Eye Plaque Brachytherapy Quality Assurance

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This thesis is presented as part of the requirements for the conferral of the degree:

Master of Science (Physics)

University of Wollongong
School of Physics, Centre for Medical Radiation Physics

March 2017
Declaration

I, Alannah M. Kejda, BMedRadPhys, declare that this thesis submitted in partial fulfilment of the requirements for the conferral of the degree Master of Science (Physics), from the University of Wollongong, is wholly my own work unless otherwise referenced or acknowledged. This document has not been submitted for qualifications at any other academic institution.

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September 4, 2017
Abstract

The increased sophistication of non-standard radiotherapy techniques have encouraged the development of precise, accurate and fast dosimetry systems for treatment verification. For eye plaque dosimetry verification, the Centre for Medical Radiation Physics (CMRP) have previously designed a diode based spectroscopic probe. This research aims at improving the original design by increasing the spatial resolution, reducing the angular dependence and minimising the distance between eye plaque and the detector sensitive volume.

GEANT4 simulations were used as a verification of the construction of the redesigned probe, with the average variation of the probes response to that of absorbed dose to water of ±1.78% for clinically applicable depths. A custom 3D-printed phantom was designed to facilitate sterile measurements whilst achieving an approximate water environment necessary for TG-43 U1 validity. Furthermore, it included a number of separate pieces and compartments designed to separate the sterile plaque and seeds from the water bath.

Experimentally, the accuracy of the new probes design and response utilising the 3D-printed water phantom in conducting spectroscopic, eye plaque brachytherapy dosimetry was ±2% compared to the projected dose rates calculated using the clinical Treatment Planning System (TPS). It was also successful in identifying seed packing errors, introduced into the plaque to test its dose rate response sensitivity.

The spectroscopic dosimetry system has been verified for use in the clinical environment as an eye plaque brachytherapy QA tool and can be used for pre-treatment verification of dose-rate.
Acknowledgments

I would like to thank the multitudes of people that have helped to make this thesis happen. Firstly, my supervisor Dr. Dean Cutajar, thank you for all of the hard work and effort you put into the making of this thesis, I appreciate all of your support. My technical support Dr. Michael Weaver, your expertise is making things automatic and for introducing me to the world of 3D printing was so much appreciated. I would like to thank both of you for your continuous hard work and for the opportunities you provided. I believe that your tutelage and guidance has made me into a better physicist!

I would like to thank Distinguished Professor Anatoly Rosenfeld for establishing this project and offering wisdom and advice into its progression. In addition to whole CMRP, both staff and students, in particular the most helpful Karen. It was a fantastic opportunity and a privilege to work as a part of one of an eminent Medical Physics research organisation, I am extremely grateful. A big thanks also goes to Dr. Susanna Guatelli for her Geant4 strategies and support, and to Dr. Marco Petasecca for sharing your unrivalled detector knowledge. I’d like to extend a huge thank you to Alex Pogossow for the development of the projects electronics.

In addition, I would like to thank and acknowledge the folk working on the UOW HPC (High Performance Cluster) network. Their high powered and fast processing, along with timely technical support was invaluable to my GEANT4 simulation work. I have to extend a huge thank you to Simon Downes and the collaboration of the Nelune Cancer Care Centre, Prince of Wales Hospital, Randwick for their collaboration on the project. In particular, I would like to thank Claire
Pagulayan and Paul White for their help in preparing and carrying out experiments, and for absorbing the lion's share of radiation in doing so.

To Lachlan Chartier, thank you for your help in realising my designs with your superior technical drawing skills. I also would like to thank you for your persistent encouragement and for being able to make me laugh in the midst of a scientific crisis. I’d like to thank my family for supporting me both emotionally and financially, for putting no time-limit on my education and for making the rest of my life as simple and as easy as was possible.

This research has been conducted with the support of the Australian Government Research Training Program Scholarship.
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Chapter 1

Introduction

A majority of ocular malignancies occur in the uveal layer of the eye, in the vascular and melanin rich choroid [1]. The most common of these is choroid melanoma, which is carcinogenic melanocytes in the choroid and has relatively high mortality rates which increase with tumour size [2]. For patients with tumours classified as large, approximately 53% die within 5 years of diagnosis [3].

Historically, the surgical removal of the eye (enucleation) or local resection has been the primary mode of treatment, eliminating any chance of preserving sight in the diseased eye. Alternative treatments such as eye plaque brachytherapy are used as an option to preserve the patient’s eye and retain their sight. This treatment method involves packing low dose rate brachytherapy seeds into an eye plaque (a concave, metallic plaque coupled with an acrylic applicator) [4]. This plaque is attached to the sclera of the patient, such that it covers the base of the tumour, and remains there for several days. The duration of the treatment is determined by the dose prescription and the activity of the seeds. The radiation emitted kills the cells in close proximity to the seeds, while the plaque acts as a shield to protect healthy tissue, such as the brain and optic nerve. Eye plaque brachytherapy is therefore considered a targeted therapy.

Currently, there is no direct verification of the dose rate of a packed plaque before its surgical insertion. The minimal quality assurance (QA) testing of this treatment modality is most likely due to the relatively low number of patients re-
ceiving the treatment annually, the technical difficulty of measuring a packed plaque prior to surgical insertion and its dependence on the availability of both brachytherapy sources and surgical theatre. The current practice is to measure the activity of the seeds as they are packed, according to the treatment plan. This, however, allows for the possibility of seed arrangement errors as there are no visual indicators of a seed’s activity, and as such, the physicist must rely on calibrated activity measurements. The efficacy of the treatment planning system (TPS) at planning the treatment must also be verified. Previous studies have legitimised silicon detectors and a spectroscopy system for prostate brachytherapy using Iodine-125 brachytherapy seeds [5].

The first generation spectroscopic dosimetry system utilised a measurement probe, mounted with a silicon diode as the detector element [6]. This probe was found to be unsuitable for eye plaque brachytherapy dosimetry due to its angular dependence, resulting in varied detector signal as a function of the incident angle of primary photons. Due to the concave shape, primary photons are incident on the detector from a large range of angles. In addition, an adjustment factor could not be applied to account for this angular dependence as depth dose measurements were conducted along the central axis of the plaque, resulting in the solid angle of primary photons changing at each depth.

The aim of this thesis is to assess the capability of the previous system to be repurposed and customised for the dosimetry of eye plaque brachytherapy.

A number of objectives have been developed in order to achieve a positive outcome:

- Develop a new spectroscopic dosimetry probe with improved angular dependence characteristics and a shallower depth of measurement for pre-treatment dose verification and quality assurance of I-125 eye plaque brachytherapy
- Verify this design using Geant4 simulations
- Design a custom eye plaque phantom allowing for ideal dosimetric conditions,
while maintaining the sterility necessary for implantation of the plaque into the patient

- Incorporate the phantom for use in the collection of automated depth dose profiles
Chapter 2

Literature Review

2.1 The Eye

The eye is a complex, sensory organ responsible for vision. Vision is the dominant sense in humans, accounting for approximately 70% of sensory receptors in the brain [1], therefore any damage or illness related to the eye will significantly impact quality of life. The adult eye is approximately spherical as depicted in Figure 2.1, averaging at 24 mm in diameter with a function is to focus environmental light onto photoreceptors found in its posterior through refraction.

Figure 2.1: Diagram of the eye showing, in particular, the anatomy of the uvea. Adapted from [1]
Due to the eyes being situated in the bony orbits of the skull [7], only approximately one sixth of the eye’s surface is externally visible with the remainder being enclosed within the orbit and surrounded by protective fat deposits and extra-ocular muscles. Within this anatomical region there are many important accessory structures of the eye that are required for its protection and maintenance, such as the conjunctiva and lacrimal glands. This area also contains very sensitive anatomy, such as the optic nerve and cornea, wherein damage to these areas could result in vision loss.

Within the layers of the eye is the uvea [1] [7]. The uvea refers to the middle of the eye and contains the iris, the ciliary body and the choroid. The choroid is very vascular as it facilitates the blood and nutrient supply to the ciliary body, the iris and to outer layers of the eye through a large number of densely packed blood vessels. The choroid is also very rich in melanin to reduce the reflection and scattering of light within the eye via the absorption of the stray photons, which would otherwise degrade and contort vision.

### 2.1.1 Ocular Melanoma

Intraocular melanomas are malignancies of the melanocytes located within the eye [2]. There are approximately 0.1 to 2.3 cases per 100 000 worldwide per year [3], with 85% of these cases occurring in Caucasian individuals. Ocular melanomas have relatively high mortality rates that increase along with the size of the tumour, as shown in Table 2.1.

<table>
<thead>
<tr>
<th>Classification</th>
<th>Tumour Size</th>
<th>Mortality Rate</th>
</tr>
</thead>
<tbody>
<tr>
<td>Small</td>
<td>&lt;3 mm</td>
<td>16%</td>
</tr>
<tr>
<td>Medium</td>
<td>3 - 8 mm</td>
<td>32%</td>
</tr>
<tr>
<td>Large</td>
<td>&gt;8 mm</td>
<td>53%</td>
</tr>
</tbody>
</table>

The poor survival rates of large masses in particular, indicate the importance of
timely diagnosis and effective treatment [8]. Melanoma of the uveal layer is likely to form metastasis disseminated through the blood stream to the liver and other organs [9]. Due to the complex structure of the eye and its many anatomic features, there are a large variety of cancers that can occur. The most common of these is choroid melanoma, making up 90% of cases. Due to the choroid being highly vascular, a relatively large amount of blood passes though it allowing for any diseased cells from that area to travel throughout the body easily, increasing the risk of spreading haematogenous metastasis to distant sites.

2.1.2 Treatment

Historically, these melanomas have been treated via surgery, by removing the eye (enucleation) or through the removal of the mass through local resection [10]. The full or partial removal of the eye not only results in the patient being effected from a physiological loss of vision, but evidence suggests that a patient may experience psychological side effects, leading to conditions such as depression [11] [12].

Alternative treatments of ocular malignancies include treatment using radiation therapy that can be delivered via external beam therapies and brachytherapy [13]. External beam radiation in Australia primarily uses high energy photons targeted onto the area of interest, however due to the delivery method of this radiation, areas in close proximity to the target can be exposed to high dose levels through the processes of scattering. Internationally, proton beams have been used to successfully treat cancer and are advantageous due to their localised energy deposition, commonly described by the Bragg peak [14]. However, there is no such treatment available in Australia as it is considered to be too expensive in contrast to other treatments.

More commonly, eye cancers are treated via low dose rate (LDR) brachytherapy where small radioactive sources are used to irradiate the target area with low energy photons over a long period of time [8]. There has been no observed difference in survival rates between enucleation and brachytherapy [15], which strengthens the
use of brachytherapy as a treatment modality due to the possibility of maintaining sight and the sparing of the eye itself.

2.2 Brachytherapy

Brachytherapy involves the use of low energy radioactive sources placed adjacent to or within the treatment volume [16]. These sources are sealed within an encapsulating material which prevents any chemical interaction or loss of the radioactive material, and also acts to shield against any charged particles emitted from the source [17]. The casing should not react with the surrounding the biological tissue and be of sufficiently high atomic number so that it can be easily identified in images.

Generally, brachytherapy is conducted in two different forms. High dose rate (HDR) brachytherapy involves sources that have a high activity and are only inserted in the body temporarily. Interstitial catheters are inserted through, or alongside the target volume. An external machine or afterloader guide the HDR source until it reaches the desired destination known as a dwell position. It is held in that position for a pre-determined time (dwell time), usually of the order of seconds, then moved by the afterloader into the next position [18]. A large dose can be delivered to the target tissue in a short amount of time without significant entrance or exit doses, as seen in external beam radiotherapy. The high dose delivered is particularly advantageous when applied to tissues with a low \( \alpha/\beta \) ratio, as they are most likely to benefit from dose escalation [19]. In low dose rate (LDR) brachytherapy the sources are most commonly radioactive seeds with a low activity that are left within the body for long periods of time. In some cases, seeds remain in the patient for just a few days to a week while other patients undergo permanent implantation. The number of seeds used varies with each treatment with larger volumes such as the prostate requiring more seeds than those necessary for eye plaques. Over this time, the radiation emitted by these sources are incident on the surrounding tissue, effectively treating the target area.
The advantages of using this radiological treatment modality include achieving a non-uniform distribution of radiation emission, steep dose gradients over small distances that can be established due to the very low energy sources, targeted irradiation of the treatment volume, and sparing the surrounding normal tissue [18]. This improves the tumour control probability and decreases the normal tissue complication probability.

2.2.1 ROPES Eye Plaque

LDR brachytherapy is employed in the treatment of ocular melanomas through the use of an eye plaque. Although many exist, the most common eye plaques used in brachytherapy are the Collaborative Ocular Melanoma Study (COMS) plaque and the Radiation and Oncology Physics and Engineering Services, Australia (ROPES) plaque, the latter of which is used and widely available in Australia. The AAPM TG-43 U1 protocol, which will be outlined below and is the current standard in brachytherapy dosimetry formalism [20] can be used to performed dosimetry of either plaque. Studies into the COMS protocol has found that the survival rates for patients with small- and medium-sized tumours are equivalent when treat with COMS brachytherapy and enucleation [21]. The ROPES plaque can be seen in Figures 2.2 and 2.4.

The 15 mm ROPES eye plaque is composed of an acrylic mould that holds a total of 10 radioactive seeds in place and a stainless steel cap or applicator. The mould also completes a secondary function placing the radioactive sources 1 mm from the surface of the eye. The applicator both encapsulates the seeds and shields the tissue around the orbit including the brain from radiation. Stainless steel is a convenient material to use as it does not react or oxidise within the body and has a relatively large effective atomic number, increasing its ability to attenuate photons and therefore act as a shield for surrounding tissue. The applicator is shaped as a spherical shell. Specially shaped plaques are required in the treatment of tumours located close to the optic nerve [22]. This allows for the treatment of tumours of
varying sizes; delivering the prescribed dose to the target area, whilst ensuring the smallest possible dose to the surrounding healthy tissue. There are several different ROPES eye plaque designs, including an 11, a 15, an 18 and a 20 mm diameter model, all of which are available with either a flanged or simple eyelets used to stitch the plaque in place.

The applicator and mould are then attached to the sclera and left for approximately 3-7 days before being removed [23]. Generally, a dose of 80-100 Gy is recommended to be delivered to the apex of the tumour at a rate of approximately 0.60 - 1.05 Gy/hr [24]. The American Brachytherapy Society has found that dose rates lower than those recommended were associated with inferior tumour control.

### 2.2.2 Iodine-125 Brachytherapy Source

There are a number of radioactive sources used in the treatment of cancer via brachytherapy. HDR brachytherapy uses high activity sources that generally emit
photons with much higher energies when compared to LDR sources, with energies of the order of a few hundred keV [25]. Due to the high activity and greater photon mean-free path of HDR sources, they have the potential to cause significant damage to healthy tissue and can only be used for very short intervals before being removed.

**Figure 2.3:** The Oncura OncoSeed model 6711 schematic with dimensions. Adapted from [26]

The seed commonly used in eye plaque brachytherapy in Australia is the Oncura OncoSeed model 6711, which contains the radioactive element Iodine-125. It consists of a cylindrical sliver core; 3 mm long and a diameter of 0.5 mm, which is coated with radioactive Iodine and encapsulated by a titanium shell, as shown in Figure 2.3. This shell is 4.5 mm long and has a diameter of 0.8 mm. It is important to note that the shell does not have uniform thickness, contributing to the anisotropy of the emitted radiation. The shell is 0.5 mm thick at the ends and 0.06 mm thick along the sides. There exists a small cavity between the silver rod and titanium capsule which does permit the source to have a small amount of moment within the seed [26].

The primary energy emission of I-125 is approximately 27 keV with a half-life of 59 days [12] however the 6711 seeds emit a variety of wavelengths both from the I-125 decay scheme and from silver core characteristic X-rays, as shown in Table 2.2. The table shows the relative frequency of the gamma ray energies emitted per 100 decays of the I-125 or the frequencies of the characteristic X-rays per 100 vacancies of that electron shell. This occurs when an I-125 photon is incident on the core and is absorbed via the photoelectric effect. A subsequent electron cascade with the
Table 2.2: Oncoseed 6711 I-125 photon spectra

<table>
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<tr>
<th>Photon</th>
<th>Mean Energy (keV)</th>
<th>Intensity per 100 decays/vacancies</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ag $K_{\alpha 1}$</td>
<td>22.16</td>
<td>45.6</td>
</tr>
<tr>
<td>Ag $K_{\beta 3}$</td>
<td>24.94</td>
<td>7.5</td>
</tr>
<tr>
<td>X $K_{\alpha 2}$</td>
<td>27.20</td>
<td>40.6</td>
</tr>
<tr>
<td>X $K_{\alpha 1}$</td>
<td>27.47</td>
<td>75.7</td>
</tr>
<tr>
<td>X $K_{\beta 3}$</td>
<td>30.94</td>
<td>6.8</td>
</tr>
<tr>
<td>$\gamma$</td>
<td>35.49</td>
<td>6.7</td>
</tr>
</tbody>
</table>

affected silver atom causes an x-ray to be emitted contributing to the overall dose delivered by the 6711 seed. These main emissions include 22 keV, 25 keV, 27 keV, 31 keV and 35 keV photons [5].

2.3 Importance of Quality Assurance

Quality assurance is of paramount importance in radiotherapy [27]. It ensures that the prescribed dose is delivered to the target volume with high levels of accuracy and reliability as well as minimising the radiation exposure of healthy tissue, thereby making the modality safe and effective [28]. Although there are both planning and dose calculation systems utilised in clinical sites to ensure the required dose is delivered, the software included in these systems is limited by the algorithms included and the time it takes to calculate the plan. Therefore, quality assurance acts as an external verification. Another important facet of QA procedures is to certify that the various components and equipment used during a treatment are functioning as expected and to accepted standards.

Eye plaque brachytherapy is a complex treatment modality that requires rigorous QA measures. The American Association of Physicists in Medicine (AAPM) formed Task Group No. 129 (TG-129) as a review of eye plaque radiotherapy [29], specifically treatments utilising COMS (Collaborative Ocular Melanoma Study)
plaques. The review found that for treatments using these COMS plaques, loaded with I-125 seeds and with a standard prescription of 85 Gy at a 5 mm depth, the dose delivered was 75 Gy. This is a discrepancy of approximately 12%. The AAPM suggest that further scrutiny be applied to the Monte Carlo algorithms used to calculate the dose delivered and that this is best achieved by ‘further dosimetric evaluations’. These algorithms assumed that the seeds would be contained in a homogeneous environment rather than a realistic heterogeneous one. The findings of the task group are significant enough to prompt the development and implementation of a new and thorough QA protocol based on externally measuring the dose at the prescription point prior to plaque insertion. Other studies specifically recommend that QA procedures help to improve positional accuracy of dose deposition and reduce the chance of incorrect placement of seeds within the plaque [30].

Figure 2.4: Picture of a 15 mm ROPES eye plaque, acrylic insert and I-125 seeds

Standard Quality Assurance procedures for eye plaque brachytherapy treatment at Prince of Wales (POW) hospital, Randwick, follow TG-129 guidelines [29]. These are modified for a ROPES plaque as the AAPM only considered COMS plaques in their report. Plaque Simulator 5 $^{TM}$, Eye Physics, is employed as the treatment planning system (TPS), which uses the activities of available seed batches to formulate a plan to best carry out the prescription. An independent check of the plan is performed using the brachytherapy module in RADCALC $^{TM}$ LSI Inc., modified for
eye plaques [31]. The discrepancy between the two plans should be no more than ± 2% [4].

There are only two batches of seeds available at any one time for use in eye plaque brachytherapy at POW, the new and the old. This reduces both the confusion and probability for mistake when determining which batch a seed comes from, as over time the sources will decay making it easy to discern between the batches.

The plan must be checked to ensure the planned dose rate is between 60 – 105 cGy/hr to the prescription point as well as to see if it has been formulated for the correct eye. If either of these conditions fail, the treatment must be re-planned. Once a plan has been accepted, the pre-insertion quality assurance is conducted. The activity of the seeds in each batch is measured individually by using a well chamber [29] and the mean activity determined. The measured mean activity of a batch should lie within ± 3% of the activity presented by the seed manufacturers certificate. The mean batch activity and the comparison calculation are then independently checked by a second physicist. The plaque and seeds are placed in the autoclave for sterilisation, while the temperature and pressure of the process monitored to ensure they are within the appropriate limits. The plaque is then packed under direct supervision of a second physicist, to ensure placement of seeds follows the plan outlined by the TPS.

Currently, there is no check to directly measure the dose rate of the loaded plaque to quantify how well it conforms to the TPS plan and if there have been any errors in the placement of the seeds missed by both physicists [4]. As the seeds themselves have no discernible features and the plaque contains no position identifiers, there is a significant probability of an error occurring in the form of a misplaced seed. Current eye plaque brachytherapy QA is not robust, which could allow the case presented by TG-129 above, where delivery of a dose 12% under the prescribed to occur. This could ultimately lead to local control failure. If the physicist was unsure of the plan, investigations using radiochromic film could be conducted, however these tests require a relatively long time to read-out results,
which is not conducive to the clinical environment. The implementation of a fast read-out, small volume, spectroscopic dosimeter, measuring the depth dose along the central axis of the probe, would further ensure the conformity of the plaque to the plan, but also determine if the plan directly delivers the dose rate indicated.

2.4 TG-43 U1 Protocol

Brachytherapy is a diverse branch of radiotherapy. Many different sources, seed designs and placement techniques can be employed in a treatment regimen. As stated previously, enucleation and brachytherapy have equivalent survival rates in the treatment of ocular cancers but only brachytherapy allows a possibility of visual preservation [21]. The only way that this outcome is achievable is through the correct execution of the treatment plan, such that the radiation interacts with the target tissue and spares the sensitive sight structures, such as the optic nerve. The AAPM formed Task Group No. 43 (TG-43) in 1995 to introduce standards and protocols into the dosimetry of brachytherapy and to address issues found in dated dose calculation formalism that did not take several important factors into account [32]. The update, TG-43 U1, was released in 2004 by the AAPM due to the dramatic increase in brachytherapy cases both with interstitial and permanent source implantation since the original report [33]. The data used in the protocol are derived from a combination of experimental and MC simulations. Thus, the formalism involves a number of assumptions. One such assumption is that the dose rate can be determined in an infinite water phantom, and thus cannot take heterogeneities into account, such as in the case of eye plaque brachytherapy. While not taking the wider environment into account, this allows for the TG-43 protocol to be pertinent in the majority of clinical brachytherapy applications.

The following is the TG-43 U1, 2-dimensional formula for the dose rate of an individual seed:
\[
\dot{D}(r, \theta) = S_K \Lambda \frac{G_L(r, \theta)}{G_L(r_0, \theta_0)} \cdot g_L(r) \cdot F(r, \theta) \tag{2.1}
\]

where \( r \) is the distance between the centre of the source and the point of interest \( P(r, \theta) \) in centimetres as shown in Figure 2.5, \( \theta \) is the polar angle subtended between \( P(r, \theta) \) and the central axis, \( r_0 \) and \( \theta_0 \) are standard measures at 1 cm and 90° from the centre of the source and the central axis respectively. \( S_K \) is the air kerma strength, \( \Lambda \) is the dose rate constant, \( G_L(r) \) is to the geometry function at \( P(r, \theta) \), \( g_L(r) \) is the radial dose function of the line source and \( F(r, \theta) \) is the 2D anisotropy function.

This formula is built around the assumption that each seed is a cylindrical source can be approximated by a line. Apart from the seed design and composition, the dose rate is primarily determined by the distance the seed is from the point of interest and the angle formed between them.

**Air-Kerma Strength:**

Air-kerma strength refers to the air-kerma rate of radiation in a vacuum, where kerma is the expected energy released per point per unit mass by incident radiation [34]. \( S_K \) has units of \( \text{cGy.cm}^2.\text{h}^{-1} \) and its formula is as follows:

\[
S_K = K_\delta(d)d^2 \tag{2.2}
\]

Where \( K_\delta(d) \) is the air-kerma rate at a distance \( d \), along the transverse plane of the seed. The point of measurement is sufficiently far from the seed such that the source can be treated as a single point and that \( S_K \) is independent of \( d \), typically 1 metre. \( K_\delta(d) \) also contains a low energy threshold denoted by \( \delta \), which is usually 5 keV. This is due to the low energy X-rays originating in the metallic capsules of the source having the potential to increase the air-kerma dose rate without contributing to the dose over distances greater than 0.1cm.
Dose Rate Constant

The dose rate constant $\Lambda$ is the ratio of the dose rate at a reference position in a medium (liquid water is recommended) to air-kerma strength. This reference position is designated as $P(r_0, \theta_0)$. The formula for the dose rate constant in water is:

$$\Lambda = \frac{\dot{D}(r_0, \theta_0)}{S_K}$$  \hspace{1cm} (2.3)

This is an absolute quantity with the units $cm^{-2}$. This constant is dependent on the source material, geometry and composition used in the treatment, which can affect the energy spectrum of the radiation emitted. This variable is important for calibration purposes of any detectors used.

Geometry Function

The geometry function is an approximation based on the characteristics of the source, specifically the distribution of radioactivity within the seed, while also improving accuracy by acting as an inverse square correction. When using the two-dimensional model the geometry function is as follows:

$$G_L(r, \theta) = \begin{cases} \\
\frac{\beta}{Lr\sin\theta}, & \text{if } \theta \neq 0^\circ \\
(r^2 - \frac{L^2}{4})^{-1}, & \text{if } \theta = 0^\circ 
\end{cases}$$

Where $\beta$ is the angle made up of the lines formed between both the edges of the line source to the point of interest as shown in the following diagram:

The active length $L$ used in this calculation can be substituted with $L_{eff}$ depending on the structure of radioactivity within the seed. An example of the different structures can be seen in Figure 2.6. $L$ can be used in the case shown in Figure 2.5, where the radioactive source is a uniform cylinder. $L_{eff}$ is required in seeds that contain uniformly distributed sources and is described by:
where $N$ is the number of sources within the seed and $\Delta S$ is the spacing between each source from their centres.

**Radial Dose Function**

The radial dose function attempts to account for the sharp gradients on the transverse plane due to attenuation and scattering. Together with the geometry function they provide realistic corrections, improving the accuracy of the calculated dose rate. The radial dose function is defined as:

$$g_L(r) = \frac{\dot{D}(r, \theta_0)}{\dot{D}(r_0, \theta_0)} \frac{G_X(r_0, \theta_0)}{G_X(r, \theta_0)}$$  \hspace{1cm} (2.5)$$

Where the subscript is denoted by an $L$ when considering the two-dimensional line-source case or a $P$ for a point-source. This quantity can be found in look-up tables included in the TG-43 U1.
Figure 2.6: Diagram of a) Oncura 6711 OncoSeed, b) Oncura 6702 I-125 brachytherapy seed. Because of the individual internal structures 6711 and 6702 would use L and $L_{eff}$ respectively in the calculation of the geometry function. Adapted from [26]

2D Anisotropy Function

There is a significant inhomogeneity around brachytherapy seeds due to the absorption and scattering of photons through the encapsulating material. In addition, the radio-opaque marker within the seeds themselves, can absorb photons, further contributing to anisotropy. The 2D anisotropy function describes this variation in dose and is defined as:

$$F(r, \theta) = \frac{\dot{D}(r, \theta) \cdot G_L(r, \theta_0)}{D(r, \theta_0) \cdot G_L(r, \theta)}$$  \hspace{1cm} (2.6)$$

To maintain consistency, the L used in $G_L(r, \theta)$ is the same as was used in the geometry function, either the active length L or effective length $L_{eff}$. A failure to do this would result in errors, particularly in measurements very close to the seed.

Some of the oversights of TG-43 U1 include that it does not consider variables that may affect the actual dose rate apart from those related to the structure and
orientation of the seed. Large portions of the body are made up of tissue that is relatively dense when compared to water, such as bone. Equivalent dose calculations must be performed to supplement TG-43 U1 in order to accurately depict the dose delivered to non-water substances and in areas of tissue inhomogeneity. Extra consideration must also be made to the applicators used to mount brachytherapy seeds and their effect on the dose delivered as the presence of an applicator can change the value of the anisotropy function [35].

2.5 Photons Interactions

Photons can interact with matter in many ways, however the likelihood of each of these interaction modes are primarily dependent on the energy of the photon and the medium it traverses [36]. It the interaction of a photon with matter that causes damage to tissue; through the ionisation of electrons and nuclear processes. Photons are electrically neutral and therefore do not continuously lose energy in matter as charged particles do, but instead have a probability of interaction following a statistical model. Although each individual radiation event occurs randomly, the average rate at which these events occur over a larger time period can be reasonably predicted and thus described with Poisson statistics [37].

2.5.1 Elastic Scattering

When an incident photon has a sufficiently low energy it can undergo elastic scattering, with only a negligible loss of energy due to the conservation of momentum [34]. Thomson scattering consists of the incident photon interacting with a free electron. This electron then undergoes classic oscillation [5]. This oscillation results in the emission of a photon in the direction as determined by the electron’s electric orientation [37].

Rayleigh scattering, like Thomson scattering, is a process where the incident photon does not transfer its energy to the medium it interacts with. When an
incident photon of sufficiently large wavelength interacts with a whole atom it can cause all of the electrons bound to that nucleus to oscillate in phase, leading to the emission of a photon with the same frequency as that of the incident photon but in a new direction [5], as shown in Figure 2.7. Photon interactions with an entire atom are known as coherent. Rayleigh scattering has a relatively low probability of occurring when considering the short wavelengths used in a clinical environment.

![Figure 2.7](image)

**Figure 2.7:** The kinematics of Rayleigh Scattering where a) shows the photon incident on an atom and b) the subsequent vibration of the atom and re-emission of the photon with no energy loss

Elastic scattering is of limited significance in radiation treatments or in diagnostic radiology as no energy is absorbed by the medium, nor is the energy transformed into electronic motion, which are the main two vehicles for cell death in radiotherapy [38]. However, it does influence the trajectory of radiation and therefore the absorption cross-section of the medium.

### 2.5.2 Energy Deposition Mechanisms

There are several mechanisms by which photons lose their energy via their interaction with matter. Considering the relatively low energy photons emitted from LDR Brachytherapy sources, only the relevant interactions will be addressed. These interactions include the Photoelectric effect and Compton Scattering, with their kinematics shown in Figures 2.8 and 2.9 respectively.

The photoelectric effect is an interaction characterised by an incident photon ejecting an inner shell electron from an atom. The incident photons energy is fully
Figure 2.8: The kinematics of the Photoelectric effect on an atom with electron shells shown. Adapted from [39]
absorbed by the electron giving it sufficient kinetic energy to overcome the binding energy of that electron shell and is ejected from the atom [34]. The kinetic energy gained by the electron is independent of its scattering angle and can be described by the following equation:

\[ T = h\nu - E_b \] (2.7)

where \( h\nu \) is the energy of the incident photon, \( E_b \) is the binding energy of the electron and \( T \) is the kinetic energy of the ejected electron. This effect occurs most readily when the incident photon has an energy the same or just higher than the binding energy of an electron shell. The K and L shells are the first and second largest contributors to photoelectric absorption, respectively.

The photoelectric effect results in a positive ion whose electrons are not in a stable configuration due to the presence of an incomplete inner shell. This can cause an electron cascade where electrons from higher energy states that are less tightly bound and further away from the nucleus, transition down to occupy the newly vacant energy levels. The electrons require a smaller amount of energy in their new positions than they did in their previous shell. The energy difference between the two energy states is then released in the form of X-rays, which is characteristic to both the element and shells effected, the result of which makes the atom become stable once again.

Compton scattering describes the interaction of a photon with sufficient energy and an unbound, stationary electron [34]. A transfer of a proportion of the incident energy and momentum to the electron results in the scattering of both the photon and electron at angles \( \theta \) and \( \phi \) respectively, relative to the photon's original direction.

As a result of the conservation of energy and momentum, the product of the interaction is the reduction in the photon's energy and the motion of the electron. The proportion of energy transferred from the photon, or the magnitude of the photon frequency shift, is determined by the initial photon energy and the angle at which the photon is scattered by the particle, as shown through the following equation [5]:

\[ \text{equation} \]
\[
h\nu' = \frac{\nu}{1 + \frac{\nu}{m_e c^2}(1 - \cos \theta)}
\]  

(2.8)

Where \( h \) is plank’s constant, \( \nu \) is the photon’s initial frequency, \( \nu' \) is its final frequency, \( c \) is the speed of light, \( m_e \) is the mass of an electron and \( \theta \) is the angle of the scattered photon relative to its original direction.

The photons energy loss reaches a maximum when \( \theta = 180^\circ \). Due to this upper energy limit of scattered photons of a single energy, a feature known as a Compton edge can be observed when organised into an energy spectrum. This feature is characterised by a low energy plateau relative to a photopeak, where a photopeak represents the events due to photoelectric absorption. The Compton edge location is before a distinctive energy gap that is approximately the size of the maximum energy loss of the photopeak energy due to Compton scattering. Thus, a proportion of the photon’s energy being lost through this process, there is a significant potential for particularly low energy photons to be scattered multiple times. This is significant
in eye plaque brachytherapy where photons are generally under 100keV.

### 2.6 Electrons Interacting with Matter

Electrons, or beta particles, traverse through matter in ways different to massless, electroneutral photons. They can collide with nuclei directly, are affected by the Coulomb force and, like photons, have the ability to excite and ionise atoms [25]. The majority of electron interactions with matter are due to the Coulomb force, which can be described in the following formula:

\[
F = \frac{Q_1 Q_2}{4\pi\epsilon_0 r^2}
\]  

(2.9)

It can be seen from the above formula that the two charges act upon one another and both the magnitude and the polarity of the charge determine the impact of the force on the beta-particle. For the purposes of clinical treatments, and the related energy range, only collisional and radiative interactions will be considered in the following paragraphs.

Electrons undergo a multitude of interactions as they move through a material, such that its behaviour can be best described through the probability of an interaction occurring. One way to predict the passage of the electron is to consider its trajectory past a nucleus, relative to the radius of the electron cloud [25] [34]. When \( b \gg a \) as shown in Figure 2.10, beta particles can undergo ‘soft’ collisions, or can be scattered elastically by the atoms electric field. The elastic process can occur a number of times, which can result in the particle travelling a relatively large distance from where it originated. This effect is particularly significant at low energies [18]. When \( b = a \), or \( b < a \), the interaction generally results in collisional or radiative process, respectively. The energy loss per unit length of an electron is described as the stopping power, which is split into two terms; collisional and radiative as shown in equation 2.10. Both of which are dependent on both the energy of the electron and the density of the traversed material [18].
\[
\left( -\frac{dE}{dx} \right)_{\text{tot}}^{\pm} = \left( -\frac{dE}{dx} \right)_{\text{col}}^{\pm} + \left( -\frac{dE}{dx} \right)_{\text{rad}}^{\pm}
\] (2.10)

These two interactions dominate at different energies within a material. Collisional interactions generally dominate at low energies, whereas radiative interactions increase proportionally to the kinetic energy of an electron, and so are the dominant interaction of high energy electrons. The following equation approximately expresses the likelihood on each interaction taking place as a function of the atomic number of the absorber medium and of the incident energy of the electron:

\[
\frac{(-dE/dx)_{\text{rad}}}{(-dE/dx)_{\text{col}}} \approx \frac{ZE}{800}
\] (2.11)

### 2.6.1 Collisional Interactions

Electrons, being massive particles, have the potential to physically collide with surrounding particles and through a single collision, can potentially lose a large portion of their energy. Generally, collisions occur when \( b \gg a \) or \( b = a \), as shown in Figure 2.10. The transfer of energy from the incident electron to an orbital electron in a collision can result in ionisation or excitation, both of which are states that lead to the emission of characteristic X-rays and/or Auger electrons. Ionisation is where an outer shell electron is permanently ejected from the cloud resulting in an atom with a net positive charge, whereas excitation is when an orbital electron gains enough energy to exist in a more energetic orbit only temporarily. The majority of electron collisional interactions result in a small net loss of energy, however, when they do transfer a large portion of their energy to an orbital electron, this electron is known as a \( \delta \)-electron, which can then go on to have collisions with other orbital electrons \[18\] \[36\] \[34\].
2.6.2 Radiative Interactions

Bremsstrahlung, or braking radiation, is the phenomenon where an electron is deflected by the Coulomb force due to the electric field of a positive nucleus, mainly occurring when $b < a$, as seen in Figure 2.10 [36] [25]. As the electron traverses the strong electric field of the positive nucleus, it undergoes both a change in direction and a reduction in kinetic energy, as shown in Figure 2.11. Any loss in kinetic energy is directly translated into the emission of electromagnetic rays following local conservation of energy, whose direction is dictated by conservation of momentum. As there is a larger force exerted on the electron coming within a smaller distance, the angle at which it is deflected is also large.
2.7 Radiation Detectors

Since the production of high purity silicon became commercially viable, radiation detectors have been fabricated based on semiconductor technology [37]. There are a number of properties semiconductors possess that make them suitable for the detection of radiation. They (silicon) have a very small ionisation energy of 3.62 keV [37], which is the amount of energy required to produce an electron-hole pair from within the crystalline lattice [40]. In comparison, a typical gas-filled radiation detector has an ionization energy of approximately 30eV. Because the number of charge carriers produced per energy deposition is much larger, semiconductor detectors have better signal to noise ratios when compared to gas-filled chambers. They also have a relatively high density (2.33g/cm$^3$) and atomic number (14), which contribute to a large energy loss per unit length for an incident particle travelling through the medium.

Silicon detectors can be made for a relatively low cost and are highly reproducible from batch to batch. They operate well at room temperature, where other semiconductors, such as germanium, require low operating temperatures to reduce thermal current. Small sensitive volumes are achievable due to a high charge collection efficiency which in turn results in excellent spatial resolution. In addition to
desirable material properties, silicon detectors are extremely practical requiring low operating voltages and are able to output data in real time.

Conversely, there are some considerations that must be made when using silicon semi-conductors in radiation detection. The pathway of radiation energy deposition has the potential to damage the silicon substrate and the silicon oxide interface [41]. Thus, increased exposure to radiation can degrade the detector’s response. Incident radiation can displace atoms within the crystal lattice, resulting in mobile holes within the substrate binding together to form a trapping site. The result of which is increased noise, decreased sensitivity, a reduction in the carrier mobility, charge collection efficiency and the detectors lifetime [42][37].

Furthermore, the application of semiconductor devices to clinical dosimetry present some complications associated with their material composition. Much of clinical dosimetry measures the energy deposited in water, which acts as an analogue to tissue. At low energies, the dose to silicon is enhanced relative to the dose in water, due to a high photoelectric cross-section [43]. The distribution of the attenuation coefficient of silicon relative to water is represented in Fig. 2.12. This makes dosimetry in that energy region somewhat erroneous.

2.7.1 Diode detectors

Within a homogenous crystal lattice, semiconductors have a well-defined electron band structure; a conduction band, a valence band and a forbidden energy gap where allowed energy states for electrons do not exist [40]. Conductors have no energy gap, while semiconductors have a relatively small energy gap compared to insulators [36]. Impurities can be introduced into the lattice such that the substrate may have an extra charger carrier. If the impurity added is a group V element, it is defined as a donor as it introduces an extra electron, forming an n-type semiconductor. If the impurity is a group III element, it is defined as an acceptor as is introduces a positive hole, forming a p-type semiconductor. The introduction of these charge carriers also alters the energy bands within the lattice, allowing energy states within
Figure 2.12: Mass absorption coefficient of water and silicon, showing the low energy discrepancies between the two (data from [43]).

Figure 2.13: Diagram outlining the band structure of dope and un-doped semiconductor material. The pink and blue bands represent the conduction and valence bands respectively while the black circles are electrons and the white, holes
the forbidden gap [44], pictured in Figure 2.13.

When two oppositely doped regions are in direct contact, there is a build-up of charge close to the interface on either side. This is due to the diffusion or drift of both donors and acceptors across the junction, leading to the establishment of an electrical potential across the junction [44]. In this region, the concentration of charge carriers is reduced due to occupied acceptor and immobilised donor sites, otherwise known as the depletion region.

Radiation detectors are generally employed using a reverse bias configuration, where an external potential difference is applied across the sensitive volume. Under reverse bias conditions, only electrons from the p+ region and holes from n-, also known as minority carriers, are attracted across the junction [34]. This serves several purposes including; lowering the leakage current and diode capacitance, forcing the current to travel in only one direction and increasing the volume of the depletion layer [45]. The concentration of charge carriers within the depletion region decreases due to the flow of only the minority carriers, resulting in a reduced leakage current. Reverse biasing also serves to increase the potential difference across the junction resulting in an extension of the depletion region [37]. This can be seen in Figure 2.14, which outlines the effects of reverse-biasing a semiconductor. Following this increase in potential difference comes the extension of the depletion region [37]. According to Poissons equation:

$$\nabla^2 \phi = -\frac{\rho}{\epsilon}$$

(2.12)

where $\phi$ is the electrical potential, $\rho$ is the net charge density and $\epsilon$ is the dielectric constant. As the potential increases, it then follows that space charge must also increase. This increases the width of the depletion region and therefore the sensitive volume of the detector.

When radiation of sufficient energy is incident on this area, electron-hole pairs are produced that have the potential to travel relatively large distances before recombining. This is achieved by the electrical potential that forms over the junction
Figure 2.14: Diagram outlining the depletion volume, electric field, electrical potential and energy bands of the junction between the n-type and p-type regions of a radiation detector with a) no external voltage applied and b) under a reverse bias configuration. Black and white circles represent electrons and hole respectively. (adapted from [44] and [37])
due to the net charge of electrons in the p-type and holes in the n-type region. It is the current formed by the motion of these charges that allows the detection of a radiation-matter interaction.

2.8 Spectroscopic Dosimetry

Spectroscopic dosimetry is defined as the determination of the dose rate at the point of a measurement based on the number of events within the photo-peaks of a measured spectrum [46]. The energy of a source is important in non-tissue equivalent detectors, such as silicon diodes [37]. The absorbed dose in a detector may have an inherent energy dependence relative to the absorbed dose to water, and therefore the read reading taken may have no clinical significance. If the radiation energy can be determined at each point along its trajectory, then the absorbed dose to water deposited along that track can be more accurately determined, using whatever material is available [34].

When considering the absorbed dose in an LDR brachytherapy regime, the effect of a source spectra is often overlooked. Currently, there are only very limited numbers of publications referring to a spectroscopic analysis of LDR brachytherapy [6][47]. However, there are a number of advantages to approaching dosimetry from a spectroscopic perspective.

Firstly, the majority of photon-emitting radioactive sources are not mono-energetic and treating it as such can lead to an incorrect estimation of the dose. Analysing each of the major photo-peaks intensity and energy to determine its relative contribution to the dose leads to an accurate net dose measurement. When conducting depth dose measurements, each photo-peak will have a reduction in size due to attenuation in the medium. This is due to the following relationship:

\[ I = I_0 e^{-\mu x}; \mu \propto 1/E^3 \]  \hspace{1cm} (2.13)

where the attenuation coefficient of a photon, \( \mu \), is inversely proportional to the cube
of the photon energy. Because of this relationship, each of the photo-peaks will be attenuated at different rates relative to one another. When performing depth dose measurements, the lower energy photons will be attenuated the most, resulting in the largest relative decrease in photo-peak intensity.

![Figure 2.15: Spectrum of an OncoSeed 6711 brachytherapy source as measured by a silicon detector and spectroscopic set-up](image)

**Figure 2.15:** Spectrum of an OncoSeed 6711 brachytherapy source as measured by a silicon detector and spectroscopic set-up

When considering the spectra collected from a source, such as in Figure 2.15, there is information other than the photo-peak height present in the data that can be investigated. The presence of any electronic noise in the detector system would be evident in a spectrum as low-energy events. This has the potential to cause inaccuracies in the overall dose measurement due to the sheer number of these counts that can be accumulated at a very high rate. There are also a large number of Compton scattering events contributing to low energy counts, particularly in the low energy radioisotopes prevalent in LDR brachytherapy. For example, approximately 40% of photons emitted from an I-125 source undergo Compton scattering [43] and all of the recorded events associated with this interaction have a very low energy. In
order to consider only the clinically relevant events in determining the actual dose deposited these low energy counts can be ignored by setting an energy or lower level threshold to remove these low energy events from the spectrum.

Converting electronic signals to energy deposited requires a number of steps. Initially, a preamplifier system with a feedback capacitor collects the charge output by the detector. As the charge increases, the potential difference of the capacitor also increases sharply [37]. This rise in voltage is proportional to the total charge collected and in turn to the energy deposited by the radiation event. Over time, the signal exponentially decreases to pre-event levels. This decay is imperative to avoid saturation of signal due to the event signals piling on top of one another. The amplifier then shapes the signal into a form that can be easily read by a multi-channel analyser (MCA). Pulse shaping can occur using differential and integral discriminator circuits [37]. The former produces an output signal that essentially acts as a high-pass filter, where low frequency signals are- and high frequency pulses are not attenuated. This acts to remove the long decay tail of the pre-amp signal while keeping the rise of the pulse intact. The integral circuit acts as a low pass filter; attenuating high frequency signals, while allowing low frequencies to pass through unchanged. This integration would smooth a pulse with a steep gradient into having a more gradual rise. However, this will only take place if the linear input pulse, from the preamplifier, has an amplitude larger than a pre-defined discriminator level. The output signal takes the shape of a Gaussian curve, which is an average of the signal pulse produce earlier. Thus, the discriminator allows the system a certain amount of functional flexibility, whereby noise can be eliminated or certain signals can be recorded. The MCA can then bin the Gaussian signal according to pulse height, and record them as counts and display that information as a spectrum.

Running the system in this manner limits the ability of the device to operate under a high count-rate environment due to shaping time. Count pile-up is where successive pulses ‘pile up’, which occurs both as outlined above due to a signal pulse occurring before the previous one has decayed and when more than one pulse is
processed within one integration time [37]. The pulse height of this event is added to the height of the previous signal, resulting in a signal that is not representative of the energy deposited by the photons. Another consequence of a high count-rate and count pile-up is clipping, where pile-up pulses saturate the amplifier, as shown in Figure 2.16. When clipping occurs, the pulse height corresponds to the maximum output voltage of the amplifier and results in events binned in the highest channel of the MCA spectrum.

2.9 Monte Carlo Simulations

Computer modelling in physics is a rapidly growing field. These systems are a means of testing hypotheses without costly equipment and technology. It can also serve to set theoretical benchmarks and standards in ideal conditions. It has the ability to track billions of individual particles’ passage through matter from high energy ions to auger electrons. The basics of simulating radiation physics is rooted in randomness and probability, and such are named ‘Monte Carlo’ simulations, after the region home to a large casino in Monaco.

The Monte Carlo method is a technique of computing algorithms through prob-
abilistic analysis [48]. Monte Carlo simulations can be used to track the trajectory of individual particles [49]. Radiation can be modelled by randomly selecting the physical interactions it undergoes, as well as any secondaries produced. The probability of certain event occurring is determined by the radiation type and energy as well as characteristics of the medium through which it travels. By simulating the tracks of a very large number of particles, the quantity of interest (often absorbed dose in media) can be calculated. Practically, the Monte Carlo method is employed in simulating radiation transport in three steps [50].

1. Firstly, an input of information describing the model must be defined. This includes defining the relevant characteristics of media included in the model, such as water and silicon. Libraries of applicable radiological characteristics, such as the probability of energy dependent interactions, attenuation coefficients and stopping power ratios, can then be referenced. The physical geometries of the simulation environment are constructed in three-dimensions, where complex bodies are formed by adding and subtracting standard shapes and are assigned a previously defined medium. Some of these volumes may be assigned as scoring volumes, where the system records predefined dosimetric data generated in that volume.

Once the geometry of the simulation environment has been demarcated, the primary radiation must be defined. This includes the radiation-type, its energy, geometric position/source within the model, the initial direction of the radiation and the number of primary particles generated. Finally, the features of the simulation itself must be input. For radiation transport models this includes, the radiological interactions to be considered and any relevant energy thresholds/cut-offs are input, while more generally, the random number generator and its initial starting point is defined.

2. The simulation generates random primary particles and performs quantitative analysis on them. This process includes determining the path length of an individual primary particle due to predefined conditions. The probability of
an interaction occurring is calculated based on the radiological characteristics of the simulation. Then, the interactions a particle undergoes is determined by the random number generator, whereby the interaction selected is from the weighted probability model. This primary particle and any other generated as it traverses a medium, are tracked until the energy of the particle is below the cut-off threshold. Each interaction of an individual particle and the subsequent energy transmission is tabulated throughout its trajectory.

3. The information tabulated from each individual particle is then summed to produce an output. This output was determined by the set-up of the scoring volume discussed above. This data can be organised in several ways, for example the energy deposited in a volume can be binned according the magnitude of energy transferred in a single interaction to produce a spectrum.

Brachytherapy dosimetry has been traditionally based on calculations assuming a homogeneous water phantom volume following TG-43 U1. Previous Brachytherapy models using Monte Carlo have compared realistic heterogeneous mediums based on CT datasets with TG-43 U1 calculations, limited by considering only homogenous, water media. Theoretically, the influence of patient geometry and heterogeneities is higher with low energy photons associated with brachytherapy, due to the high proportion of photoelectric interactions.

Comparisons between the TG-43 U1 dosimetric calculations and Monte Carlo models for breast Ir-192 plans have been conducted, utilising treatment planning CTs to create a density map [51]. A density map was used along with a look-up table to assign elemental composition to areas with specific electron densities. The comparison revealed that TG-43 U1 over-responded in areas with a low photon energy and it was assessed as being unable to accurately handle scatter radiation. While the over-response did not influence the dose calculation of the whole plan (< 2%) deviation, local dose regions showed discrepancies from 5 - 30% as the points moved further away from the implant. In particular, TG-43 U1 did not represent the lung dose well and resulted in an overestimation of the dose proximally.
of approximately 10% and then an underestimation of the dose at points further away from the sources in the order of 30%.

Monte Carlo can also be used in retrospective studies [52] [53], where the impact of inter-seed scattering and attenuation effects in prostate permanent implant therapy was investigated. The study also examined the dosimetric effects of heterogeneous materials, such as prostate calcifications, which have been shown to significantly alter doses. As above, a voxels mass density and elemental compositions are determined by converting CT numbers. The study found that, in general, TG-43 U1 was over estimating the dose delivered to the target volume compared to Monte Carlo simulations and that these doses were varying considerably between patients.

Geant4 is a particle physics toolkit powered by the C++ language [54]. It comes pre-loaded with physical data and processes as well as a framework to create realistic environments based on geometric structures and contains multiple physics models to account for photons, leptons, hadrons and ions with energies from 250 eV to the order of PeV. Geant4 can also model detector response and brachytherapy sources and output data visually, quantitatively and in terms of dose deposited. Geant4 has been used in a number of publications to model brachytherapy seeds in heterogeneous mediums. Previously, these types of calculations have been conducted using Monte Carlo, however the large computation time of such models is such that their use is not widespread [55]. Geant4 has the advantage of being designed for such uses and with the installation of packages has libraries of information for the simulation to draw from rather than re-computing parameters for each model. Cuts, or energy thresholds can be set for secondary photons, electrons and positron production. These characteristics make Geant4 well suited to modelling in medical physics.

The Livermore package is a library for photons and electrons from 10 eV to 100 GeV for photon processes and is based on polarised processes [54]. Atomic shell structure and shell cross-sectional data is directly used for calculations in this
package, as low energy interactions are particularly influenced by these elements. Libraries used for this package include the following information:

- Cross sections for photoelectric effect, Compton and Rayleigh scattering, pair production and bremsstrahlung, including integrated subshell cross-sections for the photoelectric effect and ionisation
- Energy spectra of the secondaries for electron processes
- Scattering functions for Compton effect
- Subshell binding energies and transition probabilities between the subshells for fluorescence and the Auger effect

Physical processes modelled using this package are based on the information contained in the libraries mentioned above, but also analytical evaluation of the library data.

Lemarchal et al 2015, simulated seeds by organising physical layers in nested layers. The photons were randomly placed onto the iodine layer with a random trajectory and energy, and were then sampled. The study by Lemarchal found that when modelling seeds accurately based on their physical composition and multiple layers, the dose deposited to 90% of the volume of interest (D90) decreased by 3.2% relative to TG-43 U1 formalism. With an overall difference of -8.71% in Geant4 when compared to TG-43 U1.

Previous studies have evaluated Geant4 models of Oncura 6711 brachytherapy seeds relative to the gold standard TG-43 U1 [53]. The agreement between the two calculation models were within 5% within 50 mm of the source. The Pope et al 2015 study measured the impact of prostatic calcifications on many dosimetric parameters relative to the prediction of TG-43 U1 calculations. It was found that the presence of these calcifications lead to a dose escalation proximally from the source to the calcification and a dose decrease on the distal side due to the attenuation of the relevant photon energies by the high Z components found in prostatic calcifications.
The overall effect was a decrease in D90 by 2-4% for very small calcification volumes relative to the volume of the prostate. However, the clinical significance of this decrease in D90 is minimal.

2.10 The Eye Plaque Quality Assurance System

The Eye Plaque quality assurance protocol required a number of devices to collect and interpret data and then to output the dose rate at a certain point. The following are equipment developed by the Centre for Medical and Radiation Physics (CMRP) previously.

2.10.1 Eye Plaque Probe

The Eye Plaque Probe is a small spectroscopic detector aimed at collecting the spectra of the emitted photons and measuring the dose rate of low activity sources [6], as shown in Figure 2.17. It contains a 1 mm$^3$ p-type PIN diode mounted onto a pre-amplifier board. The detector is run on a low bias of approximately 25 V. This results in the formation of an even depletion layer whose thickness is of the order of around 100 µm. The probe holds the detector rigid with a small ceramic frame which acts both to protect the detector from physical damage and to hold it in a fixed location. A Field Effect Transistor (FET) amplification system is employed to boost the original signal before reaching the pre-amp. The pre-amp integrates and amplifies the signal through to the read-out unit for analysis.

The small sensitive volume of 100 µm contributes to a high spatial resolution, which is of paramount importance when measuring depth doses. A course spatial resolution will result in less accurate depth dose profiles, particularly at the shallow depths as they contain very steep dose gradients that are of great dosimetric interest.
2.10.2 Microprocessor Unit

The microprocessor unit was also designed and developed at the CMRP and was constructed to interpret raw dosimetry data from I-125 brachytherapy seeds. It contains a shaping amplifier and an on-board microprocessor and is pictured in Figure 2.18. The shaping amplifier receives an electronic pulse from the pre-amp and further amplifies the signal while shaping it into a Gaussian curve. The signal is then passed through to the microprocessor. There is a pulse-height threshold that is set with a manual dial adjustment where if the electrical pulse does not reach that threshold, the signal will not be recorded by the system. A pulse-by-pulse counter mode is employed by the microprocessor. An induced current counting method would be inappropriate as the very low activity of I-125 sources would not allow for the build-up of charge necessary for the registration of a count. Pulse-by-pulse also has the advantage of allowing spectroscopic analysis of the signal as the energy of a photon is proportional to pulse height allowing counts to be binned per their energy.

I-125 brachytherapy seeds emit a spectrum of energy, including photons from both the fluorescence X-rays formed from the excitation of the seed’s titanium shell and γ-rays from nuclear reactions from the nuclei of the I-125 source, although the extent of the latter’s effect are minimal. This presents a difficulty in calculating the dosimetry from such a source due to each energy photon contributing a differ-
Figure 2.18: Microprocessor unit and dose calculation system

tent amount of dose. The pulse-by-pulse method allows for individual pulse height information to be used, such that counts not originating from the brachytherapy seeds, can be discarded. The microprocessor then achieves a dose calculation by using an approximation to simplify the process. Previous studies [47] concluded that the Iodine-125 brachytherapy seed spectrum can be averaged such that each count represents the energy 27 keV. Only due to the very low energies considered (in the range of 20-35 keV), the relative symmetry of the spectrum and that the average energy occurs at the most intense photopeak, can this approximation take place while calculating the dose with accuracy.
Chapter 3

Testing of the Spectroscopic Eye Plaque Probe

During the process of characterising the QA system any inconsistencies in dose measurements, angular dependencies or deviations from expected TG-43 U1 trends were investigated.

3.1 Single Seed Calibration

In order to acquire reliable and accurate measurements from the eye plaque probe, it first had to be calibrated to TG-43 U1 specifications. Without this calibration the probe would not be dosimetrically viable and therefore could not be used to gather data accurately.

A single seed was used in this calibration as the dosimetry from one source is simpler and more easily recreated when compared to 10 seed plaque measurements. The test was conducted in a custom water bath where the seed is at the bottom of the container as shown in Figure 3.1. Slabs of solid water were banked underneath the phantom so as to ensure that any scatter from below the bath would replicate that occurring in water. The detector was set-up to be directly centred and perpendicular to the seed so as to eliminate any anisotropy and angular dependencies. The initial measurement of the dose rate was taken at 3.5 mm. This 3.5 mm is relative to the
detector sensitive volume and not the tip of the detector itself. The dose rates were recorded at 0.5 mm increments up to 15 mm. At each dose rate increment, the dose rate was displayed by the read-out unit.

The statistical analysis involved in the evaluation of each of the physical dose rate measurements was achieved by acquiring a minimum of three independent readings, x. The standard deviation of the mean was used to account for the experimental uncertainties, as it expresses the spread and stability of the data acquired. The uncertainty in the average of the three readings, \( \Delta(x) \), is expressed in the following:

\[
\Delta(x) = \sqrt{\left(\frac{\sigma}{x}\right)^2};
\]  

where \( \sigma \) is the standard deviation. When normalising the data to values obtained at 10 mm, the error was calculated by expanding the above equation:

Figure 3.1: Experimental set-up for the single seed calibration
The method outlined above for determining the uncertainty of measurements was completed for all experimental data collected for the entirety of this thesis.

\[ \Delta(x) = \sqrt{\left(\frac{\sigma}{x}\right)^2 + \left(\frac{\sigma_{10mm}}{x_{10mm}}\right)^2}; \]  

\text{ (3.2)}

FIGURE 3.2: Dose rate calibration of eye plaque probe, showing the agreement between TG-43 U1 and experimental measurements

The probe was calibrated by adjusting the gain of the read-out unit such that it matched the TG-43 U1 calculated dose at that depth. This was performed at 10 mm from the detector to the seed and was not adjusted at any time after. The calibration depth dose profile very closely resembles the TG-43 U1 calculations over the same distances as represented by Figure 3.2.

3.2 Plaque Measurement

Once the single seed calibration had been completed, the full plaque preliminary testing was initiated. The setup for this experiment was almost identical to the
single seed calibration aside from utilising a plaque loaded with 10 seeds. This plaque was placed in the mould at the bottom of the water bath. The probe was then positioned to be centred and perpendicular to the middle of the plaque, not to any seed. The design of the ROPES plaque is such that there should be a uniform dose from all directions along this central axis. All of the seeds loaded into the plaque were from the same batch and should therefore have the same activity, although realistically there was likely to be negligible fluctuations in the activity of the seeds.

![Figure 3.3: Experimental results with respect to TG-43 U1. Under-response present in the shallow depths](image)

**Figure 3.3:** Experimental results with respect to TG-43 U1. Under-response present in the shallow depths

Similar to the single seed set-up, the depth dose was taken using the read-out unit, acquiring for 10 seconds. The probe was initially positioned at 0 mm from the applicator surface, which would be approximately 3.5 mm from the sensitive volume itself. The water bath was then lowered down in 1 mm increments and three measurements were taken at each depth and then averaged. The results of the depth dose were then graphed as seen in Figure 3.3. Here, there can be seen
an under-response present in the shallow depths, which was not observed with the
single seed calibration.

3.3 Shallow Depth-Dose Effects

During characterisation it was discovered that the probe displayed an under-response
in the dose rate for shallow depth dose conditions. To investigate this further, the
geometry and layout of the eye plaque was investigated by completing depth doses
with just the outer ring of the applicator filled brachytherapy seeds, followed by a
depth dose of just the inner ring, both of which have been visualised in Figure 3.4.

![Diagram of the inner and outer eye plaque circles. Seeds 1, 2 and 3
are a part of the inner circle and are in green, while seeds 4 to 10 make up the
outer circle and are shown in orange](image)

**Figure 3.4:** Diagram of the inner and outer eye plaque circles. Seeds 1, 2 and 3
are a part of the inner circle and are in green, while seeds 4 to 10 make up the
outer circle and are shown in orange

3.3.1 The Inner Ring of Seeds Only

Using the calibrated probe and setup identical to the one used for the full plaque
tests, the inner ring of seeds were loaded into the plaque before being placed into
the mould at the base of the water bath. The probe was fixed into position above
the plaque and the water bath lowered to facilitate the change in depth as done
previously. Each depth dose was recorded 3 times on 10 second acquisitions and
the results averaged to minimise random fluctuations. The depth dose was then graphed relative to TG-43 U1 calculations for the inner ring of seeds only.

![Graph comparing TG-43 U1 calculations and experimental data](image)

**Figure 3.5:** Comparison between TG43U1 calculations and inner ring experimental data

A good agreement between the inner ring of seeds and the TG-43 U1 calculation can be seen in Figure 3.5. However, the measurement taken closest to the plaque, where each seed subtends the greatest angle relative to the probe has shown an under-response. This could suggest an angular dependence of the detector, therefore the dose due the outer ring of seeds will be investigated to confirm this. The outer ring of seeds subtends an even greater angle than the inner and should therefore give a more definitive result.

### 3.3.2 The Outer Ring of Seeds Only

Using the calibrated probe and setup identical to the one used for the full plaque tests, the outer ring of seeds were loaded into the plaque before being placed into the mould at the base of the water bath. The probe was fixed into position above
the plaque and the water bath lowered to facilitate the change in depth, as done previously. Each depth dose was recorded 3 times on 10 second acquisitions and the results averaged to minimise random fluctuations. The depth dose was then graphed relative to TG-43 U1 calculations for the outer ring of seeds only.

![Figure 3.6: Comparison between TG43U1 calculations and outer ring experimental data](image)

There is a marked under-response at the shallow depth as seen in Figure 3.6, that then improves as the distance increases until the measured results are in line with what is expected, suggesting that the probe had an angular dependence. The position of the detector relative to the angles of the photons counted is visualised in Figure 3.7. There were a number of explanations for this dependence. Firstly, within the probe the detector is supported by a ceramic frame. This frame holds the detector rigid, but also could be blocking some of the incident photons coming in perpendicular with respect to the detector.

Another factor influencing the angular dependence is the orientation of the detector within the probe. The detector was mounted on it’s side such that the sen-
Figure 3.7: Range of primary photons that the detector is exposed to, by I-125 brachytherapy seeds, at differing depths

Sensitive volume has more attenuating material on one side compared to another and this would be limiting the number of incident photons from certain angles.

Separate to angular dependence, there are some issues with the accuracy and validity of TG-43 U1 at very shallow depths. The formalism does not take into account the photons that are absorbed by their encapsulating material, the seeds used in this experiment were contained in titanium. Other factors influencing this result include the absolute positioning of the detector over the plaque as a very small deviation from the central axis can cause significant differences in the dose rate measured, particularly at those small distances.
3.4 Discussion

The initial characterisation of the probe exposed a number of issues. Due to the angular dependence displayed by the probe at shallow depths it was not appropriate for use in Eye Brachytherapy quality assurance procedures. The shallow depth doses contain sharp dose gradients and therefore need to be recorded with accuracy and confidence. However, it was found to be successfully characterised following TG-43 U1 protocol for single seed sources, which by extension makes it suitable for the dosimetry of radiation sources not containing an angular component.
Chapter 4

Design and Testing of Motorised Positioning System

4.1 Implementation

The eye plaque probe that was characterised showed many signs of having an angular dependence. This would make it unfit for use as an eye plaque probe due to the wide range of angles the photons originate from within an eye plaque.

In order to confirm this was the case under more accurate circumstances; a motorised linear stage, coupled to a rigid, aluminum stand replaced the simple hand-wound stand as the depth control. This system would hold the probe in the same xy position relative to the phantom and plaque, while moving it up in steps with pre-determined sizes with micrometer accuracy. The introduction of the system pictured in Figure 4.1, removes the positioning uncertainty as a source of error.

Due to good results obtained from the single seed characterisation using the manual stage, only plaque measurements were conducted. The simple water bath was used in the following measurements, not the full eye plaque phantom as it introduced new inaccuracies and did not relate to the determination of whether the plaque had angular dependencies.
4.2 Full-plaque Tests

The set-up for the motorised stage required a similar procedure to the previous experiments, as picture in Figure 4.2. The plaque was placed at the bottom of the water bath into the mould. Solid water was banked under the water bath and the linear stage driven to it’s zero position. The probe was fitted into the gantry by placing it in the holder and positioning it such that the tip was just touching the plaque insert, the screws around the holder were then tightened to hold the probe rigid and in place. The probe was also connected to the read-out unit as before.
The gantry’s display was then used to set the step distance, which was 0.5mm, and that prompting was required before any movement would commence.

Figure 4.2: Picture of the motorised stage controlling the z position of the eye plaque probe

The dose according to the read-out unit was recorded at each depth, with three 10 second acquisitions averaged to give the dose. An MCA was also utilised so as to reduce any error stemming from the read-out unit. The results of this acquisition are seen in Figure 4.3.

These results were compared to TG-43 U1 calculations. The use of the motorised gantry has resulted in data that conforms more closely to the TG-43 U1 prediction
Figure 4.3: Depth dose profile combining the data from both experimental sources and from TG-43 U1

than what was found previously. However there can still be seen a slight under-response of the dose rate at shallow depths, consistent with what was found with the original gantry. The angular dependence of the probe would most likely contribute to this deviation and will be further investigated by analysing the depth dose profiles of the inner- and outer-seeds separately, while using the motorised gantry.

4.3 Inner-plaque

After all but the three inner-ring seeds were removed from the plaque, the same experimental conditions as seen in the full-plaque tests including the motorised positioning system were set up. Similar to the previous experiment, 0.5 mm increments for depth dose were set and the probe situated as close as possible to the plaque along it’s central axis. The method of recording the depth dose via the read-out unit and through spectral analysis also remained consistent with previous tests. The
depth dose was recorded in Figure 4.4.

![Figure 4.4: Inner Plaque depth dose profile comparing the read-out and spectral doses with TG-43 calculations](image)

The dose-rate acquisition shows a good agreement with the theoretical TG-43 U1 predictions. However, there is a slight under-response of the read-out unit on the distal part of the graph, however the results were within experiment error. Due to the dose rate being collected using the read-out unit’s microprocessor, it can be concluded that with such low-activity sources the acquisition time of 10 seconds for the read-out unit is insufficient.

Similar to the results obtained with the simple gantry, a minimal under-response was seen between the read-out dose and the TG-43 U1 calculation in the depth dose of the inner-ring of seeds. Therefore, to determine if the eye plaque probe has angular dependency issues the outer-ring of seeds were re-tested with the motorised positioning system.
4.4 Outer-plaque

The set-up and acquisition was identical to the inner-plaque experiment. A larger under-response of the system can be seen in the outer-plaque measurements. This suggests that there is an angular dependence, as the under-response was not seen when measuring the dose rate from just the inner-plaque seeds. The results are seen in Figure 4.5.

![Figure 4.5: Depth dose profile comparing the read-out and spectral doses with TG-43 U1 calculations](image)

4.5 Discussion

Even with the improved positioning system, the under-response of the probe in the shallow depths is visible. From this it was concluded that the rudimentary positioning system improved the results, making them more accurate, but did not inherently change them. It was reasonably concluded that the probe was affected by angular dependence issues.
It therefore was necessary to build a new detector system, one that was not affected by angular dependence.
Chapter 5

Design of the New Probe

A lack of accurate and reliable results obtained from initial characterisation, made clear that original eye plaque probe would not be a clinically viable device for eye plaque quality assurance testing and that the system would have to be redesigned. The new instrumentation would have to ensure angular dependence and localisation difficulties would be minimal and would not influence the overall dose reading.

5.1 Detector

The first stage in the design of the new probe was to select an appropriate detector. Similar to the one used previously, it would have to be small enough to both place at the tip of the probe and for it’s spatial resolution to be sufficiently small. It would have to be compatible with the same circuitry system in the old probe and be able to acquire radiation in pulse mode. The most important feature of the detector was that it would have to be angular independent. Not only in photon collection, but in overall probe design. It could not be made rigid using a ceramic frame as done previously.

The detector selected was an edgeless detector as shown in Figure 5.1, it is made of n-type substrate and with p+ and an n+ junctions. One of the primary features of an edgeless detector is that one of the electrodes extend around the silicon substrate volume almost encapsulating it. The result of which is a detector
Figure 5.1: Diagram of edgeless diodes where a) is the PN and b) is the NN configuration which is angular independent as the sensitive volume of the detector should be approximately uniform in thickness in all directions. The edgeless design does come with some issues, however. The relatively small distance between the two electrodes means that the detector has a high surface current. This requires the bias voltage to be low.

The diode used in the final probe was selected from the batch, all of thickness 0.1mm. Each detector had a slight variation in size or in configuration as shown in Table 6.1. Each of these detectors underwent testing to acquire I-V characteristics and spectral measurements at difference bias voltages. The diode with the most favourable characteristics, such as energy resolution and charge collection efficiency was chosen for use in the probe.
Table 5.1: Edgeless Diode Specifications

<table>
<thead>
<tr>
<th>Type</th>
<th>Configuration</th>
<th>Dimensions</th>
</tr>
</thead>
<tbody>
<tr>
<td>NN1_XSP</td>
<td>N+ on N</td>
<td>0.5 × 0.5^2</td>
</tr>
<tr>
<td>NN1-18_MS</td>
<td>N+ on N</td>
<td>1.5 × 1.5^2</td>
</tr>
<tr>
<td>NN1-18_MP</td>
<td>N+ on N</td>
<td>1.5 × 1.5^2</td>
</tr>
<tr>
<td>PN1-56_XS</td>
<td>P+ on N</td>
<td>0.5 × 0.5^2</td>
</tr>
<tr>
<td>PN1-56_S</td>
<td>P+ on N</td>
<td>1 × 1^2</td>
</tr>
<tr>
<td>PN1-56_M</td>
<td>P+ on N</td>
<td>1.5 × 1.5^2</td>
</tr>
<tr>
<td>NN2-32_XSP</td>
<td>N+ on N</td>
<td>0.5 × 0.5^2</td>
</tr>
<tr>
<td>NN2-32_SS</td>
<td>N+ on N</td>
<td>1 × 1^2</td>
</tr>
<tr>
<td>NN2-32_MP</td>
<td>N+ on N</td>
<td>1.5 × 1.5^2</td>
</tr>
</tbody>
</table>

Figure 5.2: NN1 IV Characteristics

5.1.1 I-V Characteristics

Each of the detectors in the batch were tested to obtain their I-V characteristics, investigating the detectors suitability for use in the probe by determining the impact
of surface current and noise on their output signal. The output was measured over 10V in a reverse-biased configuration as shown in Figure 5.2, 5.3 and 5.4 for the NN1, NN2 and PN1 models respectively.

All of the detectors were depleted over a very small voltage range, of the order of 0.5 V. Generally, currents lower than 10 nA are able to be used in dosimetry, as above such levels the leakage current begins to dominate the signal. The majority of the NN diodes showed a high current response over very small voltages. This unfortunately precludes them from being used in the probe as any bias voltage would result in an unsuitable current. In addition to the magnitude of the current, the shape of the detector response is significant when considering a bias voltage in which to operate the detector. A relatively flat region relative to voltage would be preferred, as it would have a leakage current less sensitive to any variation in voltage.

![Figure 5.3: NN2 IV Characteristics](image)

**Figure 5.3:** NN2 IV Characteristics
5.1.2 Spectral Characteristics

In addition to the IV characteristics, the spectral characteristics of the batch were measured using an I-125 source in a shielded container.

The spectra shown in Figures 5.5 and 5.6. The NN detectors were operated with a bias voltage of 6V, while the PN diodes 5V. Following results of the IV characteristics taken, the high currents displayed by the majority of the NN configured diodes have made them unsuitable for spectroscopy. Detectors NN2-32_XSP, NN1_XSP and NN1-32_MS/2 displayed currents of less than approximately 5nA when operated at 6V in the IV curves and were the only of the NN configured diodes to do so. They were also the only detectors that output an observable I-125 spectrum from that group. The energy of the NN diodes is course and does not show many of the features of the expected spectrum, including many of the well-defined photopeaks.

The PN configured detectors output significantly better spectra. All but PN1-56_M/1 output a current less than approximately 5nA when operated at 5V in the
Figure 5.5: NN configured diode measured spectra operated with a bias of 6V

Figure 5.6: PN configured diode measured spectra operated with a bias of 5V
IV curves, which resulted in well defined I-125 brachytherapy spectrum. The energy resolution of the spectra shown in Figure 5.6 is excellent as even low intensity photo peaks such as the 25 keV can be easily distinguished.

Figure 5.7: Diagram showing the depletion of edgeless diodes under reverse-bias conditions where a) is the PN and b) is the NN configuration

The PN-type diode was shown to be significantly superior to the NN configuration when operated at the chosen voltages. The depletion layer within the bulk N is dependent on the configuration of the diodes, for example with the NN configuration the reverse bias depletes the detector from the outer P+ region in towards the N+, while the PN is depleted from the central P+ to the outer N+, as shown in Figure 5.7. The NN configuration results in high levels of leakage current, resulting in a
very poor energy resolution of the spectrum and very high levels of noise.

There were a number of potential candidates from the PN diodes to be used in the probe. In order to ascertain the most appropriate one, both the relative efficiency and the full width half maximum (FWHM) of the 27keV peak of the detectors were compared in Table 6.2. The relative efficiency of the detector was determined by the spectral channel number of the centroid, where the higher it is, the more efficient the detector.

**Table 5.2: PN Diode Efficiency and Energy Resolution**

<table>
<thead>
<tr>
<th>Type</th>
<th>FWHM (V)</th>
<th>Centroid Voltage</th>
</tr>
</thead>
<tbody>
<tr>
<td>PN1-32_XS1</td>
<td>0.2302</td>
<td>4.075</td>
</tr>
<tr>
<td>PN1-32_XS2</td>
<td>0.2205</td>
<td>4.343</td>
</tr>
<tr>
<td>PN1-56_S1</td>
<td>0.1906</td>
<td>4.493</td>
</tr>
<tr>
<td>PN1-56_S2</td>
<td>0.2028</td>
<td>4.463</td>
</tr>
<tr>
<td>PN1-56_M2</td>
<td>0.2576</td>
<td>3.211</td>
</tr>
</tbody>
</table>

From this simple analysis, it can be seen that most appropriate diode from the batch, with both the most favourable efficiency and energy resolution is diode PN1-56_S1, closely followed by PN1-56_S2. As this detector has the smallest FWHM and highest centroid channel of the selection.

The diode chosen will therefore be operated at approximately 4.5V. This voltage gives a good combination of energy resolution and charge collection efficiency.

### 5.2 Diode Housing

In order for the sensitive volume to be uniform in all directions, the detector has to be bonded to the probe as shown. This method of bonding required the design of a completely new probe tip in order to facilitate the new detector.

A new PCB design allows the detector to sit with the p+ electrode facing outwards, as shown in Figure 5.8. This does limit how close the detector can get to the surface of the eye plaque due to simple geometry. A 3D-printed probe tip was
designed but was limited by the curvature of the plaque itself and by a minimum thickness. The radius of the inner surface of the probe tip is 10.4mm, this curved surface will restrict the proximity of the detector to the plaque, shown in Figure 5.9. The diagonal length of the PCB was calculated to determine the placement of the detector within the probe.

Figure 5.8: Maximum length dimension of PCB Detector board

\[ \text{Hypotenuse}^2 = 4.30\text{mm}^2 + 4.73\text{mm}^2 \]

\[ \text{Hypotenuse} = \sqrt{40.8629}\text{mm}^2 \]

\[ = 6.39\text{mm} \]
Figure 5.9: The detector position within the probe due to geometric limitations

Using half the maximum length of the PCB and the radius of curvature of the inner surface of the probe tip, the minimum distance the detector can sit from the tip of the probe can be calculated.

\[ b = 3.1962\, mm, \quad c = 10.4\, mm; \]

\[ c^2 = a^2 + b^2 \]
\[ a = \sqrt{c^2 - b^2} \]
\[ = 9.896\, mm \]

\[ d = c - a \]
\[ = 10.4\, mm - 9.897\, mm \]
\[ = 0.504\, mm \]

With the plastic probe cover being 5mm thick and the detector itself 0.1 mm, the minimum distance the centre of the detector can be from the outer surface of the probe tip is 0.954 mm. A rendering of the probe casing can be seen in Figure 5.10.
5.3 Probe Construction

The construction of the probe was a delicate process as the wire-bonding of the detector is very fine and fragile. Figure 5.12 shows how the PCB with diode attached was perpendicularly bonded to the probe circuit board such that once connected it would operate in a reverse bias configuration.

Next, the housing for the preamplifier was constructed including the aluminum cylindrical cover, and the 3D-printed probe tip as well as the cabling being assembled. Initial tests for the probe were conducted within a small Faraday’s cage, that also blocked any incident visible light. Good results were seen, however once the probe was removed a lot of noise was being recorded. The noise was most likely due to insufficient radio-frequency and light shielding of the probe. The 3D-printed probe tip was translucent. This was remedied by coating the plastic cover with a thin layer of nickel paint, sealed with an enamel paint, thus eliminating the rf and light interference. An aluminum strip was placed between the nickel-covered probe tip and the preamplifier housing, with a heat-shrink rubber used to hold it firmly in place. The finished product can be seen in Figure 5.11.
Figure 5.11: Photo of the finished probe, complete with the nickel layer and black enamel paint

Figure 5.12: Diagram of reverse-biased detector arrangement of the probe with a) the PCB with the diode configuration is pictured and b) the pre-amp connectors shown at the bottom
Chapter 6

Geant4 Simulations

6.1 Verification

After the design and construction of the new probe, the Geant4 toolkit was utilised in a modeling process and eye plaque simulations were conducted. This was in order to verify and ensure that our new designs would work well, particularly with photons incident from a wide range of angles. For each simulation, the Livermore low energy electromagnetic physics model was used. The simulation included models of the photoelectric and Compton effect, gamma conversion, Rayleigh scattering and Bremsstrahlung radiation, as well as a default cut value of 0.001 mm, a low energy threshold of 250 eV and a production threshold range of 0.1 mm. The output of the program was both the overall deposited dose in the silicon volume as well as a spectrum of 400 channels, with a bin width of 0.1 keV. These settings, particularly the small cut value and energy threshold, were chosen to effectively model the low energy photons emitted by I-125 brachytherapy seeds.

A simulation containing a fully formed ROPES eye plaque was used in testing simple detector volumes as seen in Figure 6.1. Initially the simulations were relatively simple; A stainless steel eye plaque applicator, acrylic mould insert and 10 Iodine-125 seeds loaded into the insert were all included in the plaque simulation code as well as a small 0.5 x 0.5 x 0.5 mm$^3$ silicon detector volume all situated within a large, cylindrical water phantom, as shown in Figure 6.1. A total of $2 \times 10^9$
incident particles originating from the seeds were modeled.

The data was then analysed by comparing the dose output and by integrating the spectra produced. Utilising the spectral component of the program was a better representation of the QA system’s read-out and included channels 20 keV to 40 keV to mimic the energy window seen in the electronics. Both doses were the normalised and compared.

The data extracted from the GEANT4 simulations used a similar error analysis as was used for experimental measurements. However, the simulations used counts assigned to a single energy to determine the dose and then normalised to the values obtained at 10 mm. The measurement uncertainty, $\Delta N_x$, was determined by the square-root of the counts in each bin, $\sqrt{N_x}$, as defined in the following equation:

$$\Delta N_x = \sqrt{\left(\frac{\sqrt{N_x}}{N_x}\right)^2 + \left(\frac{\sqrt{N_{10mm}}}{N_{10mm}}\right)^2} ; \quad (6.1)$$

It can be seen in Figure 6.2 that Geant4 and TG-43 U1 do not agree exactly at shallow depths. This could be due to some inherent shortcomings of the TG-43 U1 formalism. Firstly, the TG-43 U1 calculation only includes the I-125 brachytherapy...
Figure 6.2: Depth Dose profile using both TG-43 U1 calculations and a Geant4 model. The model includes ten I-125 seeds, a stainless steel plaque with an acrylic applicator. The detector volume is silicon seeds and not the extended environment in the Geant4 simulation. The presence of the stainless steel applicator and the acrylic insert are not taken into account, with the former contributing to a higher dose in shallow depths due to low energy photons being back-scattered from the high Z material unlike its behaviour in water. Other shortcomings of TG-43 U1 include the definition of Air KERMA being in vacuo. This means that any calculation conducted in any medium must be corrected. In addition to this, TG-43 U1 formalism is more appropriate for determine the dose at larger distances than the ones being considered above. TG-43 U1 measures the dose deposited in water, while the volume used in the simulation better approximated the material of a silicon detector, which would also contribute to a discrepancy in dose.

To confirm this analysis, the materials of the plaque applicator, insert, and the detector volume were changed to being comprised water in the following simulation.
The distances considered in this simulation were also made to be 0.25 mm increments so as to more reliably represent a depth dose. The spectral analysis of the simulation was not conducted in this test.

Figure 6.3: Depth Dose profile using both TG-43 U1 calculations and a Geant4 model. The model includes ten I-125 seeds, with all other structures defined as water.

The simulation, as seen in Figure 6.3, shows a closer agreement between the Geant4 model and TG-43 U1 calculations. This confirms that TG-43 U1 does not allow for heterogeneities such as the stainless steel plaque. Therefore the Geant4 model will be used as the calibration baseline for the probe, with TG-43 U1 being a reference tool.

6.2 PCB Model

Construction of the probe using Geant4 modeling was performed in stages in order to evaluate each components contribution to the depth dose profile. Firstly, a realistic
detector and PCB were constructed. The board itself was modeled to be made of acrylic, and each of the contacts, gold. With the ceramic scaffold being one of the main sources of angular dependence with the probes old design, the tip of the new prototype was redesigned. The new design required the detector to be mounted to a circuit board at the tip of the probe, sitting perpendicular to the PCB.

![Detector and PCB](image)

**Figure 6.4:** Geant4 Visualisation and the schematic of the PCB

Although this led to a large reduction in the angular dependence of the probe, the gold contacts could still attenuate the incident photons sufficiently to reduce the dose at shallow depths. In these simulations the eye plaque and applicator were set to stainless steel and acrylic respectively, while the iodine seed composition remained unchanged from previous tests. The detector mounted to this board was then modeled to approximate the effect that the gold components had on the edgeless diode, the visualisation of the PCB construction can be seen in Figure 6.4.

Figure 6.5 displays the close relationship between the simple water volume and the response of the PCB with mounted detector. There is a slight under-response in the shallow depths, this could be due to the gold components of the board providing a medium for attenuating incident photons, reducing the signal reaching the detector. The spectrum modeled by the simulation of the detector board was then compared to that acquired experimentally, as seen in Figure 6.6.
Figure 6.5: Depth Dose profile using Geant4 models of the PCB and the simple silicon volume

The Geant4 model seems to predict the majority of the photo peaks seen in the experimental spectrum, including those peaks formed due to the fluorescence of the gold, detector board components.

One of the drawbacks of the detector board was that the detector itself was not mounted in the centre of the board but offset. This offset was added to the code such that the detector was centred about the central axis of the plaque. The detector board also limited the proximity of the detector to the eye plaque at the corners formed by the rectangular board hit the concave surface of the plaque applicator.
Figure 6.6: Geant4 simulation of the new probe’s measured spectrum compared to that measured experimentally

6.3 New Probe Model

Finally, the detector cover was added to the Geant4 model of the new probe, as shown in 6.7. A series of concentric cylinders, with a concave tip was created to simulate the nickel-coating, the 3D-printed plastic and the air surrounding the detector board. The latter was placed within the housing such that the detector was as close to the external tip of the probe as was physically possible in real life. The nickel layer was approximated to a thickness of 0.01 mm, while the 3D-printed plastic was approximated to an acrylic material, as the exact composition of the plastic was not released by the manufacturer.

The response of the fully realised new probe was then compared against a simple water volume as shown in Figure 6.8. The probe showed little to no radiation opacity, even with the addition of the nickel paint layer. It also appeared to be minimally angularly dependent, even with the presence of gold components within the detector board. There is, however, an over-response of the probe at 1 mm. It was unclear
why this is occurring, so a second simulation was run without the nickel-paint in order to quantify its dosimetric impact.

There are multiple reasons for examining the impact of this component as a high proportion of photons in the keV low energy range have a high probability of undergoing the photoelectric effect when interacting with matter, where the extent of this effect is significantly influenced by the atomic number or Z of the medium and nickel has a significantly higher atomic number than water.

Figure 6.9 closely compares to the water volume case, along with Figure 6.8. Together they show that the response at a distance of 1 mm was due to the presence of nickel. This was definitively not from nickel fluorescence, however, as its k-shell characteristic x-rays occur at approximately 7.4 keV and was therefore not counted in the dose calculation due to an energy threshold of 20 keV. The case without nickel shows a slight over-response, potentially due to the presence of air in the probe tip above the detector. The air, being a low density material allows more photons to scatter in but not scatter out, increasing the intensity of photons incident on the sensitive volume and thus increasing the signal.
Figure 6.8: Comparison of Geant4 simulations of the depth dose from a eye plaque of the new Eye Plaque Probe and of a simple water volume.
Figure 6.9: Comparison of Geant4 simulations of the depth dose from a 10 seed eye plaque of the new eye plaque probe without a nickel coating and of a simple water volume.

Figure 6.10: Comparison of Geant4 simulations of the depth dose from the inner seeds eye plaque of the new eye plaque probe and of a simple water volume.
6.4 Angular Dependence

The potential for any angular dependence was investigated by comparing the full probe response, with and without the nickel paint, to simple water volumes, for both the inner- and outer-seed plaque cases.

![Figure 6.11: Comparison of Geant4 simulations of the depth dose from the inner seeds eye plaque of the new Eye Plaque Probe without a nickel coating and of a simple water volume](image)

Both Figures 6.10 and 6.11 show a close comparison to the dose response of a water volume. Similar to the full plaque case, the inner plaque shows the same over-response at 1 mm with the presence of nickel, but does not show slight change in response shown in Figure 6.11.

It can be seen in Figure 6.13 that the outer plaque reading with the new probe conforms well to the water volume case, despite the presence of air in the simulation, which did not appear to introduce any angular dependencies. Figure 6.12 however, shows a large deviation from the water volume. The nickel appears to attenuate the primary photons from the outer seeds more than those from the inner seeds.
Figure 6.12: Comparison of Geant4 simulations of the depth dose from the outer seeds of an eye plaque of the new eye plaque probe and of a simple water volume.

Figure 6.13: Comparison of Geant4 simulations of the depth dose from an eye plaque of the new eye plaque probe without a nickel coating and of a simple water volume.
This may be due to the photons from the outer seeds traversing through a greater thickness of nickel as a result of obliquity. Due to the relatively low dose contribution from the outer plaque photons to the full plaque, the latter still conforms closely the water volume case.

6.5 Discussion

The series of simulations performed have validated the design and components of the new probe for use in eye plaque brachytherapy dosimetry. The over-response by the 1 mm position should not influence clinical measurements, as the phantom limits the possible proximity of the probe to the plaque to 2 mm. As stated previously, the under-response due to the angular dependence introduced by the presence of nickel paints should not influence the full plaque measurements significantly. The outer plaque demonstrated a stronger effect, however it contributes relatively weakly when compared to the inner plaque. The true thickness of the nickel layer is unknown, therefore is difficult to estimate the effect it will ultimately have on experimental results.
Chapter 7

Eye Plaque QA System

7.1 Eye Plaque Phantom

During correspondence with staff at Prince of Wales Hospital it became clear that the eye plaque must remain sterile after the brachytherapy seeds are inserted so as to reduce the risk of infection to the patient. This made the previous mode of testing unable to be implemented into a hospital environment due to the plaque being exposed to water and in no way encapsulated. Therefore, a new testing method had to be formulated before the quality assurance system could be implemented clinically.

The most appropriate way of maintaining the sterility of the plaque would be to keep it in a completely separate compartment to the water. However, this caused issues in the resulting dosimetry, as there would be a layer of material between the plaque and the probe limiting the plaques ability to take surface doses. It was concluded that as important close dose measurements are, the sterility of the plaque could not be maintained in any other way.

The plaque container had to be as conformal as possible while retaining a waterproof seal. After many design attempts, the phantom depicted in Figures 7.1 and 7.2 was developed:

The new eye plaque phantom was developed with three separate pieces, designed to maintain the sterility of the eye plaque and to achieve dosimetric outcomes. In
**Figure 7.1**: Visualisation of the Eye Plaque Phantom. Including the large water tank and two sterile plaque inserts. Phantom Legs excluded in rendering.

**Figure 7.2**: Cross-section of the Eye Plaque Phantom. Displays the interlocking pieces, including the plaque release mechanism. Legs excluded.
order to encapsulate the plaque in as thin as possible material, 3D printing the
phantom and its pieces was determined to be the most viable construction option.
The main water phantom is 140 mm deep and 175 mm in diameter water tank housed
on three supporting legs. The top area is to both be filled with water and will be
where the probe rests during measurements. This is a relatively large water area
compared to previous phantoms, this was to further emulate TG-43U1 conditions
and to reduce any outside materials influence on the data collected. Supportive
ridges were added to help the container with a full load maintain its shape. The
base of the tank is solid plastic up until the centre where there is a void. The
thickness of plastic at the edge of the void is just 0.5 mm which was the limit of the
3D printing construction modality.

![Figure 7.3: Inserts for the Eye Plaque Phantom. The small cap fits on the top
of the large piece to maintain the sterility of the plaque during testing](image)

The remaining two pieces fit together to house the eye plaque, and are specifi-
cally shaped to fit as compactly as possible. Both of the smaller pieces are able to
be sterilised along with the plaque as the plastic used in the 3D printing process is
able to comfortably withstand temperatures up to approximately 60°C. The small-
est piece is the lid, shaped to the internal contour of the plaque and is just 0.5 mm
thick. Therefore when measuring using this phantom, just one millimetre of extra
distance need be accounted for, whilst maintaining sterile conditions. The second piece, or the plaque insert shown in Figure 7.3 holds the plaque, while the lid fits tightly on top. It also has notches in its sides that sit in the ridges of the water phantom to hold itself flush to the voids edge.

### 7.1.1 Spectroscopic Eye Probe Calibration Insert

In order to make the eye plaque phantom a versatile piece of equipment, a phantom accessory has been designed and fabricated. The current Eye Plaque Brachytherapy QA system requires calibrating with a single I-125 brachytherapy seed in order to output an accurate dose rate measurement, however the inserts described above only serve to hold the plaque in place. The calibration measurement is very sensitive to the seeds position relative to the probe due to the sharp dose gradients of I-125 sources. A cap was designed to hold a single seed in the water phantom such that the probe is always situated along the its central axis. The 3D printed cap, shown in Figure 7.4 is made from the same material as the phantom and sits on top of the depression analogous to the eye in water to achieve close to TG-43U1 conditions.

![Figure 7.4: Eye plaque phantom accessory cap. The cap is designed to hold a single seed for calibration purposes](image)
Chapter 8

Experimental Measurements

8.1 Characterisation

The introduction of new components when compared to the previous design, required an investigation into any characteristics of the acquired data from the probe. In particular, an investigation into the resulting measured spectrum and angular dependence of the new probe. Figure 8.1 shows the experimental set-up utilising the new probe and the eye plaque phantom and Figure 8.2 shows the spectrum measured using Oncoseed 6711 seeds with a very low activity of approximately 0.01 mCi.

The energy resolution remains superior to the old probe even within its housing, however a number of new peaks have formed in the low energy region of the spectrum. These peaks are likely due to characteristic X-rays produced in the outer nickel coating of the probe and the the gold components present on the PCB. As they did not originate from the seed itself, we may ignore these when considering their contribution to the dose to water from brachytherapy sources, which is an advantage of a spectroscopic dose calculation system and energy thresholding.
Figure 8.1: Picture showing the experimental set-up of the new probe driven by the motorised position system, the eye phantom and read-out unit.

Figure 8.2: Graph showing the spectrum of Oncoseed 6711 seeds as measured by the new probe. Low energy characteristic X-rays resulting from the probes nickel and gold components are represented by the three low energy peaks.
8.2 Single Seed

The new probe coupled with the automated positioning system had to undergo verification before testing in a clinical strength, high-activity environment. The probe was calibrated using a single seed and compared to TG-43U1 calculations. The CMRP eye plaque phantom was used for the following measurements.

For the single seed experiment, instead of using the plaque holding insert the seed was placed on top of the depression. Care was taken to ensure that the seed was properly centered relative to the probe and the probes zero position was just touching the seed. As mentioned previously the probe is limited by it’s plastic protective sheath, and so the minimum distance it can be from the seed even when zeroed, is 1.5 mm. The step size of stage was 0.2 mm for maximum sensitivity in the high dose gradient regions.

![Graph showing the normalised dose rate of a single seed measured with the new probe, compared to TG-43U1 calculations](image)

**Figure 8.3:** Graph showing the normalised dose rate of a single seed measured with the new probe, compared to TG-43U1 calculations

The experimental data shown in Figure 8.3 matches the theoretical calculations.
well and within error limits. The slight discrepancies between the two sets of data are due to the localisation of the single seed relative to the probe. It is even difficult to localised the source of radiation within the seed itself, as the silver rod implanted with the I-125 can move within the titanium capsule. Very high dose gradients exist in the area around the seed, due to the very low energy of the photons emitted, making localisation of the seed all the more important.

### 8.3 Full Plaque

As the new probe performed well measuring the dose rate of a single seed with no angular component, a full plaque measurement was conducted. The sterile inserts were used to hold the plaque in place under the phantom, bringing the closest measurement to 2.5 mm from the diode to the plaque surface. The detector was initially positioned touching outer surface of the phantom depression, each measurement was taken at intervals of 0.2 mm.

![Figure 8.4](image)

**Figure 8.4:** Graph showing the normalised dose rate of a full plaque measured with the new probe, compared to TG-43U1 calculations
Figure 8.4 shows the depth dose as measured by the new probe. Similar to the previous detector, the results are under-responding at shallow depths when compared to TG-43U1 calculations, suggesting an angular dependence is still present, albeit very small.

8.4 Clinical Strength Measurements

Eye plaque brachytherapy, although being a low dose rate modality of brachytherapy, can use relatively high activity seeds depending on the prescribed dose and duration of treatment. Due to the pulse-by-pulse photon counting method employed by the dose calculation system, there is the potential for dead time and dose pile-up to affect the dose rate as measured by the probe. The following experiments were required to verify that the calculation system could process the large dose rates from high activity sources used clinical strength eye plaques.

8.5 Calibration

The calibration of the probe required the use of both the eye plaque phantom and the single seed cap accessory in conjunction with the motorised positioning system. Firstly, a seeds activity was measured in a well chamber and recorded before being loaded into the single seed cap. The cap was then placed inside the phantom.

The probe was secured in the apparatus arm and lowered down so that it was just touching the seed. The probe tip was then positioned 8.5 mm away from the surface of the seed (the diode being 10 mm away), as this was the calibration position. TG-43U1 calculations were completed using the measured seed activity and the gain of the read-out unit adjusted, such that the dose rate output by the system matched the calculation for a point 10 mm along the central axis of a theoretical single seed.

Once the gain had been adjusted, the dose rate was recorded at increments of 0.5 mm starting at 1.5 mm.

As done before in previous measurements, the uncertainties of the experimental
data was determined by taking the average of 3 readings. When determining the positional uncertainty of measurements, the smallest increment of the positioning or measurement device can be considered to be the limiting factor of accuracy. Double this increment was used to reflect positional uncertainty in this experiment.

These results were graphed and compared to the TG-43U1 calculations.

![Graph comparing the dose rate of TG-43 U1 calculations and experimental measurements of a single seed](image)

**Figure 8.5:** Graph comparing the dose rate of TG-43 U1 calculations and experimental measurements of a single seed

Figure 8.5 shows a relatively close comparison between the measured dose rate and TG-43 U1 calculations. There is a slight discrepancy at the shallow depths which is most likely caused by uncertainty of the diode position within the probe tip. Although the plastic sheath has notches for the detector board to sit in, due to the diode being located off-centre on the square PCB, it is likely that the diode is not completely centered laterally within the probe. Any change in the distance and position of the detector relative to the seed will result in significant deviations from the expected dose-rate, due to the steep dose gradients characteristic of low-energy radioactive sources such as I-125. Nonetheless the calibration for distances of 3 mm
and above are acceptable for clinical plaque testing.

## 8.6 Full Plaque Measurement

Once the system had been calibrated using a known-activity source, it was ready for full plaque verification testing. The plaque was loaded according to a treatment plan and placed in the plaque-holding insert for the phantom, covered with the lid and placed up into position within the phantom. The probe was lowered to be just touching the upper surface of the plastic depression housing the eye plaque. The dose rate was recorded at 0.5 mm increments, as done previously. Because of the extra layers of plastic present between the plaque and the probe, the minimum distance from the plaque to the diode was 2.5 mm.

![Graph comparing the dose rate of a clinical plan and the experimental measurements of an eye plaque](image)

**Figure 8.6:** Graph comparing the dose rate of a clinical plan and the experimental measurements of an eye plaque

The experimental dose-rate was compared to the clinical plan generated for the treatment. It can be seen from Figure 8.6 that the probe response closely
matches the planned treatment. Data measured at 3 mm was found to be under responding by 5.2%, but the calibration was sufficient for distances greater than 3 mm. The majority of the other points were well within 1% of the planned dose rate. These results confirm the eye plaque probe system as being a reliable dosimeter for use in dose-rate verification for eye plaque brachytherapy.

8.7 Introduction of Errors

To further affirm the use of the system in clinical quality assurance an error was introduced into the plaque itself and the measurements re-performed to substantiate whether the system can detect set-up errors. Some of these set-up errors come in the form of a misplaced seed within the plaque. As the plaque itself has few identifying features and the seeds have none, there is potential for the physicist to place a seed incorrectly within the applicator, and for that seed to alter the dose profile of the plaque affecting treatment outcomes.

![Diagram showing placement of seeds within the ROPES 10-seed eye plaque, in particular the inner- and the outer-ring seeds numbered 1-3 and 4-10 respectively](image)

Figure 8.7: Diagram showing placement of seeds within the ROPES 10-seed eye plaque, in particular the inner- and the outer-ring seeds numbered 1-3 and 4-10 respectively

For the first measurement, a plan was used with a mixture of 3.841 mCi and 1.556 mCi seeds. A central, high activity seed was replaced with a lower-activity seed, which has approximately 40% the activity of the former. A diagram of the
seed positioning can be seen in Figure 8.7. Because of the central position of the introduced error, the probe is expected to detect the change without issue.

![Graph comparing the dose rate of a plaque packed according to the TPS and a plaque where a 3.841 mCi inner ring seed was replaced by a seed with an activity of 1.556 mCi](image)

**Figure 8.8:** Graph comparing the dose rate of a plaque packed according to the TPS and a plaque where a 3.841 mCi inner ring seed was replaced by a seed with an activity of 1.556 mCi

A significant deviation from the planned plaque can be seen in Figure 8.8 particularly at distances of 2.5-6 mm, with a maximum deviation of 18%. This would be significant enough to warrant the clinical physicist to investigate the plaque and to re-measure the activity and placement of the seeds.

Further investigations were required to see if a misplaced seed from the outer ring would register a significantly different depth dose. This time, a mixture of seeds with activities of 4.022 mCi and 1.652 mCi were used in the plaque and a higher than planned activity seed (approximately 240%), replaced an outer ring seed and the plaque was re-measured.

Figure 8.9 shows that, even with a small difference between the activity of the seed in the planned and packed plaque, the system is sensitive enough to register the variation. A significant difference between data, approximately 6-9% at each
Figure 8.9: Graph comparing the dose rate of a plaque packed according to the TPS and a plaque where a 1.652 mCi outer ring seed was replaced by a seed with an activity of 4.022 mCi

measurement point, can be seen even with the smaller contribution to the changing dose rate due to the position of the seed.

8.8 Discussion

The newly designed probe responds well in a low activity environment. The spectrum collected was superior to the previous spectra produced using the old probe and displayed a fine energy resolution where the peaks were easily resolved. The high Z components of the new probe did not produce a signal that interfered with the Oncoseed spectrum and were easily ignored using the dose threshold.

The single seed depth dose profile, as collected by the new probe closely matched the TG-43U1 calculation. Any uncertainty seen in that data can be attributed to the positioning of the seed relative to the probe. In the future the eye phantom single seed cap will be employed to improve the accuracy of measurements.
Full plaque measurements utilising the sterilisation inserts of the eye phantom and the new probe showed a fair agreement with the TG-43U1 calculation. In determining the angular effects of the probe, it can be concluded that there is a slight dependence in distances lower than approximately 3 mm. However further investigation was required with higher activity sources, as the statistics collected in this experiment were low.

The eye plaque brachytherapy QA system developed by the CMRP can be used clinically for dose rate verification of eye plaques prior to their insertion within a patient. This has been shown by the successful implementation of the phantom sterility system. In addition, the system displayed the ability to detect plaque packing errors, where incorrect seeds are loaded into either the inner- or outer-sections. While some uncertainties from detector localisation and the presence of an air gap within the probe tip exist, the results from the system once calibrated closely matched the dose of a known source and clearly registered errors when introduced.
Chapter 9

Detector Optimisation

Although the clinical strength results performed at POW hospital were within 1% of the planned dose in the calibration range, measurements acquired at small distances from the plaque showed an under-response of approximately 5%. While this does not change the efficacy of the QA system within the calibration range, an investigation into the causes of this discrepancy was initiated. As with the classic probe, mounted with a PIN diode, it is possible that the newly constructed probe has an angular dependence. This could be due to the inherent shortcomings of the diode, but this has been documented to fall within ± 2%, not the 5% evident. Other factors can be that the shielding of the probe, or the construction of the PCB within the probe could be contributing to an apparent angular dependence.

9.1 CT Scan

To assess the probes construction qualitatively, a CT scan was performed. This was to characterise both the construction of the components and the position of the diode within the probe. Figure 9.1 shows a cross-section of the probe as it is laid on the CT bed. From this image, the position of some of the vital components within the probe tip cannot be ascertained, however the general position of the preamplifier and circuit board is approximately within the centre of the outer container.

The cross-sectional position of the probe tip was also investigated. A diagram
Figure 9.1: CT image of the new probe

Figure 9.2: Diagram outlining how the position of the CT cross-section relates to the construction of the probe
outlining how the image relates to the construction of the probe can be seen in Figure 9.2. This proved difficult due to the amount of high-Z material present in the area and the small dimensions of the components. To improve the spatial resolution and reduce the impact of high-Z artifacts, the CT was set to use large exposure settings that would not be suitable for most patients as the dose delivered to them at this time would be very high. The CT settings required to acquire the image seen in Figure 9.3 include:

- 550 mAs
- 140 kV
- Pitch of 0.35 mm
- FoV of 103 mm

The nickel coating of the plastic, protective sheath can be identified as the outer circle as seen in Figure 9.3. It can be deduced from the heterogeneous intensity of that circle that the nickel coating is not completely uniform at this position on the
probe, contributing to the possibility of an angular dependence. Within the outer circle there can be seen four approximately rectangular, high intensity areas. The larger of the two, most likely represent the position of the contacts of the PCB and the tin solder adhering it to the main circuit board. The high intensity of the CT in these areas suggest that radiation can be easily attenuated and scattered, perhaps contributing to some angular dependence. However, the most significant finding of the CT image is that the central rectangular area, that is surrounding by the four previously identified, is most likely representative of the diode position. The introduction of indents into the plastic probe sheath, as discussed in Chapter 5, was to position the diode into the very centre of the probe. However, the CT image indicates that the diode is not in the centre of the probe, but to one side.

Figure 9.4: Cross-section of new probe tip showing the spatial position of the detector

Using ImageJ™, U.S. National Institute of Health, the position of the diode can be quantified as seen in Figure 9.4. The image analysis determined the diode to be approximately 0.345 mm offset from the centre of the probe, which is the same
offset of the detector from the centre of the PCB.

9.2 Distance Corrections

It can be reasonably deduced from the above investigation that the diode is not sitting in the centre of the probe, this directly effects the distance between the detector volume and the radiation source. In order to correct the displacement of the diode from the centre of the probe, the distance of the probe from the plaque was adjusted. These distance corrections were calculated using Pythagoras’ theorem, based on the distance from the centre of the sources to the detector along central axis and the radial offset. Figure 9.5 shows the corrected experimental values taken from POW clinical strength measurements.

![Figure 9.5: Depth dose showing the comparison between the TPS and experimental measurements with a distance correction applied](image_url)

The result of the distance correction shows a close relationship between the TPS and experimental results. Small discrepancies between the two data sets can
be attributed to the fact that the calibration was to 10 mm as measured by the stage, not including the correction. In order to quantify the difference between the experimental and theoretical values, an exponential fit was used to approximate the dose rate that would have been present at the integer distances, the interpolated points are shown in Figure 9.6.

Figure 9.6: Depth dose showing the comparison between the TPS and theoretical fit of experimental measurements
Table 9.1: Corrected distances from the eye plaque to detector sensitive volume.

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<th>Original distances (mm)</th>
<th>Corrected distances (mm)</th>
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9.3 Discussion

It can be seen that the difference between the theoretical dose rate and experimental is well within 2% for the full plaque measurement, which is in keeping with the limitations of the edgeless diode. It can be reasonably ascertained that there is some angular dependence in the new probe, as demonstrated by the shallow readings, but that there are significant uncertainties involved in the measurement of low energy sources such as I-125. Therefore an accurate, realistic determination of distance is required for eye plaque brachytherapy QA using this spectroscopic dosimetry system, followed by a precise calibration.
Chapter 10

Conclusion

The aim of this thesis was the development and implementation of a spectroscopic, fast read-out dosimetry system for quality assurance of I-125 eye plaque brachytherapy and that maintained sterility so required for implantation of the plaque clinically. This was achieved utilising edgeless diodes and 3D printer technology. Furthermore, the use of 3D printing has allowed the system to be robust, as a large number of the integral components can be accurately and precisely produced at a relatively low cost. The system has been successfully verified through clinical activity testing and through extensive computer modeling using the Geant4 toolkit.

Maintaining sterility of the plaque whilst maximising the range of clinically relevant distances for the probe to measure across was a challenge. The more protected the plaque was from the environment, generally the thicker its case would have to be. However, the 3D printing technology allowed just 1 mm of plastic to separate the plaque from a water environment, furthermore it protected the edgeless diode and electronics within the new probe by just 0.5 mm.

It was found during the course of this research that the probe previously developed by the CMRP mounted with a PIN diode was not appropriate for use in eye plaque brachytherapy QA due the angular dependence displayed at small probe to eye plaque distances. A number of factors contributed to this dependence such as the presence of a ceramic scaffold and the limitations of the PIN diode itself.

It was determined that a large source of error in I-125 dosimetry and qual-
ity assurance, is the definition of distances between the dosimeter and the radiation source. The very low energy of these LDR sources result in very steep dose gradients. Significant deviations of measured dose-rates from theory was often a consequence of any miscalculation of the distance between the two points. The motorised gantry and phantom calibration cap minimised the influence of positioning errors, by introducing precision and reproducibility. Furthermore, ascertaining the exact location of the detector position within the probe and taking into account the position of the radioactivity within the brachytherapy seeds themselves produced more accurate results, to within 2% of predicted.

Computer modeling using the Geant4 toolkit acted as a confirmation of experimental results and investigated the physical effects of a heterogeneous probe and environment. Initial comparisons between the gold standard, TG-43U1 and the Geant4 model for dose to water in a water showed small differences between the two calculation methods. The impact of nickel paint shielding the detector from rf radiation, proved to be minimal, contrary to Geant4 models. Most likely, the actual thickness of the nickel layer is thinner than the 0.01 mm simulated. Despite this, the close comparison confirmed the use of Geant4 as a legitimate simulation engine.

The technology produced in the course of this research can be applied more broadly than just I-125 eye plaque brachytherapy quality assurance. It can be used for dosimetry in any application utilising I-125 brachytherapy sources. It has potential to be used as a dosimeter for Palladium-103 sources with a simply recalibration and application of a new bias potential, this is due to Pd-103 spectra and relatively low energy emissions. The system is also not compatible with radioisotopes other than the two mentioned. For the dose calculation system to function properly, the spectrum must contain only low energy peaks, located in close energy proximity and whose average energy was similar to its median energy.

The future direction of this project would be the commercialisation and clinical implementation of the system. Development of a QA protocol in collaboration with POW hospital.


D. P. Kroese, T. Brereton, T. Taimre, and Z. I. Botv. Why the Monte Carlo method is so important today. WIREs Comput Stat, 6, 2014.


