TOWARDS SILICON MICRODOSIMETRY BASED VERIFICATION OF MONTE CARLO CALCULATIONS IN HADRON THERAPY

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from

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by

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CERTIFICATION

I, Iwan M. Cornelius, declare that this thesis, submitted in partial fulfilment of the requirements for the award of Doctor of Philosophy, in the Department of Engineering Physics, University of Wollongong, is wholly my own work unless otherwise referenced or acknowledged. The document has not been submitted for qualifications at any other academic institution.

Iwan M. Cornelius

June 2, 2004
This thesis is dedicated to my family, my friends, and my teachers.
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ABSTRACT

This thesis continues research into the application of silicon microdosimetry to hadron therapy applications. It proposes the use of silicon microdosimetry for the verification of Monte Carlo calculations in hadron therapy. When applied in this manner, many of the restrictions previously impeding the use of silicon microdosimeters in hadron therapy are relaxed.

An ion microprobe was used to study the charge collection properties of the silicon microdosimeters using ions commonly found in the primary and secondary radiation of hadron therapy. An experimental setup was developed and diagnostic studies were conducted to establish low beam fluence and micron beam resolution necessary for the measurements. GEANT4 Monte Carlo simulations of the measurements facilitated the quantification of the charge collection efficiency of the devices for ions with various Linear Energy Transfer values.

An ion beam analysis technique was developed to measure the collection time of charge carriers generated following ion strikes on the silicon microdosimeter. The dependence of this charge collection time was studied using the ion microprobe. A pulse shape discrimination technique was then implemented in an attempt to improve the spectral response of the microdosimeter.

Simple Monte Carlo simulations of silicon microdosimetry measurements in Fast Neutron Therapy and Proton Therapy were conducted using the GEANT4 Monte Carlo toolkit. The charge collection efficiency information obtained from microprobe experiments was incorporated into the simulation. Discrepancies between experimental and theoretical measurements was then used to suggest improvements to the simulations.

Future recommendations for the application of silicon microdosimeters in this capacity are discussed along with suggestions for other silicon based instrumentation.
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CHAPTER 1
INTRODUCTION

1.1 Interaction of Radiation with Matter

The section provides a brief discussion of the interaction of radiation with matter. It concentrates on photons, electrons, and energetic ions with energies relevant to hadron therapy.

1.1.1 Photons

For photon energies encountered in hadron therapy (X-rays and $\gamma$-rays), there are three major types of interactions whereby the photon transfers part or all of it’s energy to atomic electrons. These are: Compton scattering, photoelectric absorption, and pair production. During photoelectric absorption the photon loses it’s entire energy to the ionisation of a bound atomic electron. The vacancy produced in the ionised atom is filled by the capture of electrons from the medium or by re-arrangement of electrons within the atom, both leading to the emission of characteristic X-rays. During Compton Scattering the photon transfers a fraction of it’s energy to the atomic electron and is scattered through an angle $\theta$, the target electron recoils with an energy related to $\theta$. Pair production occurs in the presence of the electric field of a target nucleus and is energetically possible if the photon energy is twice the rest energy of the electron. All the energy greater than twice the rest energy of the electron is shared as kinetic energy of the electron and positron. The relative probability of the above interactions is a function of the target atomic number and the energy of the incoming photon [6].

1.1.2 Electrons

High energy electrons and positrons interact with matter via the ionisation interaction, bremsstrahlung production, and annihilation. Electrons may interact via the
coulomb force with atomic electrons, imparting sufficient energy to ionise the target atom. Bremsstrahlung radiation is created when the momentum of an electron changes in the electric field of the nucleus and a photon is emitted to conserve momentum. Annihilation occurs when a positron interacts with an atomic electron to produce two 511 keV annihilation photons.

1.1.3 Protons and Energetic Ions

Protons and heavier ions interact with matter predominantly via ionisation of atomic electrons, coulomb scattering on nuclei, and to a lesser extent inelastic scattering on nuclei. As these particles traverse a medium they interact with atomic electrons via the coulomb force, a process known as electronic stopping. Depending on the medium and distance of the electron from the ion path this interaction may excite the atom by raising the electron to a higher energy state or ionise the atom by transferring sufficient energy to the electron to overcome the binding energy of the nucleus. Secondary electrons produced subsequently interact with the medium by the processes described above. The maximum energy transfer in ion-electron interactions is $\frac{4m_eE_p}{M_p} \approx \frac{E_p}{500}$ of the particle energy per nucleon. Consequently the ion must undergo many such interactions prior to losing all it’s kinetic energy and coming to rest in the medium. As the velocity of the particle decreases, the time spent by electrons in the presence of it’s electric field increases and ionisation is greater. At still lower energies, charge exchange occurs where the ion picks up electrons and the effective charge decreases, reducing the number of coulomb interactions in the material. This results in the so called bragg peak of energy loss as a function of depth of penetration in the target. The ion may also transfer energy to the nuclei of the target material via elastic Coulomb collisions; referred to as nuclear stopping. The importance of nuclear stopping is negligible at high energies but is of the same order of magnitude as electronic stopping for very slow ions ($E < 1$ keV). An ion with sufficient energy may surmount the coulomb barrier of the target nucleus. In this situation inelastic reactions
may occur (see discussed below), resulting in the production of products.

1.1.4 Neutrons

Neutrons, being neutral particles, do not interact with a medium via the coulomb force. The neutron instead interacts with the nuclei of target atoms via interactions which fall into three categories: elastic scattering, inelastic scattering, and radiative capture. During elastic scatter interactions the incoming neutron may be scattered by the nuclear field, which is a potential scattering (shape elastic scattering), or captured to form a compound nucleus and emitted with the same energy (compound elastic scattering). In both cases the neutron transfers a fraction of its energy to the target nucleus (assumed to be at rest). The amount of energy transferred depends on the masses of the neutron and target nucleus and the angle of recoil. Inelastic scattering occurs when the recoiling target nucleus is elevated to an excited state during the collision. The nucleus then relaxes via emission of gamma photons, neutrons, and/or charged particles. Capture reactions dominate at low neutron energies and represent reactions which have a positive Q value. The capture of a neutron by the nucleus results in an excitation energy equal to the sum of the channel energy and the neutron binding energy. The excited nucleus relaxes via the emission of gammas, neutrons, and/or charged particles. Hence neutrons deposit energy in a medium indirectly via the production of secondary charged particles.

1.2 The Effects of Ionising Radiation on Biological Cells

When biological cells are exposed to ionising radiation a number of physical, chemical and biological changes occur. These changes are either a direct result of the ionisation of molecules in the cell or an indirect result of free radicals formed through ionisation interactions with water [1]. The biological changes may be due to the alteration of the cellular DNA, RNA, enzymes, and metabolism. As DNA is considered the key molecule for regulation of cell growth and differentiation it is considered to be the most important target molecule in the cell. The dominant mechanism for cell death is believed to be the
induction of double strand breaks (DSBs) in the DNA; however our understanding of the fundamental mechanisms behind cellular radiation response is far from complete. Recent advances in the field are discussed at such symposia as the International Congress on Radiation Research, the proceedings of which are published in the corresponding journal entitled Radiation Research.

1.3 Radiobiological Models

A number of radiobiological models have been proposed to explain biological effects of ionising radiation. What follows is a brief introduction to several models; an emphasis is placed on those specific to densely ionising radiation.

1.3.1 Linear Quadratic Model

The linear quadratic model is based on the experimentally observable quantity absorbed dose which is defined as the energy $E$, absorbed by a mass $m$;

$$ D = \frac{E}{m} = \frac{n}{V} \frac{W}{\rho} $$

(1.1)

Where $n$ is the number of ionisations in a volume $V$ of material with mass density $\rho$ and $W$ is the average energy required for each ionisation. Given an initial population of $N_o$ cells, the surviving fraction after an absorbed dose $D$ is given by:

$$ \frac{N}{N_o} = \exp \left( - (\alpha D + \beta D^2) \right) $$

(1.2)

The constants $\alpha$ and $\beta$ of equation (1.2) are determined for a particular cell line through the irradiation of clonogenic assays. A clonogenic assay is a technique for studying the effectiveness of radiation on the proliferation of cells. The term "clonogenic" refers to the fact that these cells are clones of one another. The experiment involves three major steps: the cells to be studied are "plated", an absorbed dose of ionising radiation is applied, the cells are then fixed and stained. At the conclusion of the experiment the percentage of cells which survived the treatment are counted. This experiment is repeated
for various absorbed doses of ionising radiation and percentage cell survival is plotted as a function of absorbed dose. Equation 1.2 is then fitted to the experimental data by adjusting the parameters $\alpha$ and $\beta$. A modified form of equation 1.2 exists to take into account the capacity of the cell to repair DNA damage [7]. The response of cells in vivo however, depends on the radiosensitivity of particular cell types and the complicated biochemistry and physiology of organs. In photon therapy, values of $\frac{\alpha}{\beta} = 3$ and $\frac{\alpha}{\beta} = 10$ are used for early responding tissue (marrow and gonads) and late responding tissue (kidney) respectively. The choice of these values stems from years of clinical experience.

Another model for survival following gamma irradiation is the single hit, multi-target model [8]. In this case the surviving fraction is given by:

$$\frac{N}{N_0} = 1 - (1 - \exp(-\frac{D}{D_0}))^m$$

(1.3)

Where $D_0$ is the critical absorbed dose and $m$ is the multiplicity of sub-cellular sites. These parameters are again determined by fitting to experimental cell survival curves.

The linear quadratic model may be used to accurately predict cell survival given the values of $\alpha$ and $\beta$ for a particular cell line and a measurement of absorbed dose. This however cannot be extended to predict the biological effect following an irradiation of densely ionising charged particles. In contrast to the situation for weakly ionising radiation, where the distribution of ionising is approximately random, following irradiation by charged particles ionisation is concentrated around the trajectories of the ions. In this case the values of $\alpha$ and $\beta$ are related to the composition of the charged particle field. Figure 1.1 shows the cell survival curves for alpha particles of varying energy and LET. The Linear Energy Transfer (LET) of an ion is the average energy lost in the ionisation and excitation of atomic electrons per unit path length. One sees that for alpha particles the surviving fraction of cells is much less than for X-rays for an identical absorbed dose. Moreover the biological effectiveness increases with ion LET. A quantity known as the
Figure 1.1: Survival of cultured cells derived from human kidney; Curve 1: 5.2 MeV alphas (LET 85.5 keV/μm), Curve 2: 8.3 MeV alphas (LET 60.8 keV/μm), Curve 3: 26.8 MeV alphas (LET 24.6 keV/μm), Curve 7: 200 kV X-rays (LET 2.5 keV/μm) (from reference [1]).

Relative Biological Effectiveness (RBE) was coined to describe this increased effectiveness for cell killing for high LET radiations. RBE is defined as the ratio between an absorbed dose of reference radiation (usually Co-60) and an absorbed dose of high LET radiation X resulting in the same biological endpoint;

\[
\text{RBE}_X = \frac{D_{\text{Co-60}}}{D_X} \tag{1.4}
\]

If an absorbed dose of radiation X is measured, it may then be scaled by the RBE\textsubscript{X} to obtain an equivalent absorbed dose of gamma radiation. Several radiobiological models have been proposed to calculate RBE values. These are based on the measurement of an observable quantity other than absorbed dose.
1.3.2 Microdosimetry

Microdosimetry, pioneered by the research of Kellerer and Rossi [9], was proposed in the 1960s to explain the biological effect of densely ionising radiation. Reference [10] provides a detailed review of microdosimetry. It assumes the sensitive structure is the microscopic cell nucleus; experimental microdosimetry is thus based on the measurement of a quantity that is indicative of total energy deposition on the microscopic scale. This quantity is lineal energy\(^1\) or the quotient of the total energy deposited \(\epsilon\) in a microscopic sensitive volume with the mean chord length \(< l >\) of the volume [10]:

\[
y = \frac{\epsilon}{< l >} \tag{1.5}
\]

The lineal energy is a stochastic quantity and as such possesses a probability distribution \(f(y)\). This distribution depends upon the radiation type and the geometry and composition of the sensitive volume and surrounding materials. \(f(y)\) is conventionally weighted by the lineal energy to form a dose weighted lineal energy distribution;

\[
d(y) = \frac{yf(y)}{y_f} \tag{1.6}
\]

where \(y_f\) is the frequency mean lineal energy;

\[
y_f = \int_0^\infty yf(y)dy \tag{1.7}
\]

This distribution gives the fraction of total absorbed dose delivered to the microscopic sensitive volume by lineal energy events in the interval \(y, y + dy\).

A quality factor (synonymous with RBE) may be derived from the irradiation of cells with mono-energetic ions producing a narrowly peaked lineal energy distribution of mean value \(y\). There are several formulations of this quality factor outlined in reference [2]. The equivalent absorbed dose at the point of measurement may be found by integrating the product of the dose distribution and the quality factor of the radiation;

\(^1\)Lineal energy and the Linear Energy Transfer, although having the same units, are different quantities. Lineal energy is a stochastic quantity whilst the Linear Energy Transfer is an average quantity.
Figure 1.2: Microdosimetric distributions for three neutron therapy beams produced by 65 MeV proton (p65), 4 MeV deuteron (d4), and 20 MeV deuteron (d20) irradiation of Beryllium targets (from reference [2]).

\[ H = \int_0^\infty d(y)Q(y)dy \]  

(1.8)

In a practical situation lineal energy events can span several orders of magnitude so it is customary to display the microdosimetry spectrum with a logarithmic abscissa. In order to maintain the dose to area relationship of the distribution it is necessary to scale it by the lineal energy event. The distribution is then represented as:

\[ yd(y) \text{ vs } d(\log y) \]  

(1.9)

Displaying results in this form allows a visual comparison of the radiobiological effectiveness of different radiation fields. Figure 1.2 shows an example of microdosimetry spectra for three different neutron radiotherapy facilities.

Instrumentation for measurements of microdosimetry spectra are predominantly (although not exclusively) based on the Tissue Equivalent Proportional Counter (TEPC). This device consists of a millimetre or centimetre sized proportional counter filled with a low density, tissue equivalent gas. The density of gas is chosen such that the energy deposited in the sensitive volume by a charged particle is equivalent to that for a solid,
micron diameter volume of tissue.

1.3.3 Katz Theory

Katz theory is a radiobiological model which relates cell survival following gamma irradiation to cell survival following irradiation by ions of atomic number $Z$ and energy $E$ \[11\]. It assumes the response of cells to secondary electrons from gamma irradiation is identical to that from the secondary electrons liberated by ions. If a single ion trajectory is considered, the probability of cell death will logically decrease with increasing radial distance from the ion path owing to a decrease in secondary electron fluence. The theory considers a series of concentric cylindrical shells with axis along the ion trajectory and assumes ion-nucleus scatter interactions may be ignored such that the ion trajectory is unperturbed. The number of ionisations in each shell for a single ion traversal is a statistical variable due to the stochastic nature of secondary electron production. A large number of ion traversals results in an average absorbed dose, $\bar{D}(t)$, to the shell at radial distance $t$ from the track core. The use of absorbed dose in this situation is valid as the electron fluence through any shell, on average, is approximately homogeneous.

Assuming the absorbed dose dependence of cell survival may be accurately described by the single hit multi target model, the probability $P(t)$ of cell death at a radial distance $t$ is:

$$P(t) = (1 - \exp\left(-\frac{\bar{D}(t)}{D_o}\right))^m$$ \hspace{1cm} (1.10)

Where $D_o$ and $m$ are the critical absorbed dose and multiplicity of sub-cellular sites. These parameters are found by fitting to gamma irradiation experiments \[8\]. The probability of cell death in a cylindrical shell at radial distance $t$ is then $P(t)2\pi td\tau$. This function is integrated from a radius of zero to the maximum range of secondary electrons $R_{max}$ to obtain the cross section \[12\] for cell death;

$$\sigma = \int_0^{R_{max}} P(t)2\pi td\tau$$ \hspace{1cm} (1.11)
In calculating the cross section for cell death one must determine the radial distribution of absorbed dose around the ion path. This quantity is not easily measurable and a number of analytical and Monte Carlo calculations have been developed [13].

For a fluence $\phi(E, Z)$ of ions of nuclear charge $Z$ and energy $E$, the incremental change in the number of surviving cells for an initial population $n$ is given by:

$$dn = -n \sigma(E, Z) d\phi(E, Z)$$

(1.12)

The surviving fraction as a function of fluence is then given by:

$$n = n_o \exp (-\sigma \phi)$$

(1.13)

This equation is applicable for low ion fluences. In this case the majority of cells are killed from ion traversal through or adjacent to the cell. This form of cell killing is known as “ion kill” and the probability of cell survival is given by:

$$\Pi_i = n / n_o = \exp(-\sigma \phi)$$

(1.14)

At high fluence, a significant amount of cell killing arises from the overlap of secondary electron trajectories originating from separate ion tracks. This mechanism of cell inactivation is termed “gamma kill” as it is described in the same manner as for gamma irradiation.

In a population of cells subject to heavy ion irradiation there is a contribution to cell death from both ion and gamma kill. The fraction of inactivated cells in the ion kill mode is assumed to be identical to the fraction of absorbed dose deposited in the ion kill mode such that the absorbed dose in the gamma kill mode is given by $D_\gamma = (1 - \Pi_i)\phi L$ where $\phi$ is ion fluence, $L$ is the stopping power of the ion and $\Pi_i$ is the probability of ion kill [8]. The survivors of ion kill from equation [1.17] are considered to be the initial population for gamma kill. Again, the gamma response is described by the single hit, multi target model and the probability of survival in the gamma kill mode is:
\[ \Pi_i = 1 - \left(1 - \exp\left(-\frac{D_i}{D_o}\right)\right)^n \]  

(1.15)

The surviving fraction of cells is then given by the product:

\[ \frac{N}{N_o} = \Pi_i \times \Pi_\gamma \]  

(1.16)

In a mixed radiation field, the surviving fraction of a population of cells is given by equation [1.16] where the contribution of the jth component of the radiation field is considered in ion and gamma kill modes as:

\[ \Pi_i = \exp\left(-\Sigma_j \sigma\phi\right) \]  

(1.17)

\[ \Pi_\gamma = 1 - \left(1 - \exp\left(-\frac{\Sigma_j D_\gamma}{D_o}\right)\right)^m \]  

(1.18)

In this manner the cell survival is calculated based on a knowledge of the fluence of components of the radiation field. A similar model was developed by the researchers at GSI [14] which has been applied to treatment planning in carbon ion therapy.

1.3.4 Nanodosimetry

Nanodosimetry is a radiobiological model which considers the sensitive structure to be a nanometer DSB volume. A nanodosimeter measures the number of ionisations in a volume of nanometer size per incident particle [15]. A Two-Compartment theory assumes two types of DNA damage: repairable damage resulting from 2-5 ionisations and irreparable damage from 6-10 ionisations. For ionisations greater than 10 the biological effect is assumed to be reduced as a result of charge and radical recombination.

Reparable damage may result in successful repair or permanent damage from unsuccessful repair. The induction of lethal damage from repairable multiple damage sites may be non linear with absorbed dose and is described by a second order polynomial. The irreparable multiple damage sites on the other hand are assumed to increase linearly with absorbed dose. Assuming a Poisson distribution of lethal damages the surviving
fraction of cells is given by:

\[ \frac{N}{N_0} = \exp[-(\alpha_1 q_1 D + \beta q_1^2 D^2) - \alpha_2 q_2 D] \]  

(1.19)

where D is absorbed dose and the radiation quality factors \( q_1 \) and \( q_2 \) are defined relative to a reference radiation as:

\[ q_1 = \frac{\sum_{j=2}^{j=5} P(j)}{\sum_{j=2}^{j=5} P_{\text{ref}}(j)} \]  

(1.20)

\[ q_2 = \frac{\sum_{j=6}^{j=10} P(j)}{\sum_{j=6}^{j=10} P_{\text{ref}}(j)} \]  

(1.21)

where \( P(j) \) and \( P_{\text{ref}}(j) \) are probabilities of \( j \) ionisations in the sensitive volume per unit absorbed dose or particle fluence. The particle dependence of cell survival is related to the quality factors \( q_1 \) and \( q_2 \). \( \alpha_1, \alpha_2 \) and \( \beta \) are the cell specific parameters which are assumed to be independent of particle type and energy.

This section has summarised several radiobiological models and is by no means exhaustive. Other radiobiological models for densely ionising radiation have been developed (see references [16] [17] and references therein). Each model relies upon the measurement of a characteristic of the densely ionising radiation field. In the case of microdosimetry this is the spectrum of ionisation energy deposition events in a microscopic, tissue equivalent volume. Similarly nanodosimetry considers the sensitive volume to have nanometer dimensions. Finally, Katz theory (as for Kraft-Scholz theory) requires measurement of the energy spectra of all charged particles. It is difficult to quantitatively compare the accuracy of these models as all rely upon parameters derived from cell survival experiments that are subject to large errors [18] [19]. Moreover this data is solely based on in vitro experiments. Extrapolating this to in vivo is complicated by many physiological factors. Nonetheless the development of radiobiological models is necessary for accurately predicting the biological effects of densely ionising radiation.
1.4 Hadron Therapy

Hadron therapy is a term which covers several cancer treatment modalities which utilise controlled beams of hadrons. These include: boron-neutron capture therapy, fast neutron therapy, proton therapy, and heavy ion therapy.

1.4.1 Fast Neutron Therapy

Fast Neutron Therapy (FNT) is a form of cancer treatment whereby the patient is irradiated with a collimated beam of neutrons with energies $E < 60 \text{ MeV}$. The beam is typically produced by bombarding a beryllium target with protons or deuterons. The neutron beam is physically shaped with various types of pre-collimators and collimators manufactured from iron, steel, concrete, and tungsten. The energy spectrum of the beam may be adjusted by including filters. The beam delivery system is mounted on an isocentric gantry which can rotate around the patient as shown in figure 1.3.

As discussed in section 1.1.4, neutrons of these energies interact with tissue predominantly through elastic and inelastic scattering and neutron capture reactions. The biological effect is delivered by the resulting densely ionising reaction products with a contribution from gamma radiation originating from neutron interaction in the beam.
modifying devices. By manipulating angles of irradiation and the field of irradiation, the biological effect in the tumour volume may be maximised whilst minimising damage to surrounding healthy tissue.

FNT has several advantages over conventional photon therapy. These are due to the existence of hypoxic cells in malignant tumours and a resistance of these cells to irradiation by photons. The densely ionising radiation results in a decrease in the oxygen enhancement ratio and the dependency on cell cycle [20]. There is also a reduction in the differences in radiosensitivity between cell populations, related to position in the mitotic cell cycle, and between cell lines. In terms of the absorbed dose distribution however, the modality offers no advantages over conventional radiation therapy.

1.4.2 Proton Therapy

Proton therapy (PT) is a form of cancer treatment whereby the patient is irradiated with collimated beams of protons with energies of approximately $200 \text{ MeV}$ for deep seated tumours and approximately $60 \text{ MeV}$ for ocular tumours. The advantages of proton therapy stem from the fact that protons lose the majority of their kinetic energy to the ionisation of atomic electrons close to the end of their range. This is evident in the “Bragg peak” of the absorbed dose distribution as shown in figure 1.4. This is in contrast to the absorbed dose distribution offered by conventional X-rays therapy that peaks several centimetres from the patient surface.

There are two methods of beam delivery currently used in proton therapy: passive, and dynamic mode. In passive mode, energy modulation is achieved by passing the beam through absorbing material of variable thickness; in dynamic mode this is done by tuning the synchrotron energy. In active mode the lateral dimensions of the beam are modulated by magnetic deflection; in passive mode a broad beam is produced by passing the beam through a scattering medium and a collimation system is used to define the lateral dimensions of the beam. Developments in the field of Proton Therapy are discussed at meetings of the Particle Therapy Co-Operative Group (PTCOG).
Figure 1.4: Absorbed dose distributions for a mono-energetic proton beam, a modulated proton beam, and X-rays in water (taken from www.iucf.indiana.edu).

The advantages of proton therapy over conventional X-ray therapy are a direct result of the localisation of absorbed dose to the tumour volume. As the protons are not considered to be high LET particles the radiobiological advantages obtained with FNT are not as pronounced for proton therapy.

1.5 Treatment Planning

1.5.1 Conventional Treatment Planning

The role of a treatment planning system is to calculate the configuration of beam energies, angles of incidence and lateral field shapes which minimises cell survival in the tumour volume and maximises cell survival in surrounding healthy tissue.

Figure 1.5 shows a flow diagram outlining a conventional treatment plan. In this situation patient data is obtained from computed tomography to identify the tumour volume. For a beam of known energy distribution and field shape the absorbed dose distribution in the patient is calculated by using a simple analytical model which accounts for ionisation
of atomic electrons and implements analytical corrections to account for lateral spread of the beam with depth due to scattering interactions. This may be carried out for several angles of irradiation and the resulting absorbed dose distribution compared to the optimal absorbed dose distribution. Discrepancies between the two are fed back to the treatment planning system to adjust the primary beam characteristics. This process is repeated until the planned dose distribution agrees with the optimal distribution within some limit. Figure 1.6 shows the result of such a treatment plan for protons and compares with that obtained from an X-ray treatment plan.

1.5.2 Monte Carlo Based Treatment Planning Including RBE Variations

Analytical models used in conventional treatment plans can become inaccurate when applied to inhomogeneous geometries. An alternative, albeit computationally more expensive method is to utilise Monte Carlo calculations to transport each individual particle through the patient geometry and calculate the effect of secondary particles on the target volume. Conventionally a treatment plan should be calculated in a reasonable time on a personal computer; however, advances in parallel computing are making Monte
Carlo based treatment planning systems feasible. Parallel computing involves the use of a cluster of computers, connected by a network, to behave as a single computing entity. Monte Carlo simulations are ideal for parallel computing as identical simulations can be executed on separate machines using unique seeds for the random number generators of each machine. If the Monte Carlo calculation is parallelised, the computation time simply scales with the number of computers in the cluster. In comparison to the cost of an accelerator gantry or accelerator, a cluster of a hundred computers is negligible. This added cost is also justified by the potential increase in the accuracy of treatment plans. An example of such a system is the PEREGRINE initiative at Lawrence Livermore National Laboratory [21]. Without doubt, Monte Carlo tools will provide the basis for treatment planning systems in future; whether this be using small clusters at facilities, or global grid networks.

Monte Carlo offers an opportunity to model in detail radiation transport in the patient. As mentioned in section 1.3.1 the quantity absorbed dose is not an accurate predictor of cell survival for densely ionising radiation. The increased accuracy gained by using Monte Carlo calculations should be complimented by using accurate radiobiological models as described in section 1.3. Current treatment planning systems in proton therapy are based on physical absorbed dose weighted by an RBE of 1.1. Several studies have indicated the importance of a varying RBE in proton therapy [22, 23, 24, 25] and
fast neutron therapy [26]. If absorbed dose is to be used for treatment planning, additional information is required to scale it by an RBE value. The nature of this additional information would depend upon the radiobiological model used. For instance, the lineal energy spectrum is required for microdosimetry, a nanodosimetry spectrum is required for nanodosimetry, and the energy spectrum of all primary and secondaries for Katz theory and other fluence based approaches.

Figure 1.7 shows a flow diagram of a treatment planning system for hadron therapy based on Monte Carlo and a radiobiological model. The patient geometry is imported from CT and/or MRI data taking into account the elemental composition of different materials. The tumour volume is identified. The geometry of any beam modifying devices may be incorporated in the geometry also. For a particular primary beam energy and angle of irradiation, each particle may be transported through beam modifying devices and patient geometry using detailed Monte Carlo physics models. Then, using a radiobiological model, the distribution of equivalent dose can be calculated. This can be compared to the ideal distribution and the difference fed back to the model of the primary beam and beam modifying elements to vary the angle of irradiation and beam energies. Such a technique is implemented at GSI in carbon ion therapy. This treatment planning system TRIP, utilises the Kraft-Scholz radiobiological model; however, it relies on an analytical transport equation to model the primary and secondary radiation within the patient [27].

1.5.3 Biological Dosimetry

The ideal method of treatment planning verification is to study some biological end point such as cell survival and its 3D distribution obtained in a heterogeneous phantom. A preliminary study for the verification of carbon ion therapy treatment planning with 2D cell survival distributions in a water head phantom is given in reference [28]. These studies are invaluable for verifying the entire treatment planning process, from the transport of the primary and secondary radiation, to the predictions of cell survival
by the radiobiological model. They are however time consuming, costly and susceptible to large experimental uncertainty and dosimetry measurements are still necessary to routinely verify the model of the mixed radiation field in the patient.

1.5.4 Dosimetry

Routine verification of the treatment planning system is currently achieved with dosimetric measurements within a water phantom. The spatial distribution of absorbed dose is measured, typically using an ionisation chamber, and compared to that predicted by the treatment planning system. Discrepancies between experimental and theoretical results are used to quantify the accuracy of the treatment planning system. Although usually carried out using an ionisation chamber in a homogeneous water phantom, recent work by Kohno et al. [29] has demonstrated the feasibility of measuring absorbed dose in silicon to verify a simplified Monte Carlo proton absorbed dose calculation in a heterogeneous phantom.
1.5.5 Alternative Verification Method

For the purposes of treatment plan verification in hadron therapy however, it may be necessary to base this verification on the measurement of a quantity other than absorbed dose. The Monte Carlo (or analytical model) may accurately predict absorbed dose at some point but may not accurately calculate the constituents of the primary and secondary particles. Assuming that detailed information of the primary and secondary particles is essential to calculate the equivalent dose it follows that the use of absorbed dose to verify the treatment plan is not sufficient. The ideal method of verification would be based on the measurement of the energy spectra of all charged particles along with a measurement of gamma absorbed dose, and comparing to that predicted by the treatment planning system. This should be carried out in a heterogeneous patient phantom with tissue substitutes as outline in ICRU report 44 [30]. Particle identification techniques are used extensively in nuclear physics experiments [31, 32, 33, 34] and have recently been applied to study carbon ion fragmentation in heavy ion therapy beams [35]. However these detectors assume that ions are normally incident on the detector surface, which impedes their application to in-phantom measurements in FNT and PT for which scattering of particles is significant. The large cross section area results in high levels of pileup. A simple alternative is to use silicon based microdosimeters which are sensitive to ion LET.

1.6 Silicon Microdosimetry

The history of silicon based microdosimetry was discussed in detail by Rosenfeld and Bradley et al. [36]. It covers early work by Dicello et al in the comparison of single diode devices with a spherical TEPC to the development of arrayed devices by Roth and McNulty for use as biological and electronic radiation monitoring in spacecraft and avionics. The application of silicon microdosimetry to hadron therapy was proposed by Rosenfeld [37] and investigated in detail by Bradley and Rosenfeld [3]. In this work a prototype microdosimeter was developed based on silicon on insulator (SOI) technology.

The SOI diode array was manufactured by Fujitsu research laboratories on a bonded
p-type SOI wafer of thickness 2, 5, and 10 $\mu$m. $n^+$ and $p^+$ silicon regions (see Figure 1.8) are constructed with Arsenic and Boron implantation at 30 keV and a fluence of $5 \times 10^{15} \text{ cm}^{-2}$. The impurity concentration of the p-type silicon is $1.5 \times 10^{15} \text{ cm}^{-3}$. All diodes in the array are connected in parallel (see Figure 1.9) and each pn junction has an area of $10 \times 10 \mu m^2$. The total size of each diode cell is $30 \times 30 \mu m^2$. With 120 $\times$ 40 diodes in a single device the total array area is 0.044 $cm^2$.

The microdosimeter is incorporated into a Lucite probe for microdosimetry measurements in hadron therapy. The detector is connected to an AMPTEK A250 radiation-hard charge sensitive preamplifier in close proximity to reduce input capacitance. Noise is minimised by encapsulating the entire probe in aluminium shielding which is then connected to the coaxial cable ground. This also ensures that the chip does not receive any light, producing significant noise by photo-generation of carriers. The microdosimeter collects charge generated by the traversal of an ion through the device’s sensitive volume. By measuring the charge collected one can calculate the energy lost in the sensitive volume. This is proportional to the energy lost to ionisations assuming full charge collection and assuming a mean energy of 3.6 eV to generate an electron hole pair in silicon. The preamplifier produces a voltage pulse with amplitude proportional to the energy deposited in the sensitive volume of the microdosimeter. The pulse height from each event is digitised and stored using a PC based Multi-Channel Analyser (MCA). A detailed description of the device and readout electronics is given in reference [5].

In the work by Bradley et al. the aim was to develop a silicon microdosimeter which can provide microdosimetry spectra identical to that measured by a tissue equivalent volume. These spectra may then be used to calculate equivalent absorbed dose as described in section 1.3.2. Measurements were conducted at boron neutron capture therapy [38], proton therapy [5], and fast neutron therapy [39] facilities. The silicon microdosimeter was proposed as a replacement for the tissue equivalent proportional counter (TEPC) and reference [36] provides a comparison between the TEPC and silicon microdosimeters, highlighting the cost, convenience, and accuracy of the instruments. During these studies
Figure 1.8: Cross section of the 10 μm silicon microdosimeter. All dimensions are in microns.

Figure 1.9: Scanning Electron Microscope image of the silicon microdosimeter showing an array of planar pn junctions connected in parallel.
several factors were identified which impede the widespread application of silicon microdosimeters to hadron therapy. The first issue was that of sensitive volume definition. Ideally the microdosimeter should have a well defined sensitive volume as close to spherical as possible to reduce angular dependence of the measurement of lineal energy. In the case of silicon microdosimetry the volume is cubic and charge is collected from the depletion region of the device. To quantify the sensitive volume definition it is necessary to study the variation of charge collection efficiency (CCE) with ion strike location on the device. To achieve this an ion microprobe, a finely focussed beam of ions, may be used. By scanning the beam across the microdosimeter, whilst measuring charge collected, an image of the CCE can be created; a technique known as Ion Beam Induced Charge (IBIC) imaging. Studies were carried out by Bradley et al. \[40\] to characterise the charge collection behaviour using a proton microprobe. These studies showed the charge collection behaviour varied depending on the location of the ion strike on the diode and significant charge was collected for ion strikes outside the depletion region. Subsequently, implementation of lateral isolation techniques in the design of a second generation device was suggested in order to provide a fully depleted charge collection volume \[36\]. The authors also noted the need to study the variation of these charge collection properties with ion LET.

The second issue is that of tissue equivalence; the microdosimetry spectrum measured by a silicon microdosimeter will differ from that of an equivalent tissue volume of microscopic dimensions. Consequently the microdosimetry spectrum may not be used as input to equation \[1.8\] without correction. A method was proposed whereby the mean chord length of the silicon sensitive volume was scaled \[41\]. Firstly, simple scaling of the mean chord length ignores differences in the spectral characteristics resulting from differences in geometry and composition. Secondly, any tissue equivalence correction assumes the lineal energy event spectrum is dominated by events from particles originating from outside the silicon sensitive volume. Particularly in the case of Fast Neutron Therapy, for which the cross section for inelastic scatter with silicon is significant, the validity of this
assumption requires further study. The inability of the silicon microdosimeter to provide tissue equivalent measurements remains a drawback in its application to hadron therapy and radiation protection applications.

The third issue is that of radiation damage in silicon degrading the performance of the silicon microdosimeter. The introduction of defects in the crystalline lattice caused by interactions of hadrons with silicon nuclei leads to a reduction in the charge carrier lifetime. Several studies were conducted to investigate the variation of this quantity with exposure to neutrons and protons [40]. For the case of Fast Neutron Therapy, a 10% change in the charge carrier lifetime after 3.4 patient treatment cycles was predicted. Future silicon microdosimeters, fabricated using lateral isolation techniques, will be much less susceptible to radiation damage.

1.7 Proposed Application of Silicon Microdosimeters to Hadron Therapy

This thesis proposes a reviewed application of silicon microdosimeters to hadron therapy, using the instrument to verify Monte Carlo calculations used in treatment planning. An example of how this may be achieved is shown in figure 1.10. A heterogeneous

Figure 1.10: Proposed heterogeneous phantom for silicon microdosimetry based verification of Monte Carlo calculations in Hadron Therapy.
phantom consisting of materials which mimic the elemental composition of actual tissues could be used, similar to that utilised in reference [29]. This anthropomorphic phantom could have small plugs at various positions which can be replaced by a probe containing the silicon volumes. It is feasible to incorporate the detector geometry and composition into the Monte Carlo calculation used in the treatment planning system. The phantom geometry may be obtained by CT and/or MRI imaging in a manner identical to a patient. Image analysis software can then be used to identify particular organs and assign elemental compositions. The phantom can be irradiated and energy deposition events in the microscopic silicon volumes can be measured. A Monte Carlo simulation can then be performed taking into account the charge collection efficiency of the device, the statistics of electron hole production and the influence of electronic noise. The comparison between simulated and experimental energy deposition event spectrum can be compared and a quantitative measure of the agreement used to indicate the accuracy of the calculation.

In this situation the non-tissue equivalence of the instrument does not degrade it’s usefulness. The purpose of the microdosimeter is not to provide information to be used directly with conventional microdosimetry to calculate an equivalent dose. It is simply to verify the accuracy of the Monte Carlo model of radiation interaction with the phantom and the silicon microdosimeter itself.

1.8 Thesis Outline

The aim of this thesis is to continue research into the development of silicon microdosimeters for hadron therapy. Monte Carlo calculations feature extensively throughout this thesis and Chapter 2 provides an introduction to the Monte Carlo method and a description of the GEANT4 Monte Carlo toolkit. The simulation of silicon microdosimetry measurements must also take into account the charge collection behaviour of the device. A goal of this thesis is to extend the charge collection work by Bradley et al. to IBIC studies using heavier ions. Chapter 3 describes the ANSTO ion microprobe, the development
of IBIC capabilities and preliminary experiments carried out for obtaining micron resolution and sufficiently low fluence rates necessary for IBIC. Chapter 4 describes charge collection studies conducted with a heavy ion microprobe using H, He, and C ions. A comparison of these results with GEANT4 simulations was used to determine the LET dependence of the charge collection efficiency of the devices. Chapter 5 describes an investigation of a pulse shape discrimination technique to ignore ions striking outside the ideal sensitive volume. Chapter 6 describes a preliminary study of the verification of GEANT4 Monte Carlo calculations using simple silicon microdosimetry measurements in Fast Neutron Therapy. Chapter 7 describes a similar study for simulations in Proton Therapy.
CHAPTER 2
MONTE CARLO AND THE GEANT4 TOOLKIT

2.1 Introduction

Monte Carlo is a numerical technique based on the generation of random numbers [42]. If the probability distribution of a stochastic variable is known, the monte carlo technique may be used to sample variables from this distribution based on random number generation [2]. In the case of the interaction of radiation with matter, these variables are quantities such as the distance travelled by a particle prior to interaction, the type of interaction and the target particle, the number and types of interaction products, and the recoil angles and energies of interaction products. With Monte Carlo techniques it is possible to model these interactions for complicated target geometries and material compositions; a task almost impossible using analytical methods.

The GEANT4 Monte Carlo toolkit is an open source collection of C++ classes available from the GEANT4 website [43]. It is based on object oriented C++ programming [44] and was developed by the CERN based RD44 collaboration for the simulation of high energy physics experiments. In recent years however, the capability to model low energy interactions has been added. This has seen the widespread use of GEANT4 in the medical physics community. All aspects of the simulation process have been included in the toolkit such as: geometry and materials involved, particles of interest, the generation of primary events, the tracking of particles through materials and electromagnetic fields, the physics processes governing particle interactions, the response of sensitive detector components, the generation of event data and the visualization of the detector and particle trajectories.

Developing a GEANT4 application requires the user to write his/her own C++ program using classes which inherit behaviour from GEANT4 classes. Extensive example
programs are supplied with the GEANT4 package and offer a good starting point for developing an application. The following sections describe the classes which need to be implemented in developing a GEANT4 application.

2.2 User Action Classes

There are three virtual classes whose methods must be overridden to write a GEANT4 simulation. These control the geometry of the simulation, definition of particles and physics processes to implement, and the generation of primary particles.

2.2.1 G4VUserDetectorConstruction

This base class controls definition of the detector geometry. The detector definition requires the representation of its geometrical elements, their materials and electronic properties, together with visualization attributes and other user defined properties. The geometrical representation of detector elements focuses on the solid model definition and their spatial positions, as well as their logical relations such as the relation of containment within other volumes. Solids can be formed by boolean operations with other solids and repeated structures can be easily modelled. GEANT4 also allows the geometry to be input from STEP compliant CAD systems. Materials are considered to contain elements, and elements exist as isotopes. GEANT4 provides the ability to model radiation transport in magnetic, electric, and electromagnetic fields. It also enables one to monitor ‘hits’ to sensitive regions of the geometry. A hit is a snapshot of the physical interaction of a track in the sensitive region of the detector. For each hit, a variety of information from the track can be stored such as: energy deposition, characteristics of the particle (Energy, mass, spin), geometrical information such as the volume from which the particle originated and the spatial coordinates of the hit.
2.2.2 G4VUserPhysicsList

There are three methods of this class which must be implemented. The ConstructParticle method defines all particles to be used in the simulation. The ConstructProcess method determines the models of interaction for these particles. Seven major categories of processes are provided by Geant4: electromagnetic, hadronic, decay, photolepton-hadron, optical, parameterisation and transportation. All physics processes are treated in the same manner from the tracking point of view. This method enables the user to create a process and assign it to a particle type. This openness allows the customisation of physics processes by individual users. The final method is SetCuts which determines the cut for a particle. In this case, if a particle is generated with a residual range in the material less than this cut value it will not be tracked.

2.2.3 G4VUserPrimaryGeneratorAction

The methods of this class are used to define the primary particle, it’s energy, direction cosines, and initial position. G4VUserPrimaryGeneratorAction has a pure virtual method named generatePrimaries. One must invoke the G4VPrimaryGenerator concrete class G4ParticleGun with this method. G4ParticleGun contains methods which may be used to define the particle, it’s momentum and direction, it’s energy, the time of particle, the position, it’s polarisation, and the number of particles to be considered for this event. Particles are instantiated via the generatePrimaryVertex method. There are several virtual classes whose methods may be overridden to write a GEANT4 simulation. These are responsible for actions at various stages of the simulation.

2.2.4 G4UserEventAction

An “Event” in GEANT4 starts with the initiation of tracking of a primary particle and finishes with the completion of tracking of all secondaries. This G4UserEventAction class possesses two virtual methods which are invoked at the beginning and end of each
event. The beginOfEventAction method is invoked before converting the primary particles to G4Track objects. A typical use of this method is to initialise histograms for a particular event. The method endOfEventAction is invoked at the very end of event processing. It is typically used to analyse the collection of “hits” to the sensitive region of the detector in order to extract information such as energy deposition, saving the relevant information to file.

2.2.5 G4UserRunAction

A “Run” in GEANT4 is the simulation of a number of primary particles. The G4UserRunAction class has several methods. One commonly used method 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.

2.2.6 G4UserSteppingAction

The concept of a Step in GEANT4 describes the transport of a particle between two points in space. The G4UserSteppingAction class contains methods for taking action at the end of each step and is commonly used for customisation of plotting particle trajectories if visualisation is used.

2.3 Interface Commands

GEANT4 has various built-in user interface commands. These commands can be used interactively via a user interface, in a macro file, or within C++ code with the ApplyCommand method of the G4UIManager class. G4UIManager is a base class which represents a messenger that delivers these commands to a class object. User defined commands can be implemented by inheriting behaviour from such a base class. These commands are particularly useful when the geometry, primary beam, or physics needs to be altered between simulations. In this way one may execute several simulations using a macro file which contains a number of suitable commands.
2.4 Visualisation

GEANT4 has the capability to visualise detector components, particle trajectories and tracking steps and hits of particles in detector components. GEANT4 visualisation supports interfaces to a variety of graphics systems via visualisation drivers. Although many methods of visualisation are possible, those employed in simulations of this thesis were with OpenGL drivers and the VRML (Virtual Reality Markup Language) files. The OpenGL drivers were found to be most useful for visualising, with the aim of debugging, the generation of primary events and the tracking of these events through the geometry. The use of VRML files was most useful for debugging the geometry in detail. In this situation a VRML file may be viewed with a viewing program such as the freely available “vrmlview”. VRML allows the user to navigate through the detector geometry and inspect it in detail to identify bugs in design.

For more information on the toolkit the reader is directed to the GEANT4 website that contains extensive documentation and an online user forum, from which the toolkit developers offer assistance to application developers.
CHAPTER 3
THE ANSTO HEAVY ION MICROPROBE

The experiments described in this thesis were performed using the heavy ion microprobe \[45\] at the Australian Nuclear Science and Technology Organisation (ANSTO). The microprobe is situated on a beam-line of the Australian National Tandem for Applied RESearch (ANTARES) 10 MV accelerator and was established in 1996. This chapter provides a general description of the facility and the general procedure for using the microprobe. The author was responsible for extending the capabilities of the microprobe to include low current applications necessary for conducting the experiments on silicon.

Figure 3.1: Effects of ion beam irradiation and corresponding ion beam analysis techniques commonly used with ion microprobes
microdosimeters. Beam diagnostic experiments performed are also described.

3.1 Introduction

A microprobe is a finely focussed beam of energetic ions that can be used for materials analysis and modification. The beam of ions is accelerated to MeV energies using a van de Graaf accelerator. It is then focussed to microscopic dimensions and raster scanned across a sample to be studied. The ion beam induces effects on the sample, such as characteristic X-ray emission, secondary electron emission, or induced current that are measured as a function of beam position. This effect is used to derive a two dimensional image of some characteristic of the sample, for example the elemental composition in the case of PIXE. Figure 3.1 illustrates possible effects of ion irradiation on materials together with the respective ion beam analysis techniques. For a more detailed description of these techniques the reader is referred to references [46, 47] and references therein. Microprobes are used in many fields ranging from the irradiation of single biological cells [48, 49], trace element analysis for environmental applications [50], to studies of the single event upset sensitivity of micro-electronic devices [51]. Advances in the field of microprobes are discussed at such forums as the International Conference on Nuclear Microprobe Techniques and Analysis (ICNMTA) and the Ion Beam Analysis (IBA) Conference.

3.2 ANTARES

3.2.1 Ion Sources

The general layout of the ANTARES facility is shown in figures 3.2 and 3.3. Experiments described in this thesis used ions generated by the α-tross and 860 ion sources. The α-tross ion source is used to produce alpha particles and protons. A Helium or Hydrogen gas is ionised by a radio frequency field and positive ions from the plasma are extracted from the source and pass through rubidium vapour. In doing so the ions undergo charge exchange and become negatively charged. These ions are then pre-accelerated to keV
energies. Figure 3.2 shows the electrostatic deflection plate Y1, used to vertically steer the beam into the Faraday cup FC1. Also shown is the einzel lens EL1, used to focus the beam at the position of FC1. The beam is then deflected through 45° with an analysing magnet MG1 into faraday cup FC3. Using the einzel lens EL3 the beam is focussed to the faraday cup LEFC, located at the entrance to the accelerator. The magnet MG2 is only used for slight x steering of the beam and is usually de-gaussed. Beam profile monitors (BP1 and BP2) are used, in conjunction with focussing and steering elements, to optimise beam symmetry.

The second ion source used for the experiments is the 860 Cesium sputter source
which can produce a wide range of elements from solid targets. A cathode containing
the desired ion species is bombarded with Cesium ions with energies in the order of
\( keV \). Subsequent sputter interactions produce negatively charged secondary ions which
are ejected from the surface of the cathode and pre-accelerated to \( keV \) energies. The
beam is focussed and steered to the Faraday cup FC2 using einzel lens EL2 and x and y
electrostatic steering plates X1 and Y2 respectively. The beam is analysed and transported
to the low energy Faraday cup LEFC as per above. Currents at the LEFC are typically in
the order \( 100 \, nA - 1 \, \mu A \).\(^1\)

3.2.2 Accelerator

Once the \( keV \) energy beam is focussed at the low energy flange the ions are then
accelerated by the tandem van de Graaf accelerator in two stages. The negative ions are
accelerated to the positively charged terminal where they pass through a stripping gas
which removes some or all of the electrons. This results in positively charged ions and
these are accelerated away from the terminal. The choice of the voltage setting \( V \) is
dependant on the ion energy \( E \) required and the charge state \( q \) where ion energy is given
by \( E_{ion} = V(q + 1)e \) and \( e \) is the charge on an electron. The terminal voltage stability
of this accelerator has been measured to be less than \( 1kV \) at 8\( MV \) [45]. Referring to
figure 3.2, an electrostatic quadrupole lens EL4 is situated at the low energy entrance of
the accelerator and is used to focus the beam at the high energy Faraday cup HEFC in
figure 3.3 X and Y electrostatic deflection plates (X2 and Y3 respectively) adjust for
misalignment of the beam.

Following the high energy Faraday cup, a 55° analysing magnet (MG4) is used to
select the energy and charge state of the beam and deflect it into the Faraday cup at the
entrance to the microprobe beam line MPFC1. The only focussing element used between
the accelerator and the microprobe is a quadrupole doublet (QD1) after the accelerator.
Fine tuning the beam in x and y into the object aperture of the microprobe is done using

\(^1\)Care should be taken not to exceed several \( \mu A \) as an off axis beam striking the accelerator tube can
cause an increase in vacuum pressure which can trigger the accelerator to shutdown
Figure 3.4: The ANSTO Microprobe End Station; OS: object slits, MPFC2: faraday cup, MPBV1: beam viewer, MPBP2: beam profile monitor, MPFC3: faraday cup, MPBV2: beam viewer, CS: collimating slits, SC: scanning coils, QT: quadrupole triplet, C1: forward mounted camera, SH: sample holder, C2: rear mounted camera.

the analysing magnet (MG3) and electrostatic deflection plates (Y4) respectively. A beam profile monitor BP3 is used during beam steering and symmetry optimisation.

3.3 The Microprobe

Figure 3.4 shows a schematic of the Microprobe beam line and target chamber. The object to be de-magnified is defined by a pair of x and y, cylindrical, tungsten carbide, “object” slits (OS). At five metres distance a second pair of slits, the “collimating” slits (CS), limit the divergence of the beam and remove ions scattered by the object slits. Immediately following the slits are orthogonal current carrying coils (SC). By modulating the current in each set the beam is raster scanned over the sample. The lens system
consists of a quadrupole triplet which focusses the beam onto the sample at a distance of approximately 5 cm. A wide range of ions and ion energies can be focussed and scanned by the microprobe. This is limited by that of the saturation field in the quadrupole triplet while the maximum scan area is limited by the scanning coils. The ANSTO microprobe has the ability to focus ions with a magnetic rigidity of up to \( ME/q^2 \approx 100 \text{MeVamu}^2 \) and scan these over an area of \(1 - 3 \text{mm}^2\), depending on the rigidity.

The sample holder (SH) is situated within an octagonal target chamber as shown in figure 3.4. It is mounted on a goniometer with the azimuthal angle fixed at \( \phi = 0 \) and the polar angle variable as \( \theta \in [-90:90] \). The z axis is defined by the incident beam and the xy plane is located at the focal point as shown in figure 3.4. The sample stage can be moved in x, y, and z directions using precision micrometers.

3.3.1 Focussing Procedure

The following procedure describes how to focus the ion beam assuming the beam is at the Faraday cup MPFC1. Both object and collimating slits are opened and the beam is transported to the microprobe chamber. This process involves transporting the beam to the MPFC2 Faraday cup, viewing the beam on the quartz beam viewer MPBV1 and optimise symmetry, then steering the beam to MPFC3 several metres downstream. Again the beam is viewed on the quartz viewer MPBV2. This process is carried out by adjusting the high energy steering and focussing components. A CsI crystal mounted on the target holder is moved into the beam. Fluorescence of the crystal can be observed using the rear mounted CCD camera (C2). The intensity of the beam is optimised by evenly illuminating the object aperture. Next the object slits are closed to a nominal setting. The beam must be fine steered to optimise current and symmetry at the sample holder. At this stage the view is changed to the CCD camera (C1) with optically coupled microscope mounted at 45° in the forward direction. A quartz crystal is then brought into the focal plane of the microscope via x and z translation of the sample holder. The magnetic rigidity refers to the magnetic field strength required to bend an ion of mass \( M \), charge state \( q \), and energy \( E \) through a given radius.
focusing is carried out by visually observing the fluorescence on the crystal. With both object and collimating slits set for a spot size of approximately ten microns the beam is focussed by increasing the currents in the quadrupole triplet as shown in figure 3.5. Once the beam is focussed, the scanning amplifier is activated and the beam is scanned over the crystal. The scan area, indicated by fluorescence of the crystal, is marked on the monitor. Prior to each analysis the sample must be brought into the focal plane of the microscope and the region of interest is positioned using the scan area marked on the monitor. In this situation, the sample will also be at the focal point of the quadrupole lenses.

3.3.2 Data Acquisition System

The Oxford Microbeam Data AcQuisition system (OMDAQ) consists of Windows based software, a scanning amplifier, ten, 2048 channel Analogue to Digital Converters (ADCs), and a current integrator connected to the sample holder. The scanning amplifier amplifies step pulses from the data acquisition software which determine the magnetic field strengths in the coils and hence the beam position.

The detector signal is fed into an ADC of the data acquisition system. The amplitude of the signal is proportional to the parameter of interest such as the ion energy, a
characteristic X-ray energy or the charge collected in microelectronic device. The software stores the signal height for each pulse together with the x and y beam coordinates as a data triplet (E,x,y), or “event”, in a file. This “list mode file” can be played back after the acquisition for re-analysis.

3.4 Beam Diagnostics for Low Current Applications

Charge collection measurements on microelectronic devices requires the satisfaction of two criteria. For applications such as STIM or IBIC a semiconductor detector or microelectronic sample is placed directly in the ion beam. The beam current must be sufficiently low to avoid damaging the crystalline structure of the sample yet high enough to obtain good image statistics in a reasonable time. The cumulative fluence considered acceptable for microelectronic devices is typically less than 10 ions µm$^{-2}$ [46] but is dependent on ion atomic number and energy. Moreover the number of ions impinging on the device per second should be less than several hundred to avoid pulse pile up. The second criterion is that the beam diameter is close to one micron. The best focussed beam spot-size for a given set of quadrupoles can be calculated theoretically from the quadrupole aberrations, the object size and the ratio of object distance to image distance. However, several other factors can limit the minimum attainable spot-size such as: mechanical vibration of the sample, vibration of the object and collimating slits, and electromagnetic interference leading to oscillation of the beam. Vibrations of the object slits is minimised by mounting them on a massive concrete block. Vibrations in the sample holder are minimised by dampening mechanical vibrations coming from the floor. This is done by mounting the end-station on a concrete block resting in sand. Previously the ANSTO microprobe was used for high current applications such as PIXE and RBS for which high currents and relatively large spot sizes were used. As a consequence it was necessary for the author to develop the microprobe to accommodate low current, high resolution applications such as IBIC. This section describes the method for obtaining sufficiently low beam fluence rate on the microprobe for IBIC studies and measurement of the beam
3.4.1 Methods

3.4.1.1 Stim Setup

Both tasks were carried out using a technique called Scanning Transmission Ion Microscopy (STIM). STIM is an ion beam analysis technique used for measuring the density of thin samples [52]. A charged particle detector is positioned behind the sample and the energy of the transmitted ion is measured as shown in figure 3.1. There are two modes of STIM: when the detector is placed off the beam axis the detector measures transmitted ions which scatter in the sample, this is known as dark field imaging; when the detector is placed on the beam axis the detector measures ions transmitted directly through the sample with negligible scatter, this is known as bright field imaging.

A STIM setup was designed by the author and constructed at ANSTO and is shown in figure 3.6. It is comprised of a ion implanted silicon detector encased in a steel holder.
which allows it to be placed behind the sample holder during analysis as shown in figure 3.7. The STIM detector is coupled to an ORTEC EG&G 142 charge sensitive preamplifier. The detector is fully depleted with a reverse bias voltage of $50 \text{ V}$ using an ORTEC EG&G 710 quad bias supply. Following an ion strike on the detector, a voltage pulse is produced by the preamplifier with an amplitude proportional to the energy lost in the detector. Each pulse is further amplified and shaped using an ORTEC EG&G 572 spectroscopy amplifier with shaping time of $2 \mu\text{s}$. If the ion beam is scanned over a thin sample the ion energy loss in the sample can be measured as a function of beam position.

3.4.1.2 Beam Fluence Measurements

The following procedure, based on a methodology described in reference [46] was used to establish the beam fluence rate suitable for STIM and IBIC. A copper test grid is placed in the scan area of the focussed beam and the STIM detector is placed off axis behind the grid. Each ion forward scattered off the grid into the detector induces a current pulse in the external circuit. Pulses from the spectroscopy amplifier are observed on an oscilloscope. The STIM detector is shifted closer to beam axis, and as the angular
distribution of scattered ions is peaked in the forward direction, this action results in an increase in count rate. The detector is shifted closer to the beam axis until a count rate in the order of 100 Hz is observed. The ion fluence rate may then be lowered using several methods. The first is to close the object slits. This is not suitable for spot sizes less than approximately 2 µm as thermal expansion of the slits causes the aperture to close completely. A second is to reduce the ion source output. A third is to reduce the density of stripping gas in the high voltage terminal of the accelerator. Once the count rate has decreased below several hundred counts per second the STIM detector is again shifted closer to the beam axis. The current is again lowered and this procedure is reiterated until the detector is directly in the beam and the fluence rate is several hundred counts per second. At this stage the STIM detector can remain directly in the beam and a sample can be moved into the beam scan area. If IBIC measurements are being performed, the STIM detector can be retracted and the IBIC sample placed in the scan area.

3.4.1.3 Beam Spot-size Measurements

An experiment was conducted to determine the minimum beam spot size in the x and y directions. The sample used was a copper grid with 2000 bars per inch. This corresponds to a pitch of 25 µm with bar and space dimensions of 6 µm and 19 µm respectively. It was mounted on the sample holder and brought into the focal plane of the forward mounted camera and hence the quadrupole triplet. The beam was raster scanned across a 100 × 100 µm² area of the grid and the energy of each transmitted ion was measured at each beam position. Firstly the scan area was calibrated in terms of microns per pixel. This was done by scanning the beam over a large area and counting the number of bars in the scan area in x and y. Next, the scan area was reduced and data acquired and stored in an event file. This file was then analysed to measure the statistical properties of the energy distribution at each pixel of the scan (x, y). For STIM and

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3 The latter two techniques were used in this thesis although they were time consuming owing to lag time between setting the stripping efficiency and actual change in density of gas. This could be avoided if a single ion delivery system was installed as per reference [48].
IBIC imaging the median value is used instead of the mean as it is a more useful indicator of the centre of a distribution which is skewed [53]. This is frequently the case in STIM and IBIC as extreme events can result from ions scattered by the collimating slits [46]. To obtain the median $N$ energy events, $E_1(x, y), \ldots, E_N(x, y)$ are reordered so that $Y_1(x, y) < \ldots < Y_i(x, y) < \ldots Y_N(x, y)$. $Y_i(x, y)$ is $i$th ordered statistic and the median value is given by:

$$E_{1/2}(x, y) = \begin{cases} Y_{n/2}(x, y), & \text{odd } n \\ \frac{Y_{n/2}(x, y) + Y_{(n+1)/2}(x, y)}{2}, & \text{even } n \end{cases}$$

The initial data generated by OMDAQ is a $256 \times 256$ image of the median ADC number and a frequency distribution of energy deposition events. A C++ program was written to analyse this data. The image is calibrated from ADC channel number to energy using the peak in the energy spectrum corresponding to the unattenuated ion beam and assuming zero offset in the electronics. If the spot-size in the $x$ direction is being measured this image is translated and cropped so that a single vertical bar is included in a $15 \times 15 \mu m^2$ area. The median value of all pixels in each column of this image is taken to improve statistics. A profile of this image is taken in the $x$ direction and plotted. The beam spot size may then be inferred by comparing the measured width of bar to the physical value.

3.4.2 Results and Discussion

Figure 3.8 shows the spectrum of energy deposition events for the STIM measurement. The spectrum is composed of two peaks. The high energy peak corresponds to ions passing through gaps in the grid and the low energy peak corresponds to ions passing through the copper grid and suffering energy loss. Figure 3.9 shows the full $256 \times 256$ pixel image of the median transmitted ion energy. The scan area calibration factor was calculated at $0.403 \mu m / pixel$. Black pixels indicate zero counts, a result of the low count rate.
Figure 3.8: Spectrum of transmitted ion energies from STIM measurement of copper grid with 9 \( MeV \) helium.

Figure 3.9: Bright field STIM image of copper test grid where pixel number has been calibrated in microns and ADC number in keV.
Figure 3.10: Image of single horizontal bar before (left) and after (right) calculating the median value of all x pixels.

Figure 3.10 (left) shows a translated and cropped image which covers a $15 \times 15 \, \mu m^2$ area of a single horizontal bar. Figure 3.10 (right) shows the same image after taking the median of all pixels in a row. Figure 3.11 shows the y profile across the horizontal bar. The bar width measured from this profile is $6.8 \pm 0.8 \, \mu m$. Given the bar width is $6 \, \mu m$ this indicates a maximum beam diameter in y of $0.7 \, \mu m$.

Figure 3.12 (left) shows a translated and cropped image which covers a $15 \times 15 \, \mu m^2$ area of a single vertical bar. Figure 3.10 (right) shows the same image after taking the median of all pixels in a column. Figure 3.13 shows the x profile across the vertical bar. The bar width measured from this profile is $7.6 \pm 0.8 \, \mu m$. Given the bar width is $6 \, \mu m$ this confirms a beam diameter in the order of $1 \, \mu m$.

3.4.3 Conclusions

A background to the heavy ion microprobe facility has been described. A procedure for obtaining low currents suitable for IBIC and STIM was established using Scanning Transmission Ion Microscopy. This method of beam current reduction is far from ideal. The ideal system would involve a single ion delivery system composed of electrostatic deflection plates, a high speed, high voltage amplifier, and a single ion detection system.
Figure 3.11: Y profile of median transmitted ion energy across horizontal bar of copper grid.

Figure 3.12: Image of single vertical bar before and after calculating the median value of all y pixels.
Figure 3.13: X profile of median transmitted ion energy.

The minimum beam spot-size obtainable with the current ANSTO heavy ion microprobe was measured and found to be approximately 1 $\mu$m. The main factor influencing imaging resolution was found to be vibration of the sample holder. Sources of these vibrations were the cooling fan on the turbo pump attached to the microprobe end station, the cooling fans mounted on the quadrupole magnets and the rotary roughing pumps in the vicinity of the end station. Vibrations from roughing pumps were transmitted through the bellowed hoses connecting these to the microprobe chamber. These vibrations were reduced by placing lead bricks on the hoses. It has been suggested to run all cables and hoses through large containers of sand to dampen any transmitted vibrations. The cooling fans were removed from the quadrupole triplet and overheating of the triplet did not occur for the experiments of this thesis but an alternate method of cooling should be introduced if heavier ions, and hence higher quadrupole currents, are to be used. Further improvement of beam resolution could be obtained by implementing electromagnetic shielding of the microprobe beam-line. Moreover the sample holder could be more rigidly attached to the goniometer.
CHAPTER 4
LET DEPENDENCE OF THE CHARGE COLLECTION EFFICIENCY OF SILICON MICRODOSIMETERS

4.1 Introduction

The IBIC technique is used to measure the total charge collected from a semiconducting sample following irradiation by energetic ions. The impetus for IBIC development was to locate the regions of microelectronic memory devices which are sensitive to single event upset phenomena [51]. The following discussion gives a qualitative description of charge collection following an ion strike. It refers to the simple device illustrated in figure 4.1 consisting of a p-type silicon wafer with a $n^+$ rectifying contact, a $p^+$ blocking contact, and aluminium electrical contacts for both $n^+$ and $p^+$ regions. The $p^+$ contact is grounded whilst the $n^+$ contact is held at a positive voltage. This reverse biasing results in a region of the p type wafer that contains ionised acceptor atoms. This creates a region of fixed space charge and hence an electric field (the depletion region). An equilibrium current flows due to thermal generation of carriers within the depletion region.

As an ion traverses the sample it experiences energy losses both through electronic and nuclear stopping as described in section 1.1.3. The excitation and ionisation of target atoms leads to a plasma of charge carriers along the path of the ion. The electron hole pairs then move under the influence of chemical and electrical potentials. For the silicon microdosimeters, the collection of charge is dominated by three mechanisms: drift, diffusion, and funneling. When charge is generated within the depletion region, the electric field does not penetrate the highly conductive plasma of electron hole pairs. Electrons and holes in the centre of this plasma do not “see” the electric field. Electron hole pairs on the periphery of the plasma however will drift under the influence of the electric field; electrons to the upper contact and holes toward the lower contact. This results in an erosion of the plasma and e-h pairs closer to the ion trajectory begin to “see” the electric field and drift under it’s influence. The time associated with this plasma erosion is known
Figure 4.1: pn junction showing pre-strike depletion region

as the “plasma decay” time.

When charge is generated outside the depletion region the plasma decay time is much greater as there is no electric field to aid plasma erosion. Electron hole pairs then move under the influence of the chemical potential by the process of diffusion. If they reach the depletion region they will then move via drift, otherwise they will diffuse through the crystal prior to recombination. Diffusion is a comparatively slower process and as such results in higher recombination of charge carriers.

Figure 4.2 illustrates the more complicated situation of funneling whereby charge is generated partially within the depletion region. A more detailed description of this phenomenon is given by Edmonds et al [54]. Funneling can be considered to occur in two stages. Firstly the electric field in the depletion region pulls electrons up and pushes holes down causing charge separation. The net shift of negative charge upward and positive charge downward neutralises some of the previously uncompensated ionised donors and acceptors. This neutralisation causes the depletion region to shrink, resulting in a reduced potential across the depletion region. Consequently nearly all of the applied
Figure 4.2: Pn junction following ion strike where charge is deposited partially within the depletion region.

voltage is across the substrate. This voltage enhances electron flow from the substrate to the depletion region, whilst holes are continually pushed down under negative acceptor ions with no replacements from the $n^+$ region. This creates an expanding region of space charge virtually depleted of holes. It is not depleted of electrons as there is a large supply in the p-type region.

Regardless of the mechanism for charge collection, the movement of charge constitutes a current which can be measured in an external circuit if electrical contacts are made at opposing sides of the diode. This current is integrated on a charge sensitive preamplifier to yield a voltage pulse with an amplitude that is proportional to the charge collected by the device.

In section 1.7 it was suggested to use the silicon microdosimeters to verify Monte Carlo based treatment planning systems in hadron therapy. During this simulation the experimentally measured energy deposition will be modified by charge carrier recombination. As current Monte Carlo radiation transport codes do not currently model the charge collection process it is necessary to study this experimentally. The quantity of
interest is the charge collection efficiency (CCE), or the fraction of charge carriers collected following ion traversal of the sample. This gives an indication of the amount of recombination. The aim of this chapter is to utilise the ANSTO heavy ion microprobe and the IBIC technique to study the CCE of the microdosimeters and how this quantity varies with the location of ion strike on the device, the ion LET, the applied voltage and the device thickness.

4.2 IBIC Setup and Calibration

4.2.1 Methods

A setup for IBIC measurements was designed and constructed for the ANSTO microprobe. The sample is capacitively coupled to the input of a charge sensitive amplifier as shown in figure 4.3. An AMPTEK model A-250 charge sensitive preamplifier was mounted on the purpose built PC-250 test board as shown in figure 4.4. The test board
Figure 4.4: IBIC sample holder with A250 amplifier and cables to 10 pin electrical feedthrough.

is electrically isolated from the sample holder using perspex nuts. Shielded cable is then used to convey the preamplifier power supply, the device bias, the test pulse signal, and the energy signal to the vacuum side of a 10 pin electrical feedthrough. Figure 4.5 shows the air-side of the feedthrough with an aluminium box with BNC feedthroughs. This allows the relay of signals to the NIM rack several metres from the microprobe target chamber. A battery powered voltage supply is used to provide the $\pm 6 V$ power to the A-250. Battery power is used as power supplies derived from mains, although filtered, are unacceptably noisy for these experiments. A BNC model “BH-1 tail pulse generator” was used to test the circuitry with a frequency of $1 kHz$, a pulse rise-time of $\tau_{\text{rise}} = 20\,\text{ns}$, a pulse fall time of $\tau_{\text{fall}} = 100\,\mu\text{s}$, and with a variable amplitude setting. The bias on the sample is applied using an ORTEC model 710 quad bias supply through a low pass protection filter on the PC-250. Following each ion strike the induced current is integrated

Note: when using the test pulser to test the electronics it is essential that the risetime of the pulse is less than $20\,\text{ns}$. The test input of the A-250 pre-amplifier is terminated by $50\Omega$ and has a series capacitance of $2pF$. Pulses with greater than $20\,\text{ns}$ rise times suffer considerable attenuation due to the effective high pass filter of the test input.
on the feedback capacitor of the preamplifier and produces a voltage pulse with an amplitude proportional to the total charge collected. The pulse is then amplified and shaped with CR-RC filters of a Canberra model 2022 spectroscopic amplifier with a shaping time of $\tau = 1.5 \mu s$. The output of this amplifier has a pulse shape suitable for the analogue to digital converter of the data acquisition system. The Analogue to Digital Converter (ADC) digitises the amplitude of each signal with a resolution of 2048 bins covering a full scale of $0 - 10 V$.

The ADC of the data acquisition system is calibrated in terms of ionisation energy deposited in silicon. This is achieved using a $3 MeV$ proton beam incident on a fully depleted ion implanted detector (depletion depth of $330 \mu m$). The method follows the approach used by Yang et.al. [55] and involves measuring the pulse amplitude at the output of the A250 amplifier (ie. peak channel measured with the ADC) whilst varying the angle of incidence of the proton beam. The technique assumes the calibration detector
Figure 4.6: Schematic of calibration diode and angles of irradiation used for energy calibration of IBIC electronics.

to be a fully depleted silicon volume with a thin silicon dead layer and thin aluminium contact as shown in figure [4.6]. The energy of the proton after traversing the over-layers can then be described by the following expression:

\[ E = E_p - \frac{t_{al}}{\cos \theta} \frac{dE}{dx_{al}} (E_p) - \frac{t_d}{\cos \theta} \frac{dE}{dx_{si}} (E_p) \]  

(4.1)

\( E_p \) is the initial energy of the proton, \( t_{al} \) and \( t_d \) are the thicknesses of the aluminium metallisation and silicon dead layer respectively, \( \theta \) is the angle of incidence of the proton, \( \frac{dE}{dx} (E_p) \) is the electronic stopping power of protons in material \( x \). The detector is assumed to have 100% charge collection efficiency and non ionising energy losses are ignored. In this situation the proton energy is identical to ionisation energy deposition. If we assume the relationship between channel \( Ch \) of the MCA and the energy deposited in the detector to be \( E = a \times Ch + b \), then the peak ADC channel is given by:

\[ Ch = \frac{1}{a} [E_p - \frac{t_{al}}{\cos \theta} \frac{dE}{dx_{al}} (E_p) - \frac{t_d}{\cos \theta} \frac{dE}{dx_{si}} (E_p) - b] \]  

(4.2)
A plot of peak channel versus angle of incidence was obtained experimentally. The theoretical curve from equation 4.2 is then fit to the experimental curve by adjusting the parameters \( t_{al}, t_d, a \) and \( b \). This task was done using a C++ program written by the author which is based on a genetic algorithm [56]. Genetic algorithms are based on the Darwinian theory of evolution. The program considers an “individual” to be composed of “genes” which in this case are values of \( t_{al}, t_d, a \) and \( b \). A “population” of 10000 “individuals” is generated by randomly selecting a set of genes. A “fitness” \( F \) is assigned to each individual \( j \) based on the sum of squared differences over \( n \) data points between the theoretical and experimental curves \( f_i \) and \( g_i \) respectively;

\[
F_j = \sum_{i=0}^{n} (f_i - g_i)^2
\]  

(4.3)

The population is then ranked in order of increasing fitness. The least fit 20\% of individuals do not reproduce and the remainder of the population “mate” by exchanging genes, i.e. they swap parameters. This evolutionary cycle is iterated until the population converges to within 1\%. The “most fit” individual’s genes are then output to yield the parameters \( t_{al}, t_d, a \) and \( b \).

4.2.2 Results and Discussion

The genetic algorithm resulted in the following parameters: \( a = 1.83 \text{ keV/Ch}, b = 38.5 \text{ keV} \), \( t_{al} = 1.96 \mu m, t_d = 1.26 \mu m \). The corresponding experimental and theoretical curves are illustrated in figure 4.7.

---

\[\text{A least squares method was also used to obtain these parameters. In this case the parameters obtained were: } a = 1.83 \text{ keV/Ch}, b = 45.7 \text{ keV}, t_{al} = 1.67 \mu m, t_d = 1.57 \mu m. \text{ An alternative, and perhaps more accurate, method of calibration would be to implement a silicon detector with a small region of the contact absent. In this case one may calibrate the system by measuring the ADC number corresponding to the centroid of the energy deposition peak for two beam energies.}\]
Figure 4.7: Experimental curve of ADC peak channel number versus angle of incidence of 3 MeV protons. Also shown is the fitted curve used for evaluation of calibration coefficients.

4.3 IBIC Study of Silicon Microdosimeters

4.3.1 Methods

A list of the ion atomic numbers and energies used in the IBIC experiments is shown in table 4.1. These values of LETs were chosen to cover the range relevant to the primary and secondary radiation in hadron therapy. The beam was focussed to a spot size of 1 \( \mu m^2 \) and scanned over an area of 100 \( \times \) 100 \( \mu m^2 \). The microdosimeter array was then moved into the scan area and the position of each \( n^+ \) region within this scan area was noted. Measurements were conducted on the 5 and 10 \( \mu m \) devices at reverse biases of 0, 5, 10, and 20 V. For each IBIC measurement a frequency distribution of energy deposition events was produced along with a 256 \( \times \) 256 pixel image of the median energy event. The median is used in IBIC to reduce sensitivity to outlying events such as those resulting from scattered ions [46].

It is well known that displacement damage induced by IBIC imaging can lead to
Table 4.1: Ion atomic numbers and energies used for IBIC measurements.

<table>
<thead>
<tr>
<th>Ion</th>
<th>Z</th>
<th>( E ), MeV</th>
<th>( LET_{Si} ), keV/( \mu m )</th>
<th>( R_{Si} ), ( \mu m )</th>
</tr>
</thead>
<tbody>
<tr>
<td>proton</td>
<td>1</td>
<td>3</td>
<td>19.7</td>
<td>92.1</td>
</tr>
<tr>
<td>alpha</td>
<td>2</td>
<td>9</td>
<td>97.6</td>
<td>59.1</td>
</tr>
<tr>
<td>carbon</td>
<td>6</td>
<td>25</td>
<td>820</td>
<td>25.7</td>
</tr>
</tbody>
</table>

degradation of minority carrier lifetime over the course of measurement [57]. The mechanism for this damage is the dislocation of silicon nuclei from their lattice positions from elastic scatter interactions with the incident ions. These interstitial atoms introduce mid-level defects into the band gap which act as recombination centres for electron hole pairs. This decreasing minority carrier lifetime results in a decrease in the amount of charge measured. When conducting IBIC measurements a compromise must be made between sufficient statistics to obtain the image whilst minimising damage. To minimise this effect, the ion fluence used for each analysis was limited to the point at which a noticeable change in the frequency distribution occurred. This varied from \( 3 - 32 \, \mu m^{-2} \) for carbon ions to protons. Values of the number \( N \) of ions used for each measurement are shown in Table 4.2. \( N \) decreases with an increasing device thickness and for a given bias owing to higher probability of scatter interactions. The number of defects is related to the cross section for elastic scatter of the incident ions with silicon and the amount of energy transferred to the silicon nucleus. As the mass of the ion approaches that of silicon, the energy transferred to the recoil nucleus increases and the number of silicon recoil defects increases accordingly. This fact illustrates the limited useful lifetime of silicon microdosimeters in hadron therapy applications. This is particularly true for the current device design, for which charge collection occurs predominantly via diffusion. In this case, owing to long collection times, the amount of carrier recombination is sensitive to changes in the carrier lifetime induced by radiation damage. This may be drastically reduced in future designs by utilising fully depleted microdosimeters as suggested by Bradley et
Table 4.2: Number of events per $\mu m^2$ used for IBIC measurements.

<table>
<thead>
<tr>
<th>Ion</th>
<th>Device</th>
<th>0 V</th>
<th>5 V</th>
<th>10 V</th>
<th>20 V</th>
</tr>
</thead>
<tbody>
<tr>
<td>proton</td>
<td>5 $\mu m$</td>
<td>18</td>
<td>20</td>
<td>18</td>
<td>16</td>
</tr>
<tr>
<td>proton</td>
<td>10 $\mu m$</td>
<td>16</td>
<td>16</td>
<td>18</td>
<td>16</td>
</tr>
<tr>
<td>alpha</td>
<td>5 $\mu m$</td>
<td>7</td>
<td>7</td>
<td>7</td>
<td>7</td>
</tr>
<tr>
<td>alpha</td>
<td>10 $\mu m$</td>
<td>5</td>
<td>5</td>
<td>5</td>
<td>5</td>
</tr>
<tr>
<td>carbon</td>
<td>5 $\mu m$</td>
<td>4</td>
<td>4</td>
<td>4</td>
<td>4</td>
</tr>
<tr>
<td>carbon</td>
<td>10 $\mu m$</td>
<td>4</td>
<td>3</td>
<td>3</td>
<td>3</td>
</tr>
</tbody>
</table>

Reducing the number of events in the image results in degraded image statistics. In order to improve statistics it was necessary to reduce the number of pixels in the image to $64 \times 64$ by calculating the median over an $8 \times 8$ pixel matrix. A C++ program was written to analyse this data. The error at each pixel is calculated as the standard deviation of values from the median value. We wish to derive a two dimensional image of the median charge collected for a single diode. To achieved this the entire image is translated to place a single $30 \times 30 \mu m^2$ diode in the corner of the image then cropped to produce an $X \times X$ pixel image of the median energy for a single diode, $E_{exp}(x, y)$. $X$ varies from $23 - 28$ depending upon the ion atomic numbers and energy used.

4.3.2 Results and Discussion

Figures 4.8-4.10 show images of the experimentally measured median energy deposition event as a function of beam position on a single diode of the microdosimeter. For a given beam location the experimentally measured energy deposition is related to the amount of energy deposition in the sensitive volume of the detector. This in turn is related to the thickness and composition of materials above the sensitive volume at this location. That is, the energy of the ions is reduced by the overlying materials, increasing...
the LET of the beam and resulting in an increase in energy deposition as would otherwise occur with no overlayer. Subsequently the recorded energy deposition depends upon the amount of recombination occurring during the collection of the charge created.

Figure 4.8 shows the median energy maps derived from the IBIC measurements using a 3 MeV proton beam incident on the 10 µm device. In this case, owing to the low LET of protons, the amount of energy loss in the overlayer is relatively small and variations in the measured energy deposition are dominated by the variation in charge collection efficiency with location of the ion beam. For applied voltages of 0, 5, and 10 V the image is characterised by a maximum at the centre of the diode with a subtle decrease toward the diode periphery. This effect was modelled by Bradley et al [3] using a semiconductor device simulation tool. In the case of ion strikes in the central region of the device charge collection is dominated by drift. Consequently charge collection times are low in comparison to the mean lifetime of carriers and as a result recombination is minimal. As the location of the ion beam moves further from the central region charge collection is dominated by diffusion owing to the very weak electric field in this region. As the charge collection time becomes comparable to the mean lifetime of charge carriers the amount of recombination also increases.

The maximum value of charge collection, corresponding to ion strikes at the $n^+$ region, increases with applied bias reflecting an increase in depletion layer thickness and a consequent decrease in charge collection time and hence electron hole recombination. The image characteristics change for a bias of 20 V, in which case the maximum median energy event occurs at the corners of the $n^+$ region. The probable mechanism for this enhanced charge collection is the occurrence of high electric fields in these regions at 20 V bias. This however requires further investigation.

Figure 4.9 shows the median energy images obtained with a 9 MeV helium scan of 10 µm device for various device bias. The images are very similar to the case for protons with a more pronounced amplification effect at 20 V.

Figure 4.10 shows the median energy images for 25 MeV carbon scan of 10 µm
Figure 4.8: Experimental IBIC median energy images for $3\text{ MeV}$ proton scan of $10\mu m$ device at bias of $0\text{ V}$ (top left), $5\text{ V}$ (top right), $10\text{ V}$ (bottom left) and $20\text{ V}$ (bottom right).

Figure 4.9: Experimental IBIC median energy image for $9\text{ MeV}$ helium scan of $10\mu m$ device at bias of $0\text{ V}$ (top left), $5\text{ V}$ (top right), $10\text{ V}$ (bottom left) and $20\text{ V}$ (bottom right).
Figure 4.10: Experimental IBIC median energy image for 25 MeV carbon scan of 10 μm device at bias of 0 V (top left), 5 V (top right), 10 V (bottom left) and 20 V (bottom right).

device at bias. Clearly visible in the 0, 5, and 10 V images are areas of increased energy deposition corresponding to the aluminium contact overlying the n+ region and aluminium tracks connecting these contacts. This is a direct result of the higher LET carbon ions losing energy in the aluminium, resulting in a higher LET and hence greater energy deposition in the sensitive region. These results emphasise the need to minimise overlayers in future device design. The amplification effect at 20 V is more pronounced than for 3 MeV protons and 9 MeV helium ions. This is reflected by the scale of the colour bar in figure 4.10.

4.4 GEANT4 Simulations

These IBIC measurements provide information on the charge collected from the device. In order to quantify the charge collection efficiency it is necessary to calculate the amount of charge deposited in the sensitive volume. To obtain this information the measurements were simulated using the GEANT4 Monte Carlo toolkit [43]. Images of
energy deposition for a single diode of the silicon microdosimeter were calculated for the ions and device thicknesses used in experiments.

4.4.1 Methods

Referring to the discussion of the GEANT4 toolkit of Chapter 2, a class derived from the G4VUserDetectorConstruction class was used to define the geometry. A single diode of the array was modelled by combining a system of right angle parallelepiped (RPP) volumes using the boolean (union, intersection or subtraction) operations available with the GEANT4 toolkit. The geometry is illustrated in figures 4.11 and 4.12 and is based on fabrication information and secondary ion mass spectrometry data from Fujitsu Research Laboratories [3]. The SOI wafer is modelled as a cuboid with dimensions $60 \times 60 \times t_{soi} \, \mu m^3$, where $t_{soi} = 5$ or $10 \, \mu m$ (see figure 4.12). A central $n^+$ region of dimensions $8 \times 8 \times t_{As} \, \mu m^3$ is considered where $t_{As} = 0.1 \, \mu m$ is the implantation depth of As ions. $p^+$ regions of dimensions $8 \times 8 \times t_{B} \, \mu m^3$, where $t_{B} = 0.25 \, \mu m$ is the Boron implantation depth, are considered with centres at each corner of the cell as shown in figure 4.11. The LOCOS (local consumption of silicon) silicon oxide layer is modelled as an RPP of thickness $t_{locos} = 0.200 \, \mu m$. A silicon oxide over-layer of thickness $t_{ox1} = 0.4 \, \mu m$ is considered above the SOI wafer. Aluminium contacts penetrating this layer, of dimensions $2 \times 2 \times t_{ox1} \, \mu m^3$, are placed above each $n^+$ and $p^+$ region. Aluminium volumes over these contacts of dimensions $8 \times 8 \times t_{c}$, where $t_{c} = 1 \, \mu m$, are considered and tracks connecting these volumes are described by RPPs of dimensions $11 \times 2 \times t_{c} \, \mu m^3$. Finally a thickness $t_{ox2} = 0.215 \, \mu m$ of silicon oxide is considered over all volumes.

A class was derived from the G4VUserPrimaryGeneratorAction class to model the ion beam as normally incident on the device. A method was added to simulate a raster scan across a $30 \times 30 \, \mu m^2$ area of the diode cell centred on the $n^+$ region as indicated by the white square in figure 4.11. The number of pixels in each scan is identical to the number in the corresponding experimental charge collection image, $X \times X$ from section 4.3. At each pixel, x and y coordinates of 50 ions are randomly sampled from
Figure 4.11: Geometry used for GEANT4 simulations. The white square indicates the simulated IBIC scan area. The dotted line indicates the orientation of CCE profiles.

Figure 4.12: A cross section of the device geometry used for GEANT4 simulations showing the thickness of device regions.
Figure 4.13: Simulated IBIC image for 10 $\mu$m device with 3 $MeV$ protons (top left), 9 $MeV$ helium (top right), and 25 $MeV$ carbon (bottom).

A class was derived from the G4VUserPhysicsList class to implement existing GEANT4 physics models [43] for the transport of each ion through the geometry. Ionisation of atomic electrons and scatter interactions with nuclei are taken into account. A class was derived from the G4UserEventAction class to analyse each ion strike on the device. The energy lost to ionisation in the SOI layer was calculated and output to file along with the beam location. Another C++ application was written to analyse this output file to calculate the median energy event for each pixel $E_{\text{sim}}(x, y)$. The error was determined as the standard deviation of energy events from the median value.

4.4.2 Results and Discussion

Figure 4.13 shows the median energy deposition event at each pixel of the simulated proton, helium, and carbon IBIC scans of the 10 $\mu$m device. As expected there is an...
increase in energy deposition for ion strikes over the \( p^+ \) and \( n^+ \) regions owing to energy loss in the aluminium contacts and highly doped silicon regions and a thicker SOI layer to the rest of the sensitive volume owing to the LOCOS regions. The image contrast increases with increasing LET owing to a greater sensitivity to energy loss in the overlayer structures. For 25 MeV carbon the aluminium tracks connecting the \( n^+ \) contacts are clearly visible as was seen in experimental results of figure 4.10.

4.5 Charge Collection Efficiency Profiles

Given the charge collection images and charge deposition images supplied by IBIC measurements and GEANT4 simulations respectively, it was then possible to calculate the CCE of the microdosimeters for each ion, device thickness, and device bias.

4.5.1 Methods

A charge collection efficiency map is formed from the simulated and experimental median energy images as:

\[
CCE(x, y) = \frac{E_{exp}(x, y)}{E_{sim}(x, y)}
\]

The relative error of the CCE at each pixel is calculated as:

\[
\frac{\delta CCE(x, y)}{CCE(x, y)} = \sqrt{\left(\frac{\delta E_{sim}(x, y)}{E_{sim}(x, y)}\right)^2 + \left(\frac{\delta E_{exp}(x, y)}{E_{exp}(x, y)}\right)^2}
\]

A CCE profile \( CCE(l) \) was then calculated. The orientation of this profile is along a diagonal connecting \( p^+ \) regions as indicated by the dotted line in figure 4.11.

4.5.2 Results and Discussion

Figures 4.14–4.16 show CCE profiles for the 10 \( \mu \)m device at bias of 0, 10, and 20 V. At 0 V the charge collection efficiency profile lies within the range 0.5 – 0.8, increasing to 0.6 – 0.9 for 10 V and 0.7 – 1.1 for 20 V. The CCE of greater than unity for
Figure 4.14: Charge collection efficiency profiles for 10 $\mu$m device at bias of 0 V.

Figure 4.15: Charge collection efficiency profiles for 10 $\mu$m device at bias of 10 V.
carbon ions at $20\,\text{V}$ suggest a high field avalanche multiplication effect. A similar effect in thin epitaxial pn junctions was observed by Knudson et al. [58]. The error in the CCE values increase from carbon to protons reflecting an increase in the elastic scatter cross section for lower $Z$ ions and hence larger variation in path length through the sensitive volume. The CCE values of carbon are seen to exceed those of helium, although the LET is much greater. This may be due to enhanced charge funneling from below the depletion region as at a reverse bias of $10\,\text{V}$ the devices are not fully depleted.

Figures 4.17-4.19 show CCE profiles for the $5\,\text{µm}$ device at bias of 0, 10, and $20\,\text{V}$. CCE values exceed those of the $10\,\text{µm}$ device at the same bias, a result of the increased fraction of SOI layer depleted at the same voltage. The charge amplification effect is greater than observed for the $10\,\text{µm}$ device and values exceeding unity extend approximately $5\,\text{µm}$ from the edge of the $n^+$ region.
Figure 4.17: Charge collection efficiency profiles for 5 μm device at a bias of 0 V.

Figure 4.18: Charge collection efficiency profiles for 5 μm device at bias of 10 V.
4.6 Conclusions

The CCE of a silicon microdosimeter has been calculated for a number of ions, device thickness, and reverse bias. This was achieved by comparison of experimental IBIC measurements with simulations using the Geant4 Monte Carlo toolkit. Values of CCE for the ions studied were seen to agree within statistical uncertainty with a general increase with decreasing device thickness and an increase with increasing device bias. An amplification effect was observed at a reverse bias of 20 V, a possible mechanism for this effect is high field avalanche multiplication at corners of the depletion region.

Recent studies into charge collection in SOI structures have shown that charge generated below the isolating oxide layer may induced displacement currents across the oxide leading to additional charge collection [59]. These effects were related to the oxide thickness, with an increase in induced current with a decrease in oxide thickness. This effect must be considered in the design of future SOI microdosimeters as the oxide must be sufficiently thick to avoid these effects.
A potential source of systematic error in this method of CCE calculation is from uncertainty in device composition and geometry and the approximation with a system of RPP volumes. A more accurate method of geometry definition is possible with GEANT4 by importing solid definitions from STEP compliant CAD systems.

The implications of this study for actual silicon microdosimetry measurements are that although energy deposition events must be corrected by the CCE of the device, it is not necessary to apply an LET dependent correction factor for hadron therapy applications. Moreover the charge collection images confirmed that the thickness of device overlayers should be minimised in the next generation devices as was suggested by Bradley [3]. As the amplification effect did display LET dependence, regions of high electric field should also be avoided so that amplification effects do not complicate the charge collection process. However such an effect could be used to increase the sensitive of the microdosimeter to low energy deposition events such as those resulting from high energy protons or photoelectrons which would otherwise fall below the noise threshold of electronics.

Extensive semiconductor device simulation of future devices should be performed to gain insight into the physical origins of the multiplication effect and to study the importance of induced displacement currents. The design of these devices should also follow the recommendations made by Bradley [36]. In particular the design should facilitate the collection of charge by drift alone to ensure 100 % charge collection and to obtain a cubic charge collection volume to reduce the angular dependence of the measurement. Doing so will allow one to apply a CCE correction factor independent of ion LET and location of ion strike on the device.
CHAPTER 5
A PULSE SHAPE DISCRIMINATION TECHNIQUE FOR SILICON MICRODOSIMETERS

5.1 Introduction

Work in chapter 4 and previous work by Bradley et al. [40] showed that charge collection in the silicon microdosimeter was complicated by the lateral diffusion of charge from outside the depletion region. Reference [3] suggests the fabrication of devices with lateral isolation of each diode unit as per reference [60]. It may be possible however to use the existing devices and exploit the difference in timing properties of the charge collection mechanisms. If it were possible to ignore ion strikes occurring outside the depletion region the microdosimeter performance may be improved.

Pulse shape discrimination (PSD) techniques involve measurement of the timing properties of signals produced by radiation detectors. Considerable work has been performed in the field of particle identification and the reader is directed to references [31, 61, 32] and references therein. These studies involved the simultaneous measurement of the amplitude and rise time of the preamplifier signal for fully depleted p-i-n diodes. Theoretical models have related this rise time to the plasma decay time, the transit time of charge carriers, and to a lesser extent, the intrinsic rise time of the charge sensitive preamplifier [31]. In this case the amplitude of the pulse from the amplifier is indicative of the particle energy and the charge collection time is indicative of the particle range. By plotting a 2D intensity plot of the number of counts for a particular energy and charge collection time \( N(E, T) \) one may separate the contributions from different particles. As a result, the incident particle’s energy, atomic number, and in some cases atomic mass could be determined.

Theoretical simulations of Bradley [3] looked at charge collection following 5 MeV alpha strikes for various strike locations on the microdosimeter. These studies showed the duration of the resulting current pulse increased with the distance of the ion strike from
the $n^+$ region as a result of larger transit times of charge carriers for diffusion versus drift. If this pulse is integrated on a charge sensitive preamplifier, the duration of the current pulse is directly related to the rise time of the voltage pulse produced by the charge sensitive preamplifier.

For the planar geometry of the silicon microdosimeters, the rise time of the preamplifier is expected to be indicative of the distance of ion strike from the depletion region. The aim of these experiments was to investigate the correlation between preamplifier rise time and ion strike position then, using suitable electronics, to discriminate charge collection events based on this timing information. In this manner energy deposition events resulting from ion strikes outside the depletion region could be ignored. As these events are subject to high recombination this is expected to improve the spectroscopic properties of the microdosimeter.

5.2 Timing Measurements

5.2.1 Methods

The aim of the first experiment was to investigate how the rise-time of the preamplifier signal varied with the distance of the ion strike from the pn junction. The IBIC electronics described in section 4.2 were used for these experiments. The rise-time of the voltage pulse produced by the preamplifier was measured with a technique similar to that used by Pausch et.al [61] and is illustrated in the schematic of Figure 5.1. The A250 preamplifier output signal is passed to a chain of two timing filter amplifiers (TFAs). The timing filter amplifiers used were ORTEC model 454 and had integration and differentiation times of $\tau_{int} = 5\, ns$ and $\tau_{diff} = 200\, ns$ and gains of $\times 5$ and $\times 2$. The differentiation produces a bipolar signal as illustrated in Figure 5.1. It can be shown the zero-crossing time, referred to herein as $T_{zc}$, of the resulting bipolar signal is a monotonic increasing function of the rise-time of the charge sensitive preamplifier [61].

The bipolar signal is split and fed into two constant fraction discriminators (CFDs). A constant fraction discriminator produces a logic signal when the input signal reaches a
Figure 5.1: Circuit diagram for zero-crossing time measurements; TFA: Timing filter amplifier, CFD: constant fraction discriminator, DAQ: Data Acquisition System. The voltage signal at various stages of the circuit are illustrated.
constant fraction of it’s amplitude. The general layout of the device is given in figure 5.2. Essentially it splits the signal and passes one through a delay into the non inverting input of an operational amplifier and a second through an attenuator into the inverting input of an op-amp [62]. The output of the op-amp is then passed to a zero crossing discriminator which triggers as the pulse crosses zero. If we consider the input pulse to be:

\[ V(t) = \frac{V_o}{t_r} , \quad (t < t_r) \quad (5.1) \]
\[ V(t) = V_o , \quad (t \geq t_r) \quad (5.2) \]

Then the attenuated and inverted signal will be given by:

\[ V_a(t) = -\frac{V_o t f}{t_r} , \quad (t < t_r) \quad (5.3) \]
\[ V_a(t) = -f V_o , \quad (t \geq t_r) \quad (5.4) \]

Where \( f \) is the constant fraction, set at 0.2 for commercial CFDs. The delayed pulse is given by:

\[ V_d(t) = 0 , \quad (t < t_d) \quad (5.5) \]
\[ V_d(t) = \left( \frac{t - t_d}{t_r} \right) V_o , \quad (t_d \leq t < (t_d + t_r)) \quad (5.6) \]
\[ V_d(t) = V_o , \quad t \geq (t_d + t_r) \quad (5.7) \]

If the delay time is greater than the risetime ie. \( t_d > t_r \) then the zero crossing time and hence the logic pulse of the CFD occurs for:

\[ -f V_o + \left( \frac{t - t_d}{t_r} \right) V_o = 0 \quad (5.8) \]
\[ \Rightarrow t = t_d + f t_r \quad (5.9) \]

This is independent of the pulse height. For the case of \( t_d < t_r \) however, zero crossing occurs for:
$$-\frac{V_o ft}{t_r} + \frac{V_o (t - t_d)}{t_r} = 0$$  \hspace{1cm} (5.10)

$$-ft + t = t_d$$  \hspace{1cm} (5.11)

$$\Rightarrow t = \frac{t_d}{1 - f}$$  \hspace{1cm} (5.12)

In this case the zero crossing time is independent of rise time and depends on the constant fraction setting and the delay time. When configured in this way the CFD is said to be operating in rise time compensation mode. The constant fraction discriminators used were ORTEC model 935 Quad CFD. One of the CFDs was operating in rise time compensation mode with an external delay of approximately $2\, {\text{ns}}$ provide by $10\, {\text{cm}}$ of cable. This $t_d$ value was selected so the logic pulse occurs just as the leading edge of the bipolar signal rises above the background noise. The second CFD was configured as a simple zero crossing discriminator by having an infinite delay. The threshold settings of both CFDs was decreased to a minimum at which point both CFDs would trigger on the noise. With no input signal and with a sample bias applied to simulate the experimental noise levels, both were increased until neither module produced a logic pulse. The constant fraction discriminator output logic pulses are used as start and stop signals for a time-to-amplitude converter (TAC).

The TAC used was an ORTEC model 567 TAC/SCA. The range of the TAC was set at $50\, {\text{ns}}$ with multiplication factor of $\times 10$. This results in the full scale of measurement of the TAC being $500\, {\text{ns}}$. The output of the TAC is a logic pulse with amplitude proportional to the time difference between the start and stop input signals. In this situation this time difference is proportional to the zero crossing time of the bipolar pulse. The TAC output signal $T_{zc}$ was fed into a channel of the microprobe data acquisition system along with the ion beam coordinates. During measurements a file of data triplets ($T_{zc}, x, y$) was accrued and an event histogram was formed. In order to correlate zero-crossing time with ion strike position a window of the spectrum was selected and a two dimensional image of the number of events occurring at each pixel of the scan and lying within the window was formed (ie. $N(T_{zc} + \Delta T_{zc}, x, y)$).
To test the zero crossing time circuitry a BNC model “BH-1 tail pulse generator” was used with repetition frequency of $1 \, kHz$, rise-time of $\tau_{rise} = 20 \, ns$, fall time of $\tau_{fall} = 100 \, \mu s$, an amplitude setting of $5.35$ and an attenuation factor $\times 10$.

A $25 \, MeV \, C^{+4}$ beam was used for the experiments. The beam was first focussed to a spot size of $1 \, \mu m$ using the procedure of section 3.3.1 and scanned over an area of $100 \times 100 \, \mu m^2$. The $10 \, \mu m$ SOI device was brought into the scan area and a reverse bias of $10 \, V$ was applied.

5.2.2 Results and Discussion

Figure 5.3 shows the total spectrum of $T_{zc}$ events measured and is composed of two peaks. Figure 5.4 shows the number of ion strikes registered at each pixel of the scan which lie within the given window of the $T_{zc}$ spectrum of Figure 5.3. The results clearly show the zero crossing time $T_{zc}$, and hence the rise-time, increases as the distance of the ion strike from the $n^+$ region increases. This is a direct result of the corresponding
Figure 5.4: $N(T + \Delta T_{zc}, x, y)$ maps for $T_{zc}$ windows of the spectrum of Figure 5.3. The white bar in the first map indicates a distance of 10 $\mu m$. 
increase in charge collection time. The charge collection time depends upon the mechanisms of charge collection in the device. For ion strikes close to the pn junction, the motion of charge carriers is dominated by drift under the influence of the electric field of the depletion region and charge collection is rapid. As the ion strike moves further from the junction the electric field magnitude decreases and charge collection is dominated by the diffusion of charge carriers. The charge carriers must first diffuse some distance to the depletion region prior to being collected by the electric field. As the ion strike moves further from the junction this diffusion distance increases resulting in longer collection times. From the results obtained, ion strikes are confined to the geometrical boundaries of the $10 \times 10 \mu m^2$ $n^+$ region for zero crossing times $T_{zc} \leq 284 \, ns$.

5.3 Pulse Shape Discrimination

5.3.1 Methods

To preclude non-junction strikes, a window incorporating zero-crossing times $0 \, ns < T_{zc} \leq 284 \, ns$ was set on the single channel analyzer of the TAC as shown in figure 5.5. The single channel analyzer provides a logic pulse to be used as the “enable” signal of a linear gate, through which the output of the spectroscopy amplifier is passed. The function of the spectroscopy amplifier is to shape the signal from the charge sensitive preamplifier to provide a pulse suitable for the multi channel analyser of the data acquisition system and with a pulse height proportional to energy deposition.

Using this circuit, only energy events corresponding to $T_{zc} < 284 \, ns$ would pass through the linear gate and be registered by the data acquisition system along with the beam coordinates. Energy deposition event spectra were acquired with and without the pulse shape discrimination system. Additionally, a two dimensional image of the number of events occurring at each pixel of the scan with the PSD system $N(x, y)$ was constructed.
Figure 5.5: Circuit used to gate the energy signal with a pulse shape discrimination technique. SCA: single channel analyser, SA: Spectroscopy amplifier, TAC: Time-to-amplitude converter.

Figure 5.6: Charge collection spectra obtained with (solid line) and without (broken line) pulse shape discrimination.
5.3.2 Results and Discussion

Figure 5.6 shows the total energy spectrum before and after the pulse shape discrimination technique is implemented. The low energy component of the spectrum is reduced and the energy deposition spectrum is close to Gaussian. Figure 5.7 confirms the position of all ion strikes to be within the geometrical boundaries of the $n^+$ region. The white bar indicates a distance of 10 $\mu$m. Integrating the areas of the gated and ungated spectra of figure 5.6 yields a ratio of $R_{exp} = \frac{N_{gated}}{N_{ungated}} = \frac{2727}{16281} = 0.17$. The IBIC scan has a total area $A_{scan}$ of approximately $100 \times 100 \mu m^2$ and includes approximately 15 $n^+$ regions, each having an area of $10 \times 10 \mu m^2$. Hence based on geometry alone, ignoring effects of lateral spreading of the depletion region, the expected ratio of accepted to rejected counts is $R_{geom} = \frac{A_{n^+}}{A_{scan}} = \frac{10^2 \times 15}{10^4} = 0.15$, in good agreement with the experimentally observed ratio.

5.4 Angular Dependence of Timing Measurements

5.4.1 Methods

In a practical situation ions may strike the device with varying angles of incidence. If this pulse shape discrimination technique is to be used to ignore ion strikes outside the depletion region it is essential that the $T_{zc}$ spectrum does not change with the angle of irradiation. An experiment was performed to determine the angular dependence of the
$T_{zc}$ spectrum. Experimental conditions were identical to those described in section 5.2.1. In this situation a spectrum of zero crossing time events $f(T_{zc})$ was acquired for angles of irradiation of 0, 15, 30, 45, and 60°.

5.4.2 Results and Discussion

Figure 5.8 shows zero crossing time spectra $f(T_{zc})$ for varying angles of irradiation. These results show the low $T_{zc}$ peak remains unchanged for angles of up to 30° yet the high $T_{zc}$ peak shifts considerably to higher values of $T_{zc}$. For angles of 45° and greater the low $T_{zc}$ peak shifts considerably to higher values of $T_{zc}$. The increase in zero crossing times can be explained in terms of the increase in plasma decay times. For angled strikes, a higher ion LET can be expected due to an increased energy loss in the overlayer. This increased LET leads to an increase in the plasma decay time and hence $T_{zc}$. The ratio of high to low peak areas decreases with angle reflecting the decrease in the ratio of areas (from the beam’s perspective) of non-depleted to depleted regions of the device.

5.5 Conclusions

The timing properties of a silicon-on-insulator microdosimeter for medical and space applications have been studied using an ion microprobe. This study showed that the zero crossing time $T_{zc}$ parameter, closely related to charge collection time, increased as a function of radial distance of ion strike from each pn junction. These timing properties were used to implement a pulse shape discrimination technique to render the microdosimeter insensitive to ion strikes outside the $n^+$ region.

A technique for conducting time resolved IBIC was recently applied to an ion microprobe by Vizkelethy et.al. [63, 64]. In this technique a high speed ADC is used to digitise the preamplifier output, from which detailed information on electron and hole charge collection times, mobility and lifetime can be derived. The relatively long time to digitise the waveform (0.1 – 1 s) means a beam deflector must be used to switch off the
beam during digitisation and generating 2D images is not feasible owing to long acquisition times. The technique outlined in this thesis has the advantage of acquiring data at high count rates; however, information stored in the rise of the preamplifier is lost during the pulse processing.

The ability of the pulse shape discrimination technique described in section 5.3 to ignore ion strikes outside the sensitive volume is limited by a number of factors including: a dependence of Zero Crossing Time fluctuations on signal amplitude and hence ion LET, the intrinsic rise time of the charge sensitive preamplifier places a lower limit on the measurable zero crossing time, the variation of zero crossing time with the angle of incidence, and the LET dependence of plasma decay time [65].

It is not recommended that this technique, in its present form, be applied to silicon microdosimetry measurements. It is more practical to implement lateral isolation techniques in future silicon microdosimeter design as proposed by [36]. However, the notion
of extracting additional information from the timing properties of the charge collection signal should not be ignored. One possibility which deserves further investigation is to utilise the silicon microdosimeters for particle identification.
CHAPTER 6
VERIFICATION OF MONTE CARLO CALCULATIONS IN FAST NEUTRON THERAPY

An aim of this thesis is to demonstrate the ability of silicon microdosimeters to verify Monte Carlo calculations in Hadron Therapy. This chapter outlines preliminary investigations using GEANT4 Monte Carlo simulations of silicon microdosimetry measurements in Fast Neutron Therapy [66].

6.1 Introduction

Microdosimetry has been used in Fast Neutron Therapy (FNT) to compare the radiobiological properties of different facilities [67] based on differences in measured microdosimetric spectra. As proposed in section 1.7 silicon microdosimeters may be incorporated into the Monte Carlo calculations used by future treatment planning systems. In this situation the role of silicon microdosimetry in FNT can be revised. By comparing the experimental and theoretical energy deposition spectra one can refine the model of the primary beam, components of the geometry such as beam modifying devices, the patient, or the detector, or the physical models of the interaction of the former with the latter. When applied in this manner the tissue equivalence of the detector is not required.

As a preliminary investigation into this method of application, this chapter aims to simulate silicon microdosimetry measurements performed by Bradley et.al [3]. These measurements were conducted at the Gershenson Radiation Oncology Centre, Detroit, USA. This facility uses a superconducting cyclotron to accelerate deuterons to $48.5\ MeV$ prior to impinging on a beryllium target. This produces a neutron beam with a maximum energy of $52.5\ MeV$ and a mean of approximately $20\ MeV$. It is shaped using tungsten multi-leaf collimators. During the experiment, a probe containing the microdosimeter was placed at depth in a water phantom which was irradiated by the neutron beam. Ionisation energy deposition events resulting from subsequent secondary charged particles were monitored and stored as an MCA spectrum. These measurements used the $2\ \mu m$
Figure 6.1: Comparison of microdosimetry spectra for 2 \( \mu m \) and 10 \( \mu m \) SOI devices with TEPC. For results on the left the mean chord length has been scaled to give same dose mean lineal energy as the TEPC. For results on the right the mean chord lengths have been scaled to give the same proton peak position (taken from reference [3]).

Device at 10 cm depth on the central axis of the beam and the 10 \( \mu m \) device at depths of 10 cm and 2.5 cm.

The purpose of these measurements was to compare microdosimetry spectra measured with the silicon microdosimeter to measurements performed with a 2 \( \mu m \) simulated diameter TEPC. The energy deposition event spectra recorded by the silicon microdosimeter were converted to dose weighted lineal energy spectra as described in section 1.3.2.

After scaling of the mean chord length the microdosimetric spectra were seen to reproduce certain characteristics of the TEPC spectrum as shown in figure 6.1. Differences in the spectra are a result of: the chord length distribution in the devices, the fractional variation of charge carriers generated, and the different compositions of the detectors leading to different ionisation energy losses in the sensitive volumes. In an attempt to correct these differences two methods of mean chord length scaling were used: the first scaled the mean chord length to give the same dose mean lineal energy as the TEPC and the second scaled the mean chord length to give the same peak position. This tissue
equivalence correction is valid only if secondary particles resulting from neutron interactions in the silicon volume can be ignored. This work suggested further investigation into the importance of neutron interactions to the microdosimetry spectrum. The primary aim of this chapter is to simulate the silicon microdosimetry measurements in order to investigate the feasibility of using such measurements to verify Monte Carlo calculations. A secondary goal is to use these simulations to study the importance of neutron interactions in the non tissue equivalent volume.

6.2 Methods

The simulation was written in C++ using classes which inherit behaviour from base classes of the GEANT4 Monte Carlo toolkit [43]. These classes were used to model different aspects of the simulation such as: the geometry, the primary beam, the physics of interaction, and actions carried out at the end of each particle history for analysing events. The geometry used in the simulation is given in Fig. 6.2 and follows closely the geometry of the experiment. The ion beam induced charge collection experiments of chapter [4] and reference [40] showed the charge collection efficiency of the microdosimeter varies with the location of the ion strike on the device. At an operating bias of $10V$ the average Charge Collection Efficiency (CCE) was found to be 0.8 for ions with LET in the range $20 - 820 \, keV/\mu m$. The silicon sensitive volume is consequently modelled as a single Right Angled Parallelepiped (RPP) of dimensions $4800 \times 1600 \times 10 \, \mu m^3$ with a charge collection efficiency of 0.8. The geometry of the device over-layer is simplified to a $SiO_2$ RPP of lateral dimensions identical to the sensitive volume and with a thickness of $1 \, \mu m$. The $300 \, \mu m$ air gap of the microdosimeter probe was modelled, as was the $3.5 \, mm$ perspex converter, $0.4 \, mm$ thick aluminium shield, $6 \, mm$ probe holder, and a $25 \, mm$ thickness of water corresponding to the depth of measurement. The lateral dimensions of these volumes were twice the width of the silicon sensitive volume.

\footnote{The reader should note that the complicated device geometry as modelled in chapter [4] is simplified. Also the variation of CCE with the location of ion strike on the device was ignored. This was done to conserve calculation times and memory consumption.}
All materials are defined in terms of elemental compositions and in turn all elements are defined in terms of their isotopic composition.

The field size of the neutron beam used in the experiment was $10 \times 10 \text{cm}^2$; however, for the simulation this was not feasible owing to a restriction on computation times. The neutron beam was modelled to be normally incident on the surface of the water phantom with lateral dimensions twice that of the microdosimeter array. The neutron energy distribution for the Detroit facility was obtained from a study by Kota et.al. [4] and is shown in Fig. 6.3. For each neutron history of the simulation the initial energy was randomly sampled from this distribution. The initial neutron position is randomly sampled on the surface of the phantom. This model of the primary beam ignores the production of the neutron beam and the interaction of the beam with modifying components. The ramifications of this assumption, in terms of the underestimation of the gamma component of the radiation field, is discussed later. A total of $10^8$ neutrons were used for each simulation.
Figure 6.3: Neutron energy distribution at the surface of the phantom at the Detroit FNT facility (taken from reference [4]).

Each neutron is transported (see Fig. 6.4) taking into account elastic scattering on nuclei. In the neutron energy range \( E > 19.9 \text{MeV} \) cross sections and end states are calculated using the G4LElastic model. For energies in the range \( E < 19.9 \text{MeV} \) pointwise evaluated nuclear cross section data is used and end states are calculated with the G4NeutronHPElastic model. A pre-equilibrium decay model, the G4PreCompoundModel, is used to model the inelastic scattering of neutrons with target nuclei in the energy range \( E > 19.9 \text{MeV} \). For \( E < 19.9 \text{MeV} \) the point-wise evaluated cross section data is used with the G4NeutronHPInelastic model. The final states considered for inelastic reactions are \((nA \rightarrow n\gamma s)\) (discrete continuum), \(np, nd, nt, n^3He, n\alpha, nd2\alpha, nt2\alpha, n2p, n2\alpha, np\alpha, n3\alpha, 2n, 2np, 2nd, 2n\alpha, 2n2\alpha, nX, 3n, 3np, 3\alpha, 4n, p, pd, p\alpha, 2pd, d\alpha, d2\alpha, dt, t, t2\alpha, ^3He, \alpha, 2\alpha, \) and \(3\alpha\). Capture reactions are treated similarly, for \( E > 19.9 \text{MeV} \) the G4LCapture is used and for \( E < 19.9 \text{MeV} \) the G4NeutronHPCapture model is used. Gamma photons are also transported through the simulation geometry. Interactions of gammas with atomic electrons are modelled with the G4ComptonScattering model for Compton scattering and the photoelectric effect with the G4PhotoElectricEffect model.
Figure 6.4: View from within water phantom showing several neutron histories (green) and recoil protons (blue) generated in water phantom.

Electrons are transported through the geometry taking into account scattering on atomic electrons G4MultipleScattering the model, ionisation of atomic electrons using the G4eIonisation model, and Bremsstrahlung losses using the G4eBremsstrahlung model.

Protons are transported taking into account elastic scatter interactions with the G4MultipleScattering model. Ionisation of atomic electrons is also modelled using the G4hLowEnergyIonisation model. Recoiling nuclei produced in elastic scatter interactions and reaction products from inelastic reactions also use these models. A pre equilibrium decay model, the G4PreCompoundModel, is used to model the inelastic scattering of protons. A complete description of the physics models used in the simulations may be found in reference [43].

If a charged particle is produced with a residual range in the material less than a default cut value of 10\(\mu\)m then the particle is not tracked. In this case the particle is assumed to deposit it’s entire energy at the point of generation.

If a charged particle traverses the sensitive volume of the detector then the ionisation energy lost in the sensitive volume is calculated. Energy lost in the sensitive volume is assumed to be equal to energy deposited and the energy deposition event is tallied. In
this situation the statistics of electron hole pair generation are ignored. For each particle crossing the sensitive volume the atomic number and mass of the particle is tallied. Whether the particle stopped in, crossed, or started in the sensitive volume is also determined.

An application was written in C++ to analyse the output of these simulations. A frequency distribution of all energy deposition events, $f_j$, was formed. This distribution was convolved with a Gaussian distribution of variance $\sigma = 5 \text{ keV}$ to allow for the electronic noise of the detector electronics:

$$f_j = \sum_{i=j-3\sigma}^{j+3\sigma} \frac{1}{\sqrt{2\sigma^2\pi}} f_i \exp\left(-\frac{(\epsilon_j - \epsilon_i)^2}{2\sigma^2}\right)$$  \hspace{1cm} (6.1)

Where $\epsilon_i$ is the energy value corresponding to bin $i$ of the frequency distribution.

As the magnitude of energy deposition can span several orders of magnitude the linear binned spectrum was logarithmically binned with 10 bins per decade, covering energy deposition from $1 \text{ keV} - 5 \text{ MeV}$.

### 6.3 Results and Discussion

Experimental results are only observed above the low noise threshold of $10 \text{ keV}$ of the microdosimeter probe. The AMPTEK Pocket MCA-8000 used in these experiments is known to exhibit anomalous behaviour at high count rates of pulses with amplitude below the threshold level. This results in pileup of these pulses which register as counts above the threshold. Experiment and simulation results are normalised accordingly to events greater than $20 \text{ keV}$, twice the threshold level. Fig. 6.5 shows the experimental and simulated spectra of energy deposition events for the $2 \mu m$ device. Fig. 6.6 shows spectra formed for several components of the secondary charged particle field to understand the relative contribution to the total spectrum. The spectrum is dominated by recoil protons with a smaller contribution from carbon and other recoiling nuclei (some species omitted for clarity). A continuum corresponding to recoil silicon nuclei is clearly evident. At
the peak of the distribution, the ratio of silicon recoil events to recoil proton events is approximately $1 \times 10^{-3}$.

To study the importance of neutron inelastic reactions in the silicon volume a spectrum is formed for particles which started in, stopped in, or crossed the microdosimeter. Results are shown in Fig. 6.7. These results show a significant contribution from starting and stopping particles, the significance of which is greater for higher energy deposition events. At the bin corresponding to the peak position, the ratio of stopping and starting events to crossing events is approximately $1 \times 10^{-3}$.

Fig. 6.8 shows the experimental and theoretical spectrum of energy deposition events for the 10 $\mu$m device. Again the results are normalised to the total number of events above 20 keV. In this situation there is significant discrepancy between experimental and theoretical curves in the interval 10 – 50 keV. Again this spectrum is decomposed into relative spectra for each component of the charged particle field (Fig. 6.9). This spectrum is dominated by recoil protons with a smaller contribution from carbon ions and other
Figure 6.6: Energy deposition event spectrum for 2 $\mu m$ SOI decomposed into spectra for each atomic number.

Figure 6.7: Energy deposition event spectrum for 2 $\mu m$ SOI device decomposed into spectra for crossing, starting, and stopping particles.
recoils. The recoil silicon continuum is greater than the $2 \mu m$ device. At the bin corresponding to the peak position, the ratio of silicon recoil events to recoil proton events is approximately $1 \times 10^{-2}$. This effect is a result of the increased device thickness and hence higher probability of neutron interaction in the sensitive volume. Additionally the relative contribution of carbon recoils is less than that for the $2 \mu m$ device. The discrepancy observed at lower energy deposition events may be attributed to the pileup of sub-threshold events. It may also be due to the neglect of the gamma component of the radiation field, mostly due to the fact that the production and modification of the primary beam were not modelled.

Again spectra were formed for particles which started in, stopped in, or crossed the microdosimeter. Results are shown in Fig. 6.10. At the bin corresponding to the peak position, the ratio of stopping and starting events to crossing events is approximately $1 \times 10^{-2}$. These results show an increase in the significance of starting and stopping particles compared to the $2 \mu m$ device.
Figure 6.9: Energy deposition event spectrum for 10 $\mu$m SOI device decomposed into spectra for each atomic number.

Figure 6.10: Energy deposition event spectrum for 10 $\mu$m SOI device decomposed into spectra for crossing, starting, and stopping particles.
6.4 Conclusions

Simulations of silicon microdosimetry measurements in FNT were performed using the GEANT4 Monte Carlo toolkit. Simulation results were seen to compare favourably with experimental results. In future a quantitative index of agreement should be employed. The choice of such method is complicated; one may choose a statistical test such as the Kalingorov test or some other criterion. In the case of hadron therapy however, uncertainties in other aspects of the treatment planning process (such as the biological model used or in the acquisition of patient data) may ‘smear out’ uncertainties in the radiation field. A discrepancy was observed for the 10 µm device which is most likely due to the pileup of sub-threshold events. Future simulations should be performed to model the beam production and modification in detail and to investigate the anomalous behaviour of the MCA near the threshold. To confirm the notion of pile up of sub-threshold events, a pileup model should be incorporated into the simulation. Future simulations should also model the production and collimation of the primary beam to investigate the importance of photoelectrons to the measured spectrum. Future simulations should also attempt to model the device geometry in as much detail as in Chapter 4.

Simulation results were analysed to study the importance of particles originating from neutron interactions in the silicon volume. This was seen to be significant for both 2 µm and 10 µm results. If the silicon microdosimeters are being applied in the conventional sense of microdosimetry, these results show this effect may inhibit its ability to provide tissue equivalent measurements. However if the silicon microdosimeters are being applied in the sense of verifying the Monte Carlo calculation then neutron interactions in the silicon volume are not important.

This study suggests how a non-tissue equivalent detector may be used for the verification of Monte Carlo calculations in fast neutron therapy. Future experiments should be performed in heterogeneous phantoms with materials of composition similar to tissues.
CHAPTER 7
VERIFICATION OF MONTE CARLO CALCULATIONS IN PROTON THERAPY

7.1 Introduction

Microdosimetry has been used in Proton Therapy (FNT) to investigate RBE variations in water phantoms \[68\]. Experimental silicon microdosimetry measurements in proton therapy were first performed by Bradley et.al. \([5, 3]\) at the Northeastern Proton Therapy Center (NPTC), Boston. The facility uses a cyclotron to accelerate a proton beam to an initial energy of \(230\, MeV\). The modulation of the beam energy is varied by passing the beam through a filter to create the desired spread in the Bragg peak. A double scattering system using a fixed scatterer and a second non-uniform thickness scatterer produce a beam with uniform intensity in the lateral profile over a \(10 \times 10\, cm^2\) field size. The microdosimeter was placed at a number of depths along the bragg peak in a water phantom and spectra of energy deposition events were acquired. These depths are indicated in figure 7.1 in relation to the depth dose distribution of the proton beam. Measurements were performed with \(10\, \mu m\) thick SOI devices to maximise the signal to noise ratio. Figure 7.2 shows dose weighted microdosimetry spectra at each depth of measurement. An analysis of these results was given in reference \([5]\) in terms of the changing primary proton beam energy, the changing chord length distribution due to scatter with depth, and the generation of secondary charged particles. These results demonstrated the high resolution of the silicon microdosimeter and the ability to operate at beam currents typical of therapy without serious pileup effects.

The aim of this chapter is to simulate this experiment using the GEANT4 Monte Carlo toolkit. By drawing a comparison between simulation and experimental results one may gauge the accuracy of the theoretical model of the primary beam, the phantom, the silicon microdosimeter and it’s response charge collection properties.
Figure 7.1: Depth dose data for proton beam used in experiments as measured by Markus ionisation chamber (taken from reference [5]). Depths of measurements account for water equivalent thickness of microdosimeter probe.

Figure 7.2: Microdosimetry spectra at various depths in a water phantom (taken from reference [5]).
7.2 Simulation of Beam Modifying Devices

Owing to a restriction on computation resources, it was necessary to separate the simulation of these measurements into two parts: the first simulated the passage of the primary beam through the beam modifying devices and calculated the distribution of transmitted proton energies, the second used the energy spectrum derived from the first to simulate the interaction of the beam with the phantom and microdosimeter probe.

7.2.1 Methods

Referring to the discussion of the GEANT4 toolkit [2] a class derived from the G4VUserDetectorConstruction class was used to define the geometry. A simplified model of the beam modifying devices is formed by a stack of RPP volumes as shown in figure 7.3. The materials used in this simulation are summarised in table 7.1.

\footnote{This represents a major simplification of the geometry and future simulations should follow the geometry more closely.}
Table 7.1: Beam modifying devices used in the PT simulations.

<table>
<thead>
<tr>
<th>Component</th>
<th>Material</th>
<th>$\rho$ , g cm$^{-2}$</th>
<th>Stoichiometry</th>
<th>t, mm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Scatterer</td>
<td>Lead</td>
<td>11.35</td>
<td>100 % Pb</td>
<td>1.31</td>
</tr>
<tr>
<td>Range Modulator</td>
<td>Lead</td>
<td>&quot;</td>
<td>&quot;</td>
<td>4.252</td>
</tr>
<tr>
<td>Range Modulator</td>
<td>Graphite</td>
<td>2.0</td>
<td>100 % C</td>
<td>0.7753</td>
</tr>
<tr>
<td>Beam Spreader</td>
<td>Water</td>
<td>1.0</td>
<td>$H_2O$</td>
<td>36.5</td>
</tr>
<tr>
<td>Ion Chamber</td>
<td>Water</td>
<td>&quot;</td>
<td>&quot;</td>
<td>3.04</td>
</tr>
<tr>
<td>Nozzle</td>
<td>Air</td>
<td>0.00129</td>
<td>70 % N, 30 % O</td>
<td>2550</td>
</tr>
<tr>
<td>Phantom Wall</td>
<td>Perspex</td>
<td>1.17</td>
<td>H 8 %, O 32 %, C 0.6 %</td>
<td>10</td>
</tr>
</tbody>
</table>

For the first simulation, a class was derived from `G4VUserPrimaryGeneratorAction` to model the primary beam as a pencil beam normally incident on the surface of the lead scatterer with a gaussian distribution in energy characterised by an average $\mu = 230.5 \, MeV$ and standard deviation $\sigma = 0.41 \, MeV$ \[3\]. $10^8$ primary protons were simulated.

A class was derived from the `G4VUserPhysicsList` class to define the physics of proton interaction with the beam modifying devices. The models used are identical to those for chapter \[6\]. Ions produced in inelastic reactions are only tracked if their residual range in the material is greater than 1 $\mu$m. Electrons and photons however have a cut value of 10 $cm$.

A class was derived from the `G4UserEventAction` class to analyse hits to the sensitive volume. The energies of each proton traversing an area on the beam axis at the surface of the water phantom were recorded and histogrammed.

7.2.2 Results and Discussion

The result of the simulation of the primary beam through the beam modifying devices is illustrated in figure \[7.4\] along with a gaussian fit to the distribution. The
gaussian fit is characterised by a mean of $\mu = 192.4 \text{ MeV}$ and a standard deviation of $\sigma = 1.24 \text{ MeV}$.

### 7.3 Simulation of Phantom and Microdosimeter

#### 7.3.1 Methods

A class derived from the G4VUserDetectorConstruction class was used to define the phantom and microdosimeter geometry. This geometry is shown in figure 7.5 and summarised in table 7.2. As shown in section 4, the charge collection efficiency of the microdosimeter varies with position on the device. At this operating bias the average CCE was found to be 0.8 for the range of LETs 20—820 $keV/\mu m$. The silicon sensitive volume is consequently modelled as a single right angled parallelepiped of dimensions $4800 \times 1600 \times 10 \mu m^3$ with charge collection efficiency of 0.8. In chapter 4, the complicated overlayer geometry was modelled in detail. In this simulation however, it is simplified to
Table 7.2: Phantom and microdosimeter probe components used in the PT simulations.

<table>
<thead>
<tr>
<th>Component</th>
<th>Material</th>
<th>$\rho, \text{ g cm}^{-2}$</th>
<th>Stoichiometry</th>
<th>$t, \text{ mm}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Phantom</td>
<td>Water</td>
<td>1.0</td>
<td>$\text{H}_2\text{O}$</td>
<td>Variable</td>
</tr>
<tr>
<td>Probe Holder</td>
<td>Perspex</td>
<td>1.17</td>
<td>8 % H, 32 % O, 0.6 % C</td>
<td>6</td>
</tr>
<tr>
<td>Shield</td>
<td>Aluminium</td>
<td>2.71</td>
<td>100 % Al</td>
<td>0.4</td>
</tr>
<tr>
<td>Perspex Converter</td>
<td>Perspex</td>
<td>1.17</td>
<td>H 8 %, O 32 %, C 0.6 %</td>
<td>3.5</td>
</tr>
<tr>
<td>Air Gap</td>
<td>Air</td>
<td>0.00129</td>
<td>70 % N, 30 % O</td>
<td>0.3</td>
</tr>
<tr>
<td>Device Overlayer</td>
<td>Silicon Oxide</td>
<td>2.27</td>
<td>$\text{SiO}_2$</td>
<td>0.001</td>
</tr>
<tr>
<td>Sensitive volume</td>
<td>Silicon</td>
<td>2.33</td>
<td>100 % Si</td>
<td>0.01</td>
</tr>
</tbody>
</table>

a 1 $\mu$m thickness $\text{SiO}_2$ layer. The 300 $\mu$m air gap was modelled along with the 3.5 mm perspex converter, 0.4 mm thick aluminium shield, 6 mm probe holder and a thickness of water corresponding to the depth of measurement. Simulations were made for phantom thickness of 10 mm, 188 mm, 220 mm, 227 mm, 238 mm, and 248 mm corresponding to the depths of measurements performed by Bradley et al. The lateral dimensions of the water phantom and other materials were set as twice the width of the silicon sensitive volume. Again this was done to conserve computation times.

For the second simulation a class derived from G4VUserPrimaryGeneratorAction was used to model the primary beam as a pencil beam normally incident on the surface of the phantom. This beam was assumed to have a gaussian distribution in energy determined from the first simulation as determined from the first simulation. The numbers of primary protons used in each simulation were $10^6$, $10^7$, $10^7$, $10^7$, $10^7$, $10^9$ for depths of measurements of 10 mm, 188 mm, 220 mm, 227 mm, 238 mm, and 248 mm respectively.

A class derived from the G4VUserPhysicsList class was used to define the physics of proton interaction to be used in the simulation. The models included in the simulation are similar to those for chapter[6] The exception is the inclusion of an inelastic scattering
model for protons (G4PreCompoundModel). The charged secondaries are only tracked if their residual range in material is greater than $1\,\mu m$. Electrons and gamma photons are only tracked if their residual range in the medium is greater than $10\,cm$. Figure 7.6 shows a view from within the water phantom and was generated using the vrml viewing program vrmlview.

A class was derived from the G4UserEventAction class to analyse each primary or secondary charged particle crossing the device. If a charged particle traversed the sensitive volume of the detector the ionisation energy lost in the sensitive volume was calculated. Energy lost in the sensitive volume was assumed to be equal to energy deposited and this energy deposition event was tallied. As for the simulation in chapter 6 the statistics of electron hole generation were ignored. For each particle crossing the sensitive volume the atomic number and mass of the particle was tallied. Whether the particle stopped in, crossed over, or started in the sensitive volume was also determined.
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Figure 7.6: View from within water phantom for proton therapy simulation. The proton history on the right shows an inelastic reaction with oxygen producing neutron (green) and gamma (yellow) and recoil proton (blue). Secondary electrons emanating from charged particle trajectories are shown in red.

An application was written in C++ to analyse the output of these simulations. A frequency distribution of all energy deposition events $f_j$ was formed. This distribution was convolved with a gaussian distribution of variance $\sigma = 5\, keV$ to allow for the electronic noise:

$$
f_j = \sum_{i=j-3\sigma}^{j+3\sigma} \frac{1}{\sqrt{2\sigma^2\pi}} f_i \exp\left(-\frac{(\epsilon_j - \epsilon_i)^2}{2\sigma^2}\right)
$$

(7.1)

Where $\epsilon_i$ is the energy values corresponding to bin $i$ of the frequency distribution. As the amount of energy deposition can span several orders of magnitude, the linear binned spectrum was logarithmically binned with 10 bins per decade covering energy deposition from $1\, keV - 5\, MeV$.

7.3.2 Results and Discussion

Figure 7.7 shows the spectrum of deposited energies produced by the simulation in comparison with experimental results at a depth of measurement of $10\, mm$. Firstly, the experimental results are not available below $10\, keV$ owing to the noise threshold of
the microdosimeter. Agreement between the two curves in the region $10 - 60\,keV$ is evident however the two curves depart for energies greater than this value. The simulation spectrum extends to energies of $500\,keV$ whereas the experimental results extend beyond $1\,MeV$. This may be a result of the limited statistics available for the simulated data. The beam was considered normally incident onto the phantom and at this shallow depth the angular distribution of protons may be narrower than the actual distribution. Consequently, oblique strikes to the microdosimeter and hence high energy deposition events are unlikely. In the experimental situation however the beam is not normally incident but possesses some angular distribution which would serve to broaden the angular distribution of protons at depth leading to a higher probability of high energy deposition events.

Figure 7.8 shows the spectra at the $188\,mm$ depth of measurement. In this situation the peak of the simulated spectrum is shifted to an energy of approximately $10\,keV$ in accordance with a decrease in proton energy. The simulation and experimental curves agree in the range $10 - 70\,keV$ yet disagree for higher energies. This may be the result
Figure 7.8: Spectrum of deposited energies in silicon microdosimeter at 188 mm depth in water phantom.

of a discrepancy in the angular distribution of protons at this depth.

Figure 7.9 shows the spectra at a depth of measurement of 220 mm. At this depth the peak in the simulated results has shifter further to higher energies. In this situation the curves extend to the same maximum value of energy deposition and agree reasonably well in the region $30 - 90 \text{ keV}$. The discrepancies between the curves occur at low values of energy deposition $10 - 20 \text{ keV}$ and high values of energy deposition $100 - 600 \text{ keV}$. The low energy discrepancy may be due to the fault in the MCA as mentioned in chapter 6 or due to an underestimation in the associated gamma field which in turn is due to a simplification of the beam modification process. The discrepancy at higher energy deposition events may be due to discrepancy in angular distribution of protons.

Figure 7.10 shows the spectra at the 227 mm depth of measurement. At this depth the peak in the simulated results has shifter further to higher energies. In this situation the curves agree reasonably well in the region $50 - 500 \text{ keV}$. The discrepancies between the curves exists at low values of energy deposition $10 - 50 \text{ keV}$.

Figure 7.11 shows the spectra at the 238 mm depth of measurement. This depth
Figure 7.9: Spectrum of deposited energies in silicon microdosimeter at 220 mm depth in water phantom.

Figure 7.10: Spectrum of deposited energies in silicon microdosimeter at 227 mm depth in water phantom.
Figure 7.11: Spectrum of deposited energies in silicon microdosimeter at 238 mm depth in water phantom.

of measurement corresponds to the distal edge of the Bragg peak. There are major discrepancies between simulated and experimental curves. In this region of the Bragg peak protons may have an LET around 100 keV/µm so the calculation of energy deposition events will be sensitive to any errors in device overlayer thickness, uncertainties in probe positioning and errors in primary beam energy calculations. Another interesting feature in this spectrum is the increased contribution to lower energy deposition events for the simulated data.

For each depth of measurement the contribution of starting, stopping and crossing particles was determined. The fraction of total events resulting from each type is given in table 7.3. These results show that the spectra at all depths are dominated by particles crossing the sensitive volume.

7.4 Conclusions

Simulations of silicon microdosimetry measurements in PT were performed using the GEANT4 Monte Carlo toolkit. The discrepancy between experimental and simulation
Table 7.3: Fraction \((\times 10^{-3})\) of starting, stopping and crossing particles for each simulation.

<table>
<thead>
<tr>
<th>Depth (mm)</th>
<th>crosser</th>
<th>starter</th>
<th>stopper</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>100.0</td>
<td>0.000</td>
<td>0.000</td>
</tr>
<tr>
<td>188</td>
<td>100.0</td>
<td>0.000</td>
<td>0.000</td>
</tr>
<tr>
<td>220</td>
<td>999.8</td>
<td>0.042</td>
<td>0.180</td>
</tr>
<tr>
<td>227</td>
<td>996.9</td>
<td>0.082</td>
<td>3.029</td>
</tr>
<tr>
<td>238</td>
<td>989.8</td>
<td>0.265</td>
<td>9.954</td>
</tr>
</tbody>
</table>

results for the 10\,\mu m device is most likely due to the underestimation in the gamma component of the primary beam incident on the phantom. It may also be due to the fault in the multi channel analyser. Simulation results were analysed to study the importance of particles originating from inelastic interactions in the silicon volume. These events were not seen to be important for proton therapy. In this situation, if the microdosimeter is to be applied to proton therapy with the aim of making tissue equivalent measurements, a simple scaling factor may suffice.

In general it is difficult to attribute the observed discrepancies to uncertainties in the model of the primary beam, the phantom and microdosimeter geometry and composition, or in the electrical properties of the silicon microdosimeter. To eliminate the uncertainties in the model of the primary beam, future simulations should include a detailed model of the beam modifying devices. Incorporating such a detailed model of the beam modifying devices would severely lengthen calculation time. Further work is therefore necessary to investigate parallel computing possibilities for Monte Carlo calculations for proton therapy.

Further work should be conducted to investigate the fault in the multi channel analyser. If it is attributed to pile up of sub threshold events, a theoretical model of pileup
should be incorporated into the GEANT4 simulation to model this effect. This is possible by associating a time with each particle in the PrimaryGeneratorAction class. Future experimental measurements should be performed in heterogeneous phantoms with elemental compositions close to human tissue such as those provided in reference [30]. The microdosimeter probe design should be altered to remove all material surrounding the microdosimeter with the exception of aluminium shielding.
CHAPTER 8
CONCLUSIONS

This thesis has proposed an alternative method of applying silicon microdosimeters to hadron therapy. Instead of attempting to derive a tissue equivalent microdosimetry spectrum from that measured in silicon, it is suggested to use the raw spectrum for the macroscopic verification of Monte Carlo calculations. When applied in this manner the problem of (non) tissue equivalence of the silicon microdosimeters is circumvented.

If the silicon microdosimeters are to be incorporated within a Monte Carlo simulation in hadron therapy, in addition to modelling the device composition and geometry, it is also necessary to model the charge collection properties of the microdosimeter. Consequently, a goal of this thesis was to use the IBIC technique to study these properties using the ANSTO heavy ion microprobe. To enable such measurements on the ANSTO heavy ion microprobe, a procedure for low current studies such as STIM and IBIC was established. The IBIC technique was added to the capabilities of the microprobe and the minimum beam spot size attainable with the ANSTO microprobe was measured using STIM. The IBIC technique was then used to image the amount of charge collected by the silicon microdosimeters following irradiation using various ions. These experiments were simulated using the GEANT4 Monte Carlo toolkit to calculate the amount of charge deposited in the silicon microdosimeters. These results were compared to create profiles of the charge collection efficiency of the silicon microdosimeters. The variation of the CCE with device bias, sensitive volume thickness, and ion LET was studied. For ion LETs relevant to hadron therapy, the CCE was not seen to vary dramatically with ion LET.

A problem with the current microdosimeter design is for the case of ions striking the device outside the depletion region, for which the measurement of energy deposition is severely effected by recombination. An attempt was made to solve this problem by developing a technique based on pulse shape discrimination. A system for measuring
charge collection times following ion strike was added to the microprobe. Imaging of charge collection times were carried out for silicon microdosimeters with $10 \mu m$ thick sensitive volumes. A technique for gating the charge collection signal from the microdosimeter based on the timing signal was tested on the microprobe. Using this system it was possible to ignore charge collection events from ion strikes outside the geometrical boundaries of the $pn$ junction. In this manner, those events suffering high amounts of recombination could be ignored. This technique however assumes that the charge collection time varies only with the location of the ion strike on the microdosimeter. The charge collection time however was seen to vary with the angle of incidence and ion LET. It is not recommended to apply this technique in its present form to silicon microdosimetry measurements. It is more practical to implement lateral isolation techniques in future silicon microdosimeter design as was originally proposed by [36].

Simulations of silicon microdosimetry measurements in FNT were performed using the GEANT4 Monte Carlo toolkit. Qualitatively, simulation results were seen to compare favourably with experimental results. A discrepancy was observed for the $10 \mu m$ device which may be due to the pileup of sub-threshold events. Simulation results were analysed to study the importance of particles originating from neutron interactions in the silicon volume. This was seen to be significant for both $2 \mu m$ and $10 \mu m$ results. If the silicon microdosimeters are being applied in the conventional sense of microdosimetry, these results show this effect may inhibit it’s ability to provide tissue equivalent measurements; however, if the silicon microdosimeters are being applied in the sense of verifying the Monte Carlo calculation, then neutron interactions in the silicon volume are not important.

Simulations of silicon microdosimetry measurements in PT were performed using the GEANT4 Monte Carlo toolkit. Significant discrepancies were observed for the measurement at the distal edge of the bragg peak. The source of this discrepancy may be due to a number of factors: uncertainties in the model of the primary beam, the phantom and microdosimeter geometry and composition, or in the electrical properties of the silicon
microdosimeter. It may also be due to the fault in the multi channel analyser. Simulation results were analysed to study the importance of particles originating from inelastic interactions in the silicon volume. These events were not seen to be important for proton therapy. In this situation, if the microdosimeter is to be applied to proton therapy with the aim of making tissue equivalent measurements, a simple scaling factor may suffice.

8.1 Recommendations

The development of treatment planning systems for hadron therapy, which use a Monte Carlo engine to characterise the radiation field within the patient, is strongly recommended. This treatment planning system should accommodate the use of radiobiological models for calculation of equivalent dose distributions in the patient. With respect to the verification of these treatment planning systems, further research into biological dosimetry is required. These experiments should also be conducted in heterogeneous phantoms with atomic composition similar to tissue as those provided in reference [30]. The problem of computation times should be addressed by utilising parallel computing techniques; whether this be using a dedicated beowulf cluster for the particular facility\footnote{Several excellent HOWTO pages exist, online and as part of many linux distributions, outlining the procedure for establishing a beowulf cluster.} or taking advantage grid computing. One should note that for the case of proton and carbon ion therapy, Monte Carlo based treatment planning is best suited to active systems as there is no need to simulate the beam modifying devices.

Although biological dosimetry is essential for verifying the entire treatment planning process, silicon microdosimeters offer the potential to verify the accuracy of the Monte Carlo calculation of the beam transport in the patient. In this thesis, the comparison between experimental silicon microdosimetry measurements and the predictions of Monte Carlo were entirely qualitative. A quantitative method of comparison should be used in future; one possibility is to use a Karlogrlov test. In the case of hadron therapy however, uncertainties in other aspects of the treatment planning process (such as the
biological model used or that introduced by patient movement) may 'smear out' uncertainties in the radiation field. In this respect, the required accuracy of the Monte Carlo calculations should also be investigated.

Future development of silicon microdosimeters should concentrate on the production of second generation devices using lateral isolation techniques, as was originally suggested by Bradley et al [3]. This will ensure a fully depleted cubic sensitive volume in which charge collection is dominated by the drift of charge carriers. As a result the charge collection time will be drastically reduced leading to a more radiation tolerant device and the cubic volume will result in a lowered angular dependence of the energy deposition measurement. The device overlayer should be simplified (or removed as suggested in reference [3]) to remove the distortion of the energy deposition measurement by energy loss in the overlayer, particularly for high LET ions, as was observed in chapter [4]. In chapters [6] and [7] the inclusion of the complicated overlayer and charge collection behaviour of the current device into the theoretical model were not feasible owing to computation times and simplifications were necessary. In the future design a more accurate modelling of the device will be possible.

Future simulations of the experimental microdosimetry measurements in hadron therapy should model the beam production and modification in detail. The routine discrepancy observed between experiment and simulation in the region of the threshold should be investigated. To confirm or nullify the notion of the pile up of sub threshold events, a pileup model should be incorporated into the simulation. Future simulations should also attempt to model the device geometry in more detail.

Once these devices are fabricated further work on the heavy ion microprobe is encouraged. The microprobe experiments performed in these experiments were very difficult and time consuming. The setup time for each experiment was typically 4-5 hours, from source start-up to the initiation of the IBIC measurement. Several actions may be taken to increase the efficiency of such measurements. One of the major time consuming procedures is to obtain a sufficiently low beam current suitable for IBIC measurements.
The development of a single ion delivery system would drastically reduce this time. This system uses an ion detection system (secondary electron detector) and a pair of high speed electrostatic deflection plates. Once the ion is detected, the deflection plates are charged, switching the beam off. This system is capable of delivering single ions for relatively high beam currents (nA). Another time consuming procedure is the adjustment of the object and collimating slits during the initial alignment of the beam and in the production of the focussed beam. Motorisation of object and collimating slits would greatly reduce this time. Finally, another time consuming procedure is in the transport of the beam from source to target using beam controlling devices (magnets, quadrupole and einzel lenses) owing to the large distances between devices and the need to iteratively adjust each individual element. The automation of this procedure is highly recommended.

Currently the silicon microdosimeters provide the spectrum of energy deposition events in the silicon volume. In regards to the verification of the Monte Carlo calculations in treatment planning it would be ideal to possess an instrument that performs charged particle identification. The possibility of extracting this information using future devices deserves investigation. In future devices, for which charge collection is dominated by drift, the shape of this current pulse would depend upon the initial density of charge carriers, in turn related to the ion atomic number [65]. Exactly how this information is extracted remains to be seen. One possibility is to digitise each and every current pulse following ion strike on the device; the integration of which would yield the energy deposition. By taking the fourier transform of the current pulse one may identify a relation between the fourier spectrum and the incident ion Z. Digitising every waveform is not feasible for count rates of hadron therapy and the implementation of existing hardware based fourier transform circuitry should be investigated.
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