Neutron Dosimetry for an 18 MV Medical Linear Accelerator

Vanja Gracanin
University of Wollongong

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Neutron Dosimetry for an 18 MV Medical Linear Accelerator

Vanja Gracanin

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Supervisor:
Distinguished Professor Anatoly Rosenfeld

Co-supervisor:
Dr. Linh Tran

The University of Wollongong
School of Physics - Centre for Medical Radiation Physics (CMRP)

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Declaration

I, Vanja Gracanin, declare that this thesis is submitted in partial fulfilment of the requirements for the conferral of the degree Doctor of Philosophy, 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.

Vanja Gracanin

April 10, 2019
Abstract

Electron Linear Accelerators (linacs) used in Radiotherapy treatments produce undesired photoneutrons when they are operated at energies above 8 MeV (electron mode) and at energies above 6 MV (photon mode). These neutrons contaminate the therapeutic beam and increase both in and out of field dose to patients. Dosimetric characterisations of these high energy beams, as well as investigations about neutron contamination are highly important for medical physicists working in the hospital. Primarily the need is to detect these neutrons and be able to estimate the neutron dose equivalent.

This thesis investigated three different types of neutron detection technologies for possible use in the characterisation of photoneutrons produced by an 18 MV linac. The detectors studied in this thesis include: silicon p-i-n diodes, 3D silicon detectors and a directional neutron detector based on three position sensitive $^3$He tubes within a single sphere of high density polyethylene. In addition, a Geant4 Monte Carlo simulation code was written to study the radiation field produced by an 18 MV medical linac. Separate simulations were used to study the response of each detector’s response to the 18 MV radiation field.

It was found that the silicon p-i-n diodes provided the best method for real time estimation of fast neutron absorbed dose and neutron dose equivalent in field. The 3D silicon detector ($10 \, \mu m$ ultra-thin detector) was found suitable for out of field neutron detection measurements (2m from isocentre) but could not be used in field. The directional neutron detector was found to be limited by its susceptibility to pulse pileup effects from photons, as well as the inability to acquire enough statistics in a low neutron fluence environment.

It was concluded that out of the three detectors studied, that the application of p-i-n diodes for passive real time in vivo neutron dosimetry on an 18 MV medical linac is suitable on the surface of a patient body or in cavities with constant temperature.
Acknowledgments

I would firstly like to thank my supervisors Distinguished Professor Anatoly Rosenfeld and co-supervisors Dr Linh Tran and Professor Michael Lerch from the Centre for Medical Radiation Physics (CMRP), for their ongoing support, valuable and informative discussion, patience, guidance and positive contribution made to both the project and my education in the field of Medical Physics. Professor Anatoly has been the guiding force behind this thesis, investing many hours in explaining physics concepts to me, critiquing my work and helping me develop my understanding and analytical capabilities. Dr Linh Tran has greatly assisted me by being a patient guide and understanding teacher. Her day to day help in the lab and performing experiments has been a great help and I am grateful to have worked with her.

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a directional neutron detector.

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Publications


V. Gracanin, S. Guatelli, D. Prokopovich, A.B. Rosenfeld, A. Berry., *Development of a Geant4 application to characterise a prototype neutron detector based on three orthogonal 3He tubes inside an HDPE sphere*, Physica Medica: European Journal of Medical Physics, Vol. 33, p189196, Published online: January 3, 2017
Conferences

SSD 18 - 18th International Conference on Solid State Dosimetry, Munich Germany, Presented a poster: A convenient verification method of the entrance photo-neutron dose for an 18MV medical linac using silicon p-i-n diodes, 3rd of July to 8th of July 2016


IEEE Nuclear Science Symposium and Medical Imaging Conference, San Diego USA, Presented a poster: Modeling a Directional Neutron Spectrometer Using Geant4, October 31st to November 7th 2015

MedPhys - The 9th Student Research Symposium of the NSW/ACT Branch of the ACPSEM, Sydney Australia, Oral Presentation: Neutron Dosimetry for an 18MV Medical linac, 3rd December 2015

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Chapter 1

Introduction

1.1 Vision of the Project

The vision of this project is to characterise the photoneutron energy spectrum for both in and out of treatment fields produced by an 18 MV medical linac. To do this, three different types of neutron detectors were investigated and evaluated for their potential use as an Quality Assurance (QA) device that can be used to estimate the neutron risk associated with high energy photon beam delivery for both patients and clinicians.

1.2 Motivation

At present Radiotherapy with photon and electron beams represents the most popular methods to control and treat cancerous tumours. High energy accelerators are used to improve treatment efficiency. In the past, high energy linacs were produced with beam energies between 15 MeV and 25 MeV, however most modern linacs typically work between 6 MV and 15 MV to minimise photoneutron production. There are several advantages of using high energy beams to treat cancerous tumours, including: lower skin dose, higher depth dose, smaller scattered dose to tissues outside the target volume and less rounded isodose curves. [3] However during production of high energy photon beams above 6 MV, linacs also produce undesirable photoneutrons which are a cause for concern for both patients and clinicians.

These photoneutrons are generated via nuclear reactions between photons and target nuclei, which contaminate the therapeutic beam. As a consequence, healthy tissues and radiosensitive organs will also receive a non-negligible neutron dose. The dose due to neutrons has high radiobiological effectiveness (RBE) in the energy range of 100keV to 2 MeV, as stated in the ICRP 60 publication. [1] This publication states the values of the neutron dose equivalent, per tissue dose at isocentre to be between 1 and 4.8 mSv
The value of neutron dose equivalent is dependent on the energy of the beam, accelerator components, room shielding and distance from the isocentre. The International Electrotechnical Commission recommends limits for the neutron-absorbed dose in the patient plan [2], however neutron measurements are rarely performed in radiotherapy departments due to a lack of sensitive and convenient measuring equipment for routine use.

This is of particular concern and is an issue that requires more quantitative research to both provide a detailed understanding of the neutron spectrum in a treatment room and the contribution of neutrons to patient dose. To do this a sensitive neutron detector needs to be designed, developed and tested.

1.3 Thesis Outline

The work described in this thesis focused on dosimetry considerations for photoneutron production from high energy photon beams produced from an 18 MV medical linac. The application of three neutron detectors; silicon p-i-n diodes, 3D silicon neutron detectors and a directional neutron spectrometer were investigated to assess their suitability to be used as a neutron dosimeter in the mixed photoneutron field produced by an 18 MV medical linac.

Chapter 2 of this thesis presents a literature review on the topic of neutron dosimetry in mixed field radiation. An overview of radiotherapy and the photonuclear effect is given, specifically the production of photoneutrons, energy distribution of photoneutrons and the effect of photoneutrons on biological tissue is discussed. Photoneutron dosimetry detection methodologies are also discussed, with an introductory focus on the recent development of silicon detection technology.

Chapter 3 outlines the basic user action classes used in Geant4 Monte Carlo (MC) simulation code, with a summary of the methods and techniques used to model an 18 MV medical Varian linac. To study the response of neutron detectors placed both on the surface and inside a solid water phantom, Geant4 version 10.01. p01 was adopted as MC code. An 18 MV medical Varian linac was modelled and the neutron, photon and electron energy spectra were modelled both on the surface and with depth inside a solid water phantom. These energy spectra are used to study the response of neutron detectors in chapters 4 and 6 of this thesis.

Chapter 4 of this thesis investigated two types of silicon p-i-n diodes and evaluated their potential use in a mixed photoneutron field produced by an 18 MV linac. In this chapter the silicon p-i-n diodes sensitivity to light, temperature and current voltage characteris-
tistics (I-V's) prior to irradiation were characterised. The response of these detectors was studied in mixed photon-neutron radiation fields. The silicon p-i-n diode detectors were shown to have excellent discrimination between fast neutron and photon radiation, with sensitivity to fast neutrons being $\sim 4000$ times higher than to photons from a $^{60}$Co source in terms of absorbed dose to tissue. The neutron tissue absorbed dose and neutron tissue dose equivalent were studied both on the surface and inside a cubic solid water phantom, using both experimental methods and Geant4 MC simulations.

Chapter 5 of this thesis investigated the use of 3D silicon detectors for the detection of photoneutrons from an 18 MV medical linac. The detectors were characterised by performing I-V & C-V measurements, as well as spectral characterisation using an alpha and neutron source. Three types of 3D detectors were investigated in this work: 1) 3D ultra-thin (10 $\mu$m thickness with removed silicon substrate) 2) 3D 20 $\mu$m thick (with silicon substrate) 3) 3D silicon Micro-structured detector. 3D silicon detectors have small active region of either 10 $\mu$m or 20 $\mu$m thickness which is constructed with small 3D columnar electrodes. These characteristics allow the detector to be fully depleted at very low bias voltages. The 3D ultra-thin detector with removed supportive silicon substrate and 10 $\mu$m thick active area, allowed for the rejection of most high energy photons entering the detector. The 3D detectors investigated in this study were coupled with a range of different polyethylene convertors to assess the most suitable convertor thickness for neutron detection. The 3D detectors coupled with convertor screen could present a useful method of neutron detection in a mixed photoneutron field produced by an 18 MV medical linac.

Chapter 6 of this thesis will investigate the characterisation of a directional neutron detector for possible use in a mixed photoneutron radiation field produced by an 18 MV medical linac. This study involves modelling a directional neutron detector based on three position sensitive $^3$He tubes located along three perpendicular axes within a single moderating sphere of High Density Polyethylene (HDPE) using Geant4 MC code. The simulation has been validated with respect to experimental measurements. A feasibility analysis of this detector for measurement of neutrons in mixed pulsed radiation field produced in an 18 MV medical linac was investigated.

Chapter 7 includes a detailed discussion and conclusion of this study. Future recommendations are discussed.
Chapter 2

Literature Overview

2.1 Radiotherapy

2.1.1 Introduction to X-ray Radiotherapy

X-ray Radiotherapy treatments involve using high energy X-rays with penetrating characteristics, that are required for treatment of deep-seated tumours. The electron linear accelerator (linac) is currently the modality of choice for the production of high energy X-rays in radiotherapy applications. They are currently the most popular as they have multiple electron and photon energies available, that allow the physician to tailor to the required treatment depth.

In a linac, beams of electrons collide with a tungsten target, producing Bremsstrahlung photons, which are used to deliver a therapeutic dose to the patient, whilst avoiding to damage to healthy tissues surrounding the tumour. Electron beams are utilized for tumours located near the skin. Most linac's have a sharper dose fall-off at the beam edge and can employ high dose rates (1 to 10 Gy per minute), enabling shorter treatment times. [3]

High energy accelerators are used to improve treatment efficiency, with most modern medical linacs using beam energies between 6 MV and 15 MV. There are several advantages of using high energy beams to treat cancerous tumours, including: lower skin dose, higher depth dose, smaller scattered dose to tissues outside the target volume and less rounded isodose curves [3] [4]. However during production of high energy photon beams above 6 MV, linacs also produce undesirable photoneutrons which are a cause for concern for both patients and clinicians. These photoneutrons contaminate the therapeutic beam and result in delivering a non-negligible neutron dose to the total patient dose. Knowledge of the neutron spectrum and spatial distribution, which contaminates the beam, is needed to evaluate the increase to patient equivalent dose and to ensure that the room shielding is adequate.
Alternative radiotherapy treatments include Fast Neutron Therapy that utilises fast neutrons to treat cancer patients in just a few centres world-wide. Much for frequently charged particle therapy via the use of proton beams is frequently used. Secondary particles produced from fast neutron interactions create particles, which have a very high Lineal Energy Transfer (LET). The damage is mostly direct, leading to better destruction of hypoxic cells. However the ability of these beams to spare late responding tissues via fractionation is compromised compared with X-rays.

2.1.2 Production of photoneutrons

During the production of high energy electron or photon beams, linac's also produce undesired photoneutrons. These photoneutrons have energy dependant macroscopic absorption and scattering cross-sections, which impact the mean free path of the photoneutrons. As a result neutrons may undergo many interactions before losing energy and becoming absorbed. Secondary particles produced from photoneutrons, have high LET, making them a significant secondary particle that increases patient dose. Photoneutrons are generated in nuclear reactions between photons and the target atomic nucleus, through giant dipole resonance (GDR). The figure below illustrates the general shape of the cross section for photonuclear interaction as a function of photon energy. At lower energies (up to about 8 MeV), photon-induced excitation of the nucleus is the dominant effect. Occurring at higher energies, the most prominent feature of the cross section for photonuclear absorption is the giant dipole resonance (GDR).

![Figure 2.1: Cross section for photonuclear interaction as a function of photon energy](image)

The GDR cross section exhibits quite a broad peak, centred around 20-25 MeV for low-Z media and about 12 MeV for high-Z media (Fig. 2.1), with main peak of around 3-9
The reaction cross section increases with target atomic number $Z$. In the energy region of the GDR, $(\gamma, n)$ cross section for high $Z$ elements is a factor of ten higher than for low $Z$ materials [5]. Thus materials in the linac head and beam collimation components are the main source of undesired photoneutrons. High energy photons are also able to interact with the treatment walls, treatment couch and the patient body to create photoneutrons, also affecting the neutron distribution. In particular there is relevant backscatter from the walls. The production of neutrons is a cause for concern in medical physics, and the source of hesitation when employing higher energy radiotherapy beams.

### 2.1.3 Relevance of the Photonuclear effect

For radiotherapy applications, medical linear accelerators are available for the production of bremsstrahlung photon beams up to around 25 MeV. As previously explained, this covers the giant dipole resonance region, depicting that photonuclear interactions are most likely in the energy span used in current radiotherapy treatments.

The photon beam produced by a medical linear accelerator will undergo a number of possible photon interaction processes including, Compton scatter, photoelectric effect and pair production. For a comprehensive summary of these interactions, see Knoll [21] and Podgorsak [22]. These interactions will be the most dominant processes in comparison to the photonuclear effect. This is due to the photonuclear cross section comprising only a small fraction of the total mechanisms for photon interaction.

As is evident by (Figure 2.2), the overall photonuclear effect is not negligible. For photons interacting within the linac head (high $Z$ materials) this constitutes approximately 6%
of the cross section, whilst for photons interacting in biological media (low-Z materials) the contribution is between 2-3 %. These percentages were estimated from the photon interaction cross section Data for lead and carbon from Hubbell [7].

This emission of neutrons will have an appreciable effect on the overall patient dose, depending on the energy of the neutron. Neutrons are classified according to their energy, see table 2.1

<table>
<thead>
<tr>
<th>Energy regime</th>
<th>Energy range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cold</td>
<td>$5 \times 10^{-5}$ eV ≤ $E_k$ ≤ 0.025 eV</td>
</tr>
<tr>
<td>Thermal</td>
<td>≈ 0.025 eV</td>
</tr>
<tr>
<td>Epithermal</td>
<td>1 eV &lt; $E_k$ &lt; 1 keV</td>
</tr>
<tr>
<td>Intermediate</td>
<td>1 keV &lt; $E_k$ &lt; 0.1 MeV</td>
</tr>
<tr>
<td>Fast</td>
<td>$E_k$ &gt; 0.1 MeV</td>
</tr>
</tbody>
</table>

The photoneutron energy spectrum has a peak ∼1 MeV, however at the patients plane, after the transmission through the accelerator head; neutrons have a distribution similar to that of the heavily shielded fission source [8]. Consequently healthy tissues receive a neutron contribution that can represent a non-negligible dose to the radiosensitive organs. [8]

This increase in dose is mostly due to the high Linear Energy Transfer (LET) created from fast neutrons, where LET is the amount of energy that an ionizing particle transfers to the material traversed per unit distance. Low LET radiations will cause irradiated cells to suffer sub-lethal (repairable) damage, whilst High-LET will result in malignant mutations, with DNA damage on cellular level being largely irreparable. [3] The NCRP 116 recommends a quality factor of 20 for photoneutrons in the energy range of 0.1-2.0 MeV, which is produced in radiation therapy with photon beams [9]. They are highly penetrating with high radiobiological effectiveness (RBE). Where RBE is the ratio of biological effectiveness of one type of ionizing radiation relative to another, given the same amount of absorbed energy. Their contribution in patient out of field dose is smaller than scattered photons but considering their quality factor of 20, gives a significant contribution in patient effective dose and consequently increases the risk for the development of secondary induced fatal cancer.
2.2 The Photonuclear Effect

2.2.1 The Photonuclear effect in linear accelerators (linac’s)

The majority of photoneutrons will be produced in the architecture of the linear accelerator. The main sources of production are from high atomic number components, including the target, primary collimator, secondary collimators, wedges, blocks and multi leaf collimators. Tungsten (W) and lead (Pb) have high cross sections for \((\gamma, n)\) reaction, in comparison to biological tissues and thus are the main source of production of photoneutrons in medical linac’s. Although other elements such as iron, copper and aluminium are present, their cross section for photoneutron production is negligible, in comparison to W and Pb. A schematic of the various components involved in the linac Beam Delivery System is shown in (Fig. 2.3).

![Schematic of linac Beam Delivery System](image)

**Figure 2.3:** linac Beam Delivery System in two dimensions [3]

Not all photons produced by a linac will make it to the patient plane. The number of photons interacting and depositing their energy within collimation devices is significant. Neutrons interact weakly with atoms of higher atomic number Z [21], however neutrons have a much higher interaction probability with lighter or smaller Z elements such as hydrogen and oxygen. [5] Thermal neutrons \((E_k = 0.025\text{eV})\) are produced by the interactions with such hydrogenous materials, whilst Intermediate\((0.5\text{eV} > E_k < 10\text{keV})\) and fast neutrons\((E_k\sim\text{MeV})\) are produced by interactions with medium to high-Z materials,
such as W and Pb. Thus photoneutrons reaching the patient will be mostly due to transmission through W and Pb materials, as they both have very low absorption cross sections for the energy range of neutrons produced in the linac head.

Monte Carlo methods have been extensively used to evaluate the photoneutron characteristics in radiation therapy. Studies have shown that the neutron source strength (Q value) varies between linac model, location of scoring cell, field size and modelling geometry. One author Pena [23] calculated the contribution of different linac components of a Primus linac operating in a 15MeV photon beam for a 10cm x 10cm² field size. The contribution of different linac components in neutron source strength were reported as primary collimator 52%, secondary collimator 21%, target 12%, Multi leaf collimators (MLCs) 6.6%, shielding 5% and flattening filter 0.41%. Becker [24] in a similar study with the Primus linac confirmed this result. Comparison with studies performed by Mao on a Varian 2100C/2300C linac [10], reveal a similar overall trend of each contribution but there are small discrepancies for different components. The results obtained by Mao are summarised in table 2.2, for a Varian 2100/2300 linac, for beam energy between 15-20 MeV, with closed jaws.

<table>
<thead>
<tr>
<th>Component</th>
<th>15 MeV</th>
<th>18 MeV</th>
<th>20 MeV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Target</td>
<td>9 % (W,Cu)</td>
<td>16 % (W,Cu)</td>
<td>17.2% (W,Cu)</td>
</tr>
<tr>
<td>Primary Collimator</td>
<td>38% (W)</td>
<td>41% (W)</td>
<td>36% (W)</td>
</tr>
<tr>
<td>Flattening Filter</td>
<td>22% (W)</td>
<td>9% (Fe,Ta)</td>
<td>10.4% (Fe,Ta)</td>
</tr>
<tr>
<td>Jaws</td>
<td>29% (W)</td>
<td>35% (W)</td>
<td>36% (W)</td>
</tr>
<tr>
<td>Other (shielding, etc)</td>
<td>1.20%</td>
<td>1.40%</td>
<td>1%</td>
</tr>
</tbody>
</table>

The results obtained for the target and flattening filter in a Varian linac show higher contribution relative to the Primus. Both studies have shown that the primary collimator is the highest contributor to photoneutron production amongst all other components. This is followed by the secondary collimator, target, MLC shielding, flattening filter and other components. It is apparent through all studies that the probability of photoneutron production will increase with increasing photon energy.

### 2.2.2 Photoneutron production for different linac models

In order to analyse the origin of neutrons reaching the patient, Geant4 MC methods are used to model the different components of a linac, initiate the primary electron beam
striking the target and then following the history of the particles, until they are completely gone. Information about the interactions and number of generated particles as well as the deposited energies in different parts of any defined volume can be tallied and provided at the end of the simulation. Several studies using MC and experimental methods have analysed the contribution of different parts of a linac in neutron fluence received by a patient (at isocentre), for a number of commercial linacs. [10], [11], [12], [13]

In order to compare a number of commercial linac models, a quantity for neutron production, known as the source strength, Q, is used. The source strength has been defined by McCall [12] as the number of neutrons at the isocentre, originating from the linac head per X-ray dose delivered at the isocentre. Neutron source strength is an important factor used in neutron dose calculations for both shielding purposes and patient out of field dose calculations. The results for a number of studies on neutron source strength for a number commercially used linacs is summarised in table 2.3 below. It should be noted that the Siemens and GE LINACS are no longer produced.

<table>
<thead>
<tr>
<th>Manufacturer</th>
<th>Model</th>
<th>Energy (MeV)</th>
<th>Q(n Gy⁻¹)</th>
<th>Study</th>
</tr>
</thead>
<tbody>
<tr>
<td>Siemens</td>
<td>MD</td>
<td>15</td>
<td>2E+11</td>
<td>Followill et al. (2003)</td>
</tr>
<tr>
<td>Siemens</td>
<td>KD</td>
<td>18</td>
<td>8.8E+11</td>
<td>Followill et al. (2003)</td>
</tr>
<tr>
<td>Siemens</td>
<td>Primus (MLC)</td>
<td>15</td>
<td>2.1E+11</td>
<td>Followill et al. (2003)</td>
</tr>
<tr>
<td>Varian</td>
<td>2100C/2300C</td>
<td>20</td>
<td>1.2E+12</td>
<td>Mao et al. (1997)</td>
</tr>
<tr>
<td>Varian</td>
<td>2100C/2300C</td>
<td>18</td>
<td>1.2E+12</td>
<td>Mao et al. (1997)</td>
</tr>
<tr>
<td>Varian</td>
<td>2100C/2300C</td>
<td>15</td>
<td>6.8E+11</td>
<td>Mao et al. (1997)</td>
</tr>
<tr>
<td>Varian</td>
<td>1800C</td>
<td>18</td>
<td>2.9E+12</td>
<td>McCall (1987)</td>
</tr>
<tr>
<td>Varian</td>
<td>20C</td>
<td>15</td>
<td>9.3E+11</td>
<td>McCall (1987)</td>
</tr>
<tr>
<td>GE</td>
<td>Saturne 43</td>
<td>25</td>
<td>2.4E+12</td>
<td>Fenn and McGinley (1995)</td>
</tr>
<tr>
<td>GE</td>
<td>Saturne 43</td>
<td>18</td>
<td>1.5E+12</td>
<td>Fenn and McGinley (1995)</td>
</tr>
<tr>
<td>GE</td>
<td>Saturne 41</td>
<td>15</td>
<td>4.7E+11</td>
<td>Fenn and McGinley (1995)</td>
</tr>
<tr>
<td>Elekta</td>
<td>SL-20</td>
<td>17</td>
<td>0.69E+12</td>
<td>McGinley et al. (1993)</td>
</tr>
<tr>
<td>Elekta</td>
<td>SL-25</td>
<td>22</td>
<td>2.37E+12</td>
<td>McGinley et al. (1993)</td>
</tr>
<tr>
<td>Elekta</td>
<td>SL-20</td>
<td>18</td>
<td>0.46E+12</td>
<td>Followill et al. (2003)</td>
</tr>
<tr>
<td>Elekta</td>
<td>SL-25</td>
<td>18</td>
<td>0.46E+12</td>
<td>Followill et al. (2003)</td>
</tr>
</tbody>
</table>

It is clear that neutron source strength for different linac's depends on the photon energy and linac head structure, as well as the model. There is a wide range of Q values reported for specific linac models and beam energies. For example the source strength from the Varian 2100C/2300C increased from 15MeV, Q value of 6.8x10¹¹ n Gy⁻¹ to 18 MV, with Q value 1.2x10¹² n Gy⁻¹, whilst in comparison the Siemens MD at 15 MeV showed a Q
value of $0.2 \times 10^{12}$ n Gy$^{-1}$. The differences in Q value are due to differing manufacturer, linac model and beam energy. Each manufacturer will have a different linac head design, different materials used in the construction and subsequently different shielding. All these factors influence the Q value. The differences between Q values for each model indicate that compromises ought to be made between the advantages of normal tissue sparing yielded by higher energy treatment modes and neutron doses. Careful considerations should be made when dealing with paediatric patients (children) or those with sensitive needs, as they have a higher sensitivity to radiation.

2.2.3 The Photonuclear effect in biological medium

High energy photons are able to generate neutrons in vivo. The neutron reaction cross sections are significant for the primary to constituents of tissue, which include H, C, N and O. The human body is a fairly effective moderator, due to the high hydrogen content. An example of a thermal neutron interaction in human tissue, involving neutron capture reactions by nitrogen: 14N (n, p) 14C. The kerma factor for this interaction in soft tissue is approximately 3.4 % nitrogen (ICRP, 2001)), which would be about $2.7 \times 10^{-11}$ cGy (n$^{-1}$cm$^{-2}$). This interaction involves a transferred kinetic energy of 0.62 MeV, 94% of which is given to the proton, with range in the order of microns. However, the ratio of H to N atoms in tissue is about 40 and it is ultimately hydrogen interactions that dominate the kerma, which can be seen in the table below, which shows the relative importance of the key reactions contributing to kerma in human tissue at different energies.

<table>
<thead>
<tr>
<th>Component</th>
<th>Reaction</th>
<th>Key Particle</th>
<th>0.5 MeV</th>
<th>5 MeV</th>
<th>9 MeV</th>
<th>14.5 MeV</th>
<th>19.5 MeV</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hydrogen</td>
<td>H(n,n)H</td>
<td>Proton</td>
<td>93.6</td>
<td>88.2</td>
<td>83</td>
<td>71.6</td>
<td>65.7</td>
</tr>
<tr>
<td>Carbon</td>
<td>12C(n,n)12C</td>
<td>C recoil</td>
<td>1.4</td>
<td>1.4</td>
<td>1.1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>12C(n,a)9Be</td>
<td>Alpha</td>
<td>0</td>
<td>0</td>
<td>0.7</td>
<td>0.7</td>
<td>0.5</td>
</tr>
<tr>
<td></td>
<td>12C(n,a')3a</td>
<td>Alpha</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>2</td>
<td>3.7</td>
</tr>
<tr>
<td>Oxygen</td>
<td>16O(n,a)13C</td>
<td>Alpha, C recoil</td>
<td>0</td>
<td>3.4</td>
<td>4.7</td>
<td>7.3</td>
<td>1.5</td>
</tr>
<tr>
<td></td>
<td>16O(n,p)16N</td>
<td>Proton</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>1</td>
<td>1.1</td>
</tr>
<tr>
<td></td>
<td>16O(n,p')15N</td>
<td>Proton</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0.1</td>
<td>1.5</td>
</tr>
<tr>
<td></td>
<td>16O(n,a')12C</td>
<td>Alpha, C recoil</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>7</td>
<td>13.8</td>
</tr>
<tr>
<td></td>
<td>16O(n,p')15N</td>
<td>N recoil</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0.1</td>
<td>3.7</td>
</tr>
<tr>
<td></td>
<td>16O(n,n)16O</td>
<td>O recoil</td>
<td>4.6</td>
<td>6.1</td>
<td>4.9</td>
<td>3.4</td>
<td>3.4</td>
</tr>
<tr>
<td></td>
<td>16O(n,n')16O</td>
<td>O recoil $e^{-1}e^{+}$</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0.2</td>
<td>0.5</td>
</tr>
<tr>
<td>Nitrogen</td>
<td>14N(n,n')13C</td>
<td>Proton</td>
<td>0.4</td>
<td>0.9</td>
<td>1.6</td>
<td>2.4</td>
<td>2.4</td>
</tr>
<tr>
<td>other</td>
<td>0.4</td>
<td>0.9</td>
<td>1.6</td>
<td>2.4</td>
<td>2.4</td>
<td>2.4</td>
<td></td>
</tr>
</tbody>
</table>

The cross section of neutron interaction with various components depends upon the en-
ergy of the neutron and the physical properties of the nuclei it is interacting with. Elastic scattering: \((n, n)\) reactions will occur when a neutron collides with a nucleus that recoils with an angle \(\phi\) with respect to the neutrons initial direction of motion. In this interaction, the kinetic energy and momentum are conserved. Inelastic scattering: \((n, n')\) or \((n, n'\gamma)\) reaction involves a neutron with a high enough kinetic energy first being captured by the nucleus, raising it into an excited state, then being re-emitted with a lower energy and in different direction from which it first came. The nucleus is then left in an excited state and will de-excite by emitting high-energy \(\gamma\) rays. The third most significant process is neutron capture. In this situation a thermal neutron will hit a nucleus and emit a proton or \(\gamma\) ray. The secondary charged particles generated from neutron interactions in tissue typically have short ranges and deliver a much higher dose in comparison to thermal and epithermal neutrons.

An example of a neutron absorption reaction with hydrogen: \(1H (n, \gamma) 2H\), will involve a gamma transition which does not contribute directly to kerma. It is nonetheless an important reaction given the large number of H nuclei in tissue. The energy of the photon liberated is 2.2 MeV. In a patient, photons of this energy will be absorbed at about 5% per cm in tissue. The size of the human body ultimately means that the hydrogen capture component generally dominates the dose from neutron interactions.

Studies performed by Allen and Chaudhri [15] investigated the ratio of proton/neutron and alpha/neutron particle yields as a function of the beam energy in approximate-tissue material (\(C_{5}H_{40}O_{18}N\)). They found that the production of alpha and proton particles in approximate tissue, relative to the yield of neutrons, differs strongly with energy. (Fig. 2.4)

![Figure 2.4](image-url)

**Figure 2.4:** Left: the ratio of protons and alpha particle yields to the neutron yields as a function of the beam energy. [15] Right: The energy distribution of emitted particles following photonuclear interaction in human tissues. [16]
Below 8 MeV, the number of protons and neutrons are similar. Between 10-15 MeV the ratio of protons to neutrons is approximately half, while the ratio of alphas to neutrons is 3 at 15 MeV. Above 20 MeV there are roughly double the number of protons to neutrons and only a small fraction of alphas. As stated previously in section 2.1.3 by Hubbell [7] the overall cross sections for biological materials will be generally lower than those for photonuclear interaction in linac components. The photonuclear component will however become more significant, as the beam energy is increased. Allen and Chaudhri [15] and Halbleib [16] have confirmed this result, finding that the dose equivalent due to photonuclear reactions within tissue irradiated with a 24 MeV beam is about 1% of the photon dose but at 30 MeV this dose equivalent increases to about 3% of the photon dose.

### 2.2.4 Energy distribution of neutrons by the photonuclear effect

The energy distribution of neutrons emitted by photonuclear disintegration is characterized by two components as shown in equation 2.1; the evaporation spectrum and the direct emission spectrum, as stated by Rene [4].

\[
n(E) = \frac{A E}{T^2} \exp \left( \frac{-E}{T} \right) + B \int_{E_{\text{max}}-S}^{E_{\text{max}}} \frac{\ln \left( \frac{E_{\text{max}}}{E+S} \right)}{\ln \left( \frac{E_{\text{max}}}{E+S} \right)} dE
\]

(A - normalization factor that accounts for the amount of evaporation neutrons.
B - normalization factor that accounts for the amount of fast neutrons.
T - nuclear temperature of a particular nucleus (Kelvin).
E - energy of neutrons.
E_{\text{max}} - maximum energy that neutrons can reach.
S - neutron binding energy in the nucleus.

The low energy peak at about 100 keV is due to the thermalization of primary neutrons. This peak is due to the nucleus isotropic evaporation and involves a single parameter, the nuclear temperature T (MeV) for a particular nucleus. The second component is due to fast neutrons and can be seen in figure by the large peak at approximately 1 MeV. These fast neutrons are due to direct reactions. (Fig. 2.5) Fast neutrons comprise up to 15% of the total neutrons produced by the \( (\gamma, n) \) reactions, with average energy approximately between 1-2 MeV.
In this thesis we are most interested in the direct or fast neutrons, with average energy approximately between 1-2 MeV. The table 2.5 below lists the average neutron energies for high energy linacs.

**Table 2.5:** Average photoneutron energies for high-energy (15, 18, 20 and 25 MV) accelerators. [5]

<table>
<thead>
<tr>
<th>Maximum Photon Energy (MeV)</th>
<th>Average Neutron Energy (MeV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>15</td>
<td>1.15</td>
</tr>
<tr>
<td>18</td>
<td>1.25</td>
</tr>
<tr>
<td>20</td>
<td>1.31</td>
</tr>
<tr>
<td>25</td>
<td>1.46</td>
</tr>
</tbody>
</table>

The average neutron energies listed above appear to increase with increasing incident photon energy, not exceeding 2MeV. However in the patient plane, the average energy of neutrons can vary, due to the complex distribution of neutrons as they transmit through the accelerator head, patient and treatment room.

### 2.2.5 Contribution of linac components to photoneutron field

Electrons incident upon a target in a linear accelerator, via bremsstrahlung will generate highly forward-peaked photon distributions. The greatest fraction of photonuclear interactions will occur in the collimating devices. The first shaping device is the primary collimator, which results in the most significant yield of photoneutrons. This is followed by the secondary collimators (jaws and multi-leaf collimators). The target itself yields photoneutrons, but its relatively thin physical size, means that self-absorption of high
energy photons via photonuclear interaction is less pronounced than in the collimation devices. The fourth-most dominant linac component resulting in neutrons reaching the patient plane is the flattening filter. The flattening filter is usually of a low Z type medium and also presents a smaller linear path length than the collimators, thus limiting the photo-neutron yield.

As can be seen in (Fig. 2.6), the Siemens Primus results in lower photo-neutrons in comparison to the Varian model. This is most likely due to the fact that Siemens Primus has one singular pair of jaws operating orthogonally to the MLC, whilst the Varian has two pairs of jaws, followed by an MLC. So it is the amount of secondary collimator material that will influence the production of photo-neutrons.

![Graph showing Neutron Yield](image)

**Figure 2.6:** Linac components contributing to photo-neutron yield as (%) of the total photo-neutron yield. Varian data from Mao [10] and Primus data from Becker [24]

Neutron production is dependant on the photon beam energy and the model of linac. Photoneutron production of medical linacs has not been extensively compared, however, such information is necessary to determine neutron dose equivalents for different linacs and to calculate shielding requirements for linac bunker rooms.

A study by Howell [18] was performed to measure the neutron spectra from the most up-to-date linacs from three manufacturers: Varian 21EX operating at 15, 18, and 20 MV, Siemens ONCOR operating at 15 MV and 18 MV, and Elekta Precise operating at 15 MV and 18 MV. Neutron production was measured using gold foil activation in Bonner spheres. Based on the measurements, the authors determined that the neutron spectra did not change significantly between accelerators or even as a function of treatment energy, however the neutron fluence, and therefore the ambient dose equivalent, did vary,
increasing with increasing treatment energy. The Varian linac had the highest neutron ambient dose equivalent for different treatment energies. Further studies performed on different linac models and designs is required to understand the different contribution to the photoneutron production and thus the final neutron dose delivered to the patient. By understanding the various linac components that contribute to the photoneutron yield across a wide range of linac models and beam energies, patient treatment planning can be improved.

2.2.6 Influence of treatment Energy, field size, distance from isocentre on photoneutron production

The dependence of photonuclear production is related to the energy of the incoming photon beam and will depend on the linac model. The difference between linac models, arise from differences in shielding and design, thus changing the photon spectrum in the high energy tail region. It has been observed that the field size will also have an impact upon the neutron field reaching the patient plane. Data collected from Hashemi [8] and Zabihzadeh [19] for a Saturne 20 have shown an almost linear relationship between neutron dose and field size, as seen in (Fig. 2.7).

![Figure 2.7: Neutron dose as a function of field size for a Saturne 20 linac](image)

It is seen that a smaller field size, will have a shielding function, scattering neutrons away from the patient plane. Thus using a smaller field size has the added advantage of shielding a patient, whilst increasing the field size will increase the production of photoneutrons and the patient will receive additional dose. Another study performed by Kry [20], studied the effect of field size on the neutron dose as a function of off-axis distance, for a Varian 1800 linac model. The results for 10 cm x 10 cm and 15 cm x 15 cm field sizes (Fig. 2.8).
For a 15 cm x 15 cm field size, the neutron dose for the first 10cm is fairly independent of the off-axis distance. At distances larger than 10cm from the centre of beam axis the dose decreases exponentially. This trend was repeated for the 10 cm x 10 cm field size but dropping of at about 7cm. It is evident that at distances further away from the central beam axis, the shielding effect of the jaws results in a reduced neutron dose.

2.3 Photoneutron Dosimetry

2.3.1 Dosimetry

Accurate dosimetry in radiotherapy is essential to ensure a cancerous tumor is treated, whilst also minimising the risk of severe side effects to healthy tissues and organs during the treatment. Dosimetry is the measurement of the absorbed dose delivered by ionizing radiation, which relies on measuring the energy deposited on a macroscopic scale. Absorbed dose is expressed in units of Gray (Gy) and is measured in units of J/kg (equation2.2)

\[ D = \frac{E}{m} \]  

(2.2)

Where D is the dose, E is the energy deposited in Joules and m is the mass (kg) of the material exposed. The absorbed dose is a useful measurement of the total radiation dose a person is exposed to but isn't able to determine the effect of different types of radiations. Different types of radiation can produce differing biological effects. High ionising radiation can produce electrons, which can produce secondary delta electrons that have dense ionisation tracks.
The dose equivalent (H) is given by equation 2.3 and is used to estimate the biological effects produced from different particle types of particular energy. The dose equivalent uses different quality factors as outlined by the ICRP Publication 60 to weight the different types of radiation based on the particular biological effects they will produce.

\[ H = D Q_f \]  

(2.3)

The dose equivalent is expressed as the product of the dose (D) with a quality factor (Q_f) used to weight the biological effect of the radiation. H is defined in units of Sieverts (Sv). The dimensionless quality factor Q_f is dependent on both particle type and energy, and for any radiation field, its value is an average over all components. Q_f is defined to be a function of LET. X-rays, gamma rays and electrons are known as low LET radiation. Higher LET is more destructive to biological material than low LET radiation at the same dose. All radiation ultimately manifests itself through charged particles, so LET is a good measure of localised radiation damage to materials not limited to biological structures. For photons and electrons the Q_f is defined as 1.0 and densely ionizing particles, such as alpha particles have an Q_f of 20.

Body tissues react differently to radiation and cancer-induction occurs at different rate of dose in different tissues. The effective dose (Equation 2.4) is used to describe the risk of developing fatal cancer in the tissue in question. The values of the tissue weighting factors \( \omega_T \) are given in table 2.6.

\[ E = \sum \omega_T H_T \]  

(2.4)

<table>
<thead>
<tr>
<th>Organ/tissue</th>
<th>( \omega_T )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone marrow, colon, lung, stomach, breast, remainder</td>
<td>0.12</td>
</tr>
<tr>
<td>Gonads</td>
<td>0.08</td>
</tr>
<tr>
<td>Bladder, liver, oesophagus, thyroid</td>
<td>0.04</td>
</tr>
<tr>
<td>Bone surface, skin, brain, salivary glands</td>
<td>0.01</td>
</tr>
</tbody>
</table>

Effective dose is estimated for a reference person under a given exposure situation and is individual specific. It can be used to provide a risk assessment for an individual of a specific age and gender. It is a useful tool for comparing the relative doses related to stochastic effects from different diagnostic examinations, technologies and procedures at different hospitals.
CHAPTER 2. LITERATURE OVERVIEW

The ambient dose equivalent is used to give a conservative estimate of the effective dose a person would receive when staying at the point of the monitoring instrument. Ambient does equivalent \( H^*(d) \) is the normal monitoring quantity for X-rays, gamma and neutron radiation where \( d \) is the depth at which the dose is delivered. International convention in radiation protection is to use the ambient dose equivalent at 10 mm depth i.e. \( H^*(10) \). [1]

2.3.2 Overview of neutron dosimeters used in photoneutron fields

The complexity of mixed field dosimetry arises firstly because the neutron field of interest coexists with a very high fluence photon field, and that each radiation can potentially give rise to fields of the other. The fact that essentially all neutron fields are thus contaminated to varying degrees by photons requires the use of sophisticated dosimetry, Knoll [21]. It is often a combination of multiple dosimeters that is the most useful approach to mixed gamma-neutron field dosimetry.

Photoneutron spectra, fluence (number of particles striking an area) and dose equivalence due to neutrons have been previously measured next to or inside phantoms made of ICRU tissue or water equivalent using various techniques. Previous experiments aimed at detecting neutrons have included the use of: Bubble Detectors, Activation foils and Thermo luminescent Dosimeters (TLD's), to name a few. Some bubble detectors and nuclear track detectors are insensitive to photons. The greatest challenge for photoneutron dosimetry has been achieving the correct energy dependence of the dose response.

A study by Barquero [6], found that paired TLD-600/ TLD-700 techniques can be used to calculate the neutron and photon doses in mixed gamma-neutron fields produced by a Radiotherapy linac. Paired TLD's are small crystals: one with Li-6 construction and one with Li-7 construction which are both moderated and unmoderated for detecting fast and thermal neutrons as well as gammas. The paired crystals are placed within polyethylene spheres (Fig. 2.9). After irradiation the Li-6 and Li-7 crystals are heated and the light emitted from the crystal is used to measure the ionizing radiation in the room. The paired TLD method allows for the neutron spectra to be obtained for thermal to several MeV neutrons at various positions in the room. The thermal peak remained in all zones, in agreement with Holeman et al. (1977) [34], however TLD's working inside the primary linac beam, were deemed unreliable, due to the primary photon beam intensity, hiding the neutron component on the TLD glow curve. There could also be a problem with the TLD linearity if the primary photon field becomes greater than 5 Gy. [6] This limits paired TLD method to only out of field neutron dosimetry. Other detectors not sensitive to photons, such as bubble dosimeters (Lin [35]) or activation foils (Krmar et al. 1999 [13]), are more convenient to use than TLD's for measurements out of field.
A study done by D’Errico [36] has recently developed a new approach using bubble detectors, that involves a combination of three superheated drop detectors with different neutron energy responses, to evaluate dose-equivalent and energy distributions of photon-neutrons in a phantom. Measurement of neutron dose equivalent is related to the number of bubbles created in the liquid. The neutron entrance dose equivalent was measured to be 4.5 mSv Gy$^{-1}$ of photons from an 18 MV medical linac. In accordance with Turner [25], the tissue-equivalent dose is uniform from approximately 200 keV to more than 15 MeV, thus the method used making it a reliable neutron detector. However the bubble detectors design is susceptible to damage and the time it takes to perform accurate measurements in a clinical environment, does not make it suitable for routine Quality assurance procedures.

A study by Khaled [37] used CR39 track etch detector that were processed through electrochemical etching. After etching, the detectors were thoroughly washed in running water and dried overnight. Counting of tracks was carried out on a PC-based automatic image analysis system. Tracks of nuclear recoils, results from interaction between the neutrons and the H, C, and O nuclei that form the polymer network. The induced proton tracks are predominant due to the highly elastic scattering crosssection of H (38 barns) when compared with C and O. The neutron fluence is then measured and used for the neutron strength calculation. The average neutron energies was found to be about 5.3 3.2 and 3.6 1.9 MeV for the Elekta and Varian machines, respectively, and the most probable energy is in the range of 12 MeV for both LINACs. Readout and post analysis takes a considerable amount of time, limiting the devices use in a clinical environment.

Recent studies are now focusing on Semiconductor p-i-n diodes, for mixed field neutron detection. One study by Guardiola [41] has performed measurements with ultra-thin 3D silicon sensors in a radiotherapy treatment room using a Siemens PRIMUS linac. The 3D silicon sensors (Fig. 2.10) were able to detect the spectrum of recoil alpha and Li ions from neutron interactions occurring inside the detector at a distance of 2 m along the pa-
A study by Arbor [42], investigated the real-time detection of fast and thermal neutrons in radiotherapy with Complementary Metal Oxide Semiconductor (CMOS) sensors. The CMOS sensors consist of a sensitive part of $2.56 \times 2.56 \text{ mm}^2$ made of $64 \times 64$ interconnected micro-diodes of $3 \times 3 \mu\text{m}^2$ and $40 \mu\text{m}$ pitch. The detector has a 14 micron thick sensitive layer, that is paired with polythene to convert fast neutrons and thermal neutrons into recoil protons. Most likely an exoergic capture reaction is used to detect thermal neutrons. The detector can also be paired with a $^{10}$B to detect alpha particles. It was found that CMOS sensors were unable to be used in the radiation field but were able to be placed outside the radiation field. It was proposed that a 3D neutron phase space can be experimentally estimated and used as an input for MC dose calculations on a voxelized phantom. This extrapolation can become very complex to apply for Intensity Modulated Radiation Therapy (IMRT) and would need to take into account the patient position on the couch. Further investigation is needed.

Figure 2.10: Left: photo of a wafer with ultra-thin 3D silicon detectors processed at IMB-CNM. Right: Image of a cross section of an ultra-thin 3D detector, with columnar electrodes along a strip-contact. [41]

This thesis will examine similar detectors based on 3D silicon technology (see Chapter 5) as studied by Guardiola [41] and silicon p-i-n diode technology that is based on permanent radiation damage (see Chapter 4) and a directional neutron detector to study the neutron field produced by an 18 MV linac (see chapter 6). The detectors based on 3D silicon technology have the advantage of large active area and substrate that has been etched, which allows for rejection of gamma photons. Silicon p-i-n diodes based on bulk radiation damage, have been previously applied to fast neutron dosimetry as personnel accident dosimeters and for military applications. It is believed that the semiconductor p-i-n diode will provide adequate neutron sensitivity and high gamma discrimination in mixed gamma-neutron fields produced by an 18 MV linac, for both in and out of field measurements. Experimental measurements, together with MC simulations will be used to determine the neutron ambient dose equivalent.
2.3.3 Silicon p-i-n Diodes

A silicon p-i-n diode consists of a section of doped intrinsic silicon sandwiched between ion implanted n-type section on one side and a p-type section on the other, as can be seen in (Fig. 2.11). Metallic contacts are attached to both the p and n type sections.

![Schematic drawing of long base p-i-n diode](image)

The mode of operation of p-i-n diodes relies on the characteristic behaviour of charge carriers in an intrinsic high resistivity silicon. Irradiation of the diode will induce a spectrum of radiation defects in the silicon. The types of defects introduced into the lattice will depend on the type of incident radiation and energy of the incident particles.

When a fast neutron collides with an atom in the silicon lattice, its energy is transferred to the constituent elements of the lattice and lattice defects are produced by scattered recoil nuclei. The energy required to produce electron-hole pairs in silicon is around 3.6eV and the amount of energy required to dislodge a silicon atom from its lattice site is around 25eV. [26] The displacement of atoms from their equilibrium sites will produce clustered displacement damage. An example of clustered displacement damage can be seen in (Fig.2.12) of a 50keV silicon recoil atom. Displacement damage and ionization damage will occur simultaneously. The electron-hole pairs caused by the ionization may annihilate in a relatively short time depending on the external bias applied to the silicon device. Displacement damage on the other hand is more destructive and the effects can be semi-permanent to permanent. [27]
CHAPTER 2. LITERATURE OVERVIEW

Figure 2.12: Silicon atomic cluster displacement damage. [43]

Provided that the recoil nuclei have enough kinetic energy, they are able to produce a secondary group of lattice defects. These defects are in the form of vacancy and interstitial sites, within the semiconductor. Some of these defects will be stable at room temperatures and will remain present in the structure; these defects are created by displacement damage KERMA (Kinetic Energy Released per unit Mass Absorber). The number of mobile vacancies in the semiconductor, prior to the formation of stable defects, will act as recombination (trap) centres for minority carriers. As the number of recombination centres increases, the charge carrier lifetime decreases and the resistivity in the diode base increases. [27] This is due to the larger probability of carriers recombining at lattice defect sites and acting as compensation centres.

The determination of neutron damage KERMA for silicon has been theoretically calculated and published as standard data sets for reference [28]. This data has been experimentally validated, with an uncertainty of less than 10%. The total Si KERMA, displacement damage Si KERMA and tissue KERMA are shown in (Fig. 2.13). The effect of silicon displacement damage KERMA can be correlated with tissue displacement damage KERMA, which will be discussed in section 2.3.5 in more detail.
2.3.4 Long Base Planar Silicon p-i-n diode

In applications where the neutron sensitivity is not known, a silicon p-i-n diode should have a wide dynamic range of sensitivities. In recent studies performed by Rosenfeld [29] and Nahmo [30], it has been found that a long base silicon diode based on high resistivity silicon could provide an improved sensitivity range.

In a study by Rosenfeld [29], two types of planar p-i-n diodes based on n type high resistivity silicon were developed and manufactured to measure neutrons. The two types of planar p-i-n diodes were circular type (C-1 and C-2) (Fig. 2.14) and planar p-i-n diodes consisting of a linear array of p+ contacts. Type C type diodes had radial base-length (a) of, 0.5 and 1.45 mm, the diameter of the P+ region is 1.5 mm and 0.5 mm and the thickness of n+ ring is 0.5 mm and 0.1mm for C-1 and C-2 respectively. The response of the planar p-i-n diodes of different configurations (circular and linear array) were investigated in fast neutron and 3 MeV proton fields.

Figure 2.14: Circular planar diodes C-1 and C-2 [29]
C type diodes, were found to have wide dose range sensitivity and demonstrated the ability to work in on-line mode in a neutron field. These diodes can be used for neutron measurements in on-line applications, allowing for the reduction of error associated with temperature effects. In a similar study by Lee [30] and Liu [31] also investigated planar-structure circular type p-i-n diodes with surface n+ ring. These diodes have been proposed to enhance the performance of silicon detectors without requiring excessive silicon wafer thickness. In this way, the base length of the diode (i region) is controlled by the radius of the backside n+ doping ring, which is represented in (Fig. 2.15) by the n+-Si.

![Cross Section view of the circular planar p-i-n diode](image)

**Figure 2.15:** Cross Section view of the circular planar p-i-n diode [31]

All three authors (Rosenfeld [29], Lee [30] and Liu [31]) found that the circular planar p-i-n diodes demonstrated sensitivity $\sim t^2$, where $t$ is the radial distance between p+ core and periphery regions. The results obtained agreed with the predicted theory by Swartz [27].

Studies by Nahmo [30], showed that by increasing the intrinsic layer thickness (diode base length), along with the decreasing the diode cross-sectional area, improved the detection sensitivity. The diodes with the best configuration showed promising linear response in the range from 10 to 1,000cGy tissue equivalent and no angular dependence. Neutron sensitivity of up to 13mV/cGy was achieved with a 5mA current pulse, with sensitivity three times higher than that of similar commercial neutron diodes [30]. This clearly illustrates that a long base silicon diode has improved sensitivity in comparison to singular bulk p-i-n diodes and that by changing the cross sectional area and intrinsic layer thickness [27] there is a good possibility that the sensitivity can be further increased. Further investigation of sensors with adjustable base length is required and will be investigated in chapter 4 of this thesis.

### 2.3.5 Neutron Displacement Damage KERMA

Silicon p-i-n diodes work on the principle that their forward impedance increases with neutron exposure due to damage effects introduced into the Si lattice by radiation [28]. There are two mechanisms of radiation damage in silicon devices; by the deposition of
ionizing and non-ionizing energy. The dose associated with the ionizing energy losses (IEL) is responsible for build-up charge within the silicon oxide of the Metal-Oxide-Semiconductor (MOS) devices; whilst the dose associated with non-ionizing energy losses (NIEL) is responsible for atomic displacements in bulk silicon [29] [58] [28] [59] [25]

Most predictions of the neutron energy dependence in silicon semiconductor devices have been based on the amount of damage produced. In the literature (for instance [60]), the damage effect due to initial and cascading-displacements, induced by neutrons, is expressed by the damage function (also referred to as the displacement damage KERMA function) $D(E)$ in units of MeVcm$^2$ [61]. The displacement damage function accounts for both the cross section for neutron silicon scattering and the energy released in creating displacements. $D(E)$ is defined as:

$$D(E) = \sum_k \sigma_k(E) \int f_k(E, E_R) \rho_k(E_R) dE_R$$

(2.5)

Where $E$ is the incoming neutron energy, $\sigma_k(E)$ is the cross section for the k-th reaction, $f_k(E, E_R)$ is the probability that a recoil atom is generated with kinetic energy between $E_R$ and $E_R + dE_R$ and, $\rho_k(E_R)$ is the partition energy for the recoil nucleus. From equation 2.5, the energy density $E_{dis}$ deposited through atomic displacements by neutrons characterized by the neutron spectral fluence $\phi(E)$ in n cm$^{-2}$ MeV$^{-1}$ is the defect producing energy imparted per cm$^3$ and is given equation 2.6:

$$E_{dis} = N_A \int_{E_{min}} D(E) \phi(E) dE \quad [MeV cm^{-3}]$$

(2.6)

Where $N_A = (\rho_Si N_{Av})/A_{Si}$ is the number of atoms per cm$^3$ in the bulk silicon, $\rho_Si$ and $A_{Si}$ are the density and atomic weight of the silicon medium, respectively; $N_{Av}$ is the Avogadro constant and $E_{min}$ is the minimum neutron energy for inducing displacement damage.

The photoneutron energy spectrum produced by an 18 MV medical linac is most prominent between 200 keV and 2 MeV, where neutrons have maximum biological effectiveness [62]. The advantage of using passive silicon p-i-n diodes for fast neutron dosimetry is that the silicon displacement damage KERMA determined radiation damage correlates well with tissue neutron KERMA for neutron energies above 250 keV, as shown in Fig 2.16(a) and Fig 2.16(b). [58] [28] [59]

Fig 2.16(a) is showing the silicon displacement damage KERMA, total silicon KERMA and total tissue KERMA. The silicon total KERMA is the addition of all IEL and NIEL occurring in the silicon. Fig 2.16(b) is showing the ratio of the total silicon KERMA to the
total tissue KERMA. From Fig 2.16(b) it is possible to see that there exists energy intervals where the total silicon KERMA energy response is matching well to the total tissue neutron KERMA response. In particular for Tissue Equivalent (TE) neutron dosimetry, an energy interval from 0.25MeV to 5 MeV is of interest. Here the p-i-n diode response when placed in an 18 MV medical linac field, can match the tissue dose within +15%. This energy range is of greatest relevance, as neutron interactions with tissue at these energies have a high neutron cross section of interaction, leading to greatest contribution to tissue dose.

The neutron displacement damage KERMA can be written as in equation 2.7, which is a function of the neutron fluence ($\phi_n$) and the corresponding displacement damage kerma factor ($k_n$) as a function of neutron energy, where $k_n = D(E) / \text{silicon density}$.

$$\text{neutron KERMA}\text{displacement} = \int_{E_{\text{min}}}^{E_{\text{max}}} \phi_n(E) k_n(E) \, dE$$

(2.7)

The displacement of atoms from their equilibrium sites produces defects within the lattice, which can be quantified as a change in carrier lifetime and resistivity. Defects within the lattice are produced along the tracks of secondary particles and in clusters at the end of their tracks [59] and act as recombination centres. As the number of recombination centres increase, the carrier lifetime will decrease accordingly in equation 2.8:

$$\frac{1}{\tau} = \frac{1}{\tau_0} + K\phi$$

(2.8)

Where $\tau_0$ is the initial carrier lifetime, $\tau$ is the lifetime after irradiation, $K$ is the damage
constant for the semiconductor material in a particular type of radiation field and $\phi$ is the neutron fluence. [28] [59] The decrease in the effective carrier lifetime causes a marked increase in the forward voltage across the base of a diode at constant current injection [27]. Additionally, as $\tau$ changes due to irradiation, the resistivity $\rho$, changes according to equation 2.9:

$$\rho = \rho_0 \exp\left(\frac{\phi}{K}\right)$$

(2.9)

Where $\rho_0$ is the initial resistivity before irradiation and $\rho$ is the resistivity following irradiation. [59] An increase in the resistivity of n type silicon under neutron irradiation increases the sensitivity and linearity for long based diodes [59]. The resistivity of silicon used for the fabrication of the p-i-n diodes in this study was 1 k$\Omega$cm, which is not high enough for small neutron doses to have an appreciable effect on the resistivity.

The displacement damage created in the p-i-n diode is read by periodically passing a fixed current pulse into the device and measuring the corresponding voltage across the diode [29] [28] [59]. The geometry of the diode is an important factor, influencing the detector sensitivity. Under the same readout current, the injected current density is different for different P+ area and diode cross sections. It was demonstrated that the sensitivity of a silicon p-i-n diode in forward voltage mode operation grows with the square of the base diode length ($w$). [27]. The effect of different p+ geometries, base length between different P+ areas and the resistivity of different p-i-n diodes will be investigated in this thesis.
Chapter 3

Geant4 simulation modelling an 18 MV medical linac

3.1 Introduction

Geant4 (version 10.01.p01) was adopted as MC code, to study the response of neutron detectors placed both on the surface and inside a solid water phantom, complementing the experimental measurements performed in chapters 4 and 6. This chapter outlines the basic user classes used in Geant4 MC simulation code, with a summary of the methods and techniques used to model an 18 MV medical Varian linac. An 18 MV medical Varian linac was modelled, omitting the linac head shielding and bunker walls. A linac treatment plan simulation tool by Cornelius et al. [45] was adapted by implementing geometry specifications for an 18 MV linac obtained from the vendor Varian. The neutron, photon and electron energy spectra were modelled both on the surface and with depth inside the solid water phantom.

3.2 The Geant4 Toolkit

The Geant4 toolkit is a general purpose code developed to model the interactions of particles with matter. It is able to simulate the transport of many particle types and has previously been used for a variety of applications including: radiotherapy physics, high energy physics and radiation protection in space [47].

Geant4 provides the ability to describe the detector geometry, materials and to visualise the simulation experimental set-up. It allows for particles to be tracked from the point of generation to their termination (kinetic energy of the particle is equal to zero) and the ability to store useful information of the detector response. It also provides an extensive set of physics models, to describe interactions of particles with matter across a wide energy range [48]. By using Object Oriented technology and C++ language, Geant4 is a
flexible and easy to extend simulation toolkit.

The Geant4 toolkit requires the user to write his or her own C++ program using classes which inherent behaviour from kernel Geant4 classes. The classes which were implemented in developing a Geant4 application of an 18 MV medical linac model are described in the following sections.

### 3.3 User Action Classes

User Action Classes are virtual classes whose methods control the geometry of the simulation, definition of particles as well as the generation of primary particles and their physics processes. They also control the flow of the simulation and allow the retrieval of useful information including track structure, particle interactions and energy depositions. In the simulation created to model an 18 MV medical linac the user action classes utilised were: `G4UserDetectorConstruction`, `G4VUserPhysicsList`, `G4VUserPrimaryGeneratorAction`, `G4VUserEventAction`, `G4UserRunAction`, `G4UserSteppingAction` and a number of visualisation and interface commands. Each of these user action classes is outlined below.

#### 3.3.1 G4UserDetectorConstruction

This base class controls the definition of the experimental set-up, in terms of geometrical components and material composition. A geometrical component is defined in terms of shape, material, position and rotation in the experimental set-up. Visualisation attributes can be defined at this level, to allow the visualisation of the experimental set-up. In a Geant4 simulation the detector is just a component of the experimental set-up, declared sensitive, where it is possible to retrieve information about the hits. A hit is a snapshot of the physical interactions of a track in the sensitive region of the detector. The concept of a track represents physics information (position, energy deposition, mass, spin, etc.) of the particle under propagation [48].

A linac treatment plan simulation tool by Cornelius et al. [45] was adapted by implementing geometry specifications for an 18 MV linac obtained from the vendor Varian. When modelling an 18 MV medical linac the following components were defined in the `G4UserDetectorConstruction`: bunker filled with air, electron target and applicator, ion chamber, flattening filter, X and Y Jaws and multi leaf collimators (MLCs) (Fig. 3.1).
A 3D CAD program Solidworks [51] was used to accurately model the treatment target, flattening filter (see Fig. 3.1), primary collimator, jaws and MLCs and exported as STEP files [49] before being converted to geometry description mark-up language (GDML) using the MeshLab software and imported into Geant4 via CADMESH [50]. An example of using CAD to model a flattening filter is shown in Fig. 3.2.
Figure 3.2: CAD modelling of Flattening Filter using Solidworks. [51]

The linac head shielding and bunker walls were emitted in the simulation, due to two reasons. The first reason being that the pin diode detector is most sensitive to fast neutrons and as there is extremely low probability that a fast neutron created from interactions with the shielding or walls would reach the small detector volume. The time needed to simulate enough neutrons, in the case of a treatment beam would take months and so was deemed unnecessary to simulate the bunker walls and linac head shielding.

3.3.2 G4VUserPhysicsList

There are three methods of this class which must be implemented, which include ConstructParticle(), ConstructProcess() and SetCuts().

The ConstructParticle() method defines all particles, involved in the experimental set-up. Geant4 provides the implementation of all the particles defined in the Particle Data Group Book [44], however the particles involved in the experimental set-up of the simulation must be explicitly invoked in this method.

The ConstructProcess() method determines the models of interaction for these particles. The user is able to create a process and assign it to a particular particle type.

The SetCuts() method determines the threshold of production of secondary particles, expressed in range. If a secondary particle is generated with a residual range in the material less than this value, the particle will not be generated and tracked, but its energy will be considered locally deposited in the medium.
The physics packages used to model interaction of particles in the 18 MV medical linac include the EMStandard Option3, HadronQElasticPhysics and HadronPhysicsQGSP_BIC_HP. The physics processes modelled in Geant4 electromagnetic packages include for photons, the photoelectric effect, Compton scattering, Rayleigh scattering, and pair production; for electrons, bremsstrahlung production, ionisation, multiple scattering; and for positrons, multiple scattering, ionisation and the annihilation process. 100µm range cuts were used throughout the geometry. The Geant4 hadronic physics component provides description of hadronic elastic and inelastic scattering for hadrons and ions.

3.3.3 G4VUserPrimaryGeneratorAction

This class allows modelling of the radiation field investing the experimental set-up of the simulation, in terms of type of particle, polarisation, position, direction, energy, time. The number of primary particles to be generated in one event must also be defined. An event consists of a collection of primary particles to be tracked. A field of particles was generated from subsequent phase space files (see section 3.3.5).

3.3.4 G4VUserEventAction

An event in Geant4 begins with the initiation of tracking one or more primary particle (as defined in the PrimaryGeneratorAction) and finishes with the completion of tracking all secondaries. The G4VUserEventAction class possesses two virtual methods which are invoked at the beginning and at the end of each event, the BeginOfEventAction() and the EndOfEventAction(), respectively.

3.3.5 G4UserRunAction

The concept of Run in Geant4 is to keep a set of events to be simulated using the same detector geometry, the same event-generator and the same physics processes [47]. The G4UserRunAction class has several methods. One method that is commonly used is the BeginOfRunAction() method which is invoked before entering the simulation event loop. The second is EndOfRunAction() which is invoked at the very end of the event processing.

Using the BeginOfRunAction() the simulation was able to be executed in three phases. Phase Space files were used to reduce computation times, using an approach similar to Bush et al (2008) [46], whereby the simulation was executed in three parts (Fig. 2.16).

The first part of the simulation study involved using the primary generator to shoot 18.8
MeV mono-energetic electrons, with a beam radius of 2mm directly onto the target. Photons and neutrons were scored in phase space file at a plane below the ionisation chamber (Phasespace 1) see (Fig. 3.1). This section of the geometry remains fixed during any treatment plan and needs to be only simulated once. Each particle that crossed the scoring plane was recorded in binary phase space file in terms of its direction, position, particle type and weight. In this way the phase space file could be called later in the second part of the simulation. Performing the simulation is this way allowed the space file to be reused once more and speed up the simulation.

The second part of the simulation involved sampling particles from the first phase space file, where they are transported through the jaws and MLCs, then recorded again at a second scoring plane below the MLCs (scoring plane 2, see Fig. 3.1). A total of $2 \times 10^{10}$ initial electrons hitting the target, in the first part of the simulation were required to obtain good statistics in the second phase space file.

The third part of the simulation sampled the second phase space file and transported particles through to the patient or phantom geometry. The phase space was also used to directly collect the neutron and photon energy fluence spectra separately and then each spectra was weighted into 35 energy bins (from 0 to 18 MeV). The weighted spectra were used to create two separate Geant4 General Particle Source (GPS) histogram inputs. The generation of these separate general particle source spectra are used to study the photon, electron and neutron energy deposited to the detector at different depths inside a solid water (results presented later in Chapter 4). This method is particularly useful to study the neutron energy spectrum with changing depth in a solid water phantom, as neutrons comprise of only 0.03% of the total beam reaching the second phase space file, with the total number of particles reaching the phase space being $<2\%$ of the initial number of electrons simulated in the first part of the simulation.

### 3.3.6 G4UserSteppingAction

The tracking of particles step by step as they transverse from one material to the next can be monitored using the Stepping action. A step is defined by default as the distance between two interactions points/geometrical boundaries. In the simulation of the 18 MV linac, variance reduction techniques were implemented via the introduction of kill zones (placed above the target and around the primary collimator so as to remove particles from the simulation that are unlikely to contribute to a response in the readout geometry). This approach was adapted from code previously written by Cornelius [45].

The SteppingAction also allows the Geant4 user to access information such as the en-
energy, position, direction of the particle, energy deposition, etc. In the simulation of the 18 MV linac, the stepping action was used to store the energy deposited and kinetic energy of incoming particles hitting a sensitive volume. In the simulation a scoring plane or detector was used as a sensitive volume to obtain gamma, electron and neutron energy spectra (presented in sections 3.4, 3.5, 3.6 of this chapter).

3.3.7 Interface Commands and Visualisation

Geant4 has various built-in user interface commands. These commands can be used interactively via a user interface (GUI), or in batch mode in a macro file. User defined commands can be implemented in user defined classes, inheriting behaviour from G4UIMessenger base class, which represents a messenger that delivers these commands to a class object [44]. These commands are particularly useful when the geometry, primary beam, or physics parameters need to be altered between simulations.

Geant4 has the capability to visualise detector components, particle trajectories and tracking steps and hits of particles in detector components. Although many methods of visualisation are possible, the one employed in simulation studies of this thesis was the Qt visualization driver (see Fig.3.3). The Qt driver is most useful for visualising geometries in real time, allowing the user to zoom, pan and rotate without the need to manually type in commands in the terminal. Qt is an excellent tool that was used for debugging and visualisation of the generation of primary events and the tracking of these events through the geometry.

![Figure 3.3: Visualisation of the Geant4 simulation in Qt software, with option of user commands available to be selected from the left pane.](image)
A number of user interface commands were developed for the simulation study of the 18 MV medical linac in order to allow quick changes to the setup geometry, without the need for recompilation between simulation runs. The user interface commands created include: gantry angle, collimator angle, jaw and MLC positions, phantom size and composition, phantom source-to-surface distance (SSD), detector position. An example of the Qt interface allowing for the use of these quick commands is shown in Fig.3.3.

3.4 Photon energy fluence inside a solid water phantom

Fig.3.4 shows the simulated photon energy spectra detected at different depths in a water phantom, produced from an 18 MV medical linac. A higher number of photons is detected at the surface of the phantom as expected; photons scatter and therefore less photons will be detected at greater depths, along the main axis of the phantom.

![Figure 3.4: Photon energy spectra for different depths in a water phantom.](image)

3.5 Electron energy fluence inside a solid water phantom

Figure 3.5 shows the simulated energy spectra of electrons detected at different depths in a water phantom per 1Gy of dose delivered at Dmax, produced from an 18 MV medical linac. The electron energy spectrum, main peak was observed to gradually increase in counts up to a depth of (3-4) cm which is in the region of the expected Dmax of 3.3cm for an 18 MV medical linac. The electron energy fluence then decreases in magnitude after 4cm, with increasing depth in the solid water phantom. There are not many primary 18 MeV electrons seen in the spectrum Figure 3.5, as electrons reaching the phantom surface are mostly all slowed down primary electrons.
3.6 Neutron energy fluence inside a solid water phantom

Fig.3.6 shows the simulated photoneutron fluence inside a water phantom, irradiated with an 18 MV linac, calculated by means of the Geant4 simulation. The photoneutron energy fluence is shown to be greatest at the surface for fast neutrons. As the photoneutrons loose energy inside the water phantom, the spectrum is moderated.
CHAPTER 3. GEANT4 SIMULATION MODELLING AN 18 MV MEDICAL LINAC

The intermediate and thermal neutron fluence is increasing with depth and peaking at 4cm depth, before reducing with increasing depth in the phantom, due to the complete absorption of neutrons. Fast neutron fluence in the spectrum is corresponding to the right peak in Fig.3.6. The peak is reducing with increasing depth, broadening and shifting to the higher neutron energies. The peak on the left side, in Fig.3.6 is due to thermal neutrons which is increasing up to 4cm in the phantom, before decreasing with increasing depth in the water phantom.

3.7 Validation of photon and neutron fields produced by the simulation model of an 18 MV medical linac

In this thesis care was taken when modelling the flattening filter and MLC’s using CAD models, as these are primary sources of photon-neutron production. The photon Percentage Depth Dose (PDD) curves matched well with the experimental measurements, within 5% to 10%, performed using an ionisation chamber at St George hospital, see Fig.3.7. The results were deemed reasonable for calculations performed in this thesis.

![Figure 3.7: Comparison of the MC calculated and measured Percentage Depth Dose (PDD) curves of the 18 MV Varian Clinac 2100C photon beam of field size 10cm x 10 cm and SSD 100 cm).](image)

In order to ensure that the neutrons produced by this simulation model of an 18 MV Varian linac were correct, the neutron fluence and the number of primary electrons needed to produce 1 Gray of photon dose at Dmax for a 10cm x 10cm field was compared to that
in the literature (see table 3.1). It was estimated that for this simulation of an 18 MV medical linac a total of $5 \times 10^{15}$ electrons hitting the target were needed to produce 1 Gray dose at Dmax, producing $7.14 \times 10^6$ n/cm$^2$/Gy. These values compared reasonably well with authors in 3.1, with a factor of 2 difference in the electron/photon ratio for this study, in comparison to the result obtained by Saeed [54]. The neutron fluence per cm$^2$/Gy was also lower in comparison to the study by Saeed [54] and is most likely due to not modelling the neutron head shielding and bunker walls. Also, the implemented kill zone around the MLC’s would have had an impact on the neutron fluence at Dmax. The neutron energy fluence both on the surface and in the solid water phantom matched with spectra produced by the authors in 3.1, within 5% to 10%, which is considered suitable to study the response of detectors to neutrons in this thesis.

**Table 3.1: Summary of calculated neutron fluence and no. e-/photon dose Gy for an 18 MV Varian linac**

<table>
<thead>
<tr>
<th>Author</th>
<th>Code</th>
<th>fluence (n/cm$^2$/Gy)</th>
<th>e-/photon Gy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Howell 2005 [64]</td>
<td>MCNPX</td>
<td>$1.53 \times 10^7$</td>
<td>$1.00 \times 10^{15}$</td>
</tr>
<tr>
<td>Bednarz 2009 [52]</td>
<td>MCNPX</td>
<td>$2.05 \times 10^7$</td>
<td>$1.00 \times 10^9$</td>
</tr>
<tr>
<td>Alem Bezoubiri 2014 [53]</td>
<td>MCNP5</td>
<td>$1.54 \times 10^7$</td>
<td>-</td>
</tr>
<tr>
<td>Saeed 2009 [54]</td>
<td>Geant4</td>
<td>$1.40 \times 10^7$</td>
<td>$1.25 \times 10^{15}$</td>
</tr>
<tr>
<td>This study</td>
<td>Geant4</td>
<td>$7.14 \times 10^6$</td>
<td>$5.00 \times 10^{15}$</td>
</tr>
</tbody>
</table>

### 3.8 Conclusion

An 18 MV medical linac has been modelled using Geant4 MC code and Varian geometry specifications to investigate the photon, electron and neutron energy spectra reaching the water phantom surface and inside the water phantom. Validation of the simulation to photons and neutrons was found reasonable for this study. There were differences between the simulation studies compared in 3.1. These differences in calculated neutron fluence and number of electrons needed to hit the target to produce 1 Gray at isocentre, are due to: differences in the design and materials of the accelerator, different physics models, geometry of the room, composition of the shielding, quality and detail of geometrical models (flattening filter, primary collimator and jaws) used to model each linac. The neutron energy fluence, simulated in Geant4 for both on the surface and in the solid water phantom, matched with spectra produced by authors in 3.1, within 5% to 10%, which is considered suitable to study the response of detectors to neutrons in this thesis.

In conclusion Geant4 presents a valuable tool in the study of photon, electron and neutron spectra produced by an 18 MV medical linac, which can be used to study the response of different detectors placed both on the surface and inside a solid water phantom. The
Geant4 simulation will be used to study the response of silicon p-i-n diodes to neutrons at different energies in chapter 4, for calculation of the neutron dose equivalent. The Geant4 simulation will also be used to study the directional response of a neutron detector based on three $^3$He tubes in chapter 6 of this thesis.
Chapter 4

Silicon p-i-n diode neutron dosimetry for an 18 MV medical linac

4.1 Introduction

One of the aims of this thesis is to experimentally characterise two types of silicon p-i-n diodes and evaluate their potential use for neutron dosimetry in a mixed photoneutron field produced by an 18 MV linac. This chapter characterises the silicon p-i-n diodes sensitivity to light, temperature and current voltage characteristics prior to irradiation. The silicon p-i-n diodes are also used to investigate the neutron entrance dose for an 18 MV Varian Medical linac. These detectors were calibrated in separate neutron, photon and electron fields. The silicon p-i-n diode detectors have shown excellent discrimination between fast neutron and photon radiation, with sensitivity to fast neutrons being \(\sim 4000\) times higher than to photons from a \(^{60}\)Co source in terms of absorbed dose to tissue. The neutron tissue absorbed dose and neutron dose equivalent was studied both on the surface and inside a cubic solid water phantom, both experimentally and also using Geant4 MC simulations.

4.2 Detector structures

Two different types of silicon p-i-n diodes were investigated in this study; bulk silicon p-i-n diodes with different P+ and Al area and planar silicon p-i-n diodes with wide adjustable base length.

4.2.1 Silicon bulk p-i-n diodes

Ten different types of ion-implanted p-i-n diodes have been developed by the Centre for Medical and Radiation Physics (CMRP) and manufactured by a microelectronic foundry SPA BIT, Ukraine. They were manufactured with an intrinsic silicon base, and resistivity
of approximately 1k Ωcm. The ion-implanted bulk p-i-n diodes have a base length of 1.2 mm and were fabricated from high-purity n-type silicon. The p-i-n diodes were attached to two metal pins and coated with a green epoxy resin, as shown in Fig 4.1.

Figure 4.1: Silicon bulk p-i-n diode

Each of these diodes was fabricated with different geometrical dimensions. P-i-n diodes were fabricated by the conventional planar silicon process, including photolithography, oxidation, ion implantation and metal electrode sputter deposition [39], [40]. The performance of the p-i-n diode will primarily depend on the chip geometry, fabrication process and doping of the intrinsic base region, as previously investigated by Kramer [32]. Table 4.1 identifies the fabricated p-i-n diode specifications used for experimental analysis. The symbol φ denotes a circular diameter. The 5E p-i-n was fabricated with p+ region across of cross section of the p-i-n diode similar to n+ electrode, i.e. it is without photolithography process and the 1F p-i-n diode is said to approximate a photodiode (i.e. Al electrode is deposited on circumference of P+ region only). Table 4.1 outline p+ and Al ohmic contact geometry for each type of the fabricated p-i-n diodes.

4.2.2 Planar silicon p-i-n diodes with adjustable base length

Four different planar p-i-n diodes (Fig 4.2) were fabricated, allowing for the adjustment of the base length to change neutron sensitivity. When compared to bulk pin diodes, planar p-i-n diodes have the advantage of a wider range of sensitivities due to their variable base length. The silicon wafer thickness was 350µm. A schematic of the diodes is given in figure Fig 4.2.
Table 4.1: bulk silicon p-i-n diode specifications used in experiments

<table>
<thead>
<tr>
<th>Name</th>
<th>P+ (µm)</th>
<th>Al (µm)</th>
<th>Crystal (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1Cp-i-n</td>
<td>400x400</td>
<td>600x600</td>
<td>1000x1000</td>
</tr>
<tr>
<td>1Dp-i-n</td>
<td>610x610</td>
<td>600x600</td>
<td>1000x1000</td>
</tr>
<tr>
<td>1Ep-i-n</td>
<td>610x610</td>
<td>620x620</td>
<td>1000x1000</td>
</tr>
<tr>
<td>1Gp-i-n</td>
<td>610x610</td>
<td>880x880</td>
<td>1000x1000</td>
</tr>
<tr>
<td>1Kp-i-n</td>
<td>700x700</td>
<td>880x880</td>
<td>1000x1000</td>
</tr>
<tr>
<td>1Mp-i-n</td>
<td>800x800</td>
<td>880x880</td>
<td>1000x1000</td>
</tr>
<tr>
<td>1Np-i-n</td>
<td>900x900</td>
<td>800x800</td>
<td>1000x1000</td>
</tr>
<tr>
<td>05Ep-i-n</td>
<td>φ310</td>
<td>φ300</td>
<td>500x500</td>
</tr>
<tr>
<td>5Ep-i-n</td>
<td>1000x1000</td>
<td>1000x1000</td>
<td>1000x1000</td>
</tr>
<tr>
<td>1Fp-i-n</td>
<td>φ310, φ1=570, φ2=630</td>
<td>1000x1000</td>
<td>1000x1000</td>
</tr>
</tbody>
</table>

Figure 4.2: Top view of the planar array silicon p-i-n diode structures.

4.3 Semiconductor pulsed current readout system

Previous studies by Speers and Youngblood [73], aimed at characterising silicon damage cross sections, describe readers based on constant current source. Other studies by Nagarkar [74] describe using readers with pulsed current sources, which have longer time durations between readings of up to $\sim 10$ ms. Both of these studies found that the method of readout caused heating of the diodes, thus also annealing of radiation induced defects. However the pulsed current source provided significantly lower annealing effects and proved to be more useful reader. This study involved the use of a Semiconductor Radiation damage monitoring system (Fig 4.3), which was developed at the Centre for Medical and Radiation Physics (CMRP).

The Radiation damage monitoring system has an output of a 1mA pulsed current from a LM334 pulsed current source. The output from the LM334 current source is described in Fig 4.4.
4.3.1 Linearity of pulsed current source

The linearity of the pulsed current source in the readout system developed at CMRP was investigated across a wide range of resistances in the range of 0.5 kΩ to 9 kΩ. This experiment was performed to ensure that the response of the read-out system was linear for diodes with differing forward voltage drop after neutron irradiation. The experimental setup used to perform the linearity measurements is shown in Fig 4.5.

For each resistor the voltage was read directly from the readout system and checked externally with a multimeter. The temperature coefficients of the resistors varied between (350 - 400) PPM/C. The current was also measured by connecting a current probe to the amplifier unit and reading the output directly from the Cathode Ray Oscilloscope (CRO). The results of these measurements are shown in Fig 4.6. The values of forward current...
measured using a CRO were seen to be linear for the case of 1mA reader. The output current remained constant over the resistance range, previously considered. Repeating the experiment one hour later, heralded the same results and showed it could be reproduced, however after repeating the experiment 6 hours after the first experiment, the values did not remain constant but were observed to drift by 3%, indicating that the LM334 pulsed current was not constant. A possible alternative configuration to the circuit used in the LM334 data sheet (page 8) could involve using a TLV431 configured as a pulsed constant current source.

Figure 4.6: Linearity of Pulsed Current source as measured by 1mA reader and externally attached multimeter

This non uniformity in pulsed current from the LM334 pulsed current source could be due to a number of factors including, thermal effects and lead resistance. To understand each of these effects, firstly the basic schematic of the current source and adjacent resistors should be considered, this is shown in Fig 4.7.

Calibration off the LM334 involves a gain adjustment via a one point trim that will trim both slope and zero at the same time. This one point trim is possible because the output of the LM334 extrapolates to zero at 0°C and is independent of $R_{SET}$ or any initial inaccuracy [72]. This property of the LM334 is illustrated in Fig 4.8. The line a-b-c is the initial output before trimming and the line $a'b'c'$ is the desired output after being trimmed. Trimming the gain at $T_2$ is best as it allows the change in slope from start and end points to the desired position.

Even with the gain term properly adjusted, the readings taken, continued to show drift in current over a period of time. This further difficulty in calibration could be due to many factors including thermal effects and lead resistance. Thermal effects include internal heating, which is significant for $I_{SET} > 100\mu\text{A}$ [72]. One possible way to rectify this would be introducing a heat sink, however due to the setup of the circuit components; this
Figure 4.7: The circuit schematic, for the LM334, showing the total current through the device ($I_{SET}$), as the sum of the current going through the SET resistor (IR) and the LM334’s bias current ($I_{BIAS}$) [72]

Figure 4.8: Gain Adjustment for LM334 1mA current source [72]

was not feasible.

The other difficulty in calibration could be due to lead resistance. Lead resistance effects should be minimised by locating the current setting resistor as physically close to the current source. According to instrument standards outlined in [72] it only takes 0.7Ω contact resistance to reduce output current by 1% at the 1 mA level. The current setting resistor was repositioned as close as possible to the current source, and a reduction in current drift was observed.

To improve the 1mA reader stability, there are two propositions. The first is to introduce a fixed smaller resistor and variable larger resistor, or alternatively use a fixed larger resistor and variable smaller resistor, when adjusting the current. The other is to perform linear adjustment of the gain in short periodic increases to correctly adjust for the right cur-
rent and eliminate thermal effects. Care should be taken to allow for a significant period of time between measurements to allow for cooling down of the LM334 pulsed current source and thus reduce the drift in current. A recommended time between performing measurements is 20 seconds and was employed throughout the rest of this study.

4.4 Characterisation of the un-irradiated p-i-n diodes

4.4.1 Current voltage measurements (I-V) characteristics

Current Voltage (I-V) characteristics are measured to understand the p-i-n diodes sensitivity to external radiation. The I-V characteristics will initially exhibit exponential behaviour, as the charge carrier density is increased and the forward resistance of the p-i-n diode lowers. At a particular voltage (threshold voltage) the p-i-n diode will exhibit a slow increase in current, which corresponds to constant device impedance. It is this voltage that changes in proportion with the neutron dose, which shifts the forward bias I-V characteristic curve to the right (Fig 4.9). The curve on the left represents a p-i-n diode I-V characteristic curve that has not been exposed to external radiation. The curve on the right represents the same p-i-n diode that has been irradiated with neutrons with the consequent decrease in carrier lifetime in the base section of the diode, which results in an increase in the overall voltage drop across the diode. Thus by measuring the initial I-V characteristic curves for diodes, we are able to compare changes in the detector impedance, represented as a change in the forward voltage response of each diode post irradiation.

![Figure 4.9: p-i-n diode I-V characteristic curves](image)
By using a current source of approximately 1 mA pulsed at a frequency of about 100 Hz and with a pulse period of 1 ms and a peak detector circuit it is possible to readout the diodes with a duty cycle of 0.1. This minimizes heating and annealing of the diode. Readouts of the forward bias voltage change, when measured under constant current are accurate to within 1 mV. The readout is performed by pulsing a constant current source, which is based on a operational amplifier. The use of a pulsed current source is employed to reduce ohmic heating of the p-i-n diode. The effect of this heating would be to anneal the p-i-n diode, which would in turn change the forward bias voltage response of the detector. A diagram of the readout system used to measure the forward voltage change of the p-i-n diodes used in this work is shown in 4.10. This readout system is the same used as Carolan [71].

The frequency of the pulsed current source used was about 100 Hz and the pulse period was 1 ms. Readout is able to performed every 2s without causing excessive heating of the p-i-n diodes. Pulsing minimizes heating of the diode and reduces annealing processes and changes in the forward bias voltage of the diode, allowing for a more accurate diode measurement.

A standard resistor was periodically tested to maintain constancy and check calibration. The measurements of the p-i-n diodes were performed in the same session over the course of a few hours. The I-V characteristic curves were measured directly under constant current conditions at the CMRP University of Wollongong Laboratory, using I-V tracer created in Labview.

### 4.4.2 Light and temperature sensitivity of silicon p-i-n diodes

The temperature dependence of the p-i-n diodes under investigation was examined across the temperature range of (20-50) °C. The calibration of the p-i-n diode temperature sensi-
tivity is required, in order to account for any temperature fluctuations encountered when performing measurements with these detectors. A detector that is sensitive to temperature fluctuations will have an increase in forward bias voltage, when measured under constant current [71]. Any variation in the forward bias voltage response across this range should be accounted for and a correction factor applied in order to preserve accurate dosimetry.

The voltage across the diode junctions, with a thin base is proportional to the natural logarithm of the current density. The junction diode equation affectsively describes this trend as, \( I \sim \exp \left( \frac{eV}{2kT} \right) \) [63], where \( V \) is the voltage across the junction, \( k \) is the Boltzmann constant, \( e \) is the electron charge and \( T \) is the temperature. The effect of temperature on current density can be attributed to the thermal excitation of electrons from the valence band into the conduction band. Thus a rise in room temperature in the room where experiments are being performed, will result in an increase in the number of electrons excited into the conduction band and the overall detector reading.

In order to investigate the change in forward bias voltage with change in temperature, each of the bulk silicon p-i-n diodes and planar silicon p-i-n diodes were measured using the radiation damage monitoring system (Fig 4.3), across the temperature range of (20-50) °C. This was achieved through the creation of a temperature controlled environment where a p-i-n diode was placed in a small zip lock tight bag and immersed in a water bath that was heated gradually from room temperature to approximately 50°C. The temperature of the water bath was monitored by the use of a digital thermometer with accuracy of 0.1°C. The average cooling rate was approximately 1°C/min to ensure that the p-i-n diode, water and thermometer were all in thermal equilibrium. The experimental setup used is shown in Fig. 4.11 below. Depending on the base length of each diode, different theories of the effect of temperature on the forward bias response of the p-i-n diode can be applied and are discussed in more detail by Swartz [27].

\[ \text{Figure 4.11: Experimental setup for p-i-n diode temperature dependence measurements} \]

In addition to thermal excitations, photons of visible light may also contribute to the
conductivity in the semiconductor device. About 3.6eV is required to produce an electron hole pair in a semiconductor device. Visible light has a wavelength in the range (400-700) nm, which corresponds to energies in the range (1.7-3.1) eV, which may contribute to the conductivity of the diode. The effect on the forward bias response of the p-i-n diodes when exposed to visible light is examined in this chapter in order to determine if an opaque encasing is required during experiments.

### 4.5 Results of the characterisation of the un-irradiated detectors

#### 4.5.1 Results of I-V measurements of the silicon p-i-n diodes

The I-V characteristics displayed below are for bulk silicon diodes: 0.5E's and planar p-i-n diodes A1, A2, A3, A4. The remaining bulk silicon diodes are displayed in appendix A.

![Figure 4.12: Bulk silicon 0.5E p-i-n diodes forward bias I-V characteristic.](image)

The bulk silicon 0.5E diodes, as seen in Fig. 4.12 had an average forward voltage response of 1.45V for a current of 1mA. The reverse I-V characteristics of bulk silicon 0.5E diodes are shown in Fig. 4.13, which is exhibiting a broad range of stopping voltage values. The 0.5E (ir) diode was seen to have a greater forward threshold voltage (Fig. 4.12) and lower stopping voltage (Fig. 4.13). This is due to the diode being previously irradiated, by masters student Levent.

The resistance across the base length of the 0.5E p-i-n diodes was measured using a multimeter to be ∼1KΩ. These diodes had the highest resistivity across the base, in comparison to the other 9 bulk silicon p-i-n diodes (presented in table 4.1). This is evident from the I-V
CHAPTER 4. SILICON P-I-N DIODE NEUTRON DOSIMETRY FOR AN 18 MV MEDICAL LINAC

Figure 4.13: Bulk silicon 0.5E p-i-n diodes reverse bias I-V characteristic.

Figure 4.14: Planar p-i-n diodes forward bias I-V characteristic.

Figure 4.15: Planar p-i-n diodes reverse bias I-V characteristic.
characteristics obtained (see appendix A). The differences in I-V characteristics across all p-i-n diodes is due to the different design in Al dimensions, P+, N+, crystal dimensions, as summarised in table 4.1.

The results for Planar p-i-n diodes A1, A2, A3, A4 diodes, as seen in Fig. 4.14 and Fig. 4.15. At a current of 2mA the A1 detector exhibited a voltage response of 1.25V which is similar to 0.5E detector with response at 2mA being approximately 1.46 V. As the base length of the diode increases (detectors A2, A3 and A4) the forward bias threshold voltage was seen to increase. The stopping voltage of type 2 detectors had very little change in current, with decreasing voltage (Fig. 4.15), except for the A1 planar p-i-n diode that exhibited a steep reverse bias I-V curve with stopping voltage approximately 48V.

4.5.2 Results of light sensitivity of silicon p-i-n diodes

The effect of light on the forward bias voltage response of both bulk and planar silicon p-i-n diodes when measured with a 1mA pulsed current source (Radiation monitoring system - Fig. 4.3) was examined. The measurements were repeated but with the diode encased in a light tight box. The voltage reading obtained for the diodes unexposed to visible light was subtracted from the voltage reading obtained for the diodes exposed to light. This difference was obtained for each diode and the results summarised in table 4.2 and 4.3 below.

<table>
<thead>
<tr>
<th>p-i-n type</th>
<th>Difference in Reading (V)</th>
<th>Uncertainty (V)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1K</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>1M</td>
<td>0.036</td>
<td>0.010</td>
</tr>
<tr>
<td>1N</td>
<td>0.003</td>
<td>0.007</td>
</tr>
<tr>
<td>5E</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>0.5E</td>
<td>0.002</td>
<td>0.001</td>
</tr>
<tr>
<td>1F</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>1C</td>
<td>0.002</td>
<td>0.001</td>
</tr>
<tr>
<td>1D</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>1E</td>
<td>0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>1G</td>
<td>0.002</td>
<td>0.001</td>
</tr>
</tbody>
</table>

For most bulk silicon p-i-n diodes the difference in forward voltage measurements, is no greater than the uncertainty involved in the measurement (1mV). This proves that the reader is working accurately over this voltage range and that the effect of visible light on the forward bias response of bulk silicon bulk p-i-n diodes is mostly negligible, except for 1M and 1N bulk p-i-n diodes (for which the area of the Al contact was less than the...
Table 4.3: Effect of light exposure on planar silicon p-i-n diodes

<table>
<thead>
<tr>
<th>p-i-n type</th>
<th>Difference in Reading (V)</th>
<th>Uncertainty (V)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A1</td>
<td>0.004</td>
<td>0.002</td>
</tr>
<tr>
<td>A2</td>
<td>0.012</td>
<td>0.002</td>
</tr>
<tr>
<td>A3</td>
<td>0.015</td>
<td>0.002</td>
</tr>
<tr>
<td>A4</td>
<td>0.025</td>
<td>0.002</td>
</tr>
</tbody>
</table>

P+ implanted region and where photons are absorbed by the P+ region). On the other hand planar silicon p-i-n diodes exhibited much larger change in response when exposed to light, with signal changes between 4-25 mV between readings. This is due to the i-region being uncovered by Al and thus allowing photons to be absorbed by the p-n junction. These detectors require to be always covered by a light protective sheath when performing measurements or in a dark environment, as they are sensitive to light.

4.5.3 Results of temperature sensitivity of silicon p-i-n diodes

Measurements of the change in threshold voltage were made using the neutron radiation monitoring system (Fig. 4.3), at 1mA pulsed current. Data was taken periodically over the 20-50 °C temperature range. The temperature versus voltage data for one of the bulk silicon p-i-n diodes (1M) is displayed in figure Fig. 4.16 below. A linear fit was applied in order to determine a temperature sensitivity coefficient for the detector. This process was repeated for all silicon p-i-n diodes under investigation and the temperature sensitivity coefficients determined and summarised in tables table 4.4 and table table 4.5.

![Temperature Dependence for the 1M p-i-n diode](image-url)

**Figure 4.16: Temperature Dependence for the 1M p-i-n diode**
Measurements taken for bulk silicon p-i-n diodes, had average temperature coefficient of 0.7 mV/°C. The 1M p-i-n diode was the most sensitive diode to temperature, followed by the 1N and 1E diodes. For planar silicon p-i-n diodes the temperature sensitivity coefficient was seen to be dependant on the length of the base length, with the temperature sensitivity coefficient increasing with increasing detector base length.

A temperature correction factor based on the determined temperature sensitivity coefficient was applied whenever the detectors were used to perform measurements of a radiation field and a temperature fluctuation occurred in the experimental setup.

Table 4.4: Temperature sensitivity of bulk silicon p-i-n diodes

<table>
<thead>
<tr>
<th>Detector</th>
<th>Temperature sensitivity (mV/°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1C</td>
<td>0.32</td>
</tr>
<tr>
<td>1D</td>
<td>0.27</td>
</tr>
<tr>
<td>1E</td>
<td>1.05</td>
</tr>
<tr>
<td>1F</td>
<td>0.60</td>
</tr>
<tr>
<td>1G</td>
<td>0.39</td>
</tr>
<tr>
<td>1K</td>
<td>0.87</td>
</tr>
<tr>
<td>1M</td>
<td>1.80</td>
</tr>
<tr>
<td>1N</td>
<td>1.61</td>
</tr>
<tr>
<td>5E</td>
<td>0.48</td>
</tr>
<tr>
<td>0.5E</td>
<td>0.52</td>
</tr>
</tbody>
</table>

Table 4.5: Temperature Sensitivity of planar silicon p-i-n diodes

<table>
<thead>
<tr>
<th>Detector</th>
<th>Width (mm)</th>
<th>Temperature Sensitivity (mV/°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A1</td>
<td>1.8</td>
<td>13.20</td>
</tr>
<tr>
<td>A2</td>
<td>3.2</td>
<td>19.30</td>
</tr>
<tr>
<td>A3</td>
<td>2.9</td>
<td>15.20</td>
</tr>
<tr>
<td>A4</td>
<td>4.0</td>
<td>20.10</td>
</tr>
</tbody>
</table>
CHAPTER 4. SILICON P-I-N DIODE NEUTRON DOSIMETRY FOR AN 18 MV MEDICAL LINAC

4.6 Methods of neutron, photon and electron sensitivity calibration

4.6.1 Neutron calibration of the silicon p-i-n Diodes

The p-i-n diodes were calibrated in terms of tissue fast neutron KERMA, using a 14 MeV deuterium tritium fast neutron generator at the Commonwealth Scientific and Industrial Research Organization. The neutron calibration of p-i-n diodes in terms of neutron tissue KERMA, with 14 MeV fast neutrons is valid within 15% for the neutron spectrum of a $^{252}$Cf source (average neutron energy 1 MEV), which has a average fast neutron energy spectrum to an 18 MV medical linac. [9], [55] The suitability of the calibration using 14 MeV neutrons is justified by the fact that the ratio of the silicon to tissue displacement damage KERMA coefficient is 0.00914 and 0.00859 (Fig4.1(b)) for 1MeV and 14 MeV neutrons respectively, which is within 6%. The total tissue equivalent absorbed dose (D) at the point of the diode calibration is given by (equation 4.1) [63]:

$$D = K = F_n \phi$$

(4.1)

Where $\phi$ is the neutron fluence (cm$^{-2}$) and $F_n$ is the neutron tissue KERMA factor of $6.06 \times 10^{-11}$Gy cm$^2$ for 14 MeV neutrons [25]. The detectors were taped onto a large piece of tape (Fig. 4.17) and then taped onto the surface of the tube, where the neutron flux is known to be highest. Measurements of the silicon p-i-n diodes response (mV) to neutrons was performed every hour using the semiconductor radiation monitoring system at 1mA pulsed current.

![Figure 4.17: Diodes mounted on tape, waiting to be placed on the surface of the DT tube](image)

An activation copper foil was placed alongside the diodes and following irradiation was used to measure the neutron fluence (Fig. 4.18). Due to the 9.75 min decay time, the
copper foil measurement only indicates the neutron flux over the last minutes of the irradiation. It was necessary to use a fast neutron monitor to measure the relative flux over the course of the experiment [56]. A fast neutron monitor was made from a 10cm x 10cm x 5 cm EJ-204 plastic scintillator coupled to a photomultiplier tube (PMT). This was mounted at a 130 cm distance above the tube centre. The photomultiplier anode output was amplified using an Ortec 113 preamp feeding an Ortec 472 Spectroscopy Amplifier (250 ns shaping). An APTEC 5004 multi-channel analyser (MCA) was used to collect pulse height spectra from the detector. The MCA was set to continually collect spectra of 60 s live-time under control of a script. The counts between 50% and 110% of the spectrum endpoint (corresponding to full energy deposition of a 14 MeV neutron) were used to indicate the relative neutron flux.

![Figure 4.18: Copper foil spectral measurement used to determine the neutron flux.](image)

At the end of each irradiation, the copper foil was immediately transported to the gamma detector for measurement of the annihilation gamma-rays resulting from the $^{63}$Cu(n,2n)$^{62}$Cu reaction. The cooling time was measured with a stopwatch. The Bicron 6H4/5L gamma detector uses a cylindrical 6 inch (15.24cm) x 4 inch (10.16cm) Na (TI) scintillator coupled to a Photomultiplier tube. The electronic readout of the anode signals used an Ortec 113 Preamp feeding a Amptek PX-5 digital pulse processing system. The Amptek provides a digital trapezoidal filter and subsequent pulse height analysis. The foil was sandwiched between the detector and desk to limit the range of emitted positrons. The gamma detector readout was used to convert the scintillator count-rate to a measure of the neutron flux $\phi$, at the diode location which was determined by equation 4.2 and thus allowed the total fluence delivered to the detector to be calculated.

$$\phi = \frac{r_N N_\gamma}{\rho N} \frac{1}{(1 - \exp(-r_{t_a}) \times \exp(-r_{t_c}) \times (1 - \exp(-r_{t_m}))}$$

(4.2)
Where $N_{\gamma}$ is the number of detected 511keV gammas, $N$ is the number of $^{63}\text{Cu}$ nuclei in the foil and $\sigma$ is the copper reaction cross section. $t_a$, $t_c$ and $t_m$ refer to the activation cooling (transfer time from ceasing DT irradiation and starting the measurement). $P_d$ is the probability that the gamma detector detects an emitted 511keV gamma photon. The annihilation detection efficiency was previously calculated using a Geant4 simulation with experimental verification.

### 4.6.2 Photon calibration of the silicon p-i-n diodes

Photon calibration of p-i-n diodes was carried out in free air geometry using a $^{60}\text{Co}$ source up to a dose of 10kGy at the Gamma Technology Research Irradiator (GATRI) at the Australia Nuclear and Science Technology Organization (ANSTO). The diodes were placed between two 10 cm thick Polystyrene slabs. The total photon dose delivered to the diodes in terms of the dose to water was measured by an Amber Perspex dosimeter. The diode response in free air was then converted to dose in water. Additionally the photon calibration was also carried out using a 6 MV medical linac, with detectors placed on the surface of a 30cmx30cmx30cm solid water phantom and at Dmax. The detectors were irradiated with 3000MU (30Gy at Dmax) at 600MU/min by a 10cm x 10cm field at an SSD of 100cm (surface measurement) and 98.5cm SSD for Dmax measurement.

### 4.6.3 Electron calibration of the silicon p-i-n diodes

Electron calibration of the p-i-n diodes was carried out using a (6, 9, 12, 16) MeV mono energetic electron beams from a 6 MV medical linac. Five bulk p-i-n diodes were selected and placed on the surface of a solid water phantom at the central axis (Fig. 4.19). The detectors were irradiated with a dose of 2000MU corresponding to 20Gy dose in water delivered at Dmax. The dose delivered on the surface of the phantom was estimated using the electron depth dose table provided from hospital commissioning data. The p-i-n diodes response in terms of the forward voltage shift mV/cGy (water) was plotted against the incident electron energy and a line of best fit $L(E_e)$ was determined. The response of the bulk silicon p-i-n diode with volume 1x1x1.2mm$^3$, to electrons both on the surface and inside a phantom in terms of mV was calculated using equation (4.3).

$$mV(\text{electrons}) = \int_{E_{\text{min}}}^{E_{\text{max}}} D(E_e)L(E_e)S(E_e).dE$$ \hspace{1cm} (4.3)

The dose deposited to the bulk silicon p-i-n volume by electrons of energy $E_e$ which are depositing energy $D(E_e)$ is convoluted with the measured line of best fit $L(E_e)$ and the ratio of mass stopping powers of water to silicon $S(E_e)$ [57]. The dose deposited to the 1x1x1.2mm$^3$ silicon volume was estimated using Geant4 simulation (Chapter 3). An 18 MV 10cmx10cm beam of photons was incident on the silicon volume. The kinetic energy
of electrons entering the silicon detector was recorded using the pre-step information and the total dose deposited in the detector due to the incident electron was scored. A photon dose of 30Gy delivered at Dmax was used for normalization.

### 4.6.4 Geant4 Simulation of silicon p-i-n diodes in an 18 MV Medical Varian linac beam

To study the response of the bulk silicon p-i-n diodes both on the surface and inside a solid water phantom, Geant4 version 10.01.p01 was adopted as Monte Carlo code. An 18 MV medical Varian iX Silhouette Clinac was modelled, omitting the linac head shielding and bunker walls (see chapter 3).

The response of the p-i-n diode in terms of mV to the photon component of the 18 MV beam at different depths in a phantom is not obvious to calculate using a gamma calibration coefficient $\Delta V/D_\gamma$ obtained from the $^{60}$Co calibration. This is due to the equilibrium spectra of secondary electrons for an 18 MV photon field being different to a $^{60}$Co photon field. The method described in section 4.6.3 was used instead. The neutron response of the p-i-n diode in terms of mV at any depth in a phantom was calculated as a difference between the total measured response of the p-i-n diode and the calculated response to electrons in mV (section 4.6.3).

Additionally, Geant4 was used to simulate the electron and neutron energy spectra at
CHAPTER 4. SILICON P-I-N DIODE NEUTRON DOSIMETRY FOR AN 18 MV MEDICAL LINAC

each depth in the phantom, in order to estimate the electron and neutron fluence reaching the p-i-n diode. The simulated electron energy fluence \( \phi_e(E) \) was used to calculate the p-i-n diode response in terms of silicon electron displacement damage KERMA at each depth in the phantom, using equation (4.4) and the electron displacement damage KERMA factors \( k_e(E) \) in silicon. [57]

\[
electronKERMA_{displacement} = \int_{E_{min}}^{E_{max}} \phi_e(E) k_e(E) \, dE \quad (4.4)
\]

Similarly, the neutron energy spectrum was used to calculate the p-i-n diode response in terms of silicon neutron displacement damage KERMA at each depth in the phantom using equation (2.7) and the neutron displacement damage KERMA factors [66]. In all cases, a photon dose of 30Gy delivered at Dmax was used for normalization. The neutron dose equivalent (H) was determined using equation 4.5.

\[
H = \sum_E \phi(E) K(E) Q_n(E) \quad (4.5)
\]

Where \( K(E) \) represents the neutron tissue kerma factors obtained from Caswell [14] and \( Q_n(E) \) represents the neutron quality factors [68].

4.6.5 Irradiation of silicon p-i-n diodes using an 18 MV Medical Varian linac

The experiment was carried out at the St George Cancer Care Centre, St George Hospital, Kogarah in Sydney, using an 18 MV Varian iX Silhouette Clinac. Five of the bulk p-i-n diodes of the selected type were placed at the central position of the solid water phantom and irradiated at the surface (Fig. 4.20(b) up to a depth of 10cm, at 1cm increments per each depth in the phantom. A dose of 30Gy at Dmax for a 10cmx10cm field size and 100cm SSD was delivered for each detector depth. The average response across all 5 of the bulk 0.5E p-i-n diodes placed at the centre of the field were used to obtain the total detector response at each depth and calculate the absorbed neutron dose.

Additionally to the central position, the 0.5E bulk silicon p-i-n diodes were irradiated at 4 other positions on the surface of a phantom (Fig. 4.20(a)), with radiation fields 5x5cm², 10cm x 10cm cm², 20x20cm² and 40x40cm². This geometry is used to calculate the entrance neutron dose to the target (central) diode and within close proximity to the target. The photon dose in water at the central position at each depth was provided from the hospital commissioning data using a 0.125cc ionization chamber and the dose at the other 4 surface positions was measured using CMRP MOSkin detectors [69].

The response of the p-i-n diode in terms of mV is due to photons, secondary electrons
Figure 4.20: (a) Experimental Setup for investigation of neutron KERMA across surface of phantom. (b) Experimental setup for investigation of neutron KERMA with changing depth in the phantom.

(both included in tissue dose $D_\gamma$ and neutron interactions (tissue neutron dose $D_n$), which is related by equations (4.6) and (4.7):

$$\Delta V = \alpha D_\gamma + \beta D_n$$  \hspace{1cm} (4.6)

$$D_n = \frac{\Delta V - \alpha D_\gamma}{\beta}$$  \hspace{1cm} (4.7)

Where $\Delta V$ is the total change in the voltage response following irradiation, $D_\gamma$ is the photon dose, $\alpha$ is the sensitivity of the diode to photons, $\beta$ is the diode sensitivity to neutrons and $D_n$ is the measured neutron KERMA.
4.7 Results of neutron, photon and electron sensitivity calibrations

4.7.1 Results of neutron sensitivity calibration with 14 MeV fast neutrons

The neutron fluence and flux measurements are shown in Fig. 4.21. The generator was shut down for 2 hours after being turned on for 3 hours, as can be seen by the sudden drop in both plots. After being shut down for 2 hours, the generator was turned on again and the output increased in flux and the fluence delivered to the detectors. The total fluence delivered to the detectors during the measurement was $2.89 \times 10^9 \text{ cm}^{-2}$ with an uncertainty of 5%. The fading was negligible within 20 hours after irradiation.

![Figure 4.21: Neutron flux and fluence calibration measurement.](image)

The planar p-i-n diodes demonstrated sensitivity to fast neutrons approximately 6-8 times higher than bulk p-i-n diodes with base length 1.2mm, depending on their geometry. In Fig. 4.22 planar A3 detector with the base length 2.9mm showed a higher sensitivity than the Planar A4 detector with base length 4mm. This effect is under further investigation.

Although planar diodes have good neutron/photon sensitivity, they also have much higher temperature sensitivity coefficient than bulk p-i-n diodes. Bulk silicon 0.5E p-i-n diodes when placed on the surface of the patient, will be in thermal equilibrium with a patient body and thus avoid error in dosimetry associated with temperature instability.
4.7.2 Results of photon sensitivity calibration with a $^{60}$Co Source

The response of the p-i-n diode detector to photons was tested using both a $^{60}$Co source and a 6 MV linac. The silicon p-i-n diodes response when exposed to photons from a $^{60}$Co source between two slabs of PMMA slabs is shown in (Fig. 4.23). The greatest change in voltage occurs within 1kGy of delivered dose due to introduction of the defects mostly effecting the life time degradation of the charged carriers, as predicted by equation (2.8). After 1kGy, the response of the p-i-n diodes saturates, whilst the gamma dose is not enough to change the resistivity of silicon. This is due to the detector being fabricated from high resistivity n type silicon. The life time of the charge carriers is not changing after irradiation with high gamma dose. The results of the detectors sensitivity to neutrons and photons are summarized in table 4.6.

4.7.3 Results of photon sensitivity calibration with a 6 MV medical linac

For calibration using a 6 MV linac, a total dose of 60 Gy was delivered to a single bulk 0.5E p-i-n diode, placed at the centre of a 10cmx10cm field at Dmax, and measured three times to obtain an average response. The measured photon sensitivity of the bulk 0.5E p-i-n diode was $3.3 \times 10^{-4}$mV/cGy which is within 20% of the sensitivity $3.9 \times 10^{-4}$mV/cGy obtained using the $^{60}$Co source. However these sensitivity calibrations cannot be used to accurately estimate the detector response to photons from an 18 MV linac beam, as the secondary electron spectra are different to one another. In order to estimate the p-i-n diode response to photons in the phantom, we must consider the electron KERMA as discussed in section 4.6.3.
4.7.4 Summary of the detector sensitivity to neutrons, $^{60}$Co photons and temperature

The results of the detectors sensitivity to neutrons and photons are summarized in table 4.6. The bulk 0.5E p-i-n diodes were found to have the best ratio of neutron/photon sensitivity and temperature stability, so were selected to perform measurements on an 18 MV medical linac.

**Table 4.6**: Bulk and planar p-i-n diode sensitivity to neutrons, photons from $^{60}$Co source and temperature.

<table>
<thead>
<tr>
<th>Detector</th>
<th>Base length (mm)</th>
<th>Neutron Sensitivity (mV/cGy)</th>
<th>Photon Sensitivity ($x10^{-3}$ mV/cGy)</th>
<th>Temperature Sensitivity (mV/°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1C</td>
<td>1.2</td>
<td>1.23</td>
<td>0.24</td>
<td>0.32</td>
</tr>
<tr>
<td>1D</td>
<td>1.2</td>
<td>1.06</td>
<td>0.22</td>
<td>0.27</td>
</tr>
<tr>
<td>1E</td>
<td>1.2</td>
<td>0.78</td>
<td>0.19</td>
<td>1.05</td>
</tr>
<tr>
<td>1F</td>
<td>1.2</td>
<td>1.37</td>
<td>0.20</td>
<td>0.31</td>
</tr>
<tr>
<td>1G</td>
<td>1.2</td>
<td>0.90</td>
<td>0.19</td>
<td>0.39</td>
</tr>
<tr>
<td>1K</td>
<td>1.2</td>
<td>1.16</td>
<td>0.22</td>
<td>0.87</td>
</tr>
<tr>
<td>1M</td>
<td>1.2</td>
<td>1.01</td>
<td>0.23</td>
<td>1.80</td>
</tr>
<tr>
<td>1N</td>
<td>1.2</td>
<td>0.93</td>
<td>0.19</td>
<td>1.61</td>
</tr>
<tr>
<td>5E</td>
<td>1.2</td>
<td>1.16</td>
<td>0.13</td>
<td>0.48</td>
</tr>
<tr>
<td>0.5E</td>
<td>1.2</td>
<td>1.50</td>
<td>0.35</td>
<td>0.52</td>
</tr>
<tr>
<td>A1</td>
<td>1.8</td>
<td>1.00</td>
<td>10.30</td>
<td>13.20</td>
</tr>
<tr>
<td>A2</td>
<td>3.2</td>
<td>3.63</td>
<td>20.40</td>
<td>19.30</td>
</tr>
<tr>
<td>A3</td>
<td>2.9</td>
<td>9.07</td>
<td>18.50</td>
<td>15.20</td>
</tr>
<tr>
<td>A4</td>
<td>4.0</td>
<td>6.70</td>
<td>24.40</td>
<td>19.10</td>
</tr>
</tbody>
</table>
4.7.5 Results of electron calibration of the silicon p-i-n diodes

Fig. 4.24 shows the bulk p-i-n 0.5E diode sensitivity to electrons in water, obtained from irradiation using mono-energetic electron beams from a medical linac. The best line of fit was calculated to be 0.00095±0.00010 MeV mV/cGy. It was observed that there is a linear relationship between electron energy and the detector response in this energy range. A line of best fit was determined and together with simulated electron energy spectrum at each p-i-n diode position in the phantom and was used to estimate the response of the diode in terms of mV. In equation (4.6) \( D_\gamma \) is the simulated response of the p-i-n diode to secondary electrons in mV and thus the experimental neutron dose can be calculated using equation (4.7).

![Figure 4.24: Bulk 0.5E p-i-n diode sensitivity to electrons in terms of mV per cGy (water).](image)

Additionally the response of the 0.5E p-i-n diode on the surface of the phantom, was irradiated with a 10cmx10cm field of 3000 MU (30Gy) using a 6 MV linac at 100cm SSD. The response was not measurable by the reader (voltage change was less than 1 mV). This is expected due to lack of secondary electrons and suggests that response of the p-i-n diodes on a phantom surface can be mostly associated with the fast neutron dose only, in the case of irradiation with an 18 MV linac.

4.7.6 Results of a Geant4 simulation of silicon p-i-n diodes in an 18 MV Medical Varian linac beam

The photoneutron energy fluence inside a water phantom, irradiated with an 18 MV linac, calculated by means of the Geant4 simulation (see section 4.6.4 Method) is shown in Fig. 4.25. The photoneutron energy fluence is shown to be greatest at the surface for
fast neutrons as expected. As the photoneutrons lose energy inside the water phantom the spectrum is moderated. The intermediate and thermal neutron fluence is increasing with depth and peaking at 4cm depth, before reducing with increasing depth in the phantom, due to the complete absorption of neutrons. Fast neutron fluence in the spectrum is corresponding to the right peak in Fig. 4.25. The peak is reducing with increasing depth, broadening and shifting to the higher neutron energies. The peak on the left side, in Fig. 4.25 is due to thermal neutrons which is increasing up to rcm in the phantom, before decreasing with increasing depth in the water phantom.

![Figure 4.25: photoneutron Energy fluence with respect to the neutron energy, at different depths in the water phantom.](image)

Fig. 4.26 and Fig. 4.27 show the energy spectra of photons and electrons, respectively, detected at different depths in the phantom. A higher number of photons is detected at the surface of the phantom as expected; photons scatter and therefore less photons will be detected by silicon diodes at greater depths, along the main axis of the phantom. The electron energy spectrum was observed to gradually increase up to a depth of (3-4) cm in the phantom, which is in the region of the expected Dmax of 3.3cm for an 18 MV medical linac. The electron energy spectrum then decreases in magnitude after 4cm, with increasing depth in the solid water phantom.
4.7.7 Results of irradiation using an 18 MV Medical Varian linac

The neutron entrance absorbed dose was studied on the surface of a cubic solid water phantom with size 30cm x 30cm x 30cm, irradiated by an 18 MV Varian a high energy iX Silhouette Clinac at SSD=100cm. The p-i-n diodes were irradiated with photons at 3000MU with radiation fields adjusted by jaws 10cm x 10cm, 20x20cm² and 40x40cm² for both in and out of the field positions. The bulk p-i-n diodes were placed at 5 positions, equidistant at 7.5cm from the centre of the phantom as shown in Fig. 4.28(a). Fig. 4.28(b) depicts the relative neutron absorbed dose at the centre of the field normalized to the photon dose at Dmax.
It was demonstrated that the relative neutron absorbed dose normalized to the photon dose at Dmax for a 10cm x 10cm cm² field size, decreases fast out of field, about 0.3-0.4 % for 2.5cm from the edge of 10cm x 10cm cm² field. For 20x20cm² and 40x40cm² field sizes the detectors are infield and the relative neutron dose normalized to the photon dose at Dmax is between 1.5-2.0 %. The neutron entrance absorbed dose across the surface of the phantom was seen to be slightly asymmetric as the jaws were adjusted to form a larger field size. It is important to note that the results obtained, were based on low statistics and the irradiation dose would need to be increased to improve the current results.

Figure 4.29(a) shows the simulated and experimental total response of the p-i-n diode percentage response due to effects from; IEL from photons (dose deposited from photons), NIEL from electrons and NIEL from neutrons, normalized to the maximum value at 2cm depth. The total response was seen to increase up to 2cm in the phantom before decreasing with increasing depth. The simulation results agreed within error to the experimental results.

Fig. 4.29(b) and Fig. 4.29(c) show the relative neutron displacement damage KERMA and relative displacement damage electron KERMA respectively, each plotted with respect to changing depth in the phantom. Experimental electron damage KERMA was obtained using simulated electron spectral fluence convoluted with measured electron energy response on the p-i-n diode as explained in (section 4.6.3). Experimental neutron damage KERMA in terms of mV was obtained by the subtraction of electron damage KERMA in terms of mV from the total p-i-n diode response in mV at the same depth. The experimental and simulation results agreed within error for most points, with larger errors seen for the experimental neutron displacement damage KERMA measurements at greater depths, due to lower statistics. There was a build-up of the neutron displacement damage KERMA from the surface to a maximum response at 1cm depth before decreas-
Fig. 4.29(d) is showing the partial percentage contribution of NIEL (electrons) and NIEL (neutrons) contributing to the total detector response. The detector was found to have the greatest sensitivity to neutrons on the surface and within 1cm depth in the solid water phantom, before decreasing exponentially. The detector response to electrons and photons is minimal at the surface and increasing with depth in the phantom. After 2cm the detector response becomes due to both NIEL (neutrons) and NIEL (electrons) and at 3cm the detector response becomes mostly due to NIEL (electrons), as most neutrons are rapidly losing energy and being absorbed, whilst the photon dose is reaching build-up at 3.3cm depth in the solid water phantom. The calculated mV response of the bulk p-i-n diode to electrons was estimated using the discussed methods in section 4.6.3. The results of the neutron absorbed tissue dose are summarized in table 4.7 below.
Table 4.7: Bulk silicon p-i-n diode response to neutrons, electrons and calculated tissue neutron dose

<table>
<thead>
<tr>
<th>Depth (cm)</th>
<th>Total response (mV)</th>
<th>Electron response (mV)</th>
<th>Neutron response (mV)</th>
<th>Fast neutron tissue dose (cGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>10</td>
<td>0.75</td>
<td>9.25</td>
<td>6.17</td>
</tr>
<tr>
<td>1</td>
<td>14.25</td>
<td>3.62</td>
<td>10.63</td>
<td>7.09</td>
</tr>
<tr>
<td>2</td>
<td>16</td>
<td>6.39</td>
<td>9.61</td>
<td>6.4</td>
</tr>
<tr>
<td>3</td>
<td>15.75</td>
<td>7.41</td>
<td>8.34</td>
<td>5.56</td>
</tr>
<tr>
<td>4</td>
<td>14.25</td>
<td>7.53</td>
<td>6.72</td>
<td>4.48</td>
</tr>
<tr>
<td>5</td>
<td>14</td>
<td>7.56</td>
<td>6.44</td>
<td>4.29</td>
</tr>
<tr>
<td>6</td>
<td>13</td>
<td>7.37</td>
<td>5.63</td>
<td>3.75</td>
</tr>
<tr>
<td>7</td>
<td>12.25</td>
<td>7.25</td>
<td>5</td>
<td>3.33</td>
</tr>
<tr>
<td>8</td>
<td>11.5</td>
<td>7.07</td>
<td>4.43</td>
<td>2.95</td>
</tr>
<tr>
<td>9</td>
<td>10.5</td>
<td>6.78</td>
<td>3.72</td>
<td>2.48</td>
</tr>
<tr>
<td>10</td>
<td>10</td>
<td>6.6</td>
<td>3.4</td>
<td>2.27</td>
</tr>
</tbody>
</table>

The tissue neutron absorbed dose was found to be greatest at 1 cm depth to a value of 7.09 cGy, per 30 Gy photon dose at D_max, with an uncertainty of 10%. The tissue fast neutron dose at the surface of the phantom is comprising 0.22% of the total photon dose delivered at D_max. The detector response on the surface of the phantom is driven predominantly by fast neutrons, with insignificant contributions from photons and electrons (within the measurement error of 1 mV). This is supported by the calculation of the neutron displacement damage KERMA in silicon using equation (2.7), for spectra of neutrons on the surface of the phantom (Fig. 4.25). It was calculated that for neutrons with energy of > 0.1 MeV is 96% of the total contribution from the neutron spectra is from fast neutrons. At a depth of 3 cm, the contributions from neutrons and secondary electrons are approximately equal.

The neutron dose equivalent on the surface of the phantom was determined by equation 4.5. The entrance neutron dose equivalent per 1 Gy of photon dose at D_max was calculated for an 18 MV Varian Clinac, with 10 cm x 10 cm field size at 100 cm SSD based on the neutron spectrum. The neutron dose equivalent was calculated to be 17.42 mSv/Gy, partitioned as 17.08 mSv/Gy for En > 0.1 MeV and 0.34 mSv/Gy for En < 0.1 MeV. The ratio response of neutrons to photons in terms of dose equivalent is shown in table 4.8. Differences in the simulation physics models, details of the geometry and shielding in the simulation setup, will affect the calculation of the neutron dose equivalent. This makes it difficult to compare the simulated neutron dose equivalent obtained for different linacs, confirmed by comparison in table 4 from [70]. For the purposes of this study, the Geant4 Monte Carlo code was found reasonable to model the neutron and photon production by an 18 MV medical Varian linac and predict the p-i-n diode response, to justify its application for fast neutron dosimetry on a medical linac.
Table 4.8: Ratio of neutron to photon dose equivalent, with depth in a solid water phantom

<table>
<thead>
<tr>
<th>depth (cm)</th>
<th>neutron dose equivalent (Sv)</th>
<th>photon dose equivalent (Sv)</th>
<th>ratio of neutron/photon</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>17.42</td>
<td>10.71</td>
<td>1.63</td>
</tr>
<tr>
<td>1</td>
<td>20.02</td>
<td>24.30</td>
<td>0.82</td>
</tr>
<tr>
<td>2</td>
<td>18.07</td>
<td>28.77</td>
<td>0.63</td>
</tr>
<tr>
<td>3</td>
<td>15.70</td>
<td>30.00</td>
<td>0.52</td>
</tr>
<tr>
<td>4</td>
<td>12.64</td>
<td>29.67</td>
<td>0.43</td>
</tr>
<tr>
<td>5</td>
<td>12.11</td>
<td>28.74</td>
<td>0.42</td>
</tr>
<tr>
<td>6</td>
<td>10.59</td>
<td>27.81</td>
<td>0.38</td>
</tr>
<tr>
<td>7</td>
<td>9.40</td>
<td>26.73</td>
<td>0.35</td>
</tr>
<tr>
<td>8</td>
<td>8.33</td>
<td>25.65</td>
<td>0.32</td>
</tr>
<tr>
<td>9</td>
<td>7.00</td>
<td>24.66</td>
<td>0.28</td>
</tr>
<tr>
<td>10</td>
<td>6.41</td>
<td>23.70</td>
<td>0.27</td>
</tr>
</tbody>
</table>

Figure 4.30: Neutron silicon displacement damage KERMA and neutron dose equivalent, with depth in a solid water phantom.

Fig. 4.30 is depicting the calculated relative neutron tissue dose equivalent and silicon neutron displacement damage KERMA, with changing depth inside the solid water phantom. It is observed that the two curves correspond well to each other relatively. This result allows for the quick estimation of the neutron dose equivalent at any depth, based on calibration of the p-i-n diode on the surface of the phantom in terms of neutron dose equivalent for a particular 18 MV linac.
4.8 Conclusion

Silicon p-i-n diodes have a high neutron to photon tissue equivalent dose sensitivity discrimination and utilize a simple readout method. The developed ion implanted bulk silicon 0.5E p-i-n diode were shown to be convenient for real time estimation of fast neutron absorbed dose on the surface of a solid water phantom in a pulsed mixed photoneutron field, produced by an 18 MV medical linac.

It was found that the entrance surface neutron dose can be easily determined by passive silicon p-i-n diodes, based on neutron calibration with a 14 MeV neutron source and neglecting the photon contribution. The relative neutron tissue dose equivalent and silicon neutron displacement damage KERMA compared well for different depths inside the solid water phantom and present a quick method of estimation of the neutron dose equivalent at any depth, based on knowledge of the neutron dose equivalent response of the diode on the surface of the phantom for a particular linac.

Future work will involve irradiation of these detectors using clinical beams on an anthropomorphic phantom for the estimation of the patient neutron entrance absorbed dose. Currently the 0.5E p-i-n diodes studied had sensitivity to fast neutrons of 1.5 mV/cGy and provided measurements of fast neutron absorbed dose with an overall uncertainty of 15-20% when 30 Gy photon dose was delivered at Dmax using an 18 MV linac. For in patient application when the dose delivered is 2 Gy/fraction, a more sensitive p-i-n diode is required. A p-i-n diode with a sensitivity of 50 mV/cGy for fast neutron dosimetry was reported [59] and will be very suitable for fast neutron dosimetry for simple and immediate real time readout during or after irradiation. The application of p-i-n diodes on the surface of a patient body or in cavities with constant temperature make p-i-n diodes very suitable for passive real time in vivo neutron dosimetry on a 18 MV medical linac.
Chapter 5

3D silicon detectors for neutron detection

This chapter investigated the use of 3D silicon detectors for the detection of photoneutrons from an 18 MV medical linac. Detectors were characterised by performing I-V & C-V measurements, as well as spectral characterisation using an alpha and neutron source. Three types of 3D detectors were investigated in this work: 1) 3D ultra-thin (10 µm thickness with removed silicon substrate) 2) 3D 20 µm thick (with silicon substrate) 3) 3D silicon Micro-structured detector. 3D silicon detectors have small active region of either 10 µm or 20 µm thickness which is constructed with small 3D columnar electrodes. These characteristics allow the detector to be fully depleted at very low bias voltages. The 3D ultra-thin detector with removed supportive silicon substrate and 10 µm thickness of active area, allows for the rejection of most high energy photons entering the detector. The 3D detectors investigated in this study were coupled with a range of different polyethylene convertors to assess the most suitable convertor thickness for neutron detection. The use of 3D detectors could present a useful method of neutron detection in a mixed photoneutron field produced by an 18 MV medical linac.

5.1 Introduction

Three Dimensional (3-D) detector technology was first introduced by Parker in 1995 for applications in high-energy physics and has the advantage of radiation hardness, rapid charge collection and low depletion voltage. [75] Recent studies by Guardiola [76] have investigated an active method of neutron detection in radiotherapy treatment rooms based on ultra-thin 3D silicon detectors coated with $^{10}$B thin convertor layer. These 3D silicon detectors have two main advantages; firstly they are highly insensitive to gamma radiation background [77] and they have $^{10}$B stable coatings up to thickness of 3µm. Guardiola [76] performed measurements using a Siemens Primus linac which was collimated to a field
of 10cm x 10cm cm² at the isocentre, 0 degrees gantry angle and a 15 MV photon beam. The detectors were positioned in front of the linac in the gantry axis at a distance of 2 m from the isocentre along the patient couch. Two detectors were tested in parallel, including a detector with neutron converter and a detector with no convertor. Following irradiation the response of the no convertor detector was subtracted from the detector with the neutron convertor film to deduce charge particle interactions produced from neutrons. The work was successful in showing that the 3D silicon detectors with a portable readout system, are useful to detect the thermal neutron field inside a radiotherapy treatment room.

At the Centre for Medical and Radiation Physics (CMRP) in Wollongong we have investigated the possible use of 3D silicon detectors that were designed and fabricated at the department of Silicon Technologies and Microsystems of the Institute de Microelectrónica de Barcelona (IMB-CNM, CSIC) in Bellaterra, Spain, for possible neutron dosimetry on an 18 MV medical linac using spectroscopy mode of their operation. The detectors studied were fabricated from n-Si using 3D detector technology and had an active area thickness of 10 µm or 20 µm, so named ultra thin detectors. At CMRP the combination of polyethylene thin film convertor layer with each detector was studied for detection of photoneutrons both in and out of the radiation field, produced by an 18 MV photon beam.

5.2 Detector Structures

5.2.1 3D ultra-thin detector and 3D 20 µm thick detector

Two types of 3D detectors were fabricated on an n-type silicon on insulator (SOI) wafer at the IMB-CNM (CSIC), Barcelona, Spain. The first detector, which is named 3D ultra-thin, has an active thickness of 10 µm, area of 7.52mm x 7.52mm and was originally fabricated for use in corpuscular diagnostics of high-temperature plasma to very low X-ray spectroscopy. The detector provided nearly 100% detection efficiency for ions as well as low sensitivity to neutron and gamma background due to their extremely thin sensitive substrate [77]. The detectors have been also tested for thermal neutron detection in a strong gamma field, allowing suppression of gamma response [78]. A schematic of the 3D ultra-thin detector is shown in 5.1.
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Figure 5.1: Cross-sectional view of 3D thin detector with rows of cylindrical n+ and p+ columns and etched substrate. [79]

This detector was fabricated on 10 micron thick n-Si active layer, with silicon on insulator (SOI) with resistivity 3.5 kΩ cm, followed by a thinning process which removed the 300µm supporting wafer. A matrix of 94 x 94 p+ holes and 95 x 95 n+ holes of 5µm diameter each were etched through the 10 µm silicon active layer 5.2. The n+ holes and p+ holes were filled with polysilicon (doped with phosphorous and boron respectively) to act as the 3D n+ and p+ electrodes. All p+ electrodes were connected to the contact on one side of the sensor while all n+ electrodes were connected to the contact on the opposite side. A pitch between the same types of electrodes is 80µm. Aluminium contacts of 0.5µm thick were deposited to connect electrically all n+ and p+ electrodes respectively. 5.2 shows Al tracks connecting n+ and p+ electrodes and n+ common electrode only. The ultra-thin 3D 10 µm thick detector with active area of 7.52mm x 7.52mm and mounted on a Printed Circuit Board (PCB), is shown in 5.3(A).

Figure 5.2: Schematic of 3D Thin Detector produced by CNM Spain. [79]

The second detector investigated was the 3D 20 µm thick detector with substrate of 300µm (where the back Di substrate was not etched out). This detector is shown in 5.3(B) and has a detection area of 3.36mm x 3.36mm, consisting of a matrix of 42 x 42 p+ holes and 43 x 43 n+ holes.
5.2.2 3D silicon micro-structured detector

3D silicon micro-structured semiconductor-based neutron detectors studied in this work, consist of a silicon diode with sinusoidal trenches that are filled with $^6$LiF neutron converter material and covered with capton tape, as shown in 5.4(b). The neutron detection efficiency of this design is limited between 4% and 4.5% for $^{10}$B and $^6$LiF neutron converter materials respectively [83]. The planar design of silicon detectors can be improved by increasing the contact between the silicon sensitive area and the neutron convertor film. [82] This has been achieved by IMB-CNM (CSIC) by created perforated-patterns in the silicon via honey comb or sinusoidal trenches, which are then filled with convertor material to increase neutron interaction probability. The maximum thermal neutron detection efficiency reported in literature is 26% for Micro-structured silicon sensors with straight trenches filled with $^6$LiF [83].

3D thin micro-structured detectors with sinusoidal or hexagonal trenches were designed and fabricated at IMB-CNM [79] (see 5.4). In this work the 3D silicon micro-structured detectors with sinusoidal trenches, filled with $^6$LiF neutron converter powder material, as shown in figure 5.4(a,b) were studied. The detector was mounted on a PCB and covered with Kapton tape. The neutron absorption cross section for $^6$Li for thermal neutrons is 940 barns. The capture reaction on $^6$Li produces an alpha particle and a triton (equation 5.1):

$$n + ^6 Li \rightarrow \alpha(2.05)MeV + ^3 H(2.73)MeV \quad (5.1)$$
The detector was fabricated on n-type high-resistivity wafer with active thickness of 300µm. The detectors had an active area of 1cm², which contains sinusoidal channels (150µm deep and 15µm wide respectively) with p+ deposition on an internal wall that was etched into silicon and filled with the ⁶LiF converter material, as shown in 5.4(b). The trench-filling process was done by spreading the converter material on the sensor surface and applying high pressure [80]. The detector appeared to have non-uniform deposition of ⁶LiF on the surface, as shown in 5.4(c) and was covered with kapton tape of thickness 70µm in order to stop the contamination of ⁶LiF with the surrounding environment. Such design of the detector is offering increased area covered with ⁶LiF converter in comparison with conventional detector with area covered on top of the detector only.

5.2.3 Summary of Detectors Investigated in this study

The 3D detectors that are investigated in this study are summarised in table 5.1 below.

<table>
<thead>
<tr>
<th>Detector name</th>
<th>Sensitive Area (mm²)</th>
<th>Substrate</th>
<th>Column structure</th>
</tr>
</thead>
<tbody>
<tr>
<td>3D 10 µm (ultra-thin)</td>
<td>56.55</td>
<td>with etched out Si substrate</td>
<td>Pillar Channels</td>
</tr>
<tr>
<td>3D 20 µm</td>
<td>11.29</td>
<td>with Si substrate</td>
<td>Pillar Channels</td>
</tr>
<tr>
<td>3D microstructured</td>
<td>56.55</td>
<td>with Si substrate</td>
<td>Sinusoidal Channels</td>
</tr>
</tbody>
</table>

5.3 Detector Characterisation

5.3.1 I-V Measurement of the Detector Structures

The reverse current voltage (I-V) characteristics of each of the detectors were studied at ANSTO, using a Kiethly 237 high voltage source. This setup is different to the setup
used to measure the forward biased IV curves in chapter 4. The setup at ANSTO involved using a fixed voltage supply and taking current measurements. 3D detectors with 10 \( \mu m \) thickness with substrate and without substrate (ultra-thin), I-V characteristics are shown in figure 5.5. The 10 \( \mu m \) 3D ultra-thin detectors exhibited similar response across both samples (no.1, no.2) whilst 3D 20 \( \mu m \) thick detectors had higher breakdown voltage (on average -50V) in comparison to 3D 10 \( \mu m \) thick ultra-thin detectors, which had a breakdown voltage of approximately -20V (figure 5.6). The 3D micro-structured detectors could only be biased up to 0.1V and had very high leakage current 5.7.

![Figure 5.5: Reverse I-V characteristics of 3D detectors of 10 \( \mu m \) thickness (ultra-thin)](image)

![Figure 5.6: Reverse I-V characteristics of 3D detectors of 20 \( \mu m \) thickness with substrate](image)
5.3.2 C-V Measurement of the Detector Structures

The capacitance voltage (C-V) characteristics of each of the detectors was studied at ANSTO, using a Boonton 7200 Capacitance Bridge. C-V characteristics for 3D 10 µm thick (ultra-thin) is shown in figure 5.8. 3D detectors with 20 µm thickness, C-V characteristics are shown in figure 5.9. The shape of the C-V characteristics in both figure 5.8 and figure 5.9 is related to the geometrical distribution of p+ and n+ columns in each device. The lateral depletion in the columns of each detector is causing the step like shape in the C-V characteristic curves.

3D micro-structured detectors exhibited very high capacitance (figure 5.10) of approximately 400pF at 5V, in comparison to 3D ultra-thin detectors (figure 5.8), which exhibited capacitance of approximately 140pF at 5V. This high capacitance is related to the large area of the detector and possibly the non-uniform deposition of ⁶LiF inside the sinusoidal trenches and surface regions. The preamp electronic readout system will also introduce capacitance effects to the setup, which will contribute to the overall capacitance of the system. Ideally the detector capacitance should not be higher than the preamp capacitance, otherwise the spectroscopy system will be too noisy. We are able to bias the detector to reduce the capacitance effect, but the capacitance will remain high, as the detector already has a high leakage current. In order to investigate these detectors, we have performed measurements with the detector unbiased. As the detector has an inbuilt p-n junction, we should still be able to observe charge collection.
CHAPTER 5. 3D SILICON DETECTORS FOR NEUTRON DETECTION

Figure 5.8: C-V characteristics of 3D Detectors of 10 $\mu$m thickness (ultra-thin)

Figure 5.9: C-V characteristics of 3D Detectors of 20 $\mu$m thickness with substrate
5.3.3 Spectral Characterisation using an $^{241}$Am alpha source

Alpha spectroscopy is used to provide information about the energy losses in silicon and energy resolution of the silicon detectors. In this thesis spectral characterisation was performed using an $^{241}$Am alpha emitting source. An Amptek A250 charge sensitive preamplifier and the detector under investigation was placed in a vacuum chamber and air evacuated, in order to remove the interactions of alpha particles with air. The amplifier output was connected to an Canberra AFT 2025 shaping amplifier, which has a shaping time of $1\mu$s. An Amptek Multi Channel Analyser (MCA) 8000A was used to record the spectral response of the detector.

The energy calibration of the spectral response of the detector was performed using both a pulse generator in conjunction with a Hamamatsu 300$\mu$m thick window-less planar silicon PIN diode, which has 100% charge collection efficiency. Both 3D 10 $\mu$m ultra-thin detector and 3D 20 $\mu$m thick detector response to alpha particles was investigated, however the 3D micro-structured detectors were not studied due to the possibility of contamination due to loose $^6$Li particles when placed in a vacuum chamber.
3D 10 \mu m ultra-thin detector response to an $^{241}\text{Am}$ source

![Figure 5.11](image)

**Figure 5.11:** Response of 10 \mu m no.1 ultra-thin detector at different biases to an $^{241}\text{Am}$ source

The 10 \mu m ultra-thin detector response to a $^{241}\text{Am}$ Source when operated at different negative bias is shown in figure 5.11. As the negative bias was applied to the detector, the spectral response shifted to the right and the charge collection increased.

3D 20 \mu m thick detector response to an $^{241}\text{Am}$ source

![Figure 5.12](image)

**Figure 5.12:** Response of 20 \mu m no.1 detector at different biases to an $^{241}\text{Am}$ source
The 3D 20 $\mu$m thick detector response to a $^{241}$Am source when operated at different bias voltage is shown in figures 5.12 and 5.13, with the main peak to the right with energy of approximately 3.3 MeV. As the bias was applied to the detector, the spectral response shifted slightly to the left (for detectors no1 and no3 in figures 5.12 and 5.13) and the charge collection slightly decreased. There was a peak observed at about 2MeV in figures 5.12 and 5.13), that occurred when a bias voltage of -30V was applied to each of the detectors. This peak could be due to low energy events occurring at the edges of the detector, indicating charge collection occurring at the edges of the detector [82].

5.3.4 Spectral Characterisation using an $^{252}$Cf source

$^{252}$Cf decays by alpha particle emission (97%) and spontaneous fission (3%), yielding an average of 3.75 fast neutrons per fission event. The half-life for alpha particle decay is 2.645 years [84]. The source used in the experiments for this thesis has an activity as at 12pm 15th of May 2007 : 1.70E5 Bq and a measured gamma dose rate on 28-08-14 to be 2$\mu$Sv/hour. The $^{252}$Cf source has decayed to roughly 16.75% of the initial activity over the period of 6.817 years (from 15/05/2007 until 03/09/2014) to an activity of 28,481 Bq.

The energy spectrum of $^{252}$Cf source, being a fission spectrum, is well characterized and can be described by Maxwellian distribution (equation 5.2) [85] as:

$$N(E) = \sqrt{E} \exp\left[\frac{-E}{T}\right]$$  \hspace{1cm} (5.2)

Where N(E) is the number of neutrons emitted at energy E, per unit energy interval, and T is a calibration constant estimated to be between 1.3MeV and 1.424 MeV, giving
an average energy between 2.316 and 2.348 MeV. [85] The neutron energy spectrum is shown in figure 5.14 and was created from the data table 4.V. ISO reference spectra [87].

![Energy of Cf-252 source neutrons](image)

**Figure 5.14**: Neutron energy spectrum of an $^{252}$Cf source, graphed from data table 4.V. ISO reference spectra [87]

The response of each detector coupled with a polyethylene convertor film, to a $^{252}$Cf source (Fig. 5.15) in free air was studied at ANSTO, using a Amptek A250 charge sensitive preamplifier, with both detector and preamplifier placed inside a vacuum chamber but not under vacuum. The amplifier output connection was connected to a Canberra AFT 2025 shaping amplifier, which has a shaping time of 1$ \mu $s. An Amptek Multi Channel Analyser (MCA) 8000A was used to record the spectral response of the detector.

The $^{252}$Cf source was positioned on a Perspex source holder (Fig. 5.16[A]), which was positioned on top of a low density polyethylene (LDPE) convertor film (5.16 [B]). The
detector was positioned below the polyethylene convertor film, with minimal air gap. The same experimental setup was used to test each of the detectors response to the $^{252}\text{Cf}$ source.

![Figure 5.16](image)

**Figure 5.16:** [A] The experimental setup with the $^{252}\text{Cf}$ source holder and polyethylene convertor placed on the detector. [B] The $^{252}\text{Cf}$ source positioned over the polyethylene and detector, ready for measurement.

A LDPE layer of density = 0.92 g cm$^{-3}$ was used to produce recoil protons from the incident neutrons from the $^{252}\text{Cf}$ source via $^1\text{H}(\text{n}, \text{p})n'$ elastic scattering reaction. The energy of a particle due to elastic recoil from a neutron interaction is given by the equation 5.3.

$$E_R = \frac{4A}{(1+A)^2}(\cos^2\theta)E_n$$

(5.3)

Where $E_R$ is the energy of the recoil particle, $E_n$ is the energy of the incident neutron, $A$ is the mass number of the target particle relative to the incident neutron and $\theta$ is the angle of the recoil particle relative to the original direction of the incident neutron [21]. For proton elastic recoils from the LDPE layer, $A = 1$, thus the energy of the recoil proton will be dependant on the angle of recoil, with the maximum energy of the recoil proton occurring at $\theta = 0^\circ$.

Elastic neutron recoil events occurring from heavier nuclei, such as silicon in the detector sensitive volume or $^{12}\text{C}$ nucleus in LDPE, will transfer a fraction of the energy of the incident neutron depending on the ratio of $\frac{4A}{(1+A)^2}$. According to this ratio the maximum energy transfer to a $^{12}\text{C}$ nucleus is 28.4% of the incident $E_n$. The maximum energy transfer to $^{28}\text{Si}$ is 13.3% of $E_n$. Another consideration is the energy losses that occur within the LDPE layer. A thin layer of LDPE convertor film would result in a recoil proton with high possibility of reaching the detector sensitive volume and depositing all energy. The elastic recoil cross section in LDPE is small and usually a thicker layer of LDPE is required to provide sufficient neutron detection efficiency. A thick layer of LDPE convertor film will result in protons produced close to the entrance of the LDPE, loosing consid-
erable amount of energy when traversing the LDPE. Charged particle equilibrium (CPE) will occur when the maximum range of highest energy recoil proton in the LDPE layer is equal to the thickness of the LDPE layer. In the case of $^{252}$Cf which has average energy of neutrons 1-2MeV, for CPE to occur a LDPE convertor film of thickness approximately 30-80$\mu$m thickness would be required. A 100 $\mu$m thick LDPE convertor film was deemed sufficient to be used for the detectors under investigation in this chapter.

3D 10 $\mu$m ultra-thin detector response to an $^{252}$Cf source

![Graph showing detector response with different thickness of polyethylene convertor films.](image)

**Figure 5.17:** 3D 10 $\mu$m ultra-thin detector response to $^{252}$Cf; comparison of detector response with different thickness of polyethylene convertor films.
The 3D 10 \( \mu \text{m} \) ultra-thin detector response to \( ^{252}\text{Cf} \) was found to be best used with a 100 \( \mu \text{m} \) polyethylene convertor (Fig 5.17) and was chosen for investigation. The response of the detector to \( ^{252}\text{Cf} \) with no convertor, was subtracted from the response of the de-
detector with 100 µm polyethylene convertor (Fig 5.18) the detector response is given in (Fig 5.19). It can be seen that all low energy events from photons with energy $< 0.1$ MeV have been removed and the resultant graph (Fig 5.19) is only showing recoil events from neutrons in the silicon. The reason we see no low energy events in the detector response is due to the small thickness of silicon being 10 µm, which is related to the low probability of gamma photon interactions in the detector. This ability of the detector to reject photon interactions in 10 µm thick silicon will be useful in the investigated of the neutron response of the detector, when placed in a mixed pulsed photoneutron field produced by an 18 MV linac.

3D 20 µm detector response to an $^{252}$Cf source

![Graph of detector response to $^{252}$Cf source with different thicknesses of polyethylene convertor films.](image)

**Figure 5.20:** 3D 20 µm thick detector response to $^{252}$Cf; comparison of the detector response with different thickness of polyethylene convertor films.
Figure 5.21: 3D 20 $\mu$m thick detector response to $^{252}$Cf; comparison between detector with 100 $\mu$m polyethylene and detector with no convertor film.

Figure 5.22: 3D 20 $\mu$m thick detector response to $^{252}$Cf; subtraction of detector response with no convertor from the response of the detector with 100 $\mu$m polyethylene convertor.

The 3D 20 $\mu$m thick detector response to $^{252}$Cf was found to be best used with a 100 $\mu$m polyethylene convertor film (Figure 5.20), exhibiting highest depositing energy of recoil protons in the detector. The detector had similar response for 250$\mu$m and 500$\mu$m thick
polyethylene convertor films as charge particle equilibrium had already been reached. After subtraction of the detector response to $^{252}$Cf with no convertor from the response of the detector with 100 $\mu$m polyethylene (Figure 5.21) the detector response is given in (Figure 5.22). In Figure 5.22 it can be seen that for an energy above 0.2MeV the detector response will be predominantly due to neutron events in the silicon, however now there are much higher low energy events below 0.2MeV in comparison with the 10 $\mu$m response to $^{252}$Cf after subtraction (Figure 5.19). This is due to the 20 $\mu$m thick detector having larger thickness of active area and also having silicon substrate of 300$\mu$m, which is creating many more scatter of low energy photons in the detector. This detector could possibly be useful in a mixed pulsed photoneutron field produced by an 18 MV linac, provided a large enough threshold is chosen.

**3D Micro-structured detector response to an $^{252}$Cf source**

![Graph showing detector response to $^{252}$Cf](image)

*Figure 5.23: Micro-structured neutron detector response to $^{252}$Cf; comparison of detector response to different thickness of polyethylene convertor films.*

The 3D Micro-structured neutron detector response to $^{252}$Cf was greater than both 10 $\mu$m thin detector and 20 $\mu$m with substrate detectors. A 100 $\mu$m polyethylene convertor film (Figure 5.23), exhibiting greatest energy range of detection and was chosen for investigation. There was very little difference seen in the detector response when no convertor or coupled with a 100 $\mu$m or 500$\mu$m polyethylene convertor film, as the CPE had already been reached with the detector coupled with a 100 $\mu$m polyethylene convertor film.
5.4 18 MV linac Measurements

5.4.1 3D Detectors Portable Readout System

Measurements of each of the detectors spectral response when placed in and out of fields produced from an 18 MV linac were investigated using a portable system, see Figure 5.24) The portable system included a CREMAT pre-amplifier, CREMAT shaping amplifier, portable bias supply, Multi-Channel Analyser, computer with ADMCA software and portable pulse generator. The detector was connected and positioned on a board that connects directly to the preamplifier (Figure 5.24). This board was then placed inside a faraday cage (aluminium case), which is used to shield the detector and sensitive CREMAT electronics from the external environment.

![Figure 5.24: Portable readout electronics system](image)

5.4.2 Out of field measurements

3D detectors: 10 µm ultra-thin and 20 µm thick detectors were positioned in the face up position (detectors face is oriented towards the ceiling) on the surface of the phantom and in field (Fig. 5.25).
When performing measurements with 10 µm ultra-thin detector on the surface of the phantom, at 10cm from the edge of a 2cm x 2cm field size and dose rate of 600MU/min there was a pulse pileup effect (summation of multiple input pulses) observed, with the peak of an average energy of 2.6 MeV. The response of the detector with no polyethylene convertor film is indicated in blue (Fig. 5.26) and the response of the detector with polyethylene convertor film is depicted in green.

There is a slight offset in the observed pulse pileup peaks. The green peak is shifted to
the right of the blue peak. This offset of the green peak to the right, might be due to the increased nuclear recoil events being created in the polyethylene and then reaching the detector. Applying a lower dose rate to the detectors caused less neutron events to be detected. Performing measurements with a 20 $\mu$m thick detector coupled with thin film converter, with detector placed on the surface of the phantom, (Fig. 5.25), yielded very poor statistics, with only 10 hits registered. This is expected, due to the small sensitive area of the detector. This makes the 20 $\mu$m detector unsuitable for neutron detection, when using a small field size produced by an 18 MV linac.

The 10 $\mu$m ultra-thin detector was placed at 2m from the isocentre on the patient couch (Fig. 5.27) and irradiated firstly with no convertor film and then later with a 100um polyethylene convertor film on top. Detectors were irradiated with 3000MU for field sizes of 5cmx5cm and 10cmx10cm.

![Figure 5.27: 3D ultra-thin detectors experimental setup for out of field measurements, set at a distance of 2m along the patient couch.](image)

Measurements with the 3D ultra-thin detector of 10 $\mu$m thickness in the face up position (detector face is oriented towards the ceiling), exhibited greater response to the 10 cm x 10 cm field size (Fig. 5.28(b)) in comparison to the 5 cm x 5 cm field size (Fig. 5.28(a)). Both measurements were performed with 3000MU delivered. The detector response to neutrons increases, with increasing beam size, due to more neutrons produced by scattering of the jaws and MLCs. This same trend was exhibited for 20 $\mu$m detector with
substrate (Fig. 5.29(a), Fig. 5.29(b)). The 20 µm detector had greater detection statistics, in comparison to the 10 µm ultra-thin detector. This is due to the fact that most spectra is driven by Compton electron events in silicon, with negligible contribution from recoil protons in high photon field (Fig. 5.29(a), Fig. 5.29(b)). The 10 µm detector performs better overall, with ability of being less sensitive to photons and the ability to subtract the photon component. It is important to mention that more statistics are required for each of these measurements. When subtracting the response of the detector with no convertor film from the response of the detector with convertor film, at low energies there is a greater statistical error.

Figure 5.28: (a) 3D 10 µm ultra-thin detector at 2m from the isocentre on the patient couch in face up position for 5cm x 5cm field size. (b) 3D 10 µm ultra-thin detector at 2m from the isocentre on the patient couch in face up position for 10cm x 10cm field size.

Figure 5.29: (a) 3D 20 µm detector at 2m from the isocentre on the patient couch in face up position for 5cm x 5cm field size. (b) 3D 20 µm detector at 2m from the isocentre on the patient couch in face up position for 10cm x 10cm field size.

Angular dependence was observed when the detector was rotated 90 degrees to be fac-
ing the direction of the linac head (Figures 5.30 and 5.31). Both 10 μm ultra-thin and 20 μm thick detectors also exhibited field size dependence, with more low energy events being registered for a larger field size. The 10 μm ultra-thin detector with no polyethylene convertor film exhibited much less counts in comparison to the 20 μm detector with no polyethylene convertor film, in figures 5.28 and 5.29. This is expected due to the 10 μm ultra-thin detector having no silicon substrate, allowing most gammas to pass through.

The ability of the 10 μm ultra-thin detector to reject gammas is demonstrated in figures 5.30, 5.31, where the response of the detector with no convertor film has been subtracted from the response of the detector with 100 μm polyethylene. From Fig.5.30(a) no energy threshold was set, as the detector was able to reject low energy events from gammas and thus the spectra depicted are predominately due to neutrons. From Fig. 5.30 (b), the 10 μm ultra-thin detector required a energy threshold of 0.05 MeV for the face up position and a energy threshold of 0.1MeV for the rotated position. These thresholds required are due to the high dose rate of 500MU/minute used when performing these measurements. Possibly smaller energy thresholds could be employed when being irradiated with a lower dose rate.

Figure 5.30: (a) 3D 10 μm ultra-thin detector at 2m from the isocentre, placed on the patient couch, angular response to neutrons for 5cm x 5cm field size. (b) 3D 10 μm ultra-thin detector at 2m from the isocentre, placed on the patient couch, angular response to neutrons for 10cm x 10cm field size.
A comparison between 3D 10 µm ultra-thin and 20 µm detectors with polyethylene converter film for a 5cm x 5cm field size is shown in figure 5.32. It was demonstrated that the 10 µm ultra-thin detector provides a greater discrimination of neutrons from photon events in comparison to the 20 µm detector, thus allowing more low energy events produced from neutrons to be detected.

**Figure 5.32:** Comparison of 10 µm ultra-thin and 20 µm detectors covered with 100µm polyethylene at 2m from the isocentre, placed on the patient couch, for 5cmx5cm field size, facing the linac.
5.5 Conclusion

The use of 3D radiation detectors was investigated for the possibility of neutron detection, in an 18 MV medical linac field. The charge collection characteristics of three detectors: 10 \( \mu \text{m} \) ultra-thin detector, 20 \( \mu \text{m} \) thick detector with 300\( \mu \text{m} \) substrate and Micro-structured with \(^{6}\text{LiF}\) converter material were investigated. Alpha spectroscopy was performed on these detectors and it was found that the 10 \( \mu \text{m} \) ultra-thin detector and 20 \( \mu \text{m} \) with substrate detector had the best charge collection at bias voltage of -10V.

Neutron spectroscopy was performed using a \(^{252}\text{Cf}\) source in combination with a number of different thickness of polyethylene neutron convertor films. A polyethylene convertor film of thickness 100 \( \mu \text{m} \) was found to be optimal for each of the detectors under investigation. The 10 \( \mu \text{m} \) ultra-thin detector was able to reject all low energy events from photons with energy < 0.1MeV and thus be able to predominantly measure recoil events from neutrons in silicon. This ability of the detector to reject photon interactions in 10 \( \mu \text{m} \) thick silicon is useful for the investigation of the detector response to neutrons, when placed in a mixed pulsed photoneutron field produced by an 18 MV linac.

The 10 \( \mu \text{m} \) ultra-thin detector and 20 \( \mu \text{m} \) with substrate detector were chosen to perform measurements on an 18 MV linac. It was demonstrated that at 10cm from the edge of a 2cm x 2cm field size on the surface of a solid water phantom, the 20 \( \mu \text{m} \) with substrate detector had low statistics, limiting the usefulness of the device for in-field neutron measurements on an 18 MV linac. When performing measurements with 10 \( \mu \text{m} \) ultra-thin detector on the surface of the phantom, at 10cm from the edge of a 2cm x 2cm field size, pulse pileup effects were observed, with an average energy of 2.6MeV. To reduce the pulse pileup effects, detectors were positioned at 2 metres from the centre of the beam position.

At 2 metres from the centre of the beam position the 10 \( \mu \text{m} \) ultra-thin detector, exhibited spectral response from photons and neutrons as expected, with ability to subtract response of the detectors due to photons. Angular dependence was observed for this detector, with greater number of events detected with the detector facing the linac beam, in comparison to the detector in the face up position (facing the ceiling). The effect of increasing the field size, increased the number of neutron events detected. It was concluded that the 10 \( \mu \text{m} \) ultra-thin detector is suitable for out of field neutron detection measurements on an 18 MV medical linac, at a distance of 2m from the isocentre. Future investigations could involve reduction of capacitance in the design of micro-structured detectors and new pillar design, etching of substrate for more efficient charge collection as well as investigation into the optimal LDPE convertor film thickness.
Chapter 6

Directional neutron detector

The Bonner Sphere Spectrometer (BSS) system is a well-established technique for neutron dosimetry that involves detection of thermal neutrons within a range of hydrogenous moderators. BSS detectors are often used to perform neutron field surveys in order to determine the ambient dose equivalent H*(10) and estimate health risk to personnel. There is a potential limitation of existing neutron survey techniques, since some detectors do not consider the direction of the neutron field, which can result in overly conservative estimates of dose in neutron fields.

This chapter shows the development of a Geant4 simulation application to characterise a prototype neutron detector based on three orthogonal $^3$He tubes inside a single HDPE sphere built at ANSTO. The Geant4 simulation has been validated with respect to experimental measurements performed with an Am-Be source. Simulation of the detector response to neutrons in mixed pulsed radiation field produced in an 18 MV medical linac was studied. Most of this chapter is based on a paper published by the primary author Vanja Gracanin, also author of this thesis [89].

6.1 Introduction

The Bonner sphere spectrometer (BSS) was developed more than three decades ago for neutron spectrometry and its response function has been extensively studied for measuring neutron spectra [91], [92]. Combining this response with unfolding techniques, reasonably reliable neutron spectroscopy can be achieved. However, this technique is time consuming and, as a result, it cannot be used to perform real-time neutron surveys. Existing, commonly used neutron survey instruments measuring H*(10) include the Leake Detector and the Studsvik instrument [93], which are typically single detector instruments. Recently a study by Barlett et al [90] highlighted a potential limitation of existing neutron survey techniques being due to a lack of consideration to the direction of the field, which has consequently led to overly conservative estimates of dose. Prior to the ICRP
the dose was conservative (an overestimate of the protection), however the changes in the ICRP 60 result in an underestimation of the dose for most energies and especially in directional fields. Some alternative designs have been proposed to improve neutron dose estimations and consist of multiple detectors inside a single sphere [94], which have reduced readout times and could be used as real time neutron survey meters. Most recently a study has used Geant4 to construct an analytical model to reproduce out of field neutron doses [99].

This chapter shows the development of a Geant4 simulation application to characterise a prototype neutron detector based on three $^3\text{He}$ tubes inside a HDPE sphere built at ANSTO. In this study the simulation was developed and validated against experimental measurements. The validation of the Geant4 application consisted of calculating the detector efficiency of a single $^3\text{He}$ tube exposed to monochromatic neutrons beams and comparison to experimental measurements performed at ANSTO with an Am-Be source. The feasibility of calculating neutron doses will be investigated.

### 6.2 Design of the Directional Spectrometer

The geometry of the Neutron Directional Spectrometer is illustrated in Fig.6.1. It consist of a sphere, 248 mm in diameter, made of HDPE to moderate the incident neutrons, with holes for three tubes with 14 mm diameter and sensitive tube regions of 12.7 mm diameter. The HDPE sphere is mounted on top of three stainless steel tubes that protrude into the sphere less than 1 cm and are mounted onto a stainless steel stand. The tubes were supplied by GE Reuter Stokes [100] and the HDPE sphere was made in-house at ANSTO. Both items were selected without regard to optimising the system performance. The tubes are 362 mm long and filled with $^3\text{He}$ gas at 10 atm pressure and operated at 1500 V. The sensitive region of the tube is 282 mm long, with the centre of each tube offset from the centre of the sphere by 8 mm in x, y and z direction. The stainless steel casing has a thickness of 0.25 mm.

Neutrons are detected by use of three $^3\text{He}$ filled tubes, operated at 1500V, located within a HDPE sphere Fig.6.1. The cross section of interaction for neutrons hitting a $^3\text{He}$ atom inside the tube is displayed in Fig.6.2. A neutron can interact with $^3\text{He}$ atom via an exothermic reaction, whereby a thermal neutron is captured and a triton and proton are produced and emitted at exactly 180 degrees to one another (equation 6.1).

\[ ^3\text{He} + n \rightarrow ^3\text{H} + ^1\text{H} + (764\text{keV}) \]  

\hspace{1cm} (6.1)
Such reaction has the highest cross section compared to the elastic scattering and $^3$He(n, c)$^4$He, when the neutron energy is below approximately 10 keV. The $^3$He(n,$^2$H)$^2$H reaction can only take place when the energy of the neutron is above few MeV. Based on these observations, it was assumed that the detector signal derives predominantly from the $^3$He(n,p)$^3$H reaction. Provided that the electric field is high enough, the proton and triton can ionise the surrounding gas atoms to create charges, which in turn ionise other gas atoms in an avalanche-like multiplication process [21]. The resulting charges are collected as measurable electrical pulses with the amplitudes being independent of the
neutron energy. The electrical pulses are combined to form a positional response of neutron interactions occurring along the length of the tube.

The efficiency (Eff) for a single bare $^3$He tube, with incident neutrons travelling perpendicular to the tube, is related to the incident neutron energy, gas pressure and tube dimensions, which can be described by equation 6.2.

$$E_{ff} = [1 - \exp\left(\frac{-x}{\mu}\right)] * 100$$ (6.2)

Where $x$ is the detector thickness and $\mu$ is the neutron mean free path. The mean free path is determined by the density and pressure of the gas and neutron energy [96].

### 6.3 Methodology

A Geant4 [95] simulation application was developed to investigate the directional behaviour of this neutron detector. Two validation studies were performed. First the detector efficiency of a single $^3$He tube exposed to monochromatic neutron beams was calculated by means of the Geant4 application and compared to the theoretically calculated efficiency (equation 6.2). Then, the detector response was calculated by means of Geant4 and compared to experimental measurements obtained with an Am-Be source located in different positions around the detector. Once the validation of the Geant4 simulation was performed, an investigation of the optimal tube radius to be placed inside the sphere was performed, to allow an optimal system to be built.

#### 6.3.1 Experimental Measurements with an Am-Be Source

The experimental set up consisted of a HDPE sphere and tubes as described in Section 2. The tubes were connected to a Mesytec position sensitive tube readout system [101]. This consisted of an MPSD8 preamp/position calculator and MCPD8 data control module. The electronics were housed in a NIM crate. An ISEG NHQ204M provided the high voltage bias of 1500V to the tubes. The system was controlled by software developed in-house, used to provide histogram outputs of the detectors response in real-time and for fixed acquisition duration. The directional response of each tube was investigated for a number of Am-Be source positions around the detector.

The experimental setup consisted of a 37 MBq Am-Be source, clamped and positioned on a retort stand. The source was double encapsulated in stainless steel and angled towards the detector directly on the surface of the sphere and at angles of (0, 30, 45, 90 and 300) degrees along a single axis of rotation (Z-axis). Fig.6.3 shows the neutron de-
detector with three $^3$He tubes connected to the readout electronics, inside the black HDPE sphere. The red marks indicate the positions on the plane passing through the centre of the sphere, where the Am-Be source was positioned. Each measurement was acquired for three hours with an average rate of around 20 counts per seconds and a total acquisition of around 250,000 events per measurement. The low energy threshold was chosen equal to 40 keV, in order to remove the background signal (experiment being performed in a nuclear reactor facility) before placing the source on the surface of the HDPE sphere.

![Figure 6.3: Left: the prototype neutron detector, with the HDPE Sphere and $^3$He tubes connected to the detector electronics used for readout. Right: Am-Be source angle positions investigated in the X Z plane, with source rotated about the Y-axis.]

### 6.3.2 Geant4 Monte Carlo Simulation

In this work Geant4 version 10.0.p02 was used to model the directional spectrometer. Geant4 enables the simulation of complex geometries using combinatorial geometry. It has the capability to incorporate tessellated volumes generated by computer aided design (CAD) programs [95].

Solidworks [51] was used to model the HDPE sphere and then exported as STEP files (an ISO compliant file format) before being converted to geometry description mark-up language (GDML) using the MeshLab software [98]. GDML is an XML extension available in Geant4, which gives the definition of geometry without the need for hard coding. The $^3$He tubes were modelled in Geant4 and then positioned inside the sphere (Fig.6.4). The geometry functionality of Geant4 is extensive and alternative options are offered to model the same geometry. The users decide which method to use to implement a specific geometry. For example, in this case Constructive Solid Geometry (CSG) solids com-
Figure 6.4: Left: HDPE Sphere created in Solidworks, cast into a tessellated mesh using MeshLab software [98]. Right: Geometry of the detector, as seen using Qt visualisation interface in Geant4.

Combined with Boolean operations could have been used to describe the same moderator. The CAD/STEP [102] option was instead chosen because it allows to model straightforward the current geometry of the device and to import easily more complex geometries, if required in the next stages of the project.

The neutron Geant4 High Precision Physics data libraries have been adopted to model the neutron interactions. The incident radiation field was either a broad beam of monoenergetic neutrons or an Am-Be ideal isotropic point source. The Am-Be source was modelled from reference spectra obtained from the ISO Report 8529-1 (March 2000), Table 3) and verified when used in Geant4 (Fig.6.5). Photon interactions were modelled in the simulation using the G4EmLivermore Physics List.

In order to understand the gamma contribution to the detector response we have simulated one million neutrons from an Am-Be source at 0 degrees to the surface of the sphere. The energy deposition of secondary electrons produced from gamma interactions inside the tube was scored. The study showed that on average an electron deposits energy of approximately 0.6 MeV. However, the percentage contribution of photons to the total tube positional response was calculated to be 0.1%. Based on these results, it was decided to model photon interactions in the simulation setup but not to score the hits from electrons that are created from photons interacting inside the tubes.

**Simulation set-up to calculate the detector efficiency**

The detector efficiency was studied using the simulation set-up shown in Fig.6.6. Monoenergetic neutrons with energy range between 0.001 eV and 10 MeV were incident perpendicular to one single tube. A rectangle plane of size 10 cm x 1.5 cm was modelled to cover a single tube uniformly with the incident monoenergetic neutron field. The number
CHAPTER 6. DIRECTIONAL NEUTRON DETECTOR

Figure 6.5: Am-Be Spectrum simulated using Geant4 from data given in the ISO Report 8529-1 (March 2000), Table 3).

of recoil protons (created from neutron capture events) generated inside the sensitive region of the tube was used to calculate the detector efficiency. The gas pressure and density were kept constant. A total of $10^7$ incident neutrons on the tube were simulated.

Figure 6.6: Experimental Setup adopted to study the efficiency of a single $^3$He tube.
Validation with respect to experimental measurements

The Geant4 application was validated against the experimental measurements performed at ANSTO and described in Section 6.3.1. The Am-Be source was modelled with neutrons generated from a point source within stainless steel packaging, with isotropic direction and energy spectrum as shown in Fig. 6.5, derived from the ISO Report 8529-1 (March 2000), Table 3). The Am-Be source was located on the HPDE sphere in the same positions as in the experiment, including the angles 0, 30, 45, 90, 300 degrees in the plane passing by the centre of the sphere (see Fig. 6.3).

An example of the simulation setup for a source position at 30 degrees is shown in Fig. 6.7. The response of the tubes was measured by counting the number of protons generated inside the tube. The counts were divided into 200 bins along the length of the tube. The number of bins was chosen to be the same as in the experiment. Simulation and experimental results were normalised to the integral value of counts for each tube, in order to allow a comparison of the relative positional response.

Detector Optimisation

In order to determine the ideal diameter of the $^3$He tubes used in the directional neutron detector, simulations were run for a single tube, at a pressure of 10 atm and radii of (5, 10, 15, 20, 25) mm separately. For each simulation, a monoenergetic neutron source was positioned at one of five coplanar angles (0, 30, 45, 60, 90) degrees relative to the tubes axis. An example of the setup for source position at 30 degrees is shown in Fig. 6.8. The source was simulated using Geant4s general particle source for a total of 10 million incident neutrons. The energy of the neutron source was varied by a factor of ten between 10 meV up to 10 MeV.
Detector response in an 18 MV linac photoneutron field

The Geant4 simulation of an 18 MV linac, previously discussed in chapter 3, was used to simulate the required neutron energy fields to irradiate the directional neutron detector. A simulation of 10 million neutrons emitted from a 10 cm $\times$ 10 cm field sizes from the position of the jaws was used to investigate the detector's response to neutrons when placed at the isocentre (no phantom) and out of field at 1 m from the isocentre. A schematic of the out of field experimental setup is shown in 6.9. The number of recoil protons generated inside the sensitive region of the tube were scored, in order to see if there was any directional dependence of neutrons.
6.4 Results

6.4.1 Experimental Results

Figure 6.10 shows an example of the response of the three $^3$He tubes, when the Am-Be source is located on the surface of the HPDE sphere forming an angle of 30 degrees with the Z-axis. Here the tube along the Y-axis represents the vertical tube, which is exhibiting a symmetrical detector response with respect to the centre of the tube, as expected because of the symmetry of the detector. The tube along the Z-axis has a more intense signal at one of its ends because the Am-Be source is located at 30 degrees from it. Most of the signal that was measured at the edge of the tube (source located at 0 degrees with the Z-axis) is due to the ionisation of charged recoils scattered by neutrons and not from neutron capture and then recoil events. This is due to the fast neutron spectrum originated by the AmBe source having not yet been moderated in the HDPE sphere.
The detector signal then decreases along the length of the tube because of the scattering of the neutrons inside the HDPE sphere. The detector response of the X-tube has a similar curve with respect to the Z-tube, however the signal is overall lower, because this tube forms an angle of 60 degrees with the Am-Be source. The efficiency of each tube was observed to drop off completely in the last 4 cm at both ends of the tube (see Figs. 6.10, 6.11, 6.12, 6.13), which is represented by not registering any counts at these positions. Such positions are excluded in the comparisons presented in Section 6.4.3.

Figure 6.11 shows the detector response of the X-tube when the Am-Be source is located in different positions on the HDPE sphere as shown in Fig. 6.3. The positions are identified with the angle formed with the Z-axis. The response of the X-tube is symmetrical with respect to the centre of the tube for an angle of 0 degrees, (see Fig. 6.7 for a schematic of the detector). At 90 degrees the source is closest to the right end of the X-tube (Positive X-tube) and has the greatest response, relative to all other angles. The slight difference between the two detector responses at 300 degrees and 45 degrees is related to the different thickness of HDPE in relation to the detector and tube position (see Fig. 6.7, top view).

It is clear that the top left section of the sphere has the lowest amount of HDPE separating the tube and the source, thus allowing more neutrons to interact with the sensitive region of the tube and produce an increased signal for a source position at 300 degrees. As each tube is offset 8 mm from the centre of the HPDE sphere, the detector response along the
length of each tube is unique in terms of intensity and distribution of counts detected. This is clearly shown in Fig. 6.12, with the response of the Y-tube changing in intensity
as the source is rotated around the tube. The Y-tube is displaying the same shape in the response along the length of the tube, with a dip in the response at the centre of the tube, being due to shadowing from the X-tube in front.

![Graph showing the response of the Y-tube for different source angles](image)

**Figure 6.13:** Experimental measurement obtained for the X-tube, changing the position of the Am-Be source, identified with the angle formed with the Z-axis (see Fig. 6.3)

The Y-tube response for source angles of 45 and 90 degrees were very similar to the source angle of 0 degrees and were not plotted in this case. All three angles are similar, as they have similar thickness of HDPE from the source to the tube. The Z-tube and X-tube gave unique response in the signal for each source location (Figs. 6.12 and 6.13). The intensity of the response for each tube is related to the angle formed by the source with the Z-axis. For the Y-tube, the most intense signal occurs for an angle of 300 degrees, while the lowest response is obtained for an angle of 30 degrees. This is again related to the thickness of the HDPE medium between the tube and the source, which is lowest for source angle at 300 degrees (see Fig. 6.7 top view)

### 6.4.2 Calculation of the detector efficiency by means of Geant4

The neutron detector has relatively low gamma sensitivity and is able to detect both thermal neutrons and fast neutrons up to an energy of 1 MeV. Figure 6.14 shows the efficiency of the detector with respect to the energy of the incident neutron, simulated using Geant4 and calculated using equation (6.2). There was a percentage difference between simulated and calculated results ranging between (1-3)%. $^3$He tubes show greatest detection
efficiency to thermal and cold neutrons. The efficiency drops to less than 1% for fast neutrons with energy above 1 MeV. It is shown that Geant4 reproduces the efficiency curve in detail, in the energy range under study, between 0.025 eV and 1 MeV. The efficiency was calculated also for an Am-Be source with the Geant4 simulation, using the same methodology as in Section 6.3.2.1. An efficiency of 6.4% was determined.

![Image of efficiency curve](image)

**Figure 6.14:** Relative efficiency of a single $^3$He tube for neutron detection with respect to the energy of the incident monoenergetic neutrons.

### 6.4.3 Validation of the simulation with respect to experimental measurements

The directional response of each tube was modelled for a number of Am-Be source positions relative to the neutron detector and compared to experimental measurements. The difference between the experimental data (exp_value) and simulation results (sim_value) was quantified by means of the % difference calculated as equation 6.3:

$$\%\text{Difference} = \frac{\text{sim}_{\text{value}} - \text{exp}_{\text{value}}}{\text{exp}_{\text{value}}} \times 100$$ (6.3)

Experimentally the tubes are subdivided into 200 positional bins (sensitive tube length of 282 mm, in the region of -14.1 cm to 14.1 cm) along their length to identify the position of generation of the protons. Therefore the same binning was adopted for the simulation. Then it was decided to perform the comparison between simulation and experiment reducing the number of bins to 40 (in the region of -10.06 cm to 10.06 cm), corresponding to a spatial resolution of 5 mm. The last 4 cm at both ends of the tubes were not included.
in the comparison because of the low statistics affecting the experimental measurements and of the drop of the efficiency at the tube extremities. The choice of using fewer positional bins was due to the observation of statistical fluctuations (Poisson distribution) affecting the experimental measurements.

Figures 6.15, 6.16, 6.17, 6.18, 6.19 show the comparison of the experimental and simulation results. The results refer to the experimental configurations as indicated in Fig. 6.3. The percentage difference (represented by the triangle symbol) is plotted together with the error affecting the experimental measurements, represented by the solid lines. The statistical uncertainty of the simulation results was lower than the one affecting the experiment.

![Figure 6.15](image)

**Figure 6.15:** Top: Normalised counts with respect to the position in the tube. Bottom: Percentage Difference (triangles) and experimental uncertainty (continuous line). Results of the X-tube (a), Y-tube (b) and Z-tube (c) with the Am-Be source forming an angle of 0 degrees with the Z-axis.

The simulation results for source angle at 0 degrees, Z-tube (Fig. 6.15) and source angle at 90 degrees, X-tube (Fig. 6.18), showed greater response than the experimental results at the end of the tube closest to the source. This is due to the assumption in the simulation that the tube charge collection efficiency is 100%, however the experimental results showed that the tube efficiency at 4 cm at both ends of each is lower and as a result, there are little to no counts observed in this region. As a result the simulation results of these specific source positions were normalised to a maximum point well away from the edges of each tube. Fig. 6.16 and Fig. 6.17 appear to show systematic change between exper-
CHAPTER 6. DIRECTIONAL NEUTRON DETECTOR

Figure 6.16: Top: Normalised counts with respect to the position in the tube. Bottom: Percentage Difference (triangles) and experimental uncertainty (continuous line). Results of the X-tube (a), Y-tube (b) and Z-tube (c) with the Am-Be source forming an angle of 30 degrees with the Z-axis.

Figure 6.17: Normalised counts with respect to the position in the tube. Bottom: Percentage Difference (triangles) and experimental uncertainty (continuous line). Results of the X-tube (a), Y-tube (b) and Z-tube (c) with the Am-Be source forming an angle of 45 degrees with the Z-axis.

Experimental and simulation results, this may be due to geometrical effects (misalignment or small mismatches between the Geant4 and experimental descriptions).
Figure 6.18: Top: Normalised counts with respect to the position in the tube. Bottom: Percentage Difference (triangles) and experimental uncertainty (continuous line). Results of the X-tube (a), Y-tube (b) and Z-tube (c) with the Am-Be source forming an angle of 90 degrees with the Z-axis.

Figure 6.19: Top: Normalised counts with respect to the position in the tube. Bottom: Percentage Difference (triangles) and experimental uncertainty (continuous line). Results of the X-tube (a), Y-tube (b) and Z-tube (c) with the Am-Be source forming an angle of 300 degrees with the Z-axis.

6.4.4 Detector Optimisation

The optimisation study of the tube efficiency with respect to changing the radius of a $^3$He tube was investigated under constant pressure of 10 atm. A linear trend was observed in
the detector response with increase in the tube radius (Fig 6.20).

![Graph showing response of a single $^3$He tube at 10 atm pressure with radius change](image1)

**Figure 6.20:** The response of a single $^3$He tube of 10 atm fill pressure at 5 different radii, for source at 45 degrees to the sphere, neutron energy 100keV.

A plot of efficiency as a function of tube radius for a constant pressure of 10 atm with the point source at 45 degrees to the tube axis is shown in Fig. 6.21. The detector efficiency is also shown to increase approximately linearly with radius for all energies except for 10 MeV, being due to the neutron cross section of interaction (see Figure 6.4). The

![Graph showing efficiency vs tube radius](image2)

**Figure 6.21:** A plot of efficiency as a function of tube radius for a constant pressure of 10 atm with the point source at 45 degrees to the tube axis.

optimal radius is dependent on the energy range of neutrons that is desired for detection.
For energies of neutrons of 100 eV and higher, there is a distinct advantage to using tubes of larger radius, with an average of 5% difference in efficiency between the 5 mm and 25 mm radius tubes. For energies of neutrons 10 eV and lower, the difference in efficiency between the largest and smallest radii drops to 2%, so if the user only wishes to detect cold to slow neutrons, 5 mm radii may be deemed sufficient. For optimal detection over the largest range of neutron energies, the 25 mm radius tube holds the distinct advantage, as it has the greatest efficiency over all energies.

6.4.5 Detector response in an 18 MV linac photoneutron field

There are a few limitations to using this detector in a medical scenario. The source of neutrons in these scenarios would not be point like, due to different shielding and scattering in the hospital environments. Also the source would not be close to the orthogonal detectors and thus directionality in a real application might be lost. As we are unable to easily determine the spectral information of the incoming neutron field it is difficult to determine the equivalent dose.

Simulation of the directional neutron detector when placed infield at isocentre position and exposed to the spectrum of neutrons from an 18 MV linac, showed very little response, with only 26 hits (created from neutron interactions) interacting across all three $^3$He gas tubes. Due to low statistics it was difficult to discern any directional dependence. Furthermore simulations of the detector response to neutrons when placed at 1m out of field from the isocentre position (phantom at isocentre), showed a total of 88 hits from neutron events interacting across all three $^3$He gas tubes. The neutron fluence at these positions it is too low for the directional neutron detector to be applicable, highlighting the inability to use this detector to measure neutron events in an mixed photoneutron radiation field produced by an 18 MV medical linac. The detector is more suitable for the investigation of secondary neutron production at a nuclear facility, where the number of secondary neutrons produced from neutron beam lines is significantly larger than that produced by an 18 MV medical linac, whose primary beam is predominantly high energy photons.

6.5 Conclusion

A Geant4 based simulation has been developed to investigate the directional behaviour of a neutron detector based on three $^3$He tubes located along three perpendicular axes within a single sphere of HDPE. The simulated relative neutron detection efficiency of a single $^3$He tube compared well to the calculated detector efficiency (Fig. 6.14). The $^3$He tube showed a greatest detection efficiency to thermal and cold neutrons. The efficiency was observed to drop to less than 1% for fast neutrons with energy above 1 MeV. Geant4
was shown to have adequate physics models to describe neutron interactions in the energy range of 0.025 eV up to 1 MeV. The experiment results could be improved by repeating the experiment for a longer period of time to obtain better statistics.

The simulation has been validated with respect to experimental measurements performed at ANSTO, Lucas Heights with an Am-Be source, positioned at the surface of the sphere at angles of (0, 30, 45, 90, 300) degrees in a single plane. Good agreement was observed overall between the experimental and simulation results for a number of Am-Be source positions with the Z plane in Figures 6.15, 6.16, 6.17, 6.18, 6.19, with some systematic structure between simulation and experimental results observed.

Generally the percentage difference varied on average between 10% and 20% within the tube position of -5 cm to +5 cm but with a few of the tubes exhibiting percentage difference up to 40%. The worst agreement was seen at the extremities of the tubes and for few positions at the centre of the tubes. Experimental errors in the positioning of the source on the surface of the HPDE sphere as well as accurate pulse signal processing using the Mesytec position sensitive tube readout system could be the cause of some disagreements between simulation and experimental results. The calibration of the positional gain of the electronics was not measured. The linearity of the tubes was also not investigated.

There are a few limitations to using this detector in a medical or industrial scenario. The source of neutrons in these scenarios would not be point like, due to different shielding and scattering in these environments. Also the source would not be close to the orthogonal detectors and thus directionality in a real application might be lost. As we are unable to easily determine the spectral information of the incoming neutron field it is difficult to determine the equivalent dose. It was concluded that the detector would not be suitable to be used in an mixed photoneutron radiation field produced by an 18 MV medical linac.

In order to obtain spectral information of the incoming neutron field, different HDPE spheres would need to be investigated and with different thermalisation efficiencies. From the positional data presented we have shown that the intensity of the detected events in each bin changes strongly with direction both experimentally and with simulation. As the distribution is different for each angle it should be possible to determine the direction of the incident source using matrix algebra. However the neutron thermalisation can affect the interaction depth in the HPDE sphere, therefore affecting the detector response. More in-depth studies are necessary to investigate this aspect. Different HPDE shapes, materials and tube designs will be studied to improve the current detector design.

Future studies will use Geant4 to investigate the detector directionality with monoenergetic neutrons in order to understand the angular response of the entire detector, which
we might then be able to construct a response function with the possibility to then un-
fold and obtain the energy of incident neutrons (initially an average) and then eventually
the entire energy spectrum. This could be of potential use at a nuclear beam facility to
understand the neutron energy spectrum at various points along the beam line.
Chapter 7

Conclusions and Recommendations

Electron Linear Accelerators (linacs) used in Radiotherapy treatments produce undesired photoneutrons when they are operated at energies above 8 MeV. These neutrons contaminate the therapeutic beam and increase both in an out of field dose to patients. As a consequence, healthy tissues receive a non-negligible dose and the risk of developing secondary cancer is increased. In addition, neutrons have large mean free paths and if shielding requirements are not met, there is an inherit risk in exposure to personnel working in radiotherapy facilities. As radiation therapies are increasingly using higher energy beams as standard treatment procedure, the production of photoneutrons is an issue. Dosimetric characterisations of these high energy beams, as well as investigations about neutron contamination are key interests. Primarily the need is to detect these neutrons and be able to estimate the neutron dose equivalent. To achieve this, three different types of neutron detectors (silicon p-i-n diodes, 3D silicon detectors and a directional neutron detector) were investigated in this thesis and evaluated for their potential use in clinical environment.

In this thesis the first detector under investigation was the silicon p-i-n diode. These detectors were found to have a high neutron to photon tissue equivalent dose sensitivity discrimination and utilize a simple readout method. The developed ion implanted bulk silicon 0.5E p-i-n diode were shown to be convenient for real time estimation of fast neutron absorbed dose on the surface of a solid water phantom in a pulsed mixed photoneutron field, produced by an 18 MV medical linac. It was found that the entrance surface neutron dose can be easily determined by passive silicon p-i-n diodes, based on neutron calibration with a 14 MeV neutron source and neglecting the photon contribution. The relative neutron tissue dose equivalent and silicon neutron displacement damage KERMA compared well for different depths inside the solid water phantom and present a quick method of estimation of the neutron dose equivalent at any depth, based on knowledge of the neutron dose equivalent response of the diode on the surface of the phantom for a particular linac.
CHAPTER 7. CONCLUSIONS AND RECOMMENDATIONS

Future work will involve irradiation of these detectors using clinical beams on an anthropomorphic phantom for the estimation of the patient neutron entrance absorbed dose. Currently the 0.5E p-i-n diodes studied had sensitivity to fast neutrons of 1.5 mV/cGy and provided measurements of fast neutron absorbed dose with an overall uncertainty of 15-20% when 30 Gy photon dose was delivered at Dmax using an 18 MV linac. For in patient application when the dose delivered is 2 Gy/fraction, a more sensitive p-i-n diode is required. A p-i-n diode with a sensitivity of 50 mV/cGy for fast neutron dosimetry was reported [59] and will be very suitable for fast neutron dosimetry for simple and immediate real time readout after irradiation. It was concluded that the application of p-i-n diodes on the surface of a patient body or in cavities with constant temperature make p-i-n diodes very suitable for passive real time in vivo neutron dosimetry on a 18 MV medical linac.

In this thesis the second type of detector investigated was the 3D silicon detectors. 3D silicon detectors have a small active region of either 10 µm or 20 µm thickness which is fabricated from an array of 3D, P and N columnar electrodes. These characteristics allow the detector to be fully depleted at very low bias voltages. The 3D ultra-thin detector with removed supportive silicon substrate and 10 µm thickness, allows for the reduction of most high energy photons entering the detector, thus allowing for more accurate neutron detection. The 3D detectors investigated in this study were coupled with a range of different polyethylene convertors to assess the most suitable convertor thickness for neutron detection. A 100 µm thick polyethylene convertor film was found to be most suitable. Based on the charge collection characteristics using an alpha source, the 10 µm ultra-thin detector and 20 µm thick detector with substrate, were chosen to perform measurements on an 18 MV linac. The 20 µm with substrate detector, exhibited low statistics (due to very small active area), limiting the usefulness of the device for in field neutron measurements on an 18 MV linac. When performing measurements with 10 µm ultra-thin detector on the surface of the phantom, close to the field, pulse pileup effects were observed. To reduce the pileup effects, detectors were positioned at 2 metres from the centre of the beam position.

At 2 metres from the centre of the beam position the 10 µm ultra-thin detector, exhibited spectral response from photons and neutrons as expected, with ability to subtract response of the detectors due to photons. Angular dependence was observed for this detector, with greater number of events detected with the detector facing the linac beam, in comparison to the detector in the face up position (facing the ceiling). The effect of increasing the field size, increased the number of neutron events detected. It was concluded that the 10 µm ultra-thin detector with 100 µm thick polyethylene convertor screen is suitable for out of field neutron detection measurements on an 18 MV medical linac, at a distance of
2m from the isocentre. Conversion of observed recoil proton events to dose equivalent require further consideration that is outside of this thesis. Future investigations could involve reduction of capacitance in the design of micro-structured detectors and new pillar design, etching of substrate for more efficient charge collection as well as investigation into the optimal LDPE convertor screen thickness.

In this thesis the third detector investigated was a directional neutron detector based on three $^3$He tubes located along three perpendicular axes within a single sphere of HDPE. A Geant4 based simulation was developed to investigate the directional behaviour of a neutron detector. Geant4 was shown to have adequate physics models to describe neutron interactions in the energy range of 0.025 eV up to 1 MeV.

The Geant4 simulation was validated with respect to experimental measurements performed at ANSTO, with an Am-Be source, positioned at the surface of the sphere at angles of (0, 30, 45, 90, 300) degrees in a single plane. Good agreement was observed overall between the experimental and simulation results for a number of Am-Be source positions with the Z plane. Generally, the percentage difference varied on average between 10% and 20% within the tube position of -5 cm to +5 cm but with a few of the tubes exhibiting percentage difference up to 40%. The worst agreement was seen at the extremities of the tubes and for few positions at the centre of the tubes. Experimental errors in the positioning of the source on the surface of the HPDE sphere as well as accurate pulse signal processing could be the cause of some disagreements between simulation and experimental results.

Simulation of the directional neutron detector when placed infield at isocentre position and exposed to the spectrum of neutrons from an 18 MV linac, showed very little response, with only 26 hits (created from neutron interactions) interacting across all three $^3$He gas tubes. Due to low statistics it was difficult to discern any directional dependence. Furthermore simulations of the detector response to neutrons when placed at 1m out of field from the isocentre position (phantom at isocentre), showed a total of 88 hits from neutron events interacting across all three $^3$He gas tubes. The neutron fluence at these positions it is too low for the directional neutron detector to be applicable, highlighting the inability to use this detector to measure neutron events in an mixed photoneutron radiation field produced by an 18 MV medical linac.

In culmination of this study, the silicon p-i-n diodes provided the best method for convenient real time estimation of fast neutron absorbed dose and neutron dose equivalent. It was concluded that the application of p-i-n diodes on the surface of a patient body or in cavities with constant temperature make p-i-n diodes is suitable for passive real time
in vivo neutron dosimetry on a 18 MV medical linac. The 3D silicon detector (10 µm ultra-thin detector) was found suitable for out of field neutron detection measurements on an 18 MV medical linac, at a distance of 2m from the isocentre. The directional neutron detector was limited in its use for both in and out of field neutron dosimetry in an 18 MV linac mixed photoneutron field. Its large size, sensitive electronics and susceptibility to pulse pileup effects from photons, as well as the inability to acquire enough statistics in a low neutron fluence environment, make it unsuitable for real time estimation of fast neutron absorbed dose and neutron dose equivalent in radiation fields produced by an 18 MV medical linac.
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Appendix A

A.1 I-V measurements of bulk silicon p-i-n diodes

Figure A.1: 1C silicon p-i-n diode forward bias I-V characteristic curve
Figure A.2: 1D silicon p-i-n diode forward bias I-V characteristic curve

Figure A.3: 1E silicon p-i-n diode forward bias I-V characteristic curve
Figure A.4: 1F silicon p-i-n diode forward bias I-V characteristic curve

Figure A.5: 1G silicon p-i-n diode forward bias I-V characteristic curve
Figure A.6: 1K silicon p-i-n diode forward bias I-V characteristic curve

Figure A.7: 1M silicon p-i-n diode forward bias I-V characteristic curve
Figure A.8: 1N silicon p-i-n diode forward bias I-V characteristic curve

Figure A.9: 5E silicon p-i-n diode forward bias I-V characteristic curve