Strip detector for high spatial resolution dosimetry in radiation therapy

Ashley James Cullen
University of Wollongong
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Strip Detector for High Spatial Resolution
Dosimetry in Radiation Therapy

Ashley James Cullen

A thesis submitted in partial fulfilment of the requirement for the award of the degree of:

Master of Science – Research
of the University of Wollongong
Centre for Medical Radiation Physics
Certification

I, Ashley James Cullen, declare that this thesis, submitted in partial fulfilment of the requirements for the award of Master of Science – Research, in the Centre for Medical Radiation Physics, 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.

Ashley James Cullen

8th July 2009
Abstract

Radiation therapy is an established modality in the treatment of tumours. With treatments ever evolving and increasing in terms of their complexities, the need arises to ensure the best quality treatment is delivered; the survival of the patient relies upon it. A modern treatment such as Intensity Modulated Radiation Therapy employs steep dose gradients varying dynamically to deliver complex dose profiles, whilst the experimental Microbeam Radiation Therapy (MRT) involves the delivery of an array of intense beams tens-of-microns wide separated by several hundred microns. In both cases, conventional dosimetry is inadequate in providing both spatial and temporal information about complex dose profiles.

The silicon strip detector was created to fill this void in current dosimetry techniques. Designed to withstand the intense beam of a synchrotron wriggler x-rays whilst not significantly perturbing the beam, the detector provides linear sensitive volume elements two hundred microns in pitch. This enables the ability to perform high spatial resolution dosimetry in real time.

This thesis investigates the use of the silicon strip detector as an on-line dosimetry system for MRT with applications to clinical radiotherapy. Of particular interest is the distribution and magnitude of energy deposition within the detector and the perturbative effect the strip detector has on a synchrotron wriggler x-ray beam.

Monte Carlo simulations are performed to investigate the properties of radiation incident upon the detector. These seek to understand how a beam traversing the detector interacts with it, as well as the effects the detector has on the transmitted beam and its properties.

The energy deposition spectrum within the detector was found to be predominant at low energies of below 100 keV. The deposition of dose through the detector was found to be largely constant with depth through the
central axis of a beam, dropping to $\sim 10^{-3}$ of the central value at 50 µm off the central beam axis for an infinitesimally thin pencil beam and $\sim 10^{-4}$ at 100 µm off-axis. Energy deposition laterally through water was determined as dominated by secondary electrons from the beam-edge to 150 µm, and Compton photons thereafter.

The depth dose of a MRT pencil beam was found to have an average decrease in dose of $(1.44 \pm 0.15)\%$ (95% C.I.) when the strip detector was introduced into the beam. The probability of interaction of incident photons with the detector for the MRT spectrum was determined with GEANT4 and theoretically with a comparison made. The overall interaction probability of an MRT photon is $(1.97 \pm 4.43 \times 10^{-4})\%$ (95% C.I.).

A simulation to determine the PVDR (peak-to-valley dose ratio) in a MRT field was created, however only the peak dose could be determined due to an inadequate primary photon count. The peak dose was found to decrease by $1.41 \pm 0.03\%$ (95% C.I.). Qualitative film measurements (deliberate overexposure of peak regions) displayed an increase of valley dose to film at 10 mm in water with the strip detector in the beam, but no such phenomenon at a depth of 1 mm. A simplified GEANT4 simulation was created to replicate such results with only five beamlets. Peak-to-valley dose ratio calculations from the simulation show no discernible effect at 1 mm depth, but a discernible increase in the PVDR at 10 mm depth; replicating experimental results.

Since charge collection across a semiconductor device is often complex and dynamically varies with the bias conditions. Ion beam induced charge collection (IBICC) studies seek to investigate the charge collection properties of the detector in various voltage-biasing conditions. It was found that the application of reverse bias to strips adjacent to that being read out reduced the effective sensitive volume of the read-out strip. This provides evidence for a proposal to incorporate biased guard-ring structures to prevent charge sharing, and improve the confinement of the sensitive volume.
Finally, clinical irradiations are performed with a highly collimated orthovoltage x-ray beam down to beam sizes of hundreds of microns. The spatial resolution of the detector in this configuration was found to be 500 µm. The efficacy of the detector in contemporary radiotherapy is also investigated through the use of a 6 MV linear accelerator's photon beam. Excellent agreement was found between strip detector read-outs and reference data for the linear accelerator. A quadratic relation was found between dose rate and charge per strip.
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### Glossary of Abbreviations

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<th>Description</th>
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<tr>
<td>ANSTO</td>
<td>Australian Nuclear Science and Technology Organisation</td>
</tr>
<tr>
<td>ANTARES</td>
<td>Australian National Tandem for Applied RESearch (Accelerator)</td>
</tr>
<tr>
<td>CMRP</td>
<td>Centre for Medical Radiation Physics</td>
</tr>
<tr>
<td>ESRF</td>
<td>European Synchrotron Radiation Facility</td>
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<tr>
<td>FWHM</td>
<td>Full-width at half-maximum</td>
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<tr>
<td>IBICC</td>
<td>Ion beam induced charge collection</td>
</tr>
<tr>
<td>ICCC</td>
<td>Illawarra Cancer Care Centre</td>
</tr>
<tr>
<td>IMRT</td>
<td>Intensity-modulated radiation therapy</td>
</tr>
<tr>
<td>LET</td>
<td>Linear Energy Transfer</td>
</tr>
<tr>
<td>Linac</td>
<td>Linear accelerator</td>
</tr>
<tr>
<td>MCA</td>
<td>Multi-channel analyser</td>
</tr>
<tr>
<td>MOSFET</td>
<td>Metal-oxide-silicon field-effect transistor</td>
</tr>
<tr>
<td>MRT</td>
<td>Microbeam radiation therapy</td>
</tr>
<tr>
<td>PVDR</td>
<td>Peak-to-valley dose ratio</td>
</tr>
<tr>
<td>SSD</td>
<td>Silicon strip detector</td>
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<td>UOW</td>
<td>University of Wollongong</td>
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Chapter 1 – Introduction

1.1 Interactions of Photons with Matter

At low, therapeutic energies, there are three primary ways photons interact with matter, namely: photoelectric absorption, Compton scattering and pair production. All processes involve the partial or total transfer of a photon’s energy to other particles. All aforementioned processes involve an abrupt change of the particle’s behaviour, contrasting the gradual dissipation of kinetic energy by charged particles as they traverse a medium.

![Figure 1 - Photon cross-section in water for photon energies up to 20 MeV. (Produced from data from the NIST XCOM Photon Cross Section Database [1]).](image)

1.1.1 Photoelectric Absorption

In photoelectric absorption, a photon interacts with an atomic electron in an absorber atom, imparting all of its energy upon it. An energetic photoelectron is ejected from one of the bound shells, with a kinetic energy equal to the photon energy minus the binding energy, given by:
\[ E_e = h\nu - E_b \]  \[2\]

Such that \( E_b \) is the binding energy of the photoelectron in its original shell. At therapeutic energies, the majority of energy is transferred to the electron in the form of kinetic energy.

The ejection of the electron leaves an ionised absorber atom with a vacancy in one of its shells. This vacancy is subsequently filled through the capture of a free electron, or by a rearrangement of electron levels. This may result in the emission of one or more characteristic photons. \[2\]

1.1.2 Compton Scattering
Compton scattering is a process involving the incident photon and an electron of an absorbing material. For the majority of therapeutic photon energies, this process is the primary interaction in matter. \[2\]

An incident photon is deflected through an angle \( \theta \) with respect to its initial momentum vector by scattering off an electron. A portion of energy is transferred from the photon to the electron, which recoils through an angle away from the photon. Since any scattering angle from 0 to \( \pi \) is possible, the energy transferred to the electron is a function of scattering angle. The energy of the scattered photon, \( h\nu' \) is given by:

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

1.1.3 Pair Production
Pair production is a consequence of Einstein’s energy-mass equivalence, possible when the photon energy exceeds 1.022 MeV, twice the rest energy of the electron. The interaction, which occurs in the Coulomb field of a nucleus, results in a photon being converted into an electron-positron pair. Excess energy above twice the rest energy is carried by the two newly created particles as kinetic energy. Due to the unstable nature of anti-matter in the presence of matter, the positron soon annihilates, producing a photon pair. \[2\]
1.1.4 Coherent Scattering
A fourth interaction, coherent scattering occurs when a photon scatters from an atom without the loss of energy. Since this process involves neither the excitation nor ionisation of the atom, this process contributes to virtually no energy transfer. The direction of the photon however is altered, resulting in the deflection of a photon of full energy, contrasting the less energetic scattered photon from Compton scattering. [2]

1.2 Radiation Therapy
Radiation therapy is an established set of treatment modalities primarily for the treatment of cancer. It involves the delivery of a known quantity of radiation to a specific target on or within the body. A recent study by Delaney et al. places the proportion of all diagnosed cancer patients receiving radiation therapy at 52%. [3] With such a large proportion of the population receiving radiation therapy, the importance of ensuring treatments meet ever-higher quality standards is realised.

Both photon and charged particle therapy rely on the process of ionisation to initiate lethal processes on cells within the body. To directly kill a tumour cell, the DNA of the cell must be irreversibly damaged; typically this is achieved through the production of a double strand break. However, the majority of radiotherapy-induced cell kill is not caused by the interaction of radiation with DNA, but rather through secondary chemical processes. The most important chemical being the hydroxyl radical, produced through the ionisation of a water molecule. This production of free radicals is responsible for the majority of radiation damage. When a free radical is created in close proximity to a DNA strand, there exists a probability that the free radical will interact with the strand, destroying its structure. [4]

1.2.2 External Beam Radiation Therapy
External beam radiation therapy involves the delivery of dose from an external radiation generator. Contemporary radiation therapy treatment involves the use of a linear accelerator’s photon beam. Microwaves within a wave-guide structure accelerate electrons at to high energies, which then
strike a target producing Bremsstrahlung. The resulting beam is spectrally shaped to ensure uniformity and collimated to conform to a specific treatment field.

Modern treatments involve the use of multi-leaf collimators, which provide the ability to shape the radiation field to conform to irregularly shaped targets. Since this spares healthy tissue far better, it enables the delivery of higher dose to the target volume. Additionally, since tumours are rarely of spherical geometry, it enables a fast and efficient way to treat a tumour from multiple angles without the fabrication of multiple lead blocking plates.

*Intensity modulated radiation therapy* (IMRT) is an advanced technique which involves the use of multileaf collimators in a dynamic mode whereby their position alters during delivery from a specific angle. This enables the delivery of complex dose profiles, further tailoring the treatment to specific requirements. This better spares critical structures in the treatment of tumours of the head and neck, ensuring better tumour control with lesser side effects.

The most recent linear-accelerator treatment, *intensity modulated arc therapy* involves a combination of IMRT with continuous gantry rotation. Rather than shooting a modulated beam from around five angles, this method delivers dose over a continuous arc about the target. This results in better tissue sparing than conventional treatment methods, since the entrance and exit dose are spread over a larger volume to deliver the same dose to the target. [5]

1.2.3 Microbeam Radiation Therapy

The effect radiation induces in tissue is a factor of not just dose rate, but also of the volume exposed. By minimising the volume irradiated, the tolerance of the healthy tissue is enhanced. Microbeam radiation therapy is an experimental form of radiosurgery, which exploits this effect, [6] delivering a spatially fractionated treatment with doses up to 50 times higher than conventional treatments. [7] It consists of the delivery of a parallel planar
array or cylindrical array of beamlets of very high dose to the target. This thesis will deal primarily with the parallel planar array configuration due to the nature of the physical construction of silicon strip detector. This treatment method results in the confinement of very large dose to strips of miniscule width (tens of microns) separated by a larger valley region (hundreds of microns) containing only dose due to scattering and secondary particles.

The ratio of the dose in the beamlets to the dose in the valley region is known as the peak-to-valley dose ratio or PVDR. This measure provides a quantitative assessment of the quality of the radiation field at a position in a medium. A high PVDR is desirable since it indicates good separation of the high dose in the peak region to the dose in the valley, thus maximisation of the spatial fractionation benefit. A low PVDR however would indicate significant scattering of the primary field has occurred.

To minimise the spread of the primary photon beam with depth, MRT requires the use of synchrotron wriggler X-rays due to their extremely low divergence properties. The wriggler x-rays pass through 1.5 mm of aluminium and 1.0 mm of copper to filter out lower energies. This results in an energy spectrum from 50 keV to well over 350 keV, with a peak energy of 83 keV and a median energy of 107 keV (see Figure 8). The beam then exits a beryllium window, and traverses an ionisation chamber before it is incident upon the multislit collimators 33 m from the source. A fast shutter system is placed upstream of the multislit collimators, which enables specifying how long each treatment element is delivered for. A computer-controlled goniometer provides precise movement of the subject through the radiation field. [8]

Studies of the effects of MRT on animal subjects display an unusual resistance of normal and developing tissues to the extreme treatment doses. Largely, experimentation has been on rats, nude mice and piglets. [7] The graph below shows the survival response of rats with intracranial 9L gliosarcoma to varying MRT doses.
Figure 2 - Survival curves of rats receiving tolerable and high-dose MRT, and unirradiated controls. Note: “Tolerable dose” corresponds to a valley dose of 19 Gy to less, whilst “High-dose” refers to a valley dose of greater than 19 Gy. [9]

The superiority of MRT in treating rat gliosarcomas is apparent given the much greater survival response when compared to conventional broadbeam fields in Figure 2. Whilst the high-dose certainly improved survival

Conventional radiotherapy techniques for the treatment of brain tumours typically involve the delivery of conformal fields to complex three-dimensional treatment regions. The risk of collateral damage to adjacent healthy tissues is high, particularly in photon treatments. This risk is especially pronounced in children, where such damage can result in malformation of the central nervous system. As such, there exists a tendency for paediatric oncologists to postpone or avoid treatment with radiation. However, such a strategy may negatively influence the survival outcome. [7]

Due to the minimal effect of MRT on developing tissues, it enables the treatment of such cases without delaying the start of treatment. It also
enables the treatment of previously inoperable or otherwise untreatable cases where treatment would likely prove fatal.

1.3 High Spatial Resolution Dosimetry
Several methods of high spatial resolution dosimetry exist in clinical practice and in research. Some employ passive elements, whilst others have active components that respond electrically to radiation.

1.3.1 Radiographic and Radiochromic Film
Methods of film dosimetry are favourable to many areas of radiation oncology, particularly where heterogeneity of dose is present across the field. This makes film one of the dosimeters of choice in IMRT research and verification. For both radiographic and radiochromic the absorbed dose may be deduced through changes in film’s final opacity to optical light.

Radiographic film is a film dosimeter containing photographic emulsions. Irradiated microscopic grains of silver bromide within the film’s sensitive volume remain within the film after chemical development, whilst those un-irradiated are removed. [10]

However, the quantitative evaluation of an ionising photons beam is difficult using conventional radiographic film due to large differences in the sensitivity to photon energies in the 10 - 200 keV range. In addition, the radiographic films are not tissue equivalent, in that their energy absorption and transfer properties differ from biological tissue. The sensitivity of the film to optical light and the requirement of wet chemical processing disadvantage the use of the film in a clinical setting. [11]

In radiation oncology, there exists the need for a film, which requires no development and yields absolute values of absorbed dose with a high degree of accuracy. Radiochromic films fit this category, in that they respond to irradiation through colour change directly and require no chemical development. The image formation is as a result of dye-formation or a polymerisation process, causing colour changes through radiation-induced, chemical alteration of the film. [11]
The logarithmic relation gives the photographic density of a radiographic or radiochromic film [10]:

\[ D = -\log \frac{I}{I_0} \]

Where \( I/I_0 \) is the transmission ratio of visible light at a given point on the film. Characteristic curves relate the density of a particular film to a dose of a particular particle, thereby enabling a determination of dose from the film. [10]

Both methods of dosimetry are advantageous in that the dose at many points in the film may be determined simultaneously with a single dosimeter. However, two main limitations exist, to obtain dose processing is required post-irradiation, typically through the digitization of light transmission across the film. The other is that it provides no details about the dose rate of the treatment, only the integral dose (within the film’s dynamic range). [10]

1.3.2 Solid-State Detectors

In a periodic crystalline lattice material, allowed energy bands for electrons are established within the material. An electron within the material is confined to specific energy regions known as the valance and conduction bands. The two are separated by a band gap, a quantum mechanically forbidden region, which varies depending upon the material, typically greater than 5 eV for insulators, and approximately 1 eV for semiconductors. [2]

Imparting energy on a valence electron causes its excitation, this may occur from thermal energy, or impinging radiation. With sufficient energy, it is possible for the electron to elevate to the conduction band, breaking covalent bonds and enabling it to drift through the crystal. The vacancy left in the valence band known as a hole may also drift through the crystal as electrons from adjacent atoms fill the vacancy leaving a hole of their own. The collective name for these two charge carriers is known as an electron-hole pair. [2]
The creation of these electron-hole pairs form the basis for solid-state detectors, with the negatively charged electrons and the positively charged holes both moving under the influence of an electric field, albeit in opposite directions.

1.3.2.1 Diode Detectors

Diode detectors are amongst the most basic of solid-state semiconductor detectors. They fundamentally consist of a silicon crystal of high purity, doped with a p-type region having an excess of holes, and a typically thinner n-type layer having an excess of electrons. Each region is connected electrically to an external readout system, where data of dosimetric importance is obtained. [12]

Under the application of a positive potential between the n-type to p-type regions, a depletion region is formed across the boundary of the doped regions through the removal of free charge carriers. The creation of an electron-hole is typically energetic enough to reach the boundary of the depletion region, producing a signal able to be measured with appropriate amplification. [12]

1.3.2.2 MOSFET Detectors

A MOSFET detector, specifically an n-type MOSFET consists of a doped body of n-type silicon overlaid with an oxide layer (typically silicon dioxide), which itself is overlaid by a metal gate. In the interface between the oxide layer and the silicon, at opposing sides are two regions of highly doped p-type silicon. These are known as the source and the drain. An induced n-type channel is formed between the source and the drain and is located close to the oxide layer. [2]
Figure 3 - Schematic diagram of n-MOSFET device. [13]

The silicon dioxide, being a dielectric material effectively makes the MOSFET device act like a capacitor, with the non-metallic side of the dielectric a semiconductor device. Incident ionising radiation produces electron-hole pairs in the oxide layer, with holes forced towards the oxide-silicon interface where they are trapped. The charge sheet at the interface causes a change in the current in the n-type channel which results in a change in the gate voltage required to keep the channel current constant. This current is very sensitive to the sheet charge due the channel’s location, making MOSFET detectors very good at detecting small changes in dose. The detector may be operated either in passive mode with no externally applied bias on the gate, or in active mode, which increases the sensitivity through reducing recombination of radiation-induced charge carriers. [13]

1.3.2.3 Edge-on MOSFET

Edge-on MOSFET dosimetry is a technique whereby a MOSFET radiation detector is positioned such that its smallest dimension is normal to the incident beam. However, MOSFETs are typically surrounded by an air gap formed by an epoxy bulb, which surrounds the sensitive devices. The device is still functional in the absence of this bulb, however it exposes the delicate detector and removes build-up material. Since for the REM RADFET, which contains two pairs of MOSFETs with gate oxide thicknesses of approximately 0.13 and 1 µm, this enables the ability to perform high spatial resolution dosimetry for MRT within a very confined sensitive region. [14]
This technique gives instantaneous results for the dose in the sensitive volume, however the MOSFET must be scanned across the radiation field to build a dose profile. It is also expected that the MOSFET will perturb the beam traversing it far more so than neighbouring beams. Whilst edge-on MOSFET would arguably provide an excellent method of high spatial resolution quality assurance, it is impractical for real-time dosimetry. Specifically, its limited radiation lifetime in the intense synchrotron field would be limit the total operational time of such devices.

1.3.2.4 MapCheck

MapCheck, an IMRT dosimetry tool from Sun Nuclear contains a 2-dimensional array of 445 n-type diodes over an area of $22 \times 22 \text{ cm}^2$. Two distinctive regions are present in the array, the central-most $10 \times 10 \text{ cm}^2$ region which has a diode spacing of 7.07 mm, and the outer region where the diode spacing increases to 14.14 mm. Since the device has relatively low spatial resolution, the device provides a method of verification, not dosimetry of IMRT plans. The calculated dose at each diode location is compared with the integrated current of each device over the irradiation period. The limitation of the spatial resolution however, ensures that the device is not useful for dosimetry of high dose gradients over small distances. [15] Due to the large diode cross-sections and array spacing, MapCheck is of no use for MRT dosimetry.
1.4 The Silicon Strip Detector System

1.4.1 Overview
An ideal high-resolution dosimeter incorporates the ability to determine the dose at multiple points simultaneously. It also employs a solution whereby it does not significantly perturb the spectrum of the incident beam, enabling its use as an on-line dosimeter.

The silicon strip detector (SSD) is a high spatial resolution solid-state detector designed by the Centre for Medical Radiation Physics (CMRP), at the University of Wollongong (UOW) and fabricated by SPO-BIT Detector, Ukraine. It was designed to fulfil the goal of providing on-line dosimetry for microbeam radiation therapy.

1.4.2 The Silicon Strip Detector

Figure 5 - Close up view of the silicon strip detector. The black rectangular region dominating the image is the p-type silicon wafer, lighter striations in the centre are aluminium and silicon dioxide layers, representing the active region of the detector. Visible at the bottom are wire bonds electrically connecting the detectors to read-out electronics.

The silicon strip detector (SSD) consists of 128 phosphor implanted n+ strips on a p-type silicon wafer. Each strip is 10 μm wide, 5 mm long, and with an inter-strip pitch of 200μm. The surface of each strip is coated in a 1 μm thick layer of aluminium, with a bonding pad at the base, where wire bonds electrically connect the strips to external electronics. The inter-strip region corresponding to the p-type silicon wafer is p+ implanted and overlaid with a 5 μm layer of silicon dioxide. The p+ implantation acts to minimise charge
build-up at the interface. The device also incorporates a p+ guard ring structure to minimise surface leakage current. [16]

The device is mounted on a plastic triple-layered circuit board with tracks electrically connecting the wire bonds from the SSD to two 68-pin connectors. Two distinct variants of the p-type SSD were constructed with p-type silicon of different resistivity: one of high resistivity (5kΩ·cm) and the other of low resistivity (10Ω·cm). This thesis exclusively deals with the p-type silicon strip detector.

![Figure 6 - The strip detector as mounted to connector circuit board.](image)

The use of the SSD permits the acquisition and readout of real-time dose measurements. As mentioned, the SSD that is the subject of this thesis consists of 128 individual strips, however these may be readout simultaneously. This achieved through the use of two 64-channel frequency-to-current converter chips connected to a data interface, the contents of readout are polled by a computer then analysed and output by a software interface.

The SSD is also a general-purpose dosimeter also suited to hospital-based radiation environments such as determining the dose distribution across a linear accelerator field, in particular, investigating the steep dose gradients present in the penumbral region and rapidly changing IMRT deliveries.
1.4.3 On-line Readout System

The SSD is effectively 128 dosimeters on a single piece of silicon; this introduces technical challenges regarding fast readout of the device using conventional techniques. To overcome this problem, the detector is operated in current mode, a necessity for the intense synchrotron beam, which induces relatively large currents in the PN junctions.

The TERA chip was produced at INFN, Torino to address the issue of high spatial resolution dosimetry. The chip itself measures 64 channels simultaneously. By measuring the frequency at which a capacitor is charged and discharged, the current induced in each strip may be deduced. [17]

![Figure 7 - Screenshot of the LabView tool sheet that the SSD readout electronics are controlled and readout from.](image)

The readout system runs a tool sheet in LabView, which communicates with the hardware, processes the raw data into meaningful quantities and orderings, and outputs the results to both the screen and to an output file. Two modes of acquisition are possible: single and continuous. For single mode, the total time and the time between successive reads are specified, the readout system will poll the hardware in intervals set until the total time is reached. In continuous mode, the software continuously polls the hardware in the intervals set until the user interrupts the process.
Together the silicon strip detector and readout system permit the real-time acquisition and readout of dose as a function of spatial position in 200 µm pitch. With future clinical implementation with MRT, the detector will enable a real-time determination of the characteristics of the radiation field. If parameters exceed clinical tolerances, a beam stop may closed instantaneously via software control. Post-irradiation, data may be used to determine the quality of treatment with specific dosimetric information, and to diagnose trends in operational effectiveness.

1.5 Thesis Outline
The aim of this thesis is to provide insight into how energy is deposited within the silicon strip detector. For this purpose, computational results obtained from Monte Carlo simulation are produced using the GEANT4 package in Chapter 2. However, since GEANT4 is purely a radiation transport code, it provides no insight into how deposited charge may be collected within the detector, for this purpose ion beam induced charge collection studies are performed in Chapter 3. This enables a determination of factors such as what effect varying charge collection characteristics occur the device under various biasing conditions. Chapter 5 is solely concerned with the use of the SSD in a radiation oncology clinical setting, with experiments performed using orthovoltage x-ray and megavoltage linear accelerator units to irradiate the device.
Chapter 2 – Monte Carlo Studies

Monte Carlo simulations were used to provide computational results to problems of energy deposition, scattering and perturbative properties of the SSD. Monte Carlo methods are a ray-tracing approach to particle physics modelling, whereby along the path, pseudo-random numbers are generated, with interactions and their properties occurring from the lookup of probability distributions.

The need for defining such probability distributions arises, which are typically derived from combinations of theoretical, empirical, and semi-empirical sources. As such, computational results require experimental validation to ensure close agreement before conclusions may be drawn about properties or processes inducing certain effects.

Monte Carlo simulations are unique in that a situation exists with zero electrical noise, no background radiation and total event collection. However since charge collection properties are not modelled: this poses a limitation in modelling in solid-state radiation detectors. This can, however be modelled using the software package TCAD which is beyond the scope of this thesis. Ion beam induced charge collection (IBICC) studies are performed in Chapter 3, provide some insight into how charge is collected across the detector’s surface.

Overview of GEANT4

GEANT4 is an open-source Monte Carlo package developed as part of an international collaboration lead by CERN. GEANT4 has applications in high-energy physics, medical physics, and space science. [18] Due to the largely modular nature, complementary and alternative physics models may be employed at the user's discretion. The package itself consists of a multitude of classes written in the object oriented C++. A simulation program is written through the use of classes inheriting behaviour from the core GEANT4 classes.
Of particular use to the silicon strip detector is the Low Energy electromagnetic package, providing the tracking of photons to below 1 keV. The use of the low energy electromagnetic physics package has been shown to have good agreement with NIST reference data [19].

2.1 Energy Deposition Spectrum

An investigation was performed into the response of the silicon strip detector to the ESRF MRT synchrotron beam in free-air. This involved the construction of the silicon strip detector’s geometry virtually within GEANT4.

![Figure 8 - The ESRF spectrum as used in Monte Carlo simulations.](image)

Since the detector is designed to be irradiated with synchrotron wrigglers x-rays, the spectrum is rather complex. To obtain an understanding as to the absorption of spectral components, firstly an infinitesimally thin pencil beam of monochromatic X-rays of 50, 100 and 150 keV are simulated incident normally upon the detector. The surface of a single strip was irradiated with 50, 100 and 150 keV monochromatic photons, and the ESRF MRT spectrum. An energy deposition spectrum is generated by incrementing the count in an array element. The incremented element corresponds to the total energy
deposited by a track in the sensitive volume, rounded to the nearest unit of keV.

Due to the low atomic number of silicon and the 375µm thickness of the detector, the interaction probability is low for the majority of the MRT spectrum. This necessitates a rather large primary photon count to yield reasonably noiseless data. As such, spectrums were created from $10^8$ incident primary photons at each energy configuration (monochromatic or MRT polychromatic).
Radiation was chosen to be incident upon a SiO₂ strip as opposed to an Al contact since physically, the SiO₂ layer represents a larger surface area than do Al contacts. Thus the probability of a photon of arbitrary origin traversing this region when crossing the face of the detector is far greater.

Figure 10 - Energy deposition spectrum for 50 keV primary photons.

Figure 11 - Energy deposition spectrum for 100 keV primary photons.
The spectra for monochromatic photons display the two prominent features of low energy photon spectra: the Compton continuum and photopeak. Increasing incident photon energy results in an increase in maximum Compton electron energy in addition to an increase in the proportion of Compton to photoelectric interaction.

The dip in counts at the 0 keV bin is a consequence of binning to the nearest 1 keV. This results in approximately half the counts in the 0 keV bin as compared to the 1 keV bin, assuming approximately equal counts in the 0 – 1.5 keV region.
The energy deposition spectrum due to the MRT beam has considerably less discernable structure than for the monochromatic photon simulations.

Significant energy deposition is observed below 40 keV, the cut-off at which very few photons are in the MRT beam. This is explained through partial energy deposition by a traversing photon, where Compton scattering off an electron results in the escape of the scattered photon from the detector, but results in the deposit of energy upon the electron. Such partial energy deposition may also occur through the production of a photoelectron able to escape from the detector.

A local maximum occurs at 60 keV, this corresponds to the point at the crossover of the interaction probabilities of photoelectric absorption and Compton scattering, with the latter dominating for higher energies. Due to the low interaction probability of most low energy photons with the 375 µm thick silicon layer, scattered photons have a low probability of interaction, resulting in impartial energy deposition in the majority of Compton scattering events. This results in a diminishing count for higher energies.
The low counts beyond 200 keV is accounted for due to the fact that less than 7% of the peak photon flux is present beyond these energies. In addition, decreasing overall interaction probabilities, increasing relative Compton interaction probabilities, and the greater scattered photon energies result in minimal, but mostly partial energy deposition interactions for photons at these energies.

This result is important for the use of MRT in that the strip detector will effectively *harden* the beam, that is, remove a photon component more weighted towards the lower energies. However, this experiment determines integral energy deposition from traversing photons. Partial energy deposition may result in electrons or photons exiting the distal surface of the detector and contribution towards surface and valley doses.

### 2.2 Spatial Distribution of Energy Deposition

#### 2.2.1 Within Strip Detector

A pencil beam was fired through a vacuum and incident centrally upon a single strip of the SSD, with photon flux spectra the same as that for the ESRF MRT spectrum. Energy deposited within the detector was binned according to its spatial location with depth \( z \), and radial distance off-axis distance \( \rho \) binned along with the energy deposited \( E_{\text{event}}(\rho,z) \) such that:

\[
E(\rho, z) = E'(\rho, z) + E_{\text{event}}(\rho, z)
\]

Where \( E_{\text{event}}(\rho,z) \) is the previously deposited energy at coordinates \((\rho,z)\).

At the end of the simulation, the cumulative energy deposition for all bins is output to a file along with corresponding coordinates in the following structure:

\[
\begin{array}{ccc}
\rho \ (\mu \text{m}) & z \ (\mu \text{m}) & E_{\rho, z} \ (\text{keV}) \\
\end{array}
\]

However it must be noted that this data is not binned in voxels of equal volume. To determine dose, normalisation of this data through division of the volume (hence mass) each radial voxel occupies needs to be performed. The
volume for a voxel bounded by $z$ dimensions $[z_1, z_2]$ and radial dimensions $[r_1, r_2]$ is calculated as:

$$V_{\rho,z} = (z_2 - z_1) \times \left( \pi r_2^2 - \pi r_1^2 \right)$$

$$= \Delta z \cdot \pi \left( r_2^2 - r_1^2 \right)$$

Since voxels are equally spaced in the $z$ dimension, the parameter $\Delta z$ is used to improve algorithm efficiency. The density of the silicon is then taken into consideration to determine a value for mass:

$$D_{\rho,z} = \frac{E_{\rho,z}}{\rho_{Si} \cdot V_{\rho,z}}$$

The data, now of dosimetric relevance is then graphed using MATLAB, where radial energy maps are generated and slices through data are taken.

*Figure 14 - Energy deposition per voxel by MRT beam in Si layer of SSD. Colour scale is in units of Gray per $10^{10}$ photons.*
Monte Carlo results show that energy deposition is near constant over the depth of the detector directly below the beam. A spike in energy deposition at a depth of 0 – 5 µm is present due to the Al-Si interface causing edge effects related to a lack of charged-particle equilibrium. The tail in energy deposition present at the distal edge of the detector is due to a lack of energy deposited in the absence of backwards scattering of photons, a probable occurrence at the low energy of soft x-rays.
Energy deposition off-axis via lateral scattering occurs within the detector due to Compton scattering being the dominant process for silicon for the majority of energies in the MRT beam. The off-axis energy deposition with depth with depth is relatively constant over the depth of the detector, again displaying features at the most proximal and distal regions of the detector relative to the beam. At 25 µm off-axis, the energy deposition is in the order of $10^{-2}$ times that at 0 µm, whilst at 50 and 100 µm, this value is in the order of $10^{-3}$ and $10^{-4}$ respectively. This sharp fall-off ensures that very minimal energy deposition arising from internal scattering would occur at the inter-strip pitch of 200 µm. However it must be stated that since GEANT4 models radiation transport, not electrodynamics, no account for electron drift or diffusion is made in the determination of the spatial distribution of energy deposition. Electrons, which may drift in intrinsic and applied potentials, are not accounted for and are assumed to traverse neutrally-charged silicon.

2.2.2 Within Water Phantom

A 25µm wide microbeam was fired through air a distance of 20 cm, incident upon the SSD. After traversing the SSD, the beam traverses a distance of 50cm again through air and is incident upon a $30 \times 30 \times 30$ cm cube of water. Energy deposited at a depth of 1 cm is recorded, with the final output exported to a file.

Since the geometry and beam distribution in this simulation are symmetric laterally, to double the effective acquired counts, the energy deposition on one side is added to the opposing side. The resulting data is then normalised to the maximum-recorded dose as displayed in Figure 17:
A very flat profile is observed up to 11 microns, corresponding to the region consisting mainly due to energy deposition by primary photons. A penumbra occurs from 12 to 150 µm attributed largely due to the range of secondary electrons produced through the photoelectric effect and Compton scattered photons. A second penumbra occurs from 150 µm to the limits of the measurement however this region follows a distinctively different logarithmic gradient to the previous region. Energy deposition in this region is contributed to by Compton scattered photons, with their lower interaction probability this permitting greater range.

2.3 Depth Dose Perturbation

A pencil beam with the energy spectrum of the ESRF MRT beam was fired through air a distance of 20cm onto the strip detector orthogonal to the silicon strip array. After traversing the detector, the beam was incident upon a water phantom of dimensions 10x10x10 cm.
Figure 18 - Visualisation of simulation geometries. Photons (blue) originate from the left, traversing the detector (black) through air and are incident upon the water phantom (grey) and with dose recorded in the sensitive sub-volume (red).

A $5 \times 5 \times 5$ mm sensitive sub-volume was placed at a depth contacting the phantom surface proximal and central to the beam. The average energy deposited per primary photon was recorded in 5 mm depth increments both in the presence and absence of the silicon strip detector in the beam.

Table 1 - Dose in sensitive volume with detector in beam, and without.

<table>
<thead>
<tr>
<th>Depth (mm)</th>
<th>Dose per $10^9$ Photons [95% C.I.]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No Detector (Gy)</td>
</tr>
<tr>
<td>5</td>
<td>1.825 ± 0.006</td>
</tr>
<tr>
<td>10</td>
<td>1.687 ± 0.006</td>
</tr>
<tr>
<td>15</td>
<td>1.562 ± 0.006</td>
</tr>
<tr>
<td>20</td>
<td>1.448 ± 0.005</td>
</tr>
<tr>
<td>25</td>
<td>1.335 ± 0.005</td>
</tr>
<tr>
<td>35</td>
<td>1.138 ± 0.005</td>
</tr>
</tbody>
</table>
Depth dose curves observe an exponential decay as expected from soft x-ray photons. A discernable decrease in dose is consistently present at all depths recorded when the detector is introduced into the beam line. This difference is statistically significant at all depths, being within 95% confidence intervals as represented visually in Figure 19. The average decrease in dose at all depths is \((1.44 \pm 0.15)\)%, a relatively minimal, however discernable effect on the macroscopic deposition of dose in water.

2.4 Spectral Perturbation

A simulation was performed to determine the spectral perturbation of the ESRF MRT beam by the SSD. A pencil beam of \(10^9\) photon with the MRT spectrum was fired through the detector with primary photons detected downstream of the detector. The unique property of Monte Carlo simulation in that the origin of individual photons may be discerned was exploited. That is, only photons, which had not interacted whatsoever were counted (i.e. non-inclusive of small angle scattering) and a spectrum, with energies rounded to the nearest 1 keV bin was constructed.

To obtain an understanding of experimental results, the interaction probability was also determined theoretically. However theoretical results
rely on experimental or computational results, or interpolations thereof for values of photon cross-sections. A source with reputable validity is the NIST XCOM photon cross-section database [1], which is used for these results. The probability of interaction of photons incident normal to the SSD with the range of energies of the MRT was determined. The calculation was performed through the equation:

\[ P_{\text{SiO}_2} (E) = 1 - \exp\left( -\sigma_{\text{SiO}_2} (E) \cdot t_{\text{SiO}_2} \right) \]

Where \( \sigma \) is the cross-section in \( \text{cm}^2/\text{g} \), \( \rho \) is the density in \( \text{g/cm}^3 \) and \( t \) is the thickness in cm.

From here, the interaction fraction of photons in the remaining Si layer, hence the whole device, may be calculated from:

\[ P_{\text{SSD}} (E) = 1 - P_{\text{SiO}_2} - \exp\left( -\sigma_{\text{Si}} (E) \cdot t_{\text{Si}} \right) \]

![Figure 20 - Interaction probability of MRT energies in SiO2 layer and whole device.](image)

The spectrum resulting from the simulation is subtracted from the primary photon spectrum, yielding the energies of photons that interacted within the SSD. This is graphed below along with the theoretical prediction calculated
through multiplication of the total interaction probability from Figure 20 by the fractional primary beam energy flux.

![Graph showing theoretical and Monte Carlo interactions]

**Figure 21** - Theoretical and GEANT4 interaction count in SSD for $10^9$ incident photons.

The experimental results resemble the same trend as that predicted through theoretical calculations. However, a shift towards higher photon energies is present in interaction quantities for the experimental data when compared to theoretical results.

This may be explained through differences in the way interaction events occur in the GEANT4 simulation versus the cross-section data of the NIST XCOM database. Additionally, theoretical data was calculated in a step-wise nature in 1 keV units, whereas the simulation was performed with a continuous photon energy spectrum. Data becomes significantly noisy beyond photon energies of 350 keV, this is due to poor statistics as a consequence of the low interaction probability of photons of these energies.
Experimental results show the most likely photon energy to interact from the MRT beam is at 85 keV. Significant portions of interacting photons do so in the energy range where Compton scattering is the most probable interaction. The computationally determined interaction percentage is $(1.97 \pm 4.43 \times 10^{-4}) \%$ (95\% C.I.).

### 2.5 Peak-to-Valley Dose Ratio

A homogeneous microbeam was incident upon the cathode side of the silicon strip detector. Distal to the beam, a cubic water phantom was placed with lengths 10 cm. At the surface and at a depth of 1 cm, energy deposited within the water was recorded as a function of lateral position within a sensitive volume 500 μm deep, 7mm high and in 1 μm wide bins.

In order to maximise computational efficiency, a single microbeam was simulated as opposed to a linear array. This approach enables far better statistical results to be obtained, however would be inadequate for small phantoms or inhomogeneous mediums.

At large distances from the detector, it is possible that energy may not be deposited in each voxel due to limitations on the amount of primary photons able to be feasibly simulated. Due to this, linear interpolation is used on every pixel with an integral energy deposition value of zero. Additionally, to improve statistics, planar symmetry is assumed about the beam's longitudinal axis of the microbeam. The energy deposition values over each side of this axis are summed to the other and divided by 2.

The superposition algorithm of Siegbahn et al.[20] is used, with arrays of various widths simulated with the use of a purpose written C++ program.

The result of the algorithm is output to a file in a format representing bin widths of 1 μm, the same width as for the simulated output. The data is then
analysed in order to determine peak and valley doses, and the resultant peak-to-valley dose ratio.

Since microbeam pitch and array width are variables of the superposition algorithm, there exists the ability to determine the outcome from various configurations without the need to run a simulation each time. However for varying beam widths, it necessitates a unique simulation run for each width. It must be noted that such a method assumes no temporal or spatial dependence on energy deposition, just position relative to the centre of the beam.

Firstly a simulation was run in the absence of any in-beam SSD with a total of $10^8$ primary photons incident normal to the phantom. The beam has properties of being 25 µm wide and 10 mm tall. The sensitive volume is 7 mm tall, 500µm deep and divided into 1µm individual sensitive elements over a total length of 100 mm. The total energy deposited is summed over the entire simulation, with the integral energy deposited in each sensitive element output to a file.

![Graph of normalised dose vs position for different beam pitches](image.png)

**Figure 22** - Results of microbeam summation by the superposition algorithm with various beam pitch parameters.

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Figure 23 - Peak-to-valley dose ratio versus beam pitch at a depth of 10 mm in water and in the absence of a detector.

Figure 24 - Fractional dose of peak and valley doses versus beam pitch. Both data sets are normalisation individually to their maximum dose.

As expected, increases in PVDR as a consequence of lesser beam pitches result primarily from an increase of the valley dose. The minimal increases of peak dose with decreasing beam pitches result from contributions to the peak region by scattering from adjacent beams. Intuitively, and as shown in
Figure 24, scattering from a single microbeam falls sharply with distance from the beam’s centre. The smaller valley doses significantly increase at pitches less than 150 µm, whilst the larger peak dose only minimally increases.

The experiment is then repeated with the SSD in the centre of the beam. This aims to make a determination if significant differences in features such as peak-to-valley dose ratio, peak dose and valley dose are present with the detector in-beam.

Table 2 - Simulation results without detector, with sensitive volume at surface of phantom.

<table>
<thead>
<tr>
<th>Beam Pitch (µm)</th>
<th>PVDR</th>
<th>Peak Dose (Gy)</th>
<th>Valley Dose (Gy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>5.91</td>
<td>168.98</td>
<td>28.59</td>
</tr>
<tr>
<td>60</td>
<td>8.30</td>
<td>165.98</td>
<td>19.99</td>
</tr>
<tr>
<td>75</td>
<td>12.51</td>
<td>163.43</td>
<td>13.06</td>
</tr>
<tr>
<td>100</td>
<td>21.35</td>
<td>161.42</td>
<td>7.56</td>
</tr>
<tr>
<td>150</td>
<td>43.48</td>
<td>159.76</td>
<td>3.67</td>
</tr>
<tr>
<td>200</td>
<td>67.04</td>
<td>159.05</td>
<td>2.37</td>
</tr>
</tbody>
</table>

Table 3 - Simulation results with detector in-beam, with sensitive volume at surface of phantom.

<table>
<thead>
<tr>
<th>Beam Pitch (µm)</th>
<th>PVDR</th>
<th>Peak Dose (Gy)</th>
<th>Valley Dose (Gy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>5.99</td>
<td>166.70</td>
<td>27.82</td>
</tr>
<tr>
<td>60</td>
<td>8.30</td>
<td>163.55</td>
<td>19.70</td>
</tr>
<tr>
<td>75</td>
<td>12.47</td>
<td>161.12</td>
<td>12.92</td>
</tr>
<tr>
<td>100</td>
<td>21.02</td>
<td>159.13</td>
<td>7.57</td>
</tr>
<tr>
<td>150</td>
<td>44.31</td>
<td>157.56</td>
<td>3.56</td>
</tr>
<tr>
<td>200</td>
<td>66.50</td>
<td>156.78</td>
<td>2.36</td>
</tr>
</tbody>
</table>
The PVDR in these results fluctuates with beam pitch, and therefore a direct determination as to how the SSD influences this parameter cannot be made from these computational results. One interesting phenomena observed is a near-constancy in the reduction of peak dose with the detector in-beam. This is present as a $1.41 \pm 0.03 \%$ (95% C.I.) decrease in dose compared to absence of any detector.

The valley dose has a trend towards a decrease with the detector, however this effect cannot be characterised due to poor data. It is expected a far greater run of primary photons would alleviate this issue by causing less statistical error as position extends away from the beam.

The cross-section of a planar sensitive volume varies with the inverse square of distance from a source or scattering centre in this case. Due to this, the probability of an arbitrarily scattered photon from a central location traversing a sensitive volume decreases greatly with distance. It is proposed that due to this statistical property, poor energy deposition information is presented distant from the beam. Since the superposition algorithm used to simulate a linear array of microbeams through summation of contributions

Figure 25 - Chart of percentage of variation of simulation results with detector in-line compared to without.
from even the most distant beams, when a large number of array elements are summed, a large source of error results for the smaller, hence more noise-prone valley dose. It is this error in the valley dose that presents a problem in the simulations of this section.

2.6 MRT Irradiations at ESRF

2.6.1 Experimental Results
A solid water phantom was placed in the ID27 MRT beam-line at the ESRF, Grenoble, France. Within the phantom, radiochromic film was put at depths of 1 mm and 10 mm. The film was irradiated in this configuration, and then a second set of film irradiations were performed with the strip detector in-beam before the phantom.

However, since the valley dose is of dosimetric interest regarding healthy tissue effects, this dose was brought into the dynamic range of the film by means of exposure time. Consequently, regions of film corresponding to the peak have saturated and are no longer of dosimetric value. A determination of the effects the detector has on the dosimetric properties was investigated through qualitative analysis of preliminary film data.

![Influence of strip detector measured at 1 mm depth](image)

**Figure 26** - Film irradiation data at 1 mm depth in solid water phantom.
It may be seen that a lesser valley dose is observed at a depth of 10 mm with the strip detector in-line, however such effect is not discernable at a depth of 1 mm.

2.6.2 Monte Carlo Results

A simulation was written in GEANT4 to replicate the conditions of the above experiment. A rectangular microbeam with the polychromatic spectrum of the ID27 MRT beam-line and with dimensions 1 mm high and 50 µm wide was incident on a water phantom of dimensions 10 × 10 × 10 cm.

Dose was scored in a region 100 µm deep, 1 mm high and 1.8 mm wide in lateral bins of 1 µm. To replicate the above experiment, this scoring region was located at depths of 1 mm and 10 mm, in simulations with and without the detector in-beam each with primary photon counts of 100 million.
Peak-to-valley dose ratios were determined in order to make an evaluation of the beam quality.

<table>
<thead>
<tr>
<th>Depth (mm)</th>
<th>Peak-to-Valley Dose Ratio (95% C.I.)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No Detector</td>
</tr>
<tr>
<td>1</td>
<td>540 ± 18</td>
</tr>
<tr>
<td>10</td>
<td>497 ± 20</td>
</tr>
</tbody>
</table>

This data agrees with the experimental data in that the detector has no discernible effect at a depth of 1 mm (within scientific uncertainty). However, at a depth of 10 mm, there is a clear reduction in the PVDR when the detector is introduced into the beam.
Chapter 3 – Ion Beam Induced Charge Collection Studies

Ion beam induced charge collection (IBICC) studies are a method for determining the electronic response of a radiation detector to radiation incident upon specific regions of the detector. IBICC studies involve the use of a microprobe, a particle accelerator that produces a microscopic pencil beam of energetic ions. This ion beam is raster scanned across a region of the sample being studied and the effect it induces is recorded.

3.1 The ANSTO Heavy Ion Microprobe

The heavy ion microprobe, a feature of the ANTARES facility at ANSTO employs a dual-stage van de Graff accelerator to produce MeV ion beams. An Alphatross ion source produces protons and alpha particles from hydrogen and helium gas respectively. The gas is ionised by a radiofrequency field, with positive ions from the plasma extracted and passed through rubidium vapour. This causes charge exchange to occur, producing negatively charged ions. [21]

The ions are accelerated to keV energies, and electrostatic steering plates are used to focus and steer the beam to the low energy flange of the van de Graff accelerator. Acceleration to MeV energies occurs in two stages. Negative ions are attracted towards the positively charged terminal and passed through a stripping gas, removing electrons. The newly positively charged ion is now repelled from the terminal. [21]

The energy of the beam exiting the high-energy flange of the accelerator is dependent on the charge of the ion produced by the ion source and the electron stripping at the accelerator centre. The current on an analysing magnet is precisely specified to select out the energy and charge state of the desired beam, and deflect it into the microprobe beam line, whilst excluding unwanted beam constituents. [21]
Figure 29 - Schematic of the ANSTO ANTARES microprobe beamline. [22]

A set of pre-slits, a set of adjustable four-jaw apertures are used to limit the beam current, the roughly collimated beam is finely shaped by the object slits, a set of highly polished stainless steel cylinders. A Faraday cup and beam viewer are located immediately after the object slits to optimise and align the beam as it exits the slits. A beam profile monitor, a Faraday cup and another beam viewer enable additional fine-tuning of the beam. [22]

An Oxford Microbeams magnetic quadrupole triplet raster scans the ion beam under the control of computer software. The final stage of the microprobe, the target chamber is an octagonal stainless steel vacuum chamber 165mm in internal diameter. 2 ports are located on each face, facilitating the attachment of electrical feed-throughs, gauges and detectors. A video camera enables the alignment of the target so that the ion beam may be scanned over a specific region. [22]

The target, in this case the silicon strip detector is connected to a multi-channel analyser (MCA) much like that used for conventional gamma spectroscopic measurements. From this equipment, the energy of an event may be derived. A determination of the co-ordinates of the beam at the detection of an event may be made through reading the current on the magnets at that moment. This enables the determination of both energetic and spatial information.

To order recorded information in a lossless format, a list mode file is used, such that recorded parameters are: a sequentially numbered event number, the x and y co-ordinates of the beam and the MCA channel number of the
detected event. If desired, the output from a second MCA may also be recorded (which otherwise is recorded as zero). This format enables the maximum amount of data to be extracted from measurements such as the ability to suppress low energy counts with a definable threshold post-acquisition, the creation of frequency histograms, in addition to mean and median energy maps.

The IBICC technique is particularly useful for semiconductor radiation detectors, where the response of the detector is not necessarily uniform across the sensitive region(s). This may be due to variations in material and geometry, or semiconductor physics governing the charge collection.

3.2 Analysis of Data

The author felt current methods for the analysis of IBICC data was inadequate, particularly in terms of quantitative analysis and the production of high-quality publishable images. A program was written in C++ to satisfy both of these demands.

The program takes text-converted list mode files as input and functions such as producing median energy maps with definable energy windows and outputting the spectral contributions to a charge collection image. These are discussed in more detail below.

3.2.1 Median Energy Maps

The program takes a filename as input, to direct the program to the location of an IBICC list mode file. Upon successful file reading, the program displays to the user both minimum and maximum ADC channel numbers present in the image, and requests lower and upper thresholds. An image is output in a text-image format containing the median energy produced from contributions within the window. This file format is both simple and versatile, consisting of a text array of median energy values, each element representing a single pixel of the IBICC image. This format may be simply read by image analysis software ImageJ, scientific plotting application gnuplot or numerical computing environment MATLAB.
The ability of the program to selectively perform median energy maps within defined windows is displayed below. Firstly, a median energy map is displayed containing no limits on the window.

Figure 30 - Median energy with a full window of an IBIC image of the silicon strip detector. Colour scale is in units of keV.

Figure 31 - Energy spectrum of IBICC measurement.
Figure 32 - Median images of the above data set, but with defined energy windows as displayed in each image.

### 3.3 Strip Detector IBICC Studies

Ion beam induced charge collection studies were performed on the SSD to determine the spatial dependence of charge collection under various bias conditions and position on the many devices.

#### 3.3.1 Calibration

A precision pulse generator was calibrated in the ANSTO detector characterisation laboratory using an alpha emitting source with multiple spectral peaks. A planar PIN diode was irradiated under vacuum, with the response of the detector recorded by a multi-channel analyser. The assumption of full energy absorption at the main spectral peaks was made. The pulser was calibrated such that its output at each pulse height setting matched the respective energy of detected spectral peaks.

Prior to each IBICC measurement at a particular amplifier configuration or bias, pulser measurements producing peaks at 0.2, 0.5, 1, 2 and 3 MeV were performed. By tabulating the MCA channel number corresponding to each pulser setting, calibration factors for each condition were determined through a linear line of best fit.
These calibration equations for each amplifier configuration and bias condition are applied to MCA channel numbers for each IBICC acquisition to yield energy in units of MeV. Since the energy of the incident ions are known, this enables a determination of charge collection efficiency to be made through comparison if the assumption of total energy deposition is made.
3.3.2 Neighbour Biasing Measurements

A single strip (termed the *read-out strip*) was connected to an Amptek A250 charge-sensitive pre-amplifier with the 4 most adjacent strips (termed *neighbouring strips*) without connection to any read-out electronics. A negative high voltage bias supply was connected to the common anode of the detector array with the cathodes of all neighbour strips connected to ground. The cathode of the read-out strip was also connected to a negative high voltage supply. By setting the power supply voltages to be equal in both magnitude and in sign, this enables the biasing of neighbouring strips whilst keeping the read-out strip unbiased.

Figure 35 - Circuit diagram of detector configuration for IBICC neighbour biasing studies.

The strip detector was raster scanned with a 3 MeV Helium\(^{2+}\) ion beam to determine the electrical response of a single strip. The strip measured and neighbouring strips were biased at 0 V to determine the unbiased response of the detector.
Figure 36 - IBICC image of the charge collection properties of a single strip grounded to 0V with neighbouring strips unbiased. Energy windows are displayed in each image segment.

Figure 37 - Energy spectrum of detector response.

The measured strip was retained at 0V, however neighbouring strips were biased at 30V to determine the effect this would have on the spatial extent of charge collection.
Figure 38 - IBICC image of the charge collection properties of a single strip grounded to 0V with neighbouring strips biased at 30V. Energy windows are displayed in each image segment.

Figure 39 - Energy spectrum of detector response.

The region of greatest charge collection occurs in two thin lines either side of the n+ strips. This corresponds to the region just past the aluminium strip
overlying the n+ strip. Due to the high LET of the alpha particle, large energy losses occur upon traversing minimal thicknesses of material, the region aforementioned is placed such that less material is traversed before the beam reaches the silicon bulk.

The next largest peak in charge collection efficiency occurs directly on the n-type strip region. Whilst this region requires a beam to traverse more material before reaching the silicon bulk region, the average length of charge carrier drift to collection regions is at minimum.

The application of a bias of 30 V to the neighbouring strips creates more confinement of the charge-sensitive region. This also has the effect of lessening the spectral feature at around 0.5 MeV, corresponding to the outer-most region of charge collection on the device, whilst minimally altering spectral characteristics of the central (most charge sensitive) region.

The application of reverse bias causes nearly all the applied voltage to appear across the depletion region. This is due to the resistivity of this region being much higher than normal doped silicon. As a consequence of Poisson’s equation, space charge must extend a greater distance either side of the detector. This has the effect of increasing the sensitive volume over which radiation produced charge carriers will be collected. [2] The application of the reverse voltage to adjacent detectors detector strips has the effect of extending the region over which charge deposited either side of the central strip is collected by the biased strips. It also has the effect of promoting drift towards the biased strips, not the unbiased central strip whose only voltage is the built-in intrinsic potential. Ultimately, this results in a reduction in the effective sensitive volume of the central, unbiased strip to radiation-induced charge deposition.

This experiment displays evidence that the possible incorporation of finger-like conductive strips between device strips may be an effective measure to reduce charge sharing. Such strips would act much like guard rings in other detectors, and be maintained at a constant potential with the common anode.
in a similar fashion to this experiment. It is expected this arrangement would result in greater charge confinement about the n+ region, and result in minimal charge sharing between strips. Consequently, the spatial resolving power of the detector should increase.
Chapter 4 – Clinical Irradiations

4.1 Orthovoltage X-ray Measurements

4.1.1 Overview
Using a Gulmay D3000 orthovoltage x-ray unit, measurements with various beam and beam shaping parameters were made with the silicon strip detector and read-out system. An orthovoltage unit was chosen since its low energy beam output is easily collimated with minimal thicknesses of high Z materials. However due to the short source-to-target distance, large beam divergence is one major disadvantage of this method.

4.1.2 Method
Firstly, measurements of the dark current in the room were taken by recording the counts in the absence of any generated radiation. Since the silicon strip detector is sensitive to light, and relatively insensitive to radiation; the bulk of detected dark current will have been derived from the room inadequately shielded from external light sources. After initial counts showed relatively high dark current, further measures were taken to minimise light, such as turning off the light in the control room and blocking light from door jams.

Firstly, a measurement was taken with the orthovoltage x-ray unit delivering a broad-beam radiation field. The strip detector was orientated perpendicular to the anode-cathode axis to negate the effects of the heel effect. From this data, faulty channels of the detector and readout system were identified as being significantly distant from the mean and removed from the dataset. The ratio of the mean to counts at each channel was determined and used as a correctional factor on subsequent measurements.
To simulate a microbeam, a precision micrometer-controlled tungsten collimator was employed. The 1μm precision enabled various beam width parameters to be set, namely 100μm, 200μm, 500μm, 1mm, 2mm and 10mm. However, due to beam divergence and scattering off collimator edges, the full-width at half maximum of the beams is expected to be wider than the width set.

Each set of data was calibrated by multiplying it by the correctional factor found from the broad beam irradiation, and then the average background count is subtracted from all channels. The resulting calibrated data is then graphed and analysed.

4.1.3 Results

The average background count was found to be 10.96 counts per second, relatively low given count rates of the order of 500 counts per second for peak values of irradiation data.
Figure 41 - Broadbeam irradiation dose profile.

Figure 42 - Correction factors versus strip number. Strips with no correction factor are faulty and excluded from further measurements.
Figure 43 - Calibrated dose profile for 10 mm field.

Figure 44 - Dose profile for 2 mm field.

Figure 45 - Dose profile for a 1 mm field.
Figure 46 - Dose profile for 500µm field.

Figure 47 - Dose profile for a 200µm field.

Figure 48 - Dose profile for a 100µm field.
The peak count at centroid was found for each field size and a linear interpolation performed to determine the full-width at half maximum (FWHM) for each collimator setting.

Table 4 - Peak counts and full-width at half maximum at various collimator widths.

<table>
<thead>
<tr>
<th>Collimator Width (cm)</th>
<th>Peak Count</th>
<th>FWHM (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>10067</td>
<td>0.30</td>
</tr>
<tr>
<td>0.2</td>
<td>21227</td>
<td>0.34</td>
</tr>
<tr>
<td>0.5</td>
<td>38261</td>
<td>0.56</td>
</tr>
<tr>
<td>1</td>
<td>48275</td>
<td>1.01</td>
</tr>
<tr>
<td>2</td>
<td>51401</td>
<td>2.08</td>
</tr>
<tr>
<td>10</td>
<td>53009</td>
<td>10.76</td>
</tr>
</tbody>
</table>

Figure 49 - Peak counts versus collimator width for varying settings.
4.1.4 Discussion

In an orthovoltage x-ray therapy unit, photons are generated through the interaction of accelerated electrons producing Bremsstrahlung when incident upon a high atomic-number anode. To enable the passage of radiation downwards towards the subject, the anode is angled. The position at which electrons interact with the anode gives rise to the heel effect. This is caused by photons produced with different angular distributions being more readily absorbed by the anode target. Since photons composing the field more proximal to the anode have a higher probability of a longer path length through the anode, the intensity in this region is decreased. This causes non-uniformity along the anode-cathode axis; however filters are in place to counter such heterogeneity.

The broad beam irradiation method for calibrating each channel intrinsically relies on the radiation field incident upon the detector being uniform. Ideally, an isotope flood field would give the most accurate results, with a cosine correction factor for slightly different source-to-detector distances. However this introduces challenges in the handling of high-activity sources purely for the purpose of calibration. GAFchromic film measurements of the field non-
uniformity for the ICCC Gulmay G3000 unit over the central region along the cross-axis are less than 1%, making the method utilised valid with minimal systematic error.

Broadening of the FWHM when compared to the set collimator width are consequences of the beam divergence present from x-ray tube derived radiation fields. Additionally, scatter of the beam off collimator edges and scatter internal to the detector will have resulted in effective beam broadening. Since the charge collection efficiency of the detector is at its greatest over the centre of an aluminium strip, representing a width of 10µm, deviations of the centre of the beam from this position will induce effects causing incomplete detection of a peak.

4.1.5 Conclusion
The ability of the strip detector to spatially resolve the dose profile of fields with minimal width was displayed. However, better calibration methods are required to ensure that errors derived from different responses of the detector and readout system do not compromise readout quality. The strip detector appears to have a spatial resolution of 500µm in this configuration, since the FWHM is not significantly wider than the collimator width. However, this does not take into account collimator-derived scatter.
4.2 Linear Accelerator Measurements

4.2.1 Overview

A Varian 2100C linear accelerator was used to irradiate a high resistivity silicon strip detector. Different set-up parameters were investigated, such as depth in a phantom, dose rate and field size with the silicon strip detector.

4.2.2 Method

The Perspex phantom around the strip detector, in addition to solid water was used to simulate the placement of the SSD in tissue. Firstly an irradiation was performed with a $10 \times 10$ cm field with the SSD at varying depths within the phantom. The surface of the phantom was kept at a constant 100cm source-to-surface distance through treatment couch movement.

![Figure 51](image)

*Figure 51 - The SSD and readout system set up on the treatment couch of a Varian 2100C linear accelerator.*

Second sets of irradiations were performed with the SSD at a constant depth of 10cm, but with dose rates of 90, 270, 400 and 470 monitor units per
minute. This provides a quantitative basis to assess the linearity of the SSD to varying dose rates.

A 10 × 10 mm field was employed to assess the response of the SSD to high dose gradients present in the penumbral region. An asymmetric field was also defined of the same dimensions, however offset to one side by 1 cm. The flattening filter of a linac is designed such that a flat beam profile will be present at a depth of 10 cm in tissue for 6MV photons. However, at superficial depths, horns arise at either side of the umbral region caused by a greater contribution to spectral composition by lower energy photons at the outer regions of the beam.

4.2.3 Results

![Graph showing SSD depth dose results for a 10 × 10 cm 6MV photon beam. Reference data from Metcalfe, Kron and Hoban [4].](image)

The SSD and readout system give excellent agreement with the reference data for a 6 MV Varian Linac across the range of clinically relevant depths.
Interestingly, a quadratic relation between dose rate and charge collected per strip were observed. 

The penumbral characteristics of a 10 × 10 mm field as measured by the SSD and read-out system. Some faulty channels are removed from the dataset to clarify the data.

4.2.4 Discussion
To deliver dose to the patient, a linear accelerator delivers a very high dose in very short pulses separated by a much larger period of zero dose. Typically,
the pulse timing over the treatment time would result in a larger dose than specified. To ensure the dose delivered is as specified, the linac drops pulses, regulating the total dose. This control method had to be disabled for the irradiations performed for this experiment to ensure the read-out system was not recording an irradiation during a dropped pulse, resulting in erroneous data.

The SSD is surrounded by a custom Perspex phantom to interface the irregular geometry with other phantoms such as solid water to simulate different depth parameters. Since the electron density is different to that of solid water; the heterogeneity with depth may have unwanted edge effects. In addition, the Perspex phantom is not in physical contact with the strips to prevent the possibility of damage to the delicate detector. The presence of air in this interface may induce build-up effects in the SSD surface related to a lack of charged particle equilibrium.

In addition, the presence of the electrical connectors in close proximity to the field may induce scatter into the sensitive volume. Additionally, their high atomic numbers would result in much higher interaction probabilities than for the tissue equivalent and Perspex phantoms.

A quadratic relation between dose rate and charge per strip was noted over the range of dose rates the detector was exposed to. These phenomena may be explained through the instantaneous dose rate dependence of solid-state diode detectors as presented by Shi, et al. [23]. A fraction of excess minority carriers, which are generated within the detector, are captured by recombination-generation centres and recombined with majority carriers. This fraction is dependent on the excess minority-carrier concentration and the recombination-generation centre concentration. As the dose rate increases, the recombination-generation centre concentration may not be sufficient to keep the recombination fraction constant. This has the effect of increasing the sensitivity of the detector due to a larger fraction of charge being collected within the detector. [23]
4.2.5 Conclusion

The silicon strip detector and read-out system gave excellent agreement with reference dosimetry depth dose data for the Varian 2100C linear accelerator. A quadratic relation between dose rate and charge per strip however was discovered.
Chapter 5 – Discussion

The microbeam radiation therapy spectrum as used in GEANT4 simulations was defined in a point-wise manner, namely in 1 keV steps. To minimise the effect such a primary photon spectrum would have on simulation results, linear interpolation of the input spectrum is performed between points.

It should be noted that in all GEANT4 simulations, the microbeam was simulated as being rectangular, with a homogeneous flux of photons across the entire profile. In addition, there was no angular distribution of photons, with the beam consisting entirely of photons normal to the detector’s surface.

As with any collimated beam, scatter is present due to the interaction of incident photons with the collimator. This can result in the production of electrons and scattered photons of lower energy, contributing to unwanted dose, particularly in the superficial and valley regions.

Since the beam itself is produced through a set of two multi-slit collimators, the slight angular distribution of the synchrotron beam results in variations in the beam profiles on one side of the treatment field as opposed to the other. This phenomenon is described in detail by Braüer-Krisch et al [24] and is illustrated in Figure 55.

![Figure 55 - The asymmetry of beam collimation caused through the use of two multi-slit collimators.](image)

Whilst a synchrotron beam has low divergence properties, particularly due to the large source-to-collimator distance, the narrow pitch and width of slits
ensures such minimal divergence does in fact affect beam profiles. Photons produced with an angular distribution towards the left may be transmitted, whilst photons with the exact opposite angular distribution on the right side are absorbed in the collimator.

The assumption inherent in the planar microbeam array simulations is that the superposition of one microbeam’s interaction properties are equally valid for all other microbeams assuming an interaction medium quasi-infinite in lateral dimensions. In reality, the use of a variable width collimator system creates a gradient of variation of microbeam width with position. Additionally, microbeams at the edge of the array have lower dose contribution by neighbouring microbeams due to having less proximal neighbours.
Considerations of such asymmetry phenomenon must be taken into account when the comparison of experimental data is made with theoretical data implementing superposition methods. Due to complexities inherent in modelling technical aspects of the beam line, the irradiation geometry was greatly simplified to enhance computational efficiency. It was decided that since the primary objective was to assess the effect of the detector on primarily the spectral energies, that advanced simulations were not required for this work.

Since the IBICC studies discussed in this thesis use alpha particles as a means of spatially delivering charge into the detector, the results are not directly transferable to photons. Due to the high LET of the alpha particle and the property of ions in matter, the alpha particle releases all kinetic energy up to a finite depth in the detector. The projected range of a 3 MeV alpha particle is 11.9 μm [NIST Nuclear] in silicon; since the thickness of the silicon layer of the detector is 375 μm, this results in a large portion of the detector not having involvement in energy deposition during IBICC studies.
Monte Carlo simulations into certain properties of radiation traversing the detector could be made more accurate with the use of larger primary photon numbers. However time and resources available limited the scope and depth of this thesis.
Chapter 6 - Conclusions and Future Work

The energy deposition properties within the detector were determined through the use of Monte Carlo simulations. In particular, by the spectral energies deposited by the ESRF MRT beam. It was found, largely due to the low interaction probability of soft x-rays with the detector, that energy deposition is largely constant across the detector, with the exception of edge effects present at 5 µm depth from the entrance and exit surfaces. However this may simply be attributed to a lack of charged particle equilibrium present in these regions.

In addition, the perturbative effect that an in-beam strip detector has on the MRT beam was determined both in terms of spectral changes, and in terms of variations in depth dose recorded comparatively to the absence of the strip detector.

It was found that the detector causes a $1.41 \pm 0.03 \%$ decrease in peak dose for a simulated microbeam array. However, the effect the detector had on the peak-to-valley dose ratio was unable to be determined due to an inadequate number of primary photons simulated resulting in large uncertainties in valley dose. It is anticipated that future work will improve upon these results; yielding statistically relevant data as to the effect the SSD has on microbeam arrays.

 Ion beam induced charge collection studies revealed the positive effect that biasing neighbouring strips has on charge collection confinement. The integration of biasing strips on future generations of SSD’s would most probably act to improve the device, however TCAD modelling should be performed to ensure this idea works at a smaller scale and that the design is optimised.

Additionally, to obtain a better understanding of how the strip detectors are affected under the high dose rates and integral doses experienced in the synchrotron radiation field for MRT, more IBICC studies should be performed.
on such detectors. Experiments to determine the charge collection properties resulting from additional biasing conditions could also be performed in the future.

Finally, clinical irradiations of the strip detector demonstrated the SSD's ability to spatially resolve high dose gradients in a modern clinical setting. In particular, to resolve very small fields with a minimum penumbral width of 500µm, and to resolve larger fields with sharp dose gradients.

It is anticipated that future work will further characterise the effect the detector has on a traversing radiation field, specifically, experimental determinations of the SSD's perturbation of MRT beams. Such work, along with improvements to the detector and the read-out system will advance towards the use of the SSD as an online dosimeter for MRT.
References


