts necessary for alignment were determined to be 7 pixels and -3 pixels for scanning session 3 and 5 pixels and -3 pixels for scanning session 4.
Table 4.3 shows the data obtained when determining the proper rotation angle for the CT data compared with the MRI images taken post-irradiation. Two points, point 1 and point 2 were analyzed initially for the MRI images from scanning session 3 (MRIa) and scanning session 4 (MRIb) and then compared to the images from the CT scan. These points were chosen at the fiducial marks. From the table, it is evident that the MRIb images were more aligned with the CT scan images than the MRIa images. The MRIb images were declared aligned with the CT images as the deviation in angle corresponded to a deviation of three-quarters of a pixel at the periphery of the dose matrix. Due to this fact, no rotation was performed on MRIb images. Because of the larger deviation in the MRIa images compared to the CT images in the first two points (deviations of more than 4 pixels at periphery of dose matrix), a third point was added and analyzed. By investigation, it was determined that the 2.4° rotation was the optimal angle when compared to all three fiducial points.

4.5 Calibration Curves/Dose Maps

In section 4.1.2, the dose rate at $d_{\text{max}}$ was computed to be 1.020 $\frac{cGy}{MU}$, which was 2.0% higher than the expected dose rate of 1.0 $\frac{cGy}{MU}$. This increase in expected dose rate changes the total absorbed dose delivered to the calibration vials. Remember, a particular number of monitor units was delivered to each vial to irradiate the vials to 1Gy, 4Gy, 7Gy and 9Gy. The corresponding increase in dose being delivered to the calibration vials was 1.02 Gy, 4.08 Gy, 7.14 Gy and 9.18 Gy. The axial slice in which the maximum $R_2$ value was observed was as follows: vials 2,3—slice3, vial4—slice2, vials 5,6 and 9—slice 7, vial 7 and 8—slice8.
Table 4.3 This table shows the angle of rotation determined to align the CT images with the MRI images taken post-irradiation. MRI data was compared with CT data (Diff\(_{CT}\)). MRIa refers to 5 days post-irradiation and MRIb refers to 2 months post-irradiation. Point 1 refers to the fiducial mark in MRI slice 40 (CT slice 23), while point 2 refers to the fiducial mark in MRI slice 47 (CT slice 19). Point 3 was only used in image MRIa.

<table>
<thead>
<tr>
<th>Image</th>
<th>Point1</th>
<th>Point2</th>
<th>Point3</th>
</tr>
</thead>
<tbody>
<tr>
<td>MRIa</td>
<td>6.254°</td>
<td>12.897°; Diff(_{CT})=2.034°</td>
<td>8.427°; Diff(_{CT})=2.453°</td>
</tr>
<tr>
<td>MRIb</td>
<td>8.239°; Diff(_{CT})=0.341°</td>
<td>14.982°; Diff(_{CT})=0.051°</td>
<td>N/A</td>
</tr>
<tr>
<td>CT scan</td>
<td>7.898°</td>
<td>14.931°</td>
<td>5.974°</td>
</tr>
</tbody>
</table>

R\(_2\) values and their standard deviations for vials 1-9 in scanning session 1 were 4.632 ± 0.357, 4.374 ± 0.214, 4.678 ± 0.322, 4.889 ± 0.273, 4.187 ± 0.225, 4.406 ± 0.317, 4.777 ± 0.360, 4.115 ± 0.257, and 4.893 ± 0.246. The average value for these nine vials were 4.550 ± 0.286. The R\(_2\) value and uncertainty obtained from vial 1 (unirradiated vial) in scanning session 3 was 5.947 ± 0.384. The ratio of these numbers (\(\frac{4.550}{5.947}\)) and (\(\frac{0.286}{0.384}\)) was 0.765 and 0.744, respectively. This data shows a considerable contamination occurring within the gel between scanning session 1 and scanning session 3. In order to account for this discrepancy, the values determined from IDL in scanning sessions 3 and 4 were multiplied by these ratios with this corrected data being used in the calibration curve.

The \(R_2\)–dose calibration curves for the 4 and 10 MV irradiation beams are shown in Figures 4.6 and 4.7, respectively. The 10 MV calibration curves were used to obtain dose rate (\(R_2\)) maps from the polymer gel. The unit for dose is Gy, while the unit for the relaxation is inverse seconds (\(\frac{1}{s}\)). Shown in the graphs are the slope, y-intercept and chi-squared (R\(^2\)) values for both beams. The graphs are linear up to at least 7.14 Gy (R\(^2\) = 0.989) for the 4 MV photons and 9.18 Gy (R\(^2\) = 0.994) for the 10 MV photons. The y-intercept values were similar in both graphs with an intercept of 4.497 for 4 MV photons and 4.695 for the 10 MV
Figure 4.6 4 MV-\( R_2 \) dose calibration curve obtained from the calibration vials. Units for dose are in Gy and units for relaxation rate (\( R_2 \)) are inverse seconds (1/s). Three vials were irradiated to 1.02 Gy, 4.08 Gy and 7.14 Gy, while one vial was left unirradiated for a background measurement. Also shown is the linear fit to the line and the \( R^2 \) value.

Figure 4.7 10 MV-\( R_2 \) dose calibration curve obtained from the calibration vials. Units for dose are in Gy and units for relaxation rate (\( R_2 \)) are inverse seconds (1/s). Four vials were irradiated to 1.02 Gy, 4.08 Gy, 7.14 Gy and 9.18 Gy, while one vial was left unirradiated for a background measurement. Also shown is the linear fit to the line and the \( R^2 \) value.
photons; however, the slopes were somewhat different (0.737 for 4 MV, 0.648 for 10 MV).

The uncertainties were very small with the largest uncertainty for the 4 MV (± 0.335) and 10 MV (± 0.502) beam occurring at a dose of 0 Gy. For both beams, the uncertainties decrease as the dose delivered increases. With image subtraction correction, the values for slope were changed for both sets of data with an intercept of 1.539 for the 4 MV photon beam and 1.305 for the 10 MV beam. $R_2$ value for 10 MV photon beam at a dose rate of 400 $\frac{MU}{min}$ was 6.963 ± 0.553, while this same value for a dose rate of 500 $\frac{MU}{min}$ was 6.738 ± 0.406, which corresponds to a difference of (6.963-6.738)/6.738 = 3.339%. The relaxation rate ($R_2$), relaxation rate error ($\sigma_{R_2}$) and fitting parameters ($\alpha$, $\sigma_{\alpha}$, $R_o$ and $\sigma_{R_o}$) for the 10 MV photon beam can be found in Table 4.4.

Average $R_2$ values for an unirradiated portion of the gel in the CGEL-S vessel (scanning sessions 1-4, position x =1-256, y=127 and z=65) were determined to be 4.600 s$^{-1}$ for scanning session 1 and approximately 4.800 s$^{-1}$ for scanning session 2-4. Upon comparing the average $R_2$ from the calibration vials from scanning session 1 and the unirradiated vial from scanning session 3 to CGEL-S vessel values, it was determined that the vessel values were approximately similar to the vials in scanning session 1 ($\frac{4.550}{4.600} = 0.989$ and $\frac{4.550}{4.800} = 0.948$) but considerably different from the vials in scanning session 3 ($\frac{5.947}{4.600} = 1.293$ and $\frac{5.947}{4.800} = 1.239$). There appeared to be an overall non-uniformity (waviness) in the $R_2$ values of the gel vessel post-irradiation and an artifact in proton density within the vials (Figure 4.8).
Table 4.4 Average relaxation rate ($R_2$) and error ($\sigma_{R_2}$) and fitting parameters for the 10 MV photon irradiation. Relaxation rare and error were multiplied to the dose correction factor.

<table>
<thead>
<tr>
<th>Dose delivered</th>
<th>0 Gy</th>
<th>1 Gy</th>
<th>4 Gy</th>
<th>7 Gy</th>
<th>9 Gy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Relaxation rate ($R_2$) and error ($\sigma_{R_2}$)</td>
<td>$R_2$</td>
<td>$\sigma_{R_2}$</td>
<td>$R_2$</td>
<td>$\sigma_{R_2}$</td>
<td>$R_2$</td>
</tr>
<tr>
<td>Measured values</td>
<td>4.568</td>
<td>0.502</td>
<td>5.328</td>
<td>0.411</td>
<td>7.524</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Fitting parameters</th>
<th>value</th>
<th>units</th>
</tr>
</thead>
<tbody>
<tr>
<td>slope $\alpha$</td>
<td>0.648</td>
<td>s$^{-1}$/Gy</td>
</tr>
<tr>
<td>slope error $\sigma_\alpha$</td>
<td>0.424</td>
<td>s$^{-1}$/Gy</td>
</tr>
<tr>
<td>intercet $R_0$</td>
<td>4.695</td>
<td>s$^{-1}$</td>
</tr>
<tr>
<td>intercept error $\sigma_{R_0}$</td>
<td>0.061</td>
<td>s$^{-1}$</td>
</tr>
</tbody>
</table>

Figure 4.8 T$_2$-weighted (left) and proton density (right) images obtained from MRI. Vial 1 is located on the upper left corner of the image and vial nine is located on the lower right corner. Note the artifacts inherent within the vials of the proton density images.

4.6 Comparisons—Image Subtraction, Contour Comparisons and Dose-Volume Analysis

4.6.1 Image Subtraction

Figures 4.9 and 4.10 show the dose deviations between the Pinnacle$^3$ (ADAC) data and MRI data 5 days post-irradiation and 2 months post-irradiation, respectively, for slices 4
Figure 4.9  Figure showing the global deviation in dose values between the ADAC dose and MRI dose (5 days post-irradiation). Data takes into account dose correction factor. Values are shown for MRI slices 4-50. Differences are computed by subtracting the MRI dose values from the ADAC dose values. Error bars are present.

Figure 4.10  Figure showing the global deviation in dose values between the ADAC dose and MRI dose (2 months post-irradiation). Data takes into account dose correction factor. Values are shown for slices 4-50. Differences are computed by subtracting the MRI dose values from the ADAC dose values. Error bars are present.

through 50 (95.0 SPD to 104.2 SPD). Data take into account the 0.765 dose correction factor (Chapter 4.5). Negative values correspond to a larger MRI dose when compared to the
ADAC dose. In Figure 4.9, dose differences range from –57.0389 cGy (slice 40) to 17.102 cGy (slice 19). For slices 15-37, the dose values from ADAC and MRI show little deviation (0 difference) when the uncertainty was taken into account. Uncertainties range from 20.767 cGy (slice 49) to 35.4813 cGy (slice 18).

In Figure 4.10, larger differences were observed between the ADAC data and MRI data, ranging from –168.983 cGy (slice 40) to –67.380 cGy (slice 5). Larger uncertainties were also witnessed for this scanning session, with values ranging from 65.730 cGy (slice 30) to 83.647 cGy (slice 7). In this figure, MRI images and ADAC images do not agree (difference not equal to zero) even when the uncertainty was considered.

Figure 4.11 and Figure 4.12 also show deviations between ADAC dose and MRI dose for slices 4-50 for MRI images taken both 5 days and 2 months post-irradiation; however these figures do not take into account the dose correction factor. In Figure 4.11, average dose differences range from –83.111 cGy (slice 40) to 5.460 cGy (slice 5) and their deviations range from 28.555 cGy (slice 14) to 47.751 cGy (slice 4). In Figure 4.12, average dose deviations range from –199.434 (slice 8) to –88.803 (slice 5), while average uncertainties range from 85.167 (slice 29) to 109.363 (slice 7). This set of data show higher MRI readings and a larger deviation between the ADAC data and CT data than data that take into account the correction factor. Fewer slices show an agreement (0 difference) between the ADAC and MRI data even when uncertainty is taken into account. Again, MRI data obtained 5 days post-irradiation was more similar to the CT data than the MRI data obtained 2 months post-irradiation.

Figures 4.13-4.17 show the dose deviations between ADAC and MRI for each slice within the 5 organs investigated in this report. Values are shown both with and without the
Deviation between ADAC data and MRI data: 5 days post-irradiation w/o correction

Figure 4.11 Figure showing the global deviation in dose values between the ADAC dose and MRI dose (5 days post-irradiation). Data does not take into account dose correction factor. Values are shown for MRI slices 4-50. Differences are computed by subtracting the MRI dose values from the ADAC dose values. Error bars are present.

Deviation between ADAC data and MRI data: 2 months post-irradiation w/o correction

Figure 4.12 Figure showing the global deviation in dose values between the ADAC dose and MRI dose (2 months post-irradiation). Data does not take into account dose correction factor. Values are shown for slices 4-50. Differences are computed by subtracting the MRI dose values from the ADAC dose values. Error bars are present.
Dose deviation between ADAC and MRI for prostate region

Figure 4.13 Figure shows the deviation between the calculated ADAC dose and measured MRI/polymer gel dose for the prostate. Plotted on the x-axis is the source-to-plane distance (SPD) ranging from –1.2 cm (98.8 SPD) to 0.8 cm (100.8 SPD). Lower SPDs correspond to inferior portions of the gel. Plotted on the y-axis is the dose values obtained from the MRI data subtracted from dose values obtained from the ADAC data. Units are in cGy. The upper pink line contains values that take into account the dose correction factor while the lower blue line does not consider this factor.

Dose deviation between ADAC and MRI for bladder region

Figure 4.14 Figure shows the deviation between the calculated ADAC dose and measured MRI/polymer gel dose for the bladder. Plotted on the x-axis is the source-to-plane distance (SPD) ranging from –1.2 cm (98.8 SPD) to 0.8 cm (100.8 SPD). Lower SPDs correspond to inferior portions of the gel. Plotted on the y-axis is the dose values obtained from the MRI data subtracted from dose values obtained from the ADAC data. Units are in cGy. The upper pink line contains values that take into account the dose correction factor while the lower blue line does not consider this factor.
Figure 4.15  Figure shows the deviation between the calculated ADAC dose and measured MRI/polymer gel dose for the lt. femoral head. Plotted on the x-axis is the source-to-plane distance (SPD) ranging from –0.6 cm (99.4 SPD) to 0.8 cm (100.8 SPD). Lower SPDs correspond to inferior portions of the gel. Plotted on the y-axis is the dose values obtained from the MRI data subtracted from dose values obtained from the ADAC data. Units are in cGy. The upper pink line contains values that take into account the dose correction factor while the lower blue line does not consider this factor.

Figure 4.16  Figure shows the deviation between the calculated ADAC dose and measured MRI/polymer gel dose for the rt. femoral head. Plotted on the x-axis is the source-to-plane distance (SPD) ranging from –0.6 cm (99.4 SPD) to 0.8 cm (100.8 SPD). Lower SPDs correspond to inferior portions of the gel. Plotted on the y-axis is the dose values obtained from the MRI data subtracted from dose values obtained from the ADAC data. Units are in cGy. The upper pink line contains values that take into account the dose correction factor while the lower blue line does not consider this factor.
Figure 4.17  Figure shows the deviation between the calculated ADAC dose and measured MRI/polymer gel dose for the rectum. Plotted on the x-axis is the source-to-plane distance (SPD) ranging from –1.6 cm (98.4 SPD) to 1.4 cm (101.4 SPD). Lower SPDs correspond to inferior portions of the gel. Plotted on the y-axis is the dose values obtained from the MRI data subtracted from dose values obtained from the ADAC data. Units are in cGy. The upper pink line contains values that take into account the dose correction factor while the lower blue line does not consider this factor.

dose correction factor. Units are in cGy. From the figures, it is evident that the measured and planned dose values were in closer agreement with one another when the correction factor was taken into account; only data that consider the correction factor is discussed here. Aside from the prostate, the measured MRI data was higher in value than the planned ADAC data. Within the prostate (center of the gel), the correction factor was shown to overcompensate for the excess dose delivered to the gel (ADAC dose was larger than MRI dose). The rectum shows the greatest agreement in dose values, as the difference in dose was nearly zero throughout the entire volume. The bladder and rt. femoral head also show good agreement between the dose values as some regions are within 10 cGy of one another. In the case of the lt. femoral head, MRI values were approximately 30 cGy higher than the ADAC values.
Dose within each organ was also analyzed for MRI images taken two months post-irradiation then compared to the ADAC planned data. Remember, negative numbers correspond to higher MRI readings. Dose deviation ranges for each organ were as follows: prostate— -91.0630 cGy to -73.5357 cGy, bladder— -134.951 cGy to -102.470 cGy, lt. femoral head— -161.475 cGy to –143.759, rt. femoral head— -157.901 cGy to –130.959 cGy and rectum— -115.766 cGy to –77.5087 cGy. Again, a better agreement was seen between the measured MRI/polymer gel dose and planned ADAC dose whenever the dose correction factor was considered.

### 4.6.2 Contour Analysis

Figure 4.18 shows a color contour depicted for ADAC dose distributions planned for an IMRT treatment. The colors represent different dose distributions throughout the gel (white: 210-220 cGy, gray: 200-210 cGy, red: 190-200 cGy, reddish blue: 180-190 cGy, blue: 150-180 cGy, light green: 105-150 cGy, dark green: less than 105 cGy). The anterior portion of the gel is located on top of the image. Images were taken at the central axis (z=0 plane, 100 SPD).

Figures 4.19-4.21 illustrate color contours obtained from the measured MRI dose distributions via IDL. Contours are shown for two different times post-irradiation with/without the 0.765 dose correction factor. No graph was generated for the MRI data 2 months post-irradiation without the correction factor as all MRI dose distributions for this graph was located on the periphery of the gel. The anterior portion of these images was also located on top of the image. Images were taken at the central axis (z=0 plane, 100 SPD). Overall, there was no complete agreement between the ADAC planned dose distributions and the MRI measured dose distributions. For the data obtained 5 days post-irradiation, dose
Figure 4.18 Figure shows ADAC dose contours within the BANG-3™ polymer gel for a 7-field prostate IMRT treatment. Colors are shown to represent the dose distribution throughout the gel (white: 210-220 cGy, gray: 200-210 cGy, red: 190-200 cGy, reddish-blue: 180-190 cGy, blue: 150-180 cGy, light green: 105-150 cGy, dark green: less than 105 cGy).

Figure 4.19 Figure shows MRI dose contours 5 days post-irradiation with 0.765 correction factor within the BANG-3™ polymer gel for a 7-field prostate IMRT treatment. Colors are shown to represent the dose distribution throughout the gel (white: 210-220 cGy, gray: 200-210 cGy, red: 190-200 cGy, reddish-blue: 180-190 cGy, blue: 150-180 cGy, light green: 105-150 cGy, dark green: less than 105 cGy).
Figure 4.20 Figure shows MRI dose contours 5 days post-irradiation without the correction factor within the BANG-3™ polymer gel for a 7-field prostate IMRT treatment. Colors are shown to represent the dose distribution throughout the gel (white: 210-220 cGy, gray: 200-210 cGy, red: 190-200 cGy, reddish-blue: 180-190 cGy, blue: 150-180 cGy, light green: 105-150 cGy, dark green: less than 105 cGy).

Figure 4.21 Figure shows MRI dose contours 2 months post-irradiation with 0.765 correction factor within the BANG-3™ polymer gel for a 7-field prostate IMRT treatment. Colors are shown to represent the dose distribution throughout the gel (white: 210-220 cGy, gray: 200-210 cGy, red: 190-200 cGy, reddish-blue: 180-190 cGy, blue: 150-180 cGy, light green: 105-150 cGy, dark green: less than 105 cGy).
distributions at the periphery of the gel appear to agree with the ADAC data when the dose correction factor was used. Without this factor, dose distributions near the center of the gel appear to be more identical. Also, dose distributions obtained 5 days post-irradiation were more analogous to the ADAC distributions than were the images obtained 2 months post-irradiation.

Figures 4.22-4.24 show isodose distributions for MRI data and CT data along with the organs analyzed in the study. MRI data is shown for images 5 days and 2 months post-irradiation, along with correction factors. The age correction factor estimates additional polymerization between imaging sessions and was (Chapter 3) determined for scanning session 4 to be 1.735; the effects of this factor can be seen in figure 4.24. These images show that highest dose delivered was to the prostate region. Without the correction factor, for both the planned ADAC data and measured MRI data, the 180, 190 and 200 cGy contour lines closely follow the contour of the prostate. Also, the measured and planned data show that parts of the bladder and rectum receive as much dose as 200 cGy. Lower dose distributions extend out to the femoral heads (150 cGy) and as far as the periphery of the gel phantom (100 cGy). Data show that the MRI dose contours extend more towards the periphery of the gel than the ADAC dose contours. High measured dose regions (200 cGy) can be seen in the gel images obtained 5 days post-irradiation. Finally, when the dose correction factor was used, lower measured doses can be seen at the center of the gel (larger deviation with planned data) while measured and planned data towards the edge of the gel appear to be more similar.

4.6.3 Dose-Volume Analysis

The volume for the tumor volume (prostate) and critical structures (lt. femoral, rt. femoral, bladder and rectum) computed from IDL are 46.266, 15.396, 11.466, 58.202, and
Figure 4.22 shows contours of the dose distributions along with the organs analyzed in the study. The contours are of ADAC plan (dashed line) and MRI analysis (solid line) 5 days post-irradiation. The dose correction factor is used in the MRI contours. Images are taken at 100 SPD (z=0). Organ color distributions are as follows: prostate-maroon, rt. femoral head-light blue, lt. femoral head-green, bladder-brownish orange and rectum-purple. Contour color distributions for dose are as follows: 200 cGy-red, 190 cGy-light blue, 180 cGy-white, 150 cGy-green and 100 cGy-purple.

36.231, respectively. The corresponding volumes from the ADAC plan were 45.549, 15.401, 11.403, 57.752 and 36.241. Units are in cm$^3$. Percent difference in these values are 1.574%, 0.032%, 0.552%, 0.779% and 0.028%. Figures 4.25 and 4.26 represent the ratios of the total volume per plane receiving a particular dose of interest (100 cGy, 150 cGy, 180 cGy, 190 cGy and 200 cGy) measured by MRI/polymer gel relative to the values obtained by ADAC. Dose is represented by colors, which are blue, pink, yellow, light blue and purple, respectively, for the regions mentioned above. The inferior portion of the gel is represented with negative dose regions) and under compensates for this excess dose near the periphery of the gel (ratio
4.23 Figure shows contours of the dose distributions along with the organs analyzed in the study. The contours are of ADAC plan (dashed line) and MRI analysis (solid line) 5 days post-irradiation. No dose correction factor is used in the MRI contours. Images are taken at 100 SPD (z=0). Organ color distributions are as follows: prostate-maroon, rt. femoral head-light blue, lt. femoral head-green, bladder-brownish orange and rectum-purple. Contour color distributions for dose are as follows: 200 cGy-red, 190 cGy-light blue, 180 cGy-white, 150 cGy-green and 100 cGy-purple.

was greater than 2 in lower dose regions). ADAC and MRI dose appear to be in agreement within the 150 cGy contour level as the ratio of MRI to ADAC was approximately one throughout the gel. In some regions within the 190 cGy contour level, some agreement was seen between the measured MRI data and calculated ADAC data, but more time than not, MRI values were smaller than the ADAC values. When the dose correction factor was not considered, MRI measurements were 1.5-3 times larger than the CT measurements for all dose regions throughout the gel.
Figure 4.24  Figure shows contours of the dose distributions along with the organ analyzed in the study. The contours are of ADAC plan (dashed line) and MRI analysis (solid line) 2 months post-irradiation. An age correction factor is used in the MRI contours to account for the increased polymerization of the gel with time. Images are taken at 100 SPD (z=0). Organ color distributions are as follows: prostate-maroon, rt. femoral head- light blue, lt. femoral head-green, bladder-brownish orange and rectum-purple. Contour color distributions for dose are as follows: 200 cGy-red, 190 cGy-light blue, 180 cGy-white, 150 cGy-green and 100 cGy-purple.

Figure 4.27 shows the maximum dose values measured per plane with the MRI/polymer gel measurements, along with the maximum dose per plane calculated by ADAC. Dose values are in cGy. Again, the inferior portion of the gel is represented by negative numbers (lower SPDs). MRI data is shown with and without the dose correction factor. When the correction factor was considered, planes near the central slices (-2.0 to 1.2) tend to be similar for both MRI and ADAC, with the MRI readings marginally larger than the ADAC readings. At x=0 (100 SPD), the MRI reading was determined to be 238.377, while the ADAC value was determined to be 219.943. This difference corresponds to a percent
Figure 4.25 Ratio of total dose per volume (cm\(^3\)) measured by MRI to total dose per volume (cm\(^3\)) calculated by ADAC per slice of data (MRI/CT) with dose correction factor. Data range from 98.6 SPD (-1.4) to 101.2 SPD (1.2). Colored lines represent contour regions (blue = 100cGy, pink = 150 cGy, yellow = 180 cGy, light blue = 190 cGy, purple = 200 cGy).

Figure 4.26 Ratio of MRI total dose per volume (cm\(^3\)) to ADAC total dose per volume (cm\(^3\)) per slice of data (MRI/CT) without dose correction factor. Data range from 98.6 SPD (-1.4) to 101.2 SPD (1.2). Colored lines represent contour regions (blue = 100cGy, pink = 150 cGy, yellow = 180 cGy, light blue = 190 cGy, purple = 200 cGy).

difference of 8.381%. Away from these planes, MRI readings were consistently higher than ADAC readings as ADAC measurements dropped to zero and MRI measurements dropped to
as low as 130.910 cGy. With no dose correction factor, the MRI dose and ADAC dose were not similar. At x=0, the measured MRI dose was 311.604, which corresponds to a percent difference of 41.675%.

Figures 4.28-4.30 show dose volume histograms (DVH) from the ADAC data and MRI data. The MRI results are shown both with and without the dose correction factor. The normalized volume, normalized to the total volume of each organ is plotted as a function of the absorbed dose received by the polymer gel. All four organs-at-risk (lt. femoral head, rt. femoral head, bladder and rectum) were analyzed along with the tumor volume (prostate). The ADAC data show that less than 50% of the volume of the femoral heads and bladder received a dose higher than 100 cGy, while 50% of the rectum volume received around 120 cGy. No portion of the femoral heads and only 20% of the rectum and bladder received a dose greater than 150 cGy. Also, 100% of the prostate volume received at least 190 cGy with around 75% receiving at least 200 cGy.
Figure 4.28 ADAC dose volume histogram analyzed with IDL software. Volume is normalized to the maximum volume within the organ. The normalized volume is plotted as a function of the dose delivered to the polymer gel. Colors represent the organs analyzed and are as follows: dark blue diamond—rectum, pink square—lt. femoral head, yellow triangle—rt. femoral head, light blue x—prostate and purple *—bladder.

Figure 4.29 MRI dose volume histogram with dose correction factor analyzed with IDL software. Volume is normalized to the maximum volume within the organ. The normalized volume is plotted as a function of the dose delivered to the polymer gel. Colors represent the organs analyzed and are as follows: dark blue diamond—rectum, pink square—lt. femoral head, yellow triangle—rt. femoral head, light blue x—prostate and purple *—bladder.
Figure 4.30 MRI dose volume histogram without dose correction factor analyzed with IDL software. Volume is normalized to the maximum volume within the organ. The normalized volume is plotted as a function of the dose delivered to the polymer gel. Colors represent the organs analyzed and are as follows: dark blue diamond—rectum, pink square—lt. femoral head, yellow triangle—rt. femoral head, light blue x—prostate and purple *—bladder.

Taking the dose correction factor into consideration, the DVH for the MRI data were as follows: 100% of the prostate volume received 150 cGy with only 10% receiving 200 cGy, 50% of the bladder, rectum and rt. femoral head received a dose of around 125 cGy with the value slightly higher for the lt. femoral head (135 cGy) and finally, less than 20% of these organs received a dose higher than 150 cGy. In the absence of the dose correction factor, the measured reading for every organ was higher than in the case with the correction factor. Accordingly, 95% of the prostate received at least 190 cGy, over 50% of every OAR received a dose of greater than 150 cGy and with the exception of the rt. femoral head and 10-20% of the OAR received a dose of at least 200 cGy.
Chapter 5

Discussion

5.1  Pinnacle\(^3\) Planning System

For any cancer center focused on performing a novel 3D-oriented radiotherapy modality, such as IMRT, a planning system capable of planning in 3-D with good spatial accuracy is a necessity. The planning system used at MBPCC is the Pinnacle\(^3\) planning system. Images are transferred to the Pinnacle\(^3\) workstation upon receiving the scans from CT. The Pinnacle\(^3\) planning system was very useful in this study as it provided the spatial location of the OAR and PTV, along with the dose distributions surrounding these regions and a dose volume histogram used to analyze the dose delivered to a particular organ of interest. At MBPCC, doctor’s requests are that the PTV receives at least 95% of the maximum dose and that the volume of the OAR receiving a dose above 50% of the maximum dose is kept to a minimum for prostate treatment planning. The Pinnacle\(^3\) plan (Figure 4.4) demonstrated the optimized plan for a tumor located within the prostate (200 cGy encompassing most of the prostate, 190 cGy encompassing the entire prostate (PTV) and 105 cGy encompassing approximately half of the volume of the OAR). Also, the dose volume histogram (Figure 4.5) reconfirmed this requested dose distribution; close to 100 % of the prostate volume received the maximum dose (prescription dose) of 2 cGy while the dose to the OAR were kept to a minimum (approximately 50 % of the volume of the OAR received nearly half (100 cGy) of the total dose).

5.2  Gel Irradiation

The BANG-\(^3\)\(^{TM}\) polymer gel was used in this study for comparison with the Pinnacle\(^3\) treatment planning system. In particular, focus of the study was on the response of the gel to
an IMRT treatment that was planned by the Pinnacle³ system. The gel was purchased from MGS Research, Inc. within a BAREX CGEL-S vessel and borosilicate glass vials. Nine vials were used for gel calibration and one vessel was used to obtain the images from irradiation, administered with the Varian Clinac 21EX.

5.3 Dose Verification

In performing this experiment, it was imperative to obtain an accurate $R_2$-dose calibration curve as this curve determined what dose was delivered to the main CGEL-S vessel. As the $R_2$ value was computed from the MRI/IDL readings, the proper dose delivered to the calibration vials should be accounted for. In setting up the experiment for the calibration procedure, it was observed that the bottom of the cubic water tank contained 1 cm of acrylic. It was concluded that an acrylic correction factor should be determined and applied to the dose calculation at $d_{\text{max}}$, the depth at which the dose to the calibration gel was computed. A correction factor of 0.996 was determined and, because the composition of acrylic and water are very similar, this factor is what was expected, a number near unity. The dose rate at $d_{\text{max}}$ is calibrated to be $1.0 \frac{cGy}{MU}$; however, this number should be tested for validity. Through calculations, the dose rate was determined to be $1.020 \frac{cGy}{MU}$, giving the dose at $d_{\text{max}}$ readings 2% higher than expected. Without the proper attention to these factors (i.e., acrylic correction factor and dose rate at $d_{\text{max}}$, the dose delivered to the calibration vials would have been off by as much as 2.5%, which would affect the calibration curve.

Additional dose measurements were acquired with XV film and an ionization chamber to verify the results obtained with the BANG-3 TM polymer gel. Both measurements were
compared to the Pinnacle$^3$ planning system to ensure that the radiation delivered agreed with the IMRT plan. The film-based method is the method currently used at MBPCC for IMRT QA. As can be seen from Chapter 4.1.3, for the individual fields, for the film-based measurements, five of the seven beams (RPO 295, RAO 225, AP 180, LAO 135 and LPO 65) did not meet the standard of 5% as stated by ICRU (ICRU 1976). The composite film also did not meet this 5% criteria. However, when film uncertainty (~2-3%) and misalignment issues (i.e., crooked film) were considered, the discrepancies were close enough to justify an agreement between the plan and the dose delivered. Upon observing the ionization chamber measurements (Chapter 4.1.3), smaller discrepancies were observed when compared to the Pinnacle$^3$ dose. The ionization chamber dose measurements were in close proximity to the values determined from the Pinnacle$^3$ system. These chamber measurements confirm the film measurements and it was concluded that the dose distribution delivered was similar to that planned by the Pinnacle$^3$ system.

In addition to the ionization chamber and film measurements, isodose curves and dose profiles were used to verify the similarities between the Pinnacle$^3$ plan and the irradiated region. Isodose curves at the central axis (100 source-to-axis distance) for the composite film clearly show the dose distributions for doses delivered to this axis (Figure 4.1). Curves show the expected highest dose region around center of the field (37 cGy) and lower dose regions toward the edges (20 cGy). The horizontal profile (Figure 4.2) shows a somewhat better agreement with the Pinnacle$^3$ plan than the vertical profile (Figure 4.3) and both profiles illustrate a higher Pinnacle$^3$ dose than film dose.
5.4 Leakage and Scatter Radiation

An improper step was taken during experimentation that potentially caused an error in the dose measurements. This step involved the placement of the gel (vials and vessel) in the treatment room two days prior to irradiation as suggested in the “handling of BANG™ gels” by MGS Research, Inc. (MGS website). As the temperature throughout MBPCC is approximately constant throughout the center, there was no need to place the gel within the treatment room. In the two days prior to irradiation, a dose of 23.0 cGy was measured as the leakage radiation and 46.0 cGy was approximated as the scatter radiation (Chapter 4.3). As stated in Table 3.2, the dose response is independent of photon or electron energy, thus any dose received by the gel, the response will be the same regardless of the photon energy. Therefore, we believe that the excess radiation did affect the gel. As a matter of fact, contamination was observed within the gel between scanning session 1 and scanning session 3. Unfortunately, there is no way to examine the impact of this radiation throughout the gel. One option would be to place the gel in the treatment room on Friday and perform the treatment on Sunday. As treatments are normally not performed on weekends at MBPCC, no excess leakage or scatter radiation would be received by the gel in this time.

5.5 Imaging

MRI was used to scan the polymer gel on four separate occasions (Chapter 3.5). The first scan was performed prior to the CT scan and was used for acquiring background measurements of the gel and for familiarity with the scanning system (i.e., acquisition set-up, scanning time and gel positioning). A second scan was performed after CT scanning to test for any contamination within the gel and was ultimately used for background subtraction.
within the CGEL-S vessel. Two additional scans were taken, five days and two months post-irradiation, to observe the radiation and aging effects on the gel.

An artifact was observed in the vial proton density images, along with a ring artifact (non-uniformity) within gel (Figure 4.8). The vial artifact was considered to be a result of the MRI scanner because it was in same position for each scan, while the vial positions were not. It is beyond the scope of this paper to determine what caused the artifact and the possible effects of the artifact on the MRI data. The ring artifacts within the gel vessel were assumed to be due to MRI equipment, which can be caused by failure or poor adjustment of equipment. The linearity of the system and thus the center of reconstructions are defined via gradients rather than being defined geometrically, therefore misalignment may cause ring artifacts close to the center. Also, ringing is observed to exist near high contrast interfaces in the image (Webb 1988).

5.6 Analysis with IDL

A major setback inherent in this study was the problem of transferring the images from the Pinnacle³ system and MRI to the same computer for analysis as the IDL software was on an on-campus computer at LSU. No files that were in Pinnacle³ format could be transferred to the campus computer because this computer had no Pinnacle³ software. For the MRI files, the campus computer was not able to read optical disks and the readings of the DAT tapes would take weeks to complete. Eventually, all files were sent to the CT scanner, whereby they were then sent over to the Pinnacle³ workstation and ftp’d to LSU for analysis.

To provide accurate analysis, shifts and rotations of the images were required to align the MRI and CT data. Shifts were performed on pre-irradiation MRI images (i.e., scanning session 2) relative to post-irradiation MRI images (i.e., scanning sessions 3 and 4) and on
post-irradiation MRI images relative to CT images. Rotations were only performed on the post-irradiation MRI images relative to the CT images. The gel ring artifact (chapter 5.5) could cause problems when rotating MRI scans to match one another. The problem would occur when performing background subtraction, as the artifacts would no longer be aligned. These rotations were not done and should not be performed in future studies if artifacts are present. Chapter 4.4 summarizes the shifts and rotations performed to align the images. An inconsistency in using the “rot function” to rotate the vials (e.g., rotated images were still misaligned) was observed. Due to this discrepancy and the artifact noted in Chapter 5.5, rotations of the vials were disregarded. Also, no shifts were performed on the vial images because of the random positioning of each vial in the MRI scans. In each scanning session, the vials were only taped together and placed in a curved apparatus therefore similar positioning of the vials between scans is not guaranteed. For future studies, it is suggested that the vials be placed in an apparatus that keeps the vial positions constant from scan to scan.

Ideally, it would be preferred that no shifting or rotations were necessary due to a possible error in the program or the inability to accurately determine proper rotation angles or shifts (e.g., used get xy function—uncertainty in correct pixel chosen). One suggestion would be to build a container to house the phantom and vials for scanning and irradiations. This device would be manufactured to firmly house the phantom and vials and thereby guarantee no shifting of the gel. A second suggestion would be to use a gel analysis service (e.g., Stony Brook), whereby all technical aspects of imaging and alignment would be left up to the professional analysis company. Price ranges would depend on the accuracy and resolution desired because longer necessary scanning times are necessary. An accuracy of 1-2% and
resolution of 1 mm is possible (personal communication). Stony Brook also will soon be offering their DoseQA 3-D software, possibly free of charge that is compatible with DICOM-RT (Stony Brook website).

Finally, through analysis, some uncertainty existed as to the orientation of the gel in the Pinnacle³ system (i.e., superior/inferior end). This presents a problem when comparing the Pinnacle³ data to the MRI data, as larger deviations would result if the wrong ends are compared. From this observation, it should be noted that there are optimum locations with which to place the fiducial markers. For example, at least one marker should be placed at center of gel while the other four markers should be positioned in way that they do not occupy the same slice; two of these markers could be placed at the superior/inferior gel borders and the other two could be placed on the sides of the vessel.

5.7 Calibration Curve

Calibration curves were drawn for 4 and 10 MV photon beams (Figures 4.6 and 4.7, respectively). Data from the curve fits show the similarities between the two curves as their chi-squared and y-intercept values are very similar (Chapter 4.5). The gel is expected to be linear to at least 10 Gy (maximum dose delivered to gel as requested to gel manufacturer). This data show the energy independence of the BANG-3™ gel. Dose rate independence is also demonstrated (3.339 % R² difference shown for gel irradiated to the same energy and dose). However, this dose rate independence is not conclusive, as more than one measurement should be evaluated. From here, it is concluded that the same calibration curve can be used for gels irradiated to the 4 MV or 10 MV photon beam. In this study, an energy of 10 MV is analyzed along with a dose rate of 400 $\frac{MU}{min}$. The uncertainties are very small with the largest uncertainty for the 4 MV and 10 MV beam occurring at a dose of 0 Gy.
These results are expected as the largest % error is expected to occur at lower doses (Oldham 1998). Values for background subtraction measurements are shown in Chapter 4.5. These values were used to compute the dose within the CGEL-S vessel.

5.8 Dose Analysis

To perform dose analysis, it is imperative that all data (i.e., MRI data, Pinnacle\(^3\) dose data and Pinnacle\(^3\) boundary data) be measured with same resolution; the measurement resolution in this study was 0.086 cm x 0.086 cm X 0.2 cm. These dimensions were chosen to match the acquisition parameters of MRI. The goal of this experiment is for the measurement error between the Pinnacle\(^3\) (ADAC) and MRI data to be within 5% of the maximum allowable dose that can be delivered to the gel (10Gy). This deviation corresponds to 50 cGy.

5.8.1 Image Subtraction Analysis

Image subtraction (pixel-by-pixel) was performed on the ADAC and MRI/polymer gel images, whereby the MRI images were subtracted from the ADAC images. Global deviations (i.e. deviations throughout the entire plane) both with and without a dose correction factor (Chapter 4.5) were first addressed in figures 4.9-4.12. It was concluded from these figures that scans performed 5 days post-irradiation that take the dose correction factor into consideration show the best agreement between ADAC and MRI data. In this case, the criteria for deviation (50 cGy) was met for nearly every slice. Local deviations (i.e., deviations throughout the plane within organs) both with and without correction factors were addressed in figures 4.13-4.17. Measured and planned data within organ regions were closer in agreement when the correction factor was taken into consideration. Negative values correspond to higher MRI readings. Dose deviations within each organ were all within the
tolerance value of 50 cGy: rectum (~0 Gy), rt. femoral head (~18 cGy), bladder (~11 cGy), lt. femoral head (~27 cGy) and prostate (~25 cGy). Almost all data show larger MRI dose measurements than ADAC dose measurements, which is expected as the gel was exposed to scatter and leakage radiation in treatment room for two days prior to irradiation. Figure 4.13 illustrates a problem existing in the dose correction factor, whereby near the center of the gel, this factor overcompensated for the excess dose delivered to the gel. However, with no correction factor, the deviations between the MRI and ADAC dose were larger. While the dose correction factor is necessary, it must be modified in order to contribute less near the center of the gel and more near the periphery of the gel.

As stated earlier, MRI and ADAC results were more in agreement for MRI images taken 5 days post-irradiation as opposed to 2 months post-irradiation both with and without the dose correction factor. The results taken 2 months post-irradiation are misleading as proper care of the gel was not taken in the 2 month span prior to imaging (e.g., gel exposed to sunlight and temperature was not monitored). However, these results were expected as spontaneous polymerization continues in the gel regardless of irradiation. The rate of the spontaneous $R_2$ increase depends on temperature (gels kept cold will polymerize slower than those kept warm) and is uniform throughout the gel volume. It can be concluded from these results that it is better to analyze the gel as soon as possible after irradiation (at least 30 min). There was an overall satisfaction with the image subtraction analysis between the MRI results and ADAC measurements. The data more applicable to our center (i.e., 5 days post-irradiation) meet deviation criteria requirements of less than 50 cGy.
5.8.2 Contour Analysis

Qualitative analysis of dose contours within the polymer gel is shown in figures 4.18-4.21 at a slice located at 100 SPD (z=0). Little agreement was observed between gel data and ADAC data. Again, MRI data taken 5 days post-irradiation more closely resembled the ADAC data than the MRI data taken 2 months post-irradiation. The dose correction factor was once more investigated and it was observed that when comparing the ADAC data to the MRI data taken 5 days post-irradiation, data toward the periphery of the gel appear more agreeable than near the center when this factor was considered and vice-versa. Again, a modification to the dose correction factor is needed.

Figures 4.22-4.24 show the contour regions along with the organs studied in this investigation. As with the image subtraction data and first set of contour data, MRI data were higher than ADAC data when no correction factor was used and lower MRI doses were observed at the center of the gel when the factor was used. An age correction factor was used on the MRI data taken 2 months post-irradiation. This factor appears to be correct as the dose distributions from MRI closely followed the distributions from ADAC. This procedure confirms the theory that images from the two post-irradiation MRI scans images are different because of the increased dose with age in the gel. The age factor thus corrects for the increased polymerization within the gel.

The latter set of figures demonstrates the ability of the polymer gel dosimeter to analyze dose within steep dose gradient regions. Data show contour lines for 100 cGy, 150 cGy, 180 cGy, 190 cGy and 200 cGy both for ADAC and MRI, along with the organ boundaries. Analysis that could be made with these figures, for example, is the dose surrounding the OAR and PTV when the correction factor is not considered. In this case, the
ADAC and MRI contour lines for 180-200 cGy appear to closely surround/encompass the prostate. Also, this same figure shows that the bladder and rectum receive high dose (as much as 200 cGy) and 150 cGy dose extending out to the femoral heads. More analysis could be made for data containing the dose correction factor and data taken two months post-irradiation.

Qualitative imaging is the key for the contour analysis imaging. With better results and less noise, the spatial deviation between the ADAC and MRI data could be measured. Overall, IDL analysis of the contour dose and organ volume distributions throughout the gel proved to be very reliable and should be used with quantitative analysis to provide more validity to the results.

5.8.3 Dose-Volume Analysis

The results of the dose-volume analysis measurements are discussed in Chapter 4.7.3. A comparison was performed between the volume of the organs for ADAC and IDL to ensure that similar volume were being analyzed. Deviations between the two volumes were determined to be very small.

Figures 4.25 and 4.26 represent the ratios contain ratios of the total dose per volume measured (cm$^3$) by MRI to total dose per volume (cm$^3$) within each plane with and without the dose correction factor. An ideal value in this instance would be one, as this would suggest identical dose per volume for ADAC and MRI. This analysis once again shows an overcompensation for the excess dose near the center of the gel (180 cGy to 200 cGy) and the reverse effect near the periphery of the gel (100 cGy). The 150 cGy region benefits most from the correction factor as the ratio is nearly unity throughout the gel. All regions are higher for MRI than ADAC when the dose correction factor is not considered. Once again an
argument here can be made about the necessity for the dose correction factor with modifications.

Maximum dose values per plane (Figure 4.27) also show the need for the dose correction factor as data with correction factor clearly agreed better with ADAC measurements than data without the correction factor. This figure also shows that, as expected, the excess dose due to leakage and scatter radiation was delivered mainly to the gel’s outer regions; towards the periphery, the MRI dose is higher than the ADAC dose even when the corrections factor is considered.

Dose volume histograms for ADAC and MRI are shown in figures 4.28-4.30. Aside from the prostate, the DVH utilizing the dose correction factor is more in agreement to the ADAC-DVH than the figure without the dose correction factor. The dose correction factor lowers the measured dose by MRI hence the reason that the DVH drops off for the prostate around 150 cGy. Neither DVH would be acceptable at MBPCC as the PTV receives too little dose in figure 4.28 and the OAR receive too high of a dose in figure 4.29. At MBPCC, a desired dose for the PTV is at least 95 % of the prescription dose. In the case of the prostate 190 cGy is optimal dose for the entire volume of the prostate as 200 cGy was the prescription dose. An ideal dose for the OAR is that no more than 50 % of their total volume receives a dose above 100 cGy; this tolerance is clearly exceeded in both MRI-DVHs. The MRI-DVH without the correction factor for the prostate is in better agreement with the ADAC-DVH than is the MRI-DVH with the dose correction factor (i.e., the entire prostate volume receives 185 cGy, which is pretty close to the desired value of 190 cGy). This data show that a correction factor must be used to account for the excess dose delivered to the polymer gel due to leakage/scatter radiation, but a modification to this factor is necessary.
5.9 Clinical Outlook at MBPCC

With IMRT and other novel radiotherapy techniques (e.g., SRS and prostate brachytherapy), MBPCC would greatly benefit from the use of the BANG-3™ polymer gel. To implement the gel dosimetry technique, MBPCC would first need to establish contacts with a gel manufacturing company (MGR Research, Inc.) and a gel analysis service (Stony Brook). All other equipment necessary for type of dosimetry is at hand at MBPCC. The following steps would need to be taken in order to perform the gel dosimetry technique for IMRT procedures (per patient)

1). Purchase one CGEL-S vessel, *8 calibration vials (minimum order). Gels can be manufactured to fit specific dose requirements (e.g., maximum dose received by gel). Arrival of gel should take place two days prior to irradiation so that gel can equilibrate to temperature of the cancer center. Keep shipment box and ice coolers for future shipments (see number 7).

2). Store gel away from blue light and ultraviolet light in room with a temperature in the range of 64-72 °F. Calibration gel and experimental gel should be kept at same environmental conditions.

3). Obtain x-ray CT scans of the BANG™ gel

4). Place the CGEL-S vessel on the treatment table and deliver the planned treatment to the gel. **Use no more than 400 $\text{MU} \text{ min}^{-1}$.

5). Place calibration vial on bottom of cubic water tank at CAX and deliver necessary doses for calibration curve. Doses delivered can be anywhere from 0 Gy to maximum allowable dose that the gel can receive (see number 1).

6). Perform any dose verification procedures (e.g., ionization chamber measurement and film QA) if desirable.

7). **Scan the CGEL-S vessel and calibration vials. See Table 3.8 for scanning parameters. Gels must be sent to scanner at least two days prior to scan. Can also ship gel to scanning service (Stony Brook) where results can be obtained in 3-4 days.

8). Analyze data from scans. Analysis can be done by an in-house method (e.g., IDL, Dose³-D QA) or by the gel scanning service.

9). Return gel to manufacturer when analysis completed.

* Because gel is independent of dose rate and temperature, the same calibration curve can be used for all photon beams specific to that machine.

** An MRI scan can be done prior to irradiation to account for background subtraction.
Small steps should first be taken to establish the BANG-3™ dosimetry at MBPCC in an effort to reduce manpower issues in using the technique. For example, one suggestion would be to limit the gel service to new sites treated or use gel once per month. This would allow the department to gain experience into using this new system, and at the same time, find a way to integrate the polymer gel technique with the current method used (film-based QA). Also, a judgment can be made into the usefulness and reliability of the gel. A second suggestion would be to use the scanning service. In this case, all that would be required of the physicists is the purchase, storage and irradiation of the gel and shipment to the scanning company.

5.10 Conclusion

All in all, we were very impressed with the capabilities of what the polymer gel can do; it allows one to analyze and compare planned dose and measured dose in 3D. The dose deviation can be analyzed and examined for any organ of interest. The measurement (absolute) error aimed for was 5% of 10 Gy or 50 cGy. This amount was maintained for all images taken 5 days post-irradiation that take into account the dose correction factor. The relative error depends upon the dose delivered to the gel. The relative error increases as the total dose delivered to the gel decreases. In this case, the total dose delivered was 2 Gy, thus the relative error was high. Looking at the situation 5 days post-irradiation with the correction factor this error was as high as 28.5% for global image subtraction analysis and for image subtraction within the organs was as high as 15% (left femoral head).

Using the data from the 10 MV $R_2$-dose calibration curve and the 10 MV fitting parameters (Table 4.4) along with the equation in Chapter 3.6.8, we computed a relative error of 9.2% with a gel irradiation of 10 Gy. The relative error was smaller than the value
obtained with a gel irradiation of 2 Gy, 25%. Though the percent error was larger than the anticipated error of 5%, this decrease was promising for this study as the error dropped to a reasonable value.

The noise levels were very plentiful with the MRI analysis. Noise estimations were approximated from IDL analysis to be 1% in proton density MR data and 1.3% for $T_2$ data, corresponding to a total amplifying noise of 1.6%. Noise can potentially be reduced by using the full dynamic range of the gel; would increase the SNR. Also, more acquisitions and signal averaging could be performed when scanning via the MRI scanner. A downfall to these increased acquisitions is an increase in scanning time.

Aside from the excess radiation received by the gel, other procedural errors were present. For instance, the fiducial markers were not placed at an optimal position to locate the superior/inferior edges of the CGEL-S vessel. Suggestions to the placement of these markers can be found in Chapter 5.6. Also, as can be seen in figures 4.22 and 4.23, the gel was not masked properly to eliminate regions outside of the gel (e.g., high dose regions at perimeter of gel). This error occurred at the periphery of the gel, thus did not affect the results. The image was kept to demonstrate errors/other data that may be present in analysis.

Finally, as the CT is more accessible to MBPCC than the MRI scanner, some questions may arise as to the use of CT for scanning the gel as opposed to MRI. As stated in Chapter 2.3.3, though x-ray CT has proven to be accurate and capable of localizing high dose gradients, this scanning method has a lower dose resolution and a greater dose uncertainty. This technique may be more developed in the future and when it is, MBPCC should reevaluate x-ray CT’s usefulness in polymer gel analysis.
Chapter 6

Conclusion

With the recent advances of radiotherapy treatment techniques at MBPCC, a 3D dosimetry technique was analyzed for its feasibility, precision and accuracy. The dosimetric system was based on the BANG-3™ polymer gel. This gel was analyzed for its effectiveness in quantifying the 3D dose distribution produced by an IMRT treatment. The plan for this treatment was produced by a Pinnacle³ treatment planning computer, whereby measured dose distributions determined from the polymer gel were compared to the calculated planned data. The PTV within the treatment was the prostate and the OAR were the femoral heads, rectum and bladder; a 7 field treatment technique was applied.

The polymer gel was purchased in a plastic cylindrical vessel and glass calibration vials. The vessel was used to map out the 3D dose distribution from the irradiation, while the vials were used to calibrate the vial response \( R_2 \) to the absorbed dose (Gy). A calibration curve was generated and used to determine the dose within the polymer gel. Vials were calibrated to different energies and different dose rates to test for an energy/dose rate dependence of the gel. The gel was found to be independent of both energy and dose rate.

A historical overview of gel dosimetry was discussed in detail. In this section, the two types of dosimeter systems used (e.g., polymer gels and Fricke gels) were discussed. In particular, research involved with these dosimetry systems was analyzed and the systems were compared. Also, other types of gel imaging systems were reviewed (e.g., optical-CT, x-ray CT and ultrasound), along with clinical uses of the polymer gel (e.g., SRS and brachytherapy). The materials used in this study and the methods involved with their use
were analyzed. In this case, technical specifications involved with the polymer gel, MRI, Pinnacle³ planning software, IMRT treatment delivery and IDL procedures were discussed.

Other dosimetric techniques were performed to test the validity of the distributions produced within the gel. In particular, ionization chamber measurements were compared to Pinnacle³ measurements on a point-by-point basis and film was used to compare planar dose distributions. All measurements met acceptance criteria and it was concluded that the planned dose and delivered dose were in agreement.

All scanning was performed by an MRI located at OLOLRMC. The scanner produced 3D images in the form of $T_2$-weighted and proton density weighted images that were combined to compute the relaxation rate ($R_2$). Images were produced in multiple planes and were compared to the Pinnacle³ dose values pixel-by-pixel using a commercial data-processing package (IDL). Scans were performed 5 days post-irradiation and 2 months post-irradiation.

Before analysis began, several steps were applied to assure that the Pinnacle³ and gel dose distributions were aligned. These steps included shifts, rotations and adjustments in pixel size. The pixel size used was 0.086 cm x 0.086 cm x 0.2cm. Three types of comparisons were done with the IDL system: image subtraction, dose-volume analysis and contour comparisons. Each method provided excellent analysis into the deviations between the calculated Pinnacle³ dose distributions and the measured gel dose distributions. The image subtraction and dose analysis provided global analysis (i.e., throughout plane) and local analysis (i.e., throughout plane and within organ). Also, the dose analysis provided a comparison between the maximum dose received per plane for the Pinnacle³ system and MRI. Contour analysis provided a qualitative analysis of the dose distributions of both Pinnacle³
and CT data; some figures contained the organs of interest and others did not. A dose correction factor was presented as the gels were erroneously exposed to scatter and leakage radiation prior to gel irradiation. All data indicated that the best results occurred for the scans taken 5 days post-irradiation when the dose correction factor was taken into consideration. In this case, nearly all data met the expected measurement error of 5% of the maximum dose (50 cGy). Relative errors were determined to decrease from 25% at a 2 Gy irradiation to 9.2% for a 10 Gy irradiation.

The BANG-3™ polymer gel dosimetry system proved to be an invaluable method for the quantification of the 3-D dose distribution provided by IMRT. A measurement error of less than 5% was achieved. This technique is suggested for use at MBPCC in performing the QA for an IMRT treatment in three dimensions. Cost, manpower issues, toxicity and uncertain response to particular environmental characteristics (i.e., temperature and aging) upon the gel are concerns about the use of the gel at this time. However, more experimentation is being done to investigate gel response; gel handling could be kept to a minimum due to the availability of a reliable gel manufacturer (MGS Research, Inc.) and a gel scanning service (Stony Brook). A decrease in cost with increased clinical use will soon put these concerns to rest.
References


Maryanski, M., “Response Modification in Polymer Gel Dosimeters,” Therapy Poster Session; Calibration & Quality Assurance; Dose Calculations; Dosimetry Instrumentation; Improvements in Conventional Treatment Planning; Treatment & Delivery Systems; Treatment Techniques, 44th AAPM Annual Meeting, Palais des Congres de Montreal, Montreal, 14-18 July 2002.

Maryanski, M. “Re: Graduate Student: Mary Bird Perkins Cancer Center.” E-mail to Paul A. Bruce. 14 Aug. 2002.


Research Systems, Inc. “IDL” www.rsinc.com


Appendix A

Pinnacle³ Treatment Plan

Appendix A provides a summary of the treatment plan used to irradiate the gel.

Technical information such as the beam setup parameters, prescription dose, isocenter position and beam geometries are contained within this section.

A.1 Patient Data

Patient Name: New, New
Patient ID: A000000
Plan Name: Gel
Trial Name: Trial 1
Date/Time: Mon. September 23, 2002/14:45:18
Institution: Mary Bird Perkins Cancer

A.2 Plan Summary Sheet

Beam Setup
Machine-Clinac 21EX
Energy-10 MV
Modality-Photons

Table A.1 Setup parameters for the various beams used to irradiate the polymer gel. Shown are the seven different beams used, their respective SSDs and the MU per fraction delivered from each beam.

<table>
<thead>
<tr>
<th>Beam</th>
<th>Start/Avg. SSD (cm)</th>
<th>MU per fraction</th>
</tr>
</thead>
<tbody>
<tr>
<td>RPO 295</td>
<td>91.70/91.70</td>
<td>37</td>
</tr>
<tr>
<td>Rt. Lat. 265</td>
<td>91.60/91.60</td>
<td>60</td>
</tr>
<tr>
<td>RAO 225</td>
<td>91.46/91.46</td>
<td>47</td>
</tr>
<tr>
<td>AP 180</td>
<td>91.28/91.28</td>
<td>100</td>
</tr>
<tr>
<td>LAO 135</td>
<td>91.25/91.25</td>
<td>57</td>
</tr>
<tr>
<td>Lt. Lat 90</td>
<td>91.33/91.33</td>
<td>66</td>
</tr>
<tr>
<td>LPO 65</td>
<td>91.45/91.45</td>
<td>36</td>
</tr>
</tbody>
</table>

Table A.2 Collimator, gantry and couch setup for the seven irradiation beams used to irradiate the polymer gel. Shown are the collimator positions, gantry start and stop angles, couch and collimator angles and the block type.

<table>
<thead>
<tr>
<th>Beam</th>
<th>X1/X2 (cm)</th>
<th>Y1/Y2 (cm)</th>
<th>Gantry Start/Stop</th>
<th>Couch/Collimator</th>
<th>Block</th>
</tr>
</thead>
<tbody>
<tr>
<td>RPO 295</td>
<td>4.5</td>
<td>3.0</td>
<td>295/295</td>
<td>180/180</td>
<td>MLC</td>
</tr>
<tr>
<td>Rt. Lat 265</td>
<td>4.5</td>
<td>3.0</td>
<td>265/265</td>
<td>180/180</td>
<td>MLC</td>
</tr>
</tbody>
</table>
Table A.2 cont’d.

<table>
<thead>
<tr>
<th></th>
<th>Lateral</th>
<th>Ant-Post</th>
<th>Sup-Inf</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resolution</td>
<td>0.400</td>
<td>0.400</td>
<td>0.400</td>
<td>cm</td>
</tr>
<tr>
<td>Dimension</td>
<td>47</td>
<td>46</td>
<td>28</td>
<td>Pixels</td>
</tr>
<tr>
<td>Origin</td>
<td>-8.783</td>
<td>-10.423</td>
<td>-4.119</td>
<td>cm</td>
</tr>
</tbody>
</table>

**Prescription**
Prescribe 200 cGy per fraction to 100% of dose at “iso” for 35 fractions.
Beam weights are proportional to point dose.
Actual dose at “iso” from all prescriptions beams is 7027.83 cGy.
7 beams are assigned to this prescription.

**Isocenter**
**Iso**
Position patient such that lasers line up with patient marks.
Move the table LEFT 0.46 cm (looking from foot of table.)
Move the table UP 1.67 cm
Move the table IN (toward the gantry) 1.05 cm.

**Plan Setup**
Data Set Name: PAULS PROJ
Data Set Dimensions: 54 slices, 512 X 512 pixels
CT to Density Table Name: MBPCC
Patient Position On back (supine) Head First
Couch: Removed at Y=-9.91
Body Board Angle: None
Number of Photon Beams: 7
Number of Stereo Beams: 0
Number of Electron Beams: 0
Number of Brachy Sources: 0
Outside-Patient Air Threshold: 0.60g/cm^3
Top Slice of CT Extended: 0.00cm
Bottom Slice of CT Extended: 0.00cm
Region of Interest Overrides: No ROI density overrides in use.

Table A.3 This table contains the dose grid geometry used in the treatment. Contained within the table is the resolution, dimension and origin at the Lateral, Ant-Post and Sup-Inf positions.
Dose
Dose Engine: Adaptive Convolve
Density Correction: Heterogeneous
Reference Point: iso
Collimator Output Factor (OFc): 0.990
MLC Transmission Factor: 0.0440
Total Transmission Factor (TTF): 1.000
SPD/OAD (cm): 100.00/0.00
Number of Fractions: 35

Table A.4 Dose parameters for the seven fields are shown in this table. Data includes the relative weight for each beam, the normalized dose at the reference point, the reference point depth/the effective depth, the equivalent square of the unblocked region along with the percentage that it blocked and the dose delivered to the reference point per fraction.

<table>
<thead>
<tr>
<th>Beam</th>
<th>Relative Weight (%)</th>
<th>Normalized Dose at Ref Pt</th>
<th>Ref Pt Depth/Eff Depth (cm)</th>
<th>Unblk Equiv Sq cm/% Blkd</th>
<th>Dose at Ref/Fraction (cGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>RPO 295</td>
<td>14.90</td>
<td>0.922</td>
<td>8.30/8.58</td>
<td>7.2/53.2%</td>
<td>30.2</td>
</tr>
<tr>
<td>Rt. Lat. 265</td>
<td>7.50</td>
<td>0.283</td>
<td>8.40/8.67</td>
<td>7.2/46.4%</td>
<td>15.0</td>
</tr>
<tr>
<td>RAO 225</td>
<td>19.40</td>
<td>0.938</td>
<td>8.54/8.84</td>
<td>7.2/89.8%</td>
<td>39.0</td>
</tr>
<tr>
<td>AP 180</td>
<td>16.20</td>
<td>0.366</td>
<td>8.72/9.02</td>
<td>7.2/86.1%</td>
<td>32.4</td>
</tr>
<tr>
<td>LAO 135</td>
<td>19.70</td>
<td>0.781</td>
<td>8.75/9.04</td>
<td>7.2/84.7%</td>
<td>39.4</td>
</tr>
<tr>
<td>Lt. Lat. 90</td>
<td>7.50</td>
<td>0.258</td>
<td>8.67/8.96</td>
<td>7.2/48.2%</td>
<td>15.1</td>
</tr>
<tr>
<td>LPO 65</td>
<td>14.80</td>
<td>0.933</td>
<td>8.55/8.84</td>
<td>7.2/68.1%</td>
<td>29.7</td>
</tr>
</tbody>
</table>

Point of Interest Dose Information
Trial Name: Trial_1
Absolute Maximum Dose: 7656.33 cGy at 2.42, 0.38, -0.52
Trial_1, iso
(Lateral, Ant-Post, Sup-Inf) = (0.44, -1.70, 0.15) cm

Table A.5 The absolute dose (cGy) and the weighting factor for each beam in the irradiation are shown in this table. (The amount shown is for 35 fractions.)

<table>
<thead>
<tr>
<th>Beam</th>
<th>Abs Dose (cGy)</th>
<th>Pct</th>
</tr>
</thead>
<tbody>
<tr>
<td>RPO 295</td>
<td>1056.36</td>
<td>15.0%</td>
</tr>
<tr>
<td>Rt. Lat. 265</td>
<td>526.18</td>
<td>7.5%</td>
</tr>
<tr>
<td>RAO 225</td>
<td>1364.90</td>
<td>19.4%</td>
</tr>
<tr>
<td>AP 180</td>
<td>1134.56</td>
<td>16.1%</td>
</tr>
<tr>
<td>LAO 135</td>
<td>1378.79</td>
<td>19.6%</td>
</tr>
<tr>
<td>Lt. Lat. 90</td>
<td>527.20</td>
<td>7.5%</td>
</tr>
<tr>
<td>LPO 65</td>
<td>1039.84</td>
<td>14.8%</td>
</tr>
<tr>
<td>Total</td>
<td>7027.83</td>
<td>100.0%</td>
</tr>
</tbody>
</table>
Region of Interest Dose Statistics

Table A.6 This table contains dose statistics to the regions of interest within the irradiated polymer gel. Shown in the table are the regions of interest along with their volume, minimum, maximum and mean values of the dose received and the standard deviation of the dose. Units are in cGy.

<table>
<thead>
<tr>
<th>ROI</th>
<th>Volume(cm^3)</th>
<th>Min Value</th>
<th>Max Value</th>
<th>Mean Value</th>
<th>Std. Dev.</th>
<th>Units</th>
<th>% outside volume</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prostate</td>
<td>45.5489</td>
<td>6257.4</td>
<td>7616.89</td>
<td>7091.09</td>
<td>98.0561</td>
<td>cGy</td>
<td>0.00%</td>
</tr>
<tr>
<td>Lt. Femoral</td>
<td>15.4014</td>
<td>2105.7</td>
<td>5131.85</td>
<td>3398.73</td>
<td>588.644</td>
<td>cGy</td>
<td>0.00%</td>
</tr>
<tr>
<td>Rt. Femoral</td>
<td>11.4029</td>
<td>1874.64</td>
<td>5077.09</td>
<td>3246.62</td>
<td>630.134</td>
<td>cGy</td>
<td>0.00%</td>
</tr>
<tr>
<td>Bladder</td>
<td>57.7522</td>
<td>1629.04</td>
<td>7433.86</td>
<td>4191.36</td>
<td>1332.77</td>
<td>cGy</td>
<td>0.00%</td>
</tr>
<tr>
<td>Rectum</td>
<td>36.241</td>
<td>1056.02</td>
<td>7165.08</td>
<td>3393.58</td>
<td>1657.9</td>
<td>cGy</td>
<td>0.00%</td>
</tr>
<tr>
<td>Ring</td>
<td>717.644</td>
<td>0</td>
<td>6008.06</td>
<td>2395.7</td>
<td>1070.03</td>
<td>cGy</td>
<td>0.00%</td>
</tr>
</tbody>
</table>
Appendix B

IDL Programming Codes

This appendix contains the various IDL programs and routines used throughout this study in order to analyze the polymer gel and compare the dose distribution from the ADAC planning computer and the MRI/gel. Also contained in this appendix are the various programs written prior to analysis in which rotations and shifts were performed.

B.1 Read_PlanROI

PRO Read_PlanROI, SLICE2SHOW = slice2show, PRINT=print, SAVE=save
;reads in region of interest (boundary data) and performs routines upon these parameters

;beginning lines used to read in boundary file and defines cindex and regiontemp
ROIfile = 'plan.roi' ;defines ROIfile
;red, green, blue, yellow, purple, lavendar (approximately)
colors = [ 255L, 255L*256L, 255L*256L^2, 255L*256L+255L*256L^2, $ 255L+255L*256L^2, 255L+127L*256L^2 ]
cindex = 0
regiontemp=bytarr(256,256) ;matrix size identical to MRI matrix

IF NOT KEYWORD_SET( slice2show ) THEN slice2show = 0
;set up the plotting window
PLOT, [0,0], XRANGE=[-14,14], YRANGE=[-14,14], TITLE='100SPD', /NODATA, $ XSTYLE=1, YSTYLE=1
OPENR, inLUN, ROIfile, /GET_LUN ;open file, allocate LUN

REPEAT BEGIN ;until end of file
;##### looking for the ROI name
readstring = '
searchstring = 'name: '
searchlength = STRLEN( searchstring )
REPEAT BEGIN
READF, inLUN, readstring
p = STRPOS( readstring, searchstring )
ENDREP UNTIL ( p NE -1 )
;extract the ROI name
ROIname = STRCOMPRESS( STRMID( readstring, p+searchlength ), /REMOVE_ALL )

;##### looking for the number of curves
searchstring = 'num_curve = '

114
searchlength = STRLEN( searchstring )
REPEAT BEGIN
READF, inLUN, readstring
p = STRPOS( readstring, searchstring )
ENDREP UNTIL ( p NE -1 )
READS, STRMID( readstring, p+searchlength ), ncurves, FORMAT='( I, ";" )'
ncurves = FIX( ncurves )

;this array stores the area contained in each region
area = FLTARR( ncurves )
region3D=bytarr(256,256,ncurves)
sliceorder=intarr(ncurves)

FOR i = 0, ncurves - 1 DO BEGIN
    ; #=#=#=#=#=# looking for the number of points
    searchstring = 'num_points = '
    searchlength = STRLEN( searchstring )
    REPEAT BEGIN
        READF, inLUN, readstring
        p = STRPOS( readstring, searchstring )
    ENDREP UNTIL ( p NE -1 )
    READS, STRMID( readstring, p+searchlength ), npoints, FORMAT='(I,";")'
npoints = FIX( npoints )

    ;read the next (garbage) line. If we wanted to error-check, it
    READF, inLUN, readstring ; should be "points={"

    ;allocate array and temp variables
    ROIdata = FLTARR( 2, npoints ) ;x is dim 0, y is dim 1
    x = 0.0 & y = 0.0 & z = 0.0

    ;read the x,y,z data points
    FOR j = 0, npoints - 1 DO BEGIN
        READF, inLUN, x, y, z
        ROIdata[* ,j] = [x, y]
    ENDFOR ;npoints

    ;end of file reading

    ;converts z number into integer value then defines sliceorder
    slice_num = ROUND( z / 0.3 )
sliceorder(i)=slice_num

    ;converts from cm to pixel indexes
    ROIdatap=FLOOR(ROIdata/0.0859375);same pixel size as dose data

    ;shift center so center at same coordinate at center of ADAC dose
ROIdatap[0,*]=ROIdatap[0,*]+123
ROIdatap[1,*]=ROIdatap[1,*]+148

; saves boundary information
; IF ( slice_num EQ 0) THEN BEGIN
; outfile = ROIname + STRCOMPRESS(npoints,/remove)
; OPENW, outLUN, outfile, /GET_LUN
; WRITEU, outLUN, ROIdatap
; FREE_LUN, outLUN
; ENDIF

; fills in boundaries
indexes=Polyfillv(roidatap[0,*],roidatap[1,*],256,256) ; gives region pixels in terms of MRI matrix size
regiontemp(*)=0b ; reinitialize array
regiontemp(indexes)=1b ; value of 1 means point is in region
region3D(*,*,i)=regiontemp

; compute the area contained within this region
area[i] = poly_area( roidata[0,0:npoints-2], roidata[1,0:npoints-2] )

IF ( slice_num EQ slice2show ) THEN BEGIN ; plots the organ regions
OPLLOT, ROIdata[0,*], ROIdata[1,*], THICK=5, COLOR=colors[cindex]
XYOUTS, ROIdata[0,0], 0.2 + ROIdata[1,0], $ 
    STRUPCASE( ROIname ), CHARSIZE = 2, CHARTHICK=2, $ 
    ALIGNMENT = 0.5, COLOR = colors[ cindex ]
; IF ( ROIname EQ 'prostate' ) THEN $
; POLYFILL,ROIdata, color = colors[ cindex ] ; fills in contours
ENDIF

; the following lines are for guess-timating pixel size
; diff = FLTARR( npoints - 2 )
; FOR j=0,npoints-3 DO diff[j] = ABS( roidata[1,j] - roidata[1,j+1] )
    ; IF KEYWORD_SET( print ) THEN $
    ; PRINT, ROIName, slice_num, npoints, $
    ; AVERAGE( diff[ WHERE( diff NE 0.0 ) ] ), ' cm/pixel'
ENDFOR ; ncurves
print,sliceorder
sliceorder=sort(sliceorder) ; returns subscripts in ascending order
print,sliceorder
region3D=region3D(*,*,sliceorder) ; redefines 3D function with corrected slice order

; the following lines of code copies data from 3mm slices into 2mm slices to be used in DVH and image subtraction analysis
new3D=bytarr(256,256,16)
IF ( ROIname EQ 'prostate' OR ROIname EQ 'bladder') THEN begin

;new slice 0-2 and 14-15 contain no prostate or bladder
new3D(*,*,3)=region3D(*,*,0); new slice at -0.8cm
new3D(*,*,4)=region3D(*,*,1); new slice at -0.6cm
new3D(*,*,5)=region3D(*,*,1); new slice at -0.4cm
new3D(*,*,6)=region3D(*,*,2); new slice at -0.2cm
new3D(*,*,7)=region3D(*,*,3); new slice at 0.0cm
new3D(*,*,8)=region3D(*,*,3); new slice at 0.2cm
new3D(*,*,9)=region3D(*,*,4); new slice at 0.4cm
new3D(*,*,10)=region3D(*,*,5); new slice at 0.6cm
new3D(*,*,11)=region3D(*,*,5); new slice at 0.8cm
new3D(*,*,12)=region3D(*,*,6); new slice at 1.0cm
new3D(*,*,13)=region3D(*,*,7); new slice at 1.2cm

outfile = ROIname + '_2mm.roi'
OPENW, outLUN, outfile, /GET_LUN
WRITEU, outLUN, new3D
FREE_LUN, outLUN
ENDIF

IF ( ROIname EQ 'ltfemoral' OR ROIname EQ 'rtfemoral') THEN begin

;new slice 0-2 and 11-15 contain no femoral heads
new3D(*,*,3)=region3D(*,*,0); new slice at -0.8cm
new3D(*,*,4)=region3D(*,*,1); new slice at -0.6cm
new3D(*,*,5)=region3D(*,*,1); new slice at -0.4cm
new3D(*,*,6)=region3D(*,*,2); new slice at -0.2cm
new3D(*,*,7)=region3D(*,*,3); new slice at 0.0cm
new3D(*,*,8)=region3D(*,*,3); new slice at 0.2cm
new3D(*,*,9)=region3D(*,*,4); new slice at 0.4cm
new3D(*,*,10)=region3D(*,*,5); new slice at 0.6cm
new3D(*,*,11)=region3D(*,*,5); new slice at 0.8cm
new3D(*,*,12)=region3D(*,*,6); new slice at 1.0cm
new3D(*,*,13)=region3D(*,*,7); new slice at 1.2cm

outfile = ROIname + '_2mm.roi'
OPENW, outLUN, outfile, /GET_LUN
WRITEU, outLUN, new3D
FREE_LUN, outLUN
ENDIF

IF ( ROIname EQ 'rectum') THEN begin

new3D(*,*,0)=region3D(*,*,0); new slice at -1.4cm
new3D(*,*,1)=region3D(*,*,1); new slice at -1.2cm
new3D(*,*,2)=region3D(*,*,1); new slice at -1.0cm
new3D(*,*,3)=region3D(*,*,2); new slice at -0.8cm
new3D(*,*,4)=region3D(*,*,3); new slice at -0.6cm
new3D(*,*,5)=region3D(*,*,3); new slice at -0.4cm
new3D(*,*,6)=region3D(*,*,4); new slice at -0.2cm

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new3D(*,*,7)=region3D(*,*,5); new slice at 0.0cm
new3D(*,*,8)=region3D(*,*,5); new slice at 0.2cm
new3D(*,*,9)=region3D(*,*,6); new slice at 0.4cm
new3D(*,*,10)=region3D(*,*,7); new slice at 0.6cm
new3D(*,*,11)=region3D(*,*,7); new slice at 0.8cm
new3D(*,*,12)=region3D(*,*,8); new slice at 1.0cm
new3D(*,*,13)=region3D(*,*,9); new slice at 1.2cm
new3D(*,*,14)=region3D(*,*,9); new slice at 1.4cm
new3D(*,*,15)=region3D(*,*,10); new slice at 1.6cm

outfile = ROIname + '_2mm.roi'
OPENW, outLUN, outfile, /GET_LUN
WRITEU, outLUN, new3D
FREE_LUN, outLUN
ENDIF

; saves boundary file data
IF KEYWORD_SET( save ) THEN BEGIN
    outfile = ROIname + '_2mm.roi'
    OPENW, outLUN, outfile, /GET_LUN
    WRITEU, outLUN, region3D
    FREE_LUN, outLUN
ENDIF

; prints info if print keyword is used
IF KEYWORD_SET( print ) THEN $
   PRINT, ROIname, ' has volume of ', 0.3 * TOTAL( area ), ' cm^3'

; read three more lines to get to the end of this ROI stanza
READF, inLUN, readstring
READF, inLUN, readstring
READF, inLUN, readstring

cindex = cindex + 1 ; increment the color index for the next ROI
ENDREP UNTIL EOF( inLUN )
    FREE_LUN, inLUN ; close file, deallocate LUN

END

B.2 Difference

Pro difference, filename1, filename2, input1, input2
; performs analysis: image subtraction, dose volume analysis and contour analysis

; mask procedure used to eliminate regions outside gel region
mask=bytarr(256,256)
for x=0L,255 do $
   for y=0L,255 do begin
      temp = (x-127)^2 + (y-124)^2
      if (temp LE 93L^2) then mask[x,y]=1
endfor
indexes=where(mask EQ 1)

con3D=bytarr(256,256,16)
; opens mr dose file
openr,1,'dose/'+filename1+'.dose1'; filename1 is sub1,sub2,nosub1,nosub2
mrdose=fltarr(256,256,70)
readu,1,mrdose
close,1
;mrdose=reverse(mrdose,2) ; rotates image in y plane

; opens ADAC dose file

openr,1,'/home/paul/ADACdose/*+filename2;filename 2 is QAbeam94.4SPD--QAbeam104.4SPD
str="
for i=0,10 do begin readf,1,str
endfor
adose=fltarr(256,256)
temp=fltarr(256)
x=fltarr(256)
readf,1,x,format=('",256(f0.3,"\n")')
y=0.0
for i=0,255 do begin readf,1,y,temp,format=’(f0.3,256(f0.3,"\n"))’
adose[*,*]=temp
endfor
;adose=reverse(adose,2) ; rotates image in y plane

; use next lines of code for image subtraction
;;oname=['prostate',"ltfemoral',"rtfemoral',"bladder',"rectum']
;;onum=[253,185,151,335,230]
;;For i=0,4 do Begin
;;con3D=bytarr(256,256,16)
;;openr,5,'/home/paul/boundaries/*+oname(i)+'_2mm.roi';+STRCOMPRESS(onum(i),/remove ) ; opens boundary file
;;readu,5,con3D
;;close,5

openr,2,'/home/paul/boundaries/*+rectum'+_2mm.roi'
readu,2,con3D
close,2
;; want to subtract at -0.8,-0.6,...,1.2 cm for bladder and prostate
;; correspond to 100.8 SPD to --98.8 SPD
indexesa=Where(con3D(*,*,input2) EQ 1b)
;diff=fltarr(256,256)
\[ \text{diff} = (\text{adose[indexesa]} - (\text{mrdose(*,*,input1)}[\text{indexesa}])); \text{input1 is 0--50; input2 is 0-15} \]
\[ \text{print,average(diff)} \]
\[ \text{print,stderr(diff)} \]

; use next 6 lines to determine maximum dose for ADAC and MRI
\[ \text{max} = \text{max(\text{adose[indexesa]})}; \text{defined by filename2} \]
\[ \text{print,max} \]
\[ \text{max} = \text{max}((0.765 \times \text{mrdose(*,*,input1)})[\text{indexesa}]); \text{defined by number used for input, takes into account dose correction factor} \]
\[ \text{print,max} \]
\[ \text{max} = \text{max}((\text{mrdose(*,*,input1)})[\text{indexesa}]); \text{defined by number used for input, does not take into account dose correction factor} \]
\[ \text{print,max} \]

; use next several lines of code for contour analysis
\[ !x.style = 1; \text{sets x-axis to exact range} \]
\[ !y.style = 1; \text{sets y-axis to exact range; 115L+75*256L^2} \]
\[ \text{color} = [135L+255L*256L^2,255*256L,255+255L*256L,255*256L+255L*256L^2,255L] \]
\[ \text{level} = [1,100,150,180,190,200] \]
\[ \text{contour,smooth(mrdose(*,*,input),5),level=level,c_color=color;/noerase;1.735 correction for 2 months post-irradiation} \]
\[ \text{contour,adose,level=level,c_linestyle=[2,2,2,2],c_color=color;/noerase} \]

; next 4 line of code are for determining number of pixels that contain a particular dose within gel
\[ \text{FOR d=0,5 DO BEGIN} \]
\[ \text{print,level(d)} \]
\[ \text{print,n_elements(where(adose[\text{indexesa}] \geq level(d))),n_elements(where((0.765 \times \text{mrdose(*,*,input1)})[\text{indexesa}] \geq \text{level(d)})]} \]
level(d))(n_elements(where((mrdose(*,*,input1))[indexesa] GE level(d)))
ENDFOR
; conplot=tvrd(/true)
; write_jpeg,'contour2.jpg',conplot,/true ;saves image as a jpeg file

;use next lines for color contouring
;loadct,12 ;loads color table
;max=max(.765*mrdose(*,*,input))
;restore,'myct.save' ;restores color table
;tvlct,r,g,b
;tv,bytscl(smooth(.765*mrdose(*,*,input),3),min=0,max=220),0
;tv,bytscl(adose,min=0,max=220) ;;1
;color table used (saved in a file)
;r(0)=0
;g(0)=0
;b(0)=0
;r(0:174)=0
;r(175:209)=120
;r(210:255)=255
;g(0:174)=120
;g(59:116)=255
;g(175:232)=0
;g(245:255)=255
;b(0:58)=0
;b(59:116)=120
;b(117:209)=255
;b(210:220)=120
;b(221:232)=0
;b(233:244)=220
;b(245:255)=255
;write statements are used to write output to file; later transferred to email
;write_tiff,'colorcontour3.tif',bytscl(smooth(.765*mrdose(*,*,input),5),min=0,max=220),red=r,green=g,blue=b
;write_tiff,'contour.tif',bytscl(adose,min=0,max=220),red=r,green=g,blue=b
;write_jpeg,'colorcontour.jpg',mrdose(*,*,input),red=r,green=g,blue=b

close,1 ;closes ADAC dose file

end

B.3 SPD Dose

Function SPDdose,filename ;opens and reads ADAC SPDdose information

openr,1,'/home/paul/ADACdose/'+filename ;opens file
str="" ;accounts for words at beginning of file
for i=0,10 do begin readf,1,str ;reads in first 10 lines of data
endfor
b=fltarr(256,256)
temp=fltarr(256)
x=fltarr(256)

readf,1,x,format='(",",256(f0.3,"\"\"))' ; reads in additional data
y=0.0
for i=0,255 do begin readf,1,y,temp,format='(f0.3,256(f0.3,"\"\"))'
b[*,i]=temp
endfor

; following lines were used to develop contours for the regions then saved the contours
; contour,b
; MIN_CURVE_SURF smoothes contours
; CONTOUR, b,COLORS = [100,150,200]
; openw,1,'filename' ; opens file
; writeu,1,b
; close,1
return,b ; returns b to be used in idl
close,1 ; closes file
end

### B.4 Average

; +
; NAME:
; AVERAGE
; PURPOSE:
; This function receives an array and returns the average of that array.
; For multi-dimensional arrays, a single value representing the average
; of all elements, is returned, if DIM is not set; if DIM is set, then
; the result is an array of one less dimension, containing the averages
; over the eliminated dimension.
; CATEGORY:
; Math function.
; CALLING SEQUENCE:
; AVERAGE( array [, dim] )
; INPUTS:
; array = An array of values (can be multi-dimensional) to be averaged.
; OPTIONAL INPUT PARAMETERS:
; dim = the dimension of the array to average over. DIM must, of course,
; be greater than zero.
; KEYWORDS:
; None.
; OUTPUTS:
; The average (a single numerical value) of the input array, if DIM was
; not set, or an array (with one less dimension than the original) of 
; averages if DIM was set.

; OPTIONAL OUTPUT PARAMETERS:
; COMMON BLOCKS:
; None.

; SIDE EFFECTS:
; None known.

; RESTRICTIONS:
; None known.

; PROCEDURE:
; all of any-D array: average_value = AVERAGE( array ).
; part of any-D array: average_value = AVERAGE( array(a:b,c:d,e:f,...) ).
; over a dimension D of array: average_value = AVERAGE( array, d )

; MODIFICATION HISTORY:
; Created by Kenneth Matthews, June 1992.

FUNCTION AVERAGE, array, dim

; If DIM is not set, return the average for the entire array.
IF N_ELEMENTS( dim ) LE 0 THEN RETURN, TOTAL( array )/N_ELEMENTS( array ) $

; If DIM is set, but not greater than zero, default to computing the
; average for the entire array.
ELSE IF dim LE 0 THEN RETURN, TOTAL( array )/N_ELEMENTS( array ) $

; If DIM is set, and greater than zero, then compute the averages over
; the specified dimension of the array.
ELSE BEGIN
sz = size( array )
IF dim LE sz(0) THEN RETURN, TOTAL( array, dim )/sz( dim ) $
ELSE BEGIN
    PRINT, 'Value exceeds array size (', $
    STRCOMPRESS( STRING( sz(0) ) ),' dimensions ).'
    PRINT, 'Defaulting to average for entire array.'
    RETURN, TOTAL( array )/N_ELEMENTS( array )
ENDELSE
ENDELSE

END

B.5 Compute 3DR2

Pro Compute3DR2, filename ;computes R2 values within calibration vials
nslice=37 ;number of slices to be analyzed; nslice changes depending on which file being analyzed

123
openr,1,'3D/'+filename+'.T2' ;opens T2 file
img3Da=intarr(256,256,nslice)
readu,1,img3Da
close,1

openr,1,'3D/'+filename+'.pd' ;opens proton density file
img3Db=intarr(256,256,nslice)
readu,1,img3Db
close,1

;echo times used with MRI
TE2=116
TE1=21.8
;computes R2
R2=fltARR(256,256,nslice) ;variable for final R2 values
R2[*] = 0 ;initializes array
temp = fltARR(256,256,nslice) ;working variable during calculations
temp[*] = 0 ;initializes
indexes = Where(img3Da NE 0, count) ;excludes zero in denominator
temp[indexes] = float(img3Db[indexes])/img3Da[indexes] ;proton density/T2
indexes = Where(temp GT 1, count) ;excludes negative results from ln
R2[indexes]=(alog(temp[indexes])/(TE2-TE1))*1000 ;calculate R2 at valid points

;print,R2(*,*,7) ;used to view slices of interest

;used when determining average values and standard deviations between slices
ave=fltarr(3)
std=fltarr(3)

;plotted image to be used to locate vial boundaries and center of gel
For i=0,2 Do begin
tvscl,img3Db[*,*,i+1]
getxy,x,y

;3X3 pixel region of interest in which R2 values were measured
roi=R2[x-1:x+1,y-1:y+1,i+1]
ave[i]=average(roi) ;average between slices
std[i]=stdev(roi) ;standard deviation between slices
print,ave[i],std[i]
Endfor

print,average(ave),average(std) ;test print
B.6 Compute 3DR2 gel

Pro Compute3DR2gel,pdfile,T2file,filename ;computes R2 values for CGEL-S vessel

case filename of ;masks according to which filename used
  '28925':img3Df=img3Df*mask;'28925
  '28943':img3Df=img3Df*mask;28943
  '28943b':img3Df=img3Df*shift(mask,1,0,0);2rotpd
  '17263':img3Df=img3Df*shift(mask,1,0,0);17263
  '28948':img3Df=img3Df*shift(mask,1,0,0);'28948'
endcase

;computes R2
TE2=116
TE1=21.8
R2=fltARR(256,256,70) ;variable for final R2 values
R2[*] = 0 ; initializes array
indexes = Where(img3Df NE 0, count) ; excludes zero in denominator
indexes = Where(temp GT 1, count) ; excludes negative results from \ln
R2[indexes] = (alog(temp[indexes])/(TE2-TE1)) * 1000; calculate R2 at valid points

; saves file to dose directory
openw,1,'dose/+'+filename+'.'R2'
writeu,1,R2
close,1

END

B.7 Dose comp

Pro DoseCompute, filename1, filename2; filename1--nosub1.R2 then nosub2.R2, filename2--sub1.R2 then sub2.R2
; program reads in R2 files and determines dose for each pixel in 256X256 matrix

openr,1,'dose/+'+filename1+'.'R2'; read nosub1.R2 and nosub2.R2
img3D=fltarr(256,256,70)
readu,1,img3D
close,1

Dose=fltarr(256,256,70)
Dose=(img3D*100-4.6953)/.6478
openw,1,'dose/+'+filename1+'.'dose1'
writeu,1,dose
close,1

openr,1,'dose/+'+filename2+'.'R2'; read in sub1.R2 and sub2.R2
img3D=fltarr(256,256,70)
readu,1,img3D
close,1

Dose=img3D*100/1.3054
openw,1,'dose/+'+filename2+'.'dose1'
writeu,1,dose
close,1

end

B.8 Get xy

PRO getxy, x, y, DATA = data, DEVICE = device

IF ( KEYWORD_SET( data ) EQ 0 AND KEYWORD_SET( device ) EQ 0 ) THEN device = 1

PRINT, 'Mark point with cursor'
CURSOR, x, y, DEVICE = device, DATA = data
PRINT, x, y

END

B.9 Load CT data

Function loadCTdata ;function used to read in CT data which was later used for analysis

spawn,'ls -1 /home/paul/ctdata/new/17258.*.img',list
CT3D=intarr(512,512,54)
For i=0,53 Do begin
temp=read_dicom(list[i])
CT3D[*,*,*]=temp
ENDFOR
return, CT3D
end

B.10 Read my file

Function readmyR2file,filename ;used to read in R2 file information and used for analysis

openr,1,'R2/'+filename+'*.r2'
R2=fltarr(256,256)
readu,1,R2
close,1
return,R2
end

B.11 Resize

;+
; NAME:
; resize
;
; PURPOSE:
; Resizes a 2-d array
;
; CATEGORY:
; Function
;
; CALLING SEQUENCE:
; nwarr = resize( array, MX=mx, MY=my )
;
; INPUTS:
; array 2-d array to resize
;
; OPTIONAL INPUT PARAMETERS:
; mx new x-dimension
; my new y-dimension

; OUTPUTS:
; nwarr resized array with dimension mx x my
;
; MODIFICATION HISTORY:
; 13-Nov-92 C.E. Ordonez
; - This function implements simple linear interpolation
; to resize a 2-d array. This seems to work better than
; the user routine CONGRID.
;
;------------------------------------------------------------------------
FUNCTION resize, array, MX=mx, MY=my

; Are you trying to trick me?
change=0
IF keyword_set(MX) NE 0 then change=change+1
IF keyword_set(MY) NE 0 then change=change+2
IF change EQ 0 then return,array

b=size(array)
IF (b(0) NE 2 ) then BEGIN
print,'% Input is not a 2-D array'
return, 1
ENDIF

nx=b(1)
y=b(2)
rx=1.0
ry=1.0
xresize=0
yresize=0
CASE change OF
1: BEGIN
IF (mx EQ nx) then return,array
xresize=1
rx = float(mx) / float(nx)
END
2: BEGIN
IF (my EQ ny) then return,array
  yresize=1
  ry = float(my) / float(ny)
END

3: BEGIN
  IF (mx NE nx) then BEGIN
    xresize=1
    rx = float(mx) / float(nx)
    END ELSE mx = nx
  IF (my NE ny) then BEGIN
    yresize=1
    ry = float(my) / float(ny)
    END ELSE my = ny
  END
ENDCASE

; Requested output dimensions same as input's
IF (xresize+yresize) EQ 0 then return,array

; Prepare for rescaling
p=fltarr(2,2)
q=fltarr(2,2)
p(0,1)=1.
q(1,0)=1.
IF xresize EQ 1 then p(0,1)=p(0,1)/rx
IF yresize EQ 1 then q(1,0)=q(1,0)/ry
tmp=poly_2d(array, p, q, 1, mx, my )

; Rescale total counts
tmp=tmp/(rx*ry)

return,tmp
end

B.12 Rots

;+  
; ROTS is used by BCKROT.PRO AND FWDROT.PRO to rotate 
; the image matrix during backprojection/forward-projection. 
;- 
;------------------------------------------------------------------------
; FUNCTION ROTS 
;------------------------------------------------------------------------
FUNCTION rots, a, angle
b = SIZE( a )

xc = ( b( 1 ) - 1. ) / 2.
yc = ( b( 2 ) - 1. ) / 2.

theta = -angle / !RADEG

c = COS( theta )
s = SIN( theta )

kx = -xc + c * xc - s * yc
ky = -yc + s * xc + c * yc
kk = 1. / ( 1. + ( s / c )^2 )

cx = kk * [ s / c^2 * ky + kx / c, s / c^2 , 1 / c, 0. ]
cy = kk * [ -s / c^2 * kx + ky / c, 1 / c, -s / c^2, 0. ]

RETURN, POLY_2D( a, cx, cy, 1, MISSING = 0 )

END

B.13 Save my 3D file

Pro savemy3Dfile,filename ;program reads in images and saves them as proton density and T2 files

spawn,'ls -1 ' + filename+'*.img',list
img3D=intarr(256,256,70)
For i=0,69 Do begin
temp=read_dicom(list[i+70])
img3D[*,*,i]=temp
ENDFOR
openw,1,'3D/'+filename+'.pd'
writeu,1,img3D
close,1

For i=0,69 Do begin
temp=read_dicom(list[i])
img3D[*,*,i]=temp
ENDFOR
openw,1,'3D/'+filename+'.T2'
writeu,1,img3D
close,1

;used to analyze effects of rebinning
tvscl,rebin(img3D(*,*,13),512,512)
tvscl,rebin(img3D(*,*,40),512,512)
B.14 Save my 3D rot file

Pro savemy3Drotfile,file2; file 2 is pre-irradiation ;program reads in images, rotates them relative to post-irradiation images and saves them as proton density and T2 files

img3Dc=intarr(256,256,70)
img3Dd=intarr(256,256,70)
img3De=intarr(256,256,70)
img3Df=intarr(256,256,70)

spawn,'ls -1 '+file2+'.*.img',list

For i=0,69 Do begin
  temp=read_dicom(list[i+70])
  img3Dc[*,*,i]=temp
ENDFOR

For i=0,69 Do begin
  temp=read_dicom(list[i])
  img3Dd[*,*,i]=temp
ENDFOR

For i=0,69 do begin img3De[*,*,i]=rebin(shift(rebin(img3Dc(*,*,i),512,512),-3,2),256,256)
  openw,1,'3D/\'+file2+'.2rotpd';2 is used for subtraction from the data 2 months postirradiation
  writeu,1,img3De
  close,1
ENDFOR

For i=0,69 do begin img3Df[*,*,i]=rebin(shift(rebin(img3Dd(*,*,i),512,512),-3,2),256,256)
  openw,1,'3D/\'+file2+'.2rotT2'
  writeu,1,img3Df
  close,1
ENDFOR

end

B.15 Show all images

Pro showallimages,filename,position=index ;shows gel images
if (Not Keyword_set(index)) then index=0 ;! = not
filename = filename + '\*.img'
spawn, 'ls -1 ' + filename, list
print, list
;window,xsize = 1200,ysize = 512
n = n_elements(list)
for i=0,n-1 do tvscl, rebin(read_dicom(list[i]),64,64), i+index
end

B.16 Show image

Pro showimage,filename ;shows an image
img=read_dicom(filename)
tvscl,img
end

B.17 Subtraction

Pro subtraction,postfile,prefile,filename,DOROT=dorot; sub1=5 days post, sub2=2 months post
;performs images subtraction procedure; subtracts pre-irradiation from post-irradiation

openr,1,'dose/'+postfile;28925.R2 for 5 days post,17263.R2 for 2 months post
postR2=fltarr(256,256,70)
readu,1,postR2
close,1

openr,1,'dose/'+prefile;28943.R2 when 28925.R2 is used, 28943b.R2 when 17263.R2 is used
preR2=fltarr(256,256,70)
readu,1,preR2
close,1

R2sub=fltarr(256,256,70)
R2sub=postR2-preR2; subtracts pre-irradiation values from post-irradiation values
Result=fltarr(256,256,70)

If keyword_set(DOROT) then $
FOR z=0,69 Do $
next line rotates 5 days post-irradiation MRI image (post-pre) with respect to CT image
(determined in analysis)
result[*,*]=shift(ROT(R2sub[*,*],2,4,1,120,127,/INTERP,/pivot,missing=0),7,-3) $ else result=shift(R2sub,5,-3,0) ;no rotation for 2 months post image

openw,1,'dose/+'+filename+'.R2'; saves the file to dose file under sub1 or sub2
writeu,1,result
close,1
end
Paul Anthony Bruce was born on October 20, 1977 in Raceland, Louisiana, making him the third child of four for Daniel and Mary Bruce. His three siblings are Daniel M. Bruce, Jr., Donna Bruce Cheramie and a twin, Paula Bruce. Paul was raised in Galliano, Louisiana, for seventeen years and received an honors diploma from South Lafourche High School in May 1995. Throughout high school, he was affiliated with the marching and concert bands as a member of the percussion section and also served as the drum co-captain for his senior year. Upon receiving his high school diploma, Paul then went on to pursue his college career at Louisiana State University where he received a Bachelor of Science degree in physics in May 2000. During his undergraduate career, Paul was very active in the college community with membership in Circle K National Honor Society, Gamma Beta Phi National Honor Society and the Society of Physics Students. Paul is currently a graduate student at Louisiana State University where he will receive his master’s degree in medical physics in May 2003. After college, Paul plans on working as a medical physicist and would like to be an active participant in the fight against cancer. Paul is very outgoing and enjoys spending time with his family and friends and enjoys watching football.