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A Monte Carlo Investigation in External Beam Radiotherapy

A THESIS

Submitted by

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in partial fulfilment of the requirements for the degree of

DOCTOR OF PHILOSOPHY

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School of Physics
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Dr. Mark Foley

September 2011
Monte Carlo techniques have been shown to be the most accurate method of modelling radiation transport and dose deposition. With increasingly complex radiotherapy techniques being employed in the continuing fight against cancer, Monte Carlo techniques provide a definitive method to assess the accuracy of these techniques and the accuracy of the procedures used to test them. This thesis established a method to tune a linear accelerator model, in an efficient process. This model was then employed to examine two techniques used currently in the measurement of the absolute dose for the quality assurance of individual patient IMRT plans.
Acknowledgements

I would like to thank my supervisor Dr. Mark Foley for all his support and guidance. I wish to express my gratitude to Prof. Jan Seuntjens for providing me with the opportunity to work with him and to carry out a large portion of the work in this PhD at McGill university. Special thanks to Andrew Alexander for his patient advice and friendship over the last few years. Thank you to my fellow students past and present in the MPRC for a lot of assistance with this work (Seán Walsh, Ronan Havelin, Tuathan O’Shea, and Paddy Downes). I would like to express my gratitude towards the clinical staff in Montreal who provided me with measured data for my work (Gabriella Stroian, Russell Ruo) and to Eunah Chung for helpful discussions on plan complexity metrics. I would like to express my special gratitude and thanks to Prof. Wil van der Putten and all the staff in the UHG radiotherapy department, especially Margaret Moore, David Rajasekar, and Christoph Kleefeld. I would like to acknowledge John and James McDonald for help with computational issues. My thanks and appreciations also go to the coffee crew for their friendship and motivation (Kev, Kev, Jev, Dan, and Fiona). Not forgetting my best friends who have always been there for me and have put up with me while I completed this work. I must also convey thanks to all colleagues and staff in the School of Physics who played a role in my education, not only in the last four and a half years, but throughout my undergraduate degree as well. Last but not least I wish to express my love and gratitude to my family. In particular my Mother and Father for all their encouragement and support throughout my life. My fiancé John for his understanding and love throughout the duration of my studies. I would also like to thank my sisters (Maria, Orla, Emma and Fiona).

Elaine Conneely
September 2011
*Dissemination of work*

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I would like to dedicate this to my fiancé John. There is no doubt in my mind that without his continued love and support I could not have completed this process.
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The objective of this thesis was to develop an accurate linear accelerator model in an efficient manner and use this model along with the recently developed MMCTP code to investigate the accuracy of two different methods of absolute dose measurement, for individual patient plan intensity modulated radiotherapy (IMRT) quality assurance techniques.

1.1 Aims

In order to use the MMCTP code to perform a clinical investigation, the system first had to be successfully beta tested and set up on a Windows platform and its functionality tested. Once this was completed for all the features of the system required for the research, work began on tuning the Varian accelerator model (Varian Medical Systems, Inc., Palo Alto, CA) to an accurate level in an efficient manner. This method was then tested on a Siemens accelerator model before the work moved on to employing the Varian model in a clinical investigation for a
cohort of 20 patients. This investigation involved examining the collapsed (all gantry angles set to zero) and rotated (gantry angles kept as planned) IMRT quality assurance technique for absolute dose measurement of the dose in a high-dose, low-dose-gradient region.

In carrying out this investigation of IMRT quality assurance techniques, different complexity metrics were also examined to try to determine why one measurement technique may produce more accurate results than another. The different complexity metrics looked at the homogeneity of the dose within the chamber, the conformity of the high dose region to the chamber, the leaf sequence and area variability, and the fraction of the beams in a given plan to directly intersect the chamber.

1.2 Overview

- **Chapter 1** gives the aims of this work and an outline of the thesis.

- **Chapter 2** looks at the background theory and some of the literature for the topics covered. A brief introduction to radiotherapy and specifically intensity modulated radiotherapy is given. Some background on the use of Monte Carlo methods in modelling radiation transport is given. The EGSnrc code is also introduced.

- **Chapter 3** discusses the MMCTP code which was used extensively in this work in both tuning the models and in modelling the IMRT QA dose distributions. The implementation of the MMCTP system in a Windows environment was tested during this phase of the work. The efficacy of the majority of the functions of the system were also tested, focusing primarily on the functions applicable to this work.
1. Introduction

- **Chapter 4** focuses on trying to develop an efficient tuning process for an EGSnrc linac model as this is the first step in implementing any Monte Carlo investigation. The method is developed using data for the Varian C2100 based in the clinic in Montreal General Hospital (1650 Cedar Avenue, Montreal, Quebec, H3G 1A4). This linac is the focus of the investigation into IMRT QA in chapter 6 of this work.

- **Chapter 5** aims to test the method devised in chapter 4, by using it to tune the Siemens Oncor linac which is based in the clinic in University Hospital Galway.

- **Chapter 6** is the main focus of this investigation. Here the method of performing the absolute dose measurement for individual patient IMRT plans during IMRT QA is investigated. The doses obtained for both collapsed and rotated delivery methods was investigated as this had not been done previously and research shows that a significant number of clinics use a collapsed method despite not knowing the effects of this on the outcome of the measurements.
1. Introduction

Chapter 1
Introduction

Chapter 2
Monte Carlo Modelling in Radiotherapy

Chapter 3
Implementing the MMCTP code

Chapter 4
Optimisation of tuning (Varian tuning)

Chapter 5
Tuning of the Siemens model
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Chapter 6
IMRT phantom QA

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Conclusions

Figure 1.1: A flow chart outlining the organisation of this thesis.
Chapter 2

Monte Carlo Modelling in Radiotherapy

In this chapter the theoretical background to many areas of the thesis are outlined. A brief introduction to external beam radiotherapy is given focusing more specifically on IMRT. As the subject of this thesis is the issue of modelling treatment plans using the Monte Carlo code, EGSnrc, to analyse IMRT quality assurance techniques, the main emphasis of this chapter goes towards detailing these areas. To this end, the quality assurance of IMRT is discussed and the fundamentals of radiotherapy and Monte Carlo techniques are given.

2.1 Radiotherapy

Radiotherapy is the use of ionising radiation to treat cancer within the body, to damage cancerous cells via alterations to the cellular nuclear DNA. This ionising radiation can come from external sources such as photons, electrons or protons,
aimed at the body or from internal sources such as a sealed or unsealed radioactive source placed inside the body. In general, fractionated ionising radiation produces more damage in malignant than non-malignant tissues (Hall and Giaccia, 2006). One of the clinical requirements for radiation delivery is to eradicate cancer by accurately delivering a high radiation dose to the tumour site with minimum delivery of dose to the surrounding normal tissue.

Radiation therapy was developed shortly after the discovery of x-rays at the end of the 19th century. Subsequent clinical and technological advances have made it one of the most successful modalities in the treatment of patients with cancer, both in curative and palliative settings. The true origin of radiotherapy is not really known, though it is often credited to Leopold Freund (Kogelnik, 1997). He is the first physician documented to have used ionising radiation for therapeutic purposes. In 1896, a year after the discovery of X-rays, he successfully treated a five year old girl in Vienna suffering from moles covering her back. One of the earliest images of radiotherapy treatment is shown in figure 2.1. This was the first treatment carried out using the linear accelerator at Stanford University and the first documented treatment for retinoblastoma with ionising radiation.

There are two main types of radiotherapy:

1. External beam radiotherapy (delivered by radiotherapy machines such as orthovoltage units or linear accelerators).

2. Brachytherapy (sealed source radiation and unsealed source or radioisotope therapy).

The most common treatment modality today is the use of the high energy linear accelerator (linac) machine (Podgorsak, 2005).
2. Monte Carlo Modelling in Radiotherapy

Figure 2.1: The first patient to receive radiation therapy from the medical linear accelerator at Stanford was a 2-year-old boy. This is Gordon Isaacs, the first patient treated with radiation therapy for retinoblastoma in 1957. Gordon’s right eye was removed January 11, 1957 because the cancer had spread. His left eye, however, had only a localized tumour that prompted his doctor, Henry Kaplan to try to treat it with the electron beam. (Image from Stanford University, Stanford Report, April 18, 2007).

2.1.1 Linear accelerator - accelerator treatment head

For the purpose of this work, focus was placed on the treatment head as this was the part of the linear accelerator which was actually modelled, as it is the last juncture before the radiation exits the machine for delivery. The components of the complete linear accelerator including the treatment head are shown in figure 2.2.

For radiotherapy applications, the beam must obey certain specifications. It must:

- have a narrow energy specification
- be homogeneous
- be symmetric
In order to obtain a steep dose gradient between the target volume and healthy tissue, the penumbra of the beam has to be as small as possible. After exiting the wave guide, the beam does not have these properties, so control and focusing are required. The treatment head contains the beam production system, consisting of the scattering foil and beam shaping elements which help to achieve the required specifications for electron beam production and the x-ray target and flattening filter for photon beam production (Podgorsak, 2005).

2.1.1.1 Electron treatment head

The electron beam, when it exits the window of the accelerator tube, is a narrow pencil beam, about 3 mm in diameter (Khan, 2003). The scattering foil spreads out this small beam of electrons and provides a flat uniform electron treatment field (Johns and Cunningham, 1983). An appropriate scattering foil for the electron energy being used is placed in the beam line. As electrons penetrate this foil they are scattered and form a broadened electron beam. Most linacs employ
a pair of metallic foils to scatter the beam output from the bending magnet as
dual foil scattering improves electron beam flatness and also allows for larger
field sizes as this widens the beam (Khan, 2003). The thickness of the foil is de-
signed specifically so that most of the electrons are scattered instead of suffering
bremsstrahlung. However, a small fraction of the total energy is still converted
into bremsstrahlung and appears as x-ray contamination of the electron beam.
This photon contamination appears as the tail in the percent depth dose (PDD)
curve (Khan et al., 1991). The dose chamber monitors the dose, energy, and
flatness or symmetry of this beam (Podgorsak, 2005). An electron applicator,
for the field size chosen, defines the final electron field size and shape (Ma et al.,
1997). This detachable electron applicator is attached to the accessory mount of
the treatment head. Since electrons are readily scattered in air, the collimation
of the electron beam must be achieved close to the skin surface of the patient.
For electron beam field shaping, an auxiliary collimator takes the form of an
applicator consisting of scrapers down to within 5 cm of the skin surface.

Figure 2.3: The accelerator treatment head for electron (a) and photon (b) beam
2. Monte Carlo Modelling in Radiotherapy

2.1.1.2 Photon treatment head

For the generation of photon treatment beams the electron beam is incident on an x-ray target. This produces a forward peaked photon beam which can then be flattened using a flattening filter, which is energy appropriate (Mayles et al., 2007). This flattened beam then passes through an ion chamber which monitors the dose being administered to the patient (Johns and Cunningham, 1983). The beam must then be shaped to the required field size using jaws and a multileaf collimator (MLC). Other field shaping devices such as wedges, blocks and compensators may also be used.

For intensity modulated treatments the field is shaped using MLC’s for either a step-and-shoot delivery or a dynamic delivery method (figure 2.4). The step-and-shoot method delivers a number of individual beams each shaped by the MLC to deliver a modulated dose distribution incident from the one position for the one IMRT beam. The dynamic method uses an MLC which is moving continuously to deliver an intensity modulated dose distribution with the linear accelerator head in the one position for the one IMRT beam.

2.1.2 External beam radiotherapy

In external beam radiotherapy a source of radiation external to the patient is used to target a disease site within the patient body. This is in contrast to brachytherapy where the radiation source is placed either, within or onto, the target volume (Podgorsak, 2005). Most external beam therapies are carried out with photon beams, though some are carried out with electron beams and an increasing amount with protons (Jones and Errington, 2000) the determination of which treatment is employed is decided by the referring clinician and multidisciplinary discussion.
2. Monte Carlo Modelling in Radiotherapy

Figure 2.4: A comparison of the implementation of step and shoot IMRT (a) and dynamic IMRT (b). Reproduced from Schlegel and Mahr (2001). For the step and shoot technique the mlc’s define decreasingly smaller fields such that the areas in the final field receive the most dose as they are irradiated each time, with the beam turning on for each field but turning off in between while the mlc leaves move. For the dynamic technique however, the mlc’s move continuously throughout treatment, with the beam on continuously.

2.1.2.1 Electron beam radiotherapy

There are some cases in radiotherapy where electron beams are favoured over photon beams due to the differing properties of each type of radiotherapy. For example, electron beams can be used to treat head and neck cancers to avoid irradiation of the spinal cord and to treat chest walls so as to limit the irradiated volume of the lung. The most clinically useful energy range for electrons is 6 MeV to 20 MeV. At these energies, the electron beams can be used for treating superficial tumours (those less than 5 cm deep) with a characteristically sharp dropoff in dose beyond the tumour site, which helps to spare more healthy tissue which lies deep relative to the tumour site (Khan, 2003). The principal applications are:

1. The treatment of skin and lip cancers
2. Chest wall irradiation for breast cancer
3. Administering a boost dose to nodes
4. The treatment of head and neck cancers.

As electrons travel through a medium, they interact with atoms (Attix, 1986), resulting in the deposition of their dose, through different processes due to Coulomb force interactions. The processes are:

1. Inelastic collisions with nuclei (bremsstrahlung)
2. Elastic collisions with atomic electrons
3. Inelastic collisions with atomic electrons (ionisation and excitation)
4. Elastic collisions with nuclei.

These collisions cause the electrons to undergo multiple scatterings, resulting in their displacement from their original direction of motion. Due to the relatively small mass of electrons they undergo a large amount of scattering (Podgorsak, 2005).

![Figure 2.5: A comparison of the percentage depth dose curves for electrons (a) and photons (b). Reproduced from Podgorsak (2005).](image)
2. Monte Carlo Modelling in Radiotherapy

2.1.2.2 Photon beam radiotherapy

Most external beam radiotherapy is carried out with photon beams as they have greater penetration when compared with kilovoltage beams and electron beams while also providing greater skin sparing capabilities. This is brought about by dose build-up due to a lack of electronic equilibrium at shallow depths. Today, photon beams are primarily generated using linear accelerators which gives the advantage of being able to choose different beam energies and smaller penumbra at the edge of the beam (Mayles et al., 2007). Photons are attenuated and scattered when passing through a material through interactions (Attix, 1986) which include:

1. Photoelectric effect
2. Coherent (Rayleigh) scattering
3. Compton effect (incoherent scattering)
4. Pair production
5. Photonuclear reactions.

External beam radiotherapy is usually carried out with more than one beam. This is done to allow an increased dose to the target while keeping the dose to the surrounding healthy tissues to a minimum. In addition most photon beams are delivered using conformal radiotherapy where the radiation field or beam is shaped using MLC’s to further conform the dose to the target and minimise the dose to the surrounding healthy tissue.
2. Monte Carlo Modelling in Radiotherapy

2.1.3 IMRT

IMRT is a significant technological advance in the field of radiotherapy which allows the dose to tumours to be maximised while concurrently minimising the dose to the surrounding healthy tissues by sculpting a high dose field around the disease site with hitherto unachievable precision (Webb, 2003). This increases the chances of controlling the spread of the cancer while reducing the morbidities associated with the side effects of radiotherapy treatment. IMRT is most commonly implemented in a hospital radiotherapy department using a linear accelerator with a multi-leaf collimator (MLC). It is different to conformal radiotherapy as it is not just the field shape that is varied but also the dose across the field is varied (figure 2.7). The step-and-shoot IMRT technique does, however, use many small conformal fields incident from the one angle to generate an intensity modulated field.

The “dose painting” technique of IMRT makes it ideal for use in areas where it can be difficult to shape the dose accurately to the target volume i.e. in areas
with many heterogeneities such as in complex head and neck cancers or tumour bulks that invade and encapsulate delicate structures. It is becoming increasingly more popular and is being applied to more and more disease sites (Nguyen et al., 2011) particularly in situations where conformal plans fail to meet the necessary dose restrictions to the organs at risk, as is the case for some prostate patients (figure 2.8).

In order for IMRT treatments to achieve their potential it is necessary to be able to accurately predict the radiation dose that will be administered to the patient in these complex treatment techniques. Previous studies have shown that standard treatment planning methods in certain clinical IMRT configurations are limited in the prediction of the radiation dose and differences of over 20% have been observed (Ma et al., 2000).

In inverse treatment planning, the treatment planner sets the limits or clinical objectives of an acceptable plan, such as:

- maximum and minimum dose to the target
• maximum dose to the organs at risk

• dose volume objectives i.e. set a maximum dose limit for a certain percentage of an organ at risk

• uniformity objectives

• weightings or importance factors

The treatment planning program then tries to come up with a treatment plan which fits all of these criteria, however, all the criteria cannot usually be satisfied and the algorithm must come up with ones which cover as many of the objectives as possible or covers the most important ones (based on the weighting factors). It is then up to the treatment planner to select the most suitable plan (Mayles et al., 2007).

Figure 2.8: Image showing the principles of IMRT, with the dose intensity of the beams varying to increase the dose to the target (prostate in this case) while minimising the dose to the organs at risk (bladder and rectum in this case). Reproduced from Webb (2003).
2. Monte Carlo Modelling in Radiotherapy

2.1.4 Quality assurance

Quality assurance (QA) in radiotherapy is designed to provide optimum patient care and to ensure the safety of patients and staff (Podgorsak, 2005). As radiotherapy is a multidisciplinary treatment modality, the quality assurance programs must be carried out by all disciplines involved with each having clearly defined responsibilities. The physicists generally have responsibility for the equipment in the department and any measurements and calculations which must be performed and general radiation safety. The quality assurance program for the equipment in the department should cover commissioning of the equipment before it is first used comprising of extensive checks and analysis of all aspects of the machine. Then there should be maintenance quality assurance procedures to be carried out daily, weekly, monthly, quarterly and annually. Where the daily QA would comprise of brief checks, working up to the annual QA which would be similar to the initial commissioning process. These QA checks should also include the treatment planning system in use in the department. In addition to QA procedures for the equipment and systems, there should be treatment delivery QA checks in place. These should encompass dosimetric and geometric tests (e.g. using TLD’s and EPID’s).

Dosimetric checks, i.e. in-vivo dose measurements, are generally only carried out for specific patient groups or for more complex treatments. However, for IMRT treatments, as they are more complex treatments, a more comprehensive QA procedure needs to be in place (James et al., 2008). IMRT treatments require plan verification and pre-treatment checks in addition to the periodic testing of the system. Checks of planning, delivery and data transfer should be performed prior to treatment for each patient. Pre-treatment verification involves measurement of point doses, fluence maps, and combined dose distributions as well as any
additional IMRT specific checks to be made to the linac as part of the normal linac QA procedures (James et al., 2008).

2.1.5 Treatment planning and dose calculation algorithms

There is no single classification method for dose calculation algorithms; one option is to divide them into two groups:

- Correction based methods
- Model based methods

For the purpose of this discussion, focus will be placed on the model based methods as they are the most accurate and most widely used.

The correction based methods are not commonly employed as inaccuracies of 20%-70% have been reported, especially in low density mediums and at the boundaries of heterogeneities, specifically in situations where electronic equilibrium is not fully achieved and increased lateral electron scatter is not fully taken into account (Du Plessis et al., 2001). The correction based methods are based primarily on correction factors obtained from measurements. They are semi-empirical algorithms which apply corrections for heterogeneities to calculate dose distributions in a patient. The dose at any one point is usually calculated in two separate steps, the dose due to the primary direct beam and the scattered dose; these two values are then added to obtain the total dose (Khan, 2003). Model methods are based more on the physical modelling of beam interactions and actual radiation transport. They are more fundamental methods and can calculate the dose in heterogeneities directly (Khan, 2003). One particular type of model based algorithm, is the convolution-superposition method.
The convolution equation (Ahnesjo and Aspradakis, 1999) separately considers the transport of primary photons (primary dose) and that of the scattered photon and electron emerging from the primary photon interaction (secondary dose). The primary dose is calculated from ray-tracing and the secondary dose is calculated from a convolution of Terma (defined below) and precalculated dose kernels (Khan, 2003).

The convolution equation: the dose $D(\vec{r})$ at a point $\vec{r}$ is given by,

$$D(\vec{r}) = \int \frac{\mu}{\rho} \psi(\vec{r}) A(\vec{r} - \vec{r'}) d^3\vec{r'}$$

(Khan, 2003)
Where $\frac{\mu}{\rho}$ is the mass attenuation coefficient, $\Psi(\vec{r}')$ is the primary photon energy fluence, (the product of these two, is the quantity known as Terma) and $A(\vec{r} - \vec{r}')$ is the convolution kernel. The convolution kernel is a matrix of dose distribution deposited by scattered photons and electrons set in motion at the interaction site of the primary photon.

The convolution-superposition equation (Ahnesjo and Aspradakis, 1999) is a convolution with a spatially variant kernel, with distance corrected for using electron density relative to water for inhomogeneities.

$$D(\vec{r}) = \int \frac{\mu}{\rho} \Psi(\rho \vec{r} \cdot \vec{r}') A(\rho \vec{r} - \vec{r}' \cdot (\vec{r} - \vec{r}')) d^3\vec{r}' \quad (2.2)$$

(Khan, 2003)

Where $\rho \vec{r} \cdot \vec{r}'$ is the radiological path length from the source to the primary photon interaction site and $\rho \vec{r} - \vec{r}' \cdot (\vec{r} - \vec{r}')$ is the radiological path length from the site of primary photon interaction to the site of dose deposition. $A(\rho \vec{r} - \vec{r}' \cdot (\vec{r} - \vec{r}'))$ is the dose kernel.

In the work by Krieger and Sauer (2005), a simple multi-layer phantom composed of styrofoam and white polystyrene was irradiated with a 10 x 10 cm$^2$ field. It was found that in the polystyrene (figure 2.10), which is of higher density, both the Monte Carlo and Collapsed Cone algorithms agreed well with the measured data (within 3%). In this region, the Pencil Beam values were up to 14% higher than the measurements. In the low density material (figure 2.10), the Collapsed Cone algorithm underestimated the dose by an average of 10% while Monte Carlo agreed well within the error limits with the measurements at all points. The values from the Pencil Beam calculations were not comparable (about 30% too high) in the low density regions.
2. Monte Carlo Modelling in Radiotherapy

2.1 Figure 2.10: Dose profiles calculated with different algorithms for a 10 x 10 cm² field. Reproduced from (Krieger and Sauer, 2005). Algorithms investigated were, Monte Carlo (MC), collapsed cone (CC), and pencil beam (PB).

2.2 Monte Carlo methods

The result of otherwise unsolvable problems can often be estimated by using random numbers. This technique for obtaining an answer to a problem is called Monte Carlo simulation. Monte Carlo simulation presents estimated solutions to a range of mathematical problems by carrying out modelling experiments on a computer (Fishman, 1996).

Monte Carlo simulation is a numerical method for solving mathematical problems, it is essentially a statistical package where only the probability of the values can be known, and which randomly generates values for variables repeatedly, to simulate scenarios of a problem. These values are obtained from within a fixed range and chosen to suit a probability distribution (Rubinstein, 1981). With Monte Carlo modelling, a large system can be sampled in a number of ways, and that information can be used to model the system as a whole. Although simulation was often seen as a method of last resort, advances in modelling methods, software, and technical developments made simulation a more broadly used and accepted
The use of Monte Carlo simulation enables us to investigate more complex problems than we otherwise could using traditional methods.

The name Monte Carlo comes, unsurprisingly, from Monte Carlo, Monaco and was employed in the 1940’s by scientists working on the nuclear weapons project to assign a label to the use of random numbers in solving problems. This name was chosen since the primary attractions in Monte Carlo at the time were casinos containing games of chance such as roulette wheels, dice, and slot machines, which exhibit random behaviour. During the development of atomic energy after World War II scientists needed to solve problems of particle transport through materials. These problems proved to be too difficult to solve by hand with the difference equation. American mathematicians John von Neumann and Stanislaw Ulam suggested that if sampling experiments were performed on the newly developed electronic computer, they could provide practical estimations to the desired solutions (Mayles et al., 2007). The theoretical foundation of the method had been known long before this as certain problems in statistics were sometimes solved by means of random sampling. The simulation of random numbers by hand is a laborious process and for this reason, it was only after digital computers were developed that Monte Carlo technique began to be studied in detail (Fishman, 1996).

The Monte Carlo method can be used to model the transport of radiation particles as a random series of free flights that end with an interaction where the particle changes its direction of motion, loses energy, and in some cases, produces secondary particles. The main drawback of the Monte Carlo method is that all the results are affected by statistical uncertainties, which can be made as small as we like, provided we can perform enough Monte Carlo calculations to increase the sampled population but this also increases the computation time. The Monte Carlo method is a commonly used technique for examining complex problems and applications can be found in many fields.
Monte Carlo simulation of radiotherapy units is one of the most accurate methods for predicting absorbed dose distributions in radiation therapy (Ma and Jiang, 1999). The Monte Carlo method can precisely model the transport of radiation for radiation therapy and is powerful in dealing with any complex geometry. Monte Carlo dose calculations can potentially reduce the errors that may be present in the dose calculation algorithms which are currently used in treatment planning. However, Monte Carlo calculations also contain random errors, or statistical uncertainty, the level of which decreases inversely with the square root of the computation time (Keall et al., 2000). The fact that Monte Carlo is so computationally intensive has become much less of a problem due to the rapid increase in speed and decrease in cost of computers (Rogers, 2006), and the employment of innovative variance reduction techniques (Verhaegen and Seuntjens, 2003 & Chetty et al., 2007).

Monte Carlo simulations are being used increasingly in medical physics. The work of Ma and Jiang (1999) investigated the use of Monte Carlo methods in modelling electron beams from linacs. This paper provides an extensive review of Monte Carlo modelling, from the initial modelling of accelerator components to modelling complete accelerator treatment heads to calculate percentage depth dose and dose profile curves. The origin for the idea of component modules, such as those used in the BEAMnrc code (Rogers et al., 1995), is often accredited to Udale (1988).

It should be noted that the uncertainty in dose distributions measured clinically is reportedly about 2% (Ma and Jiang, 1999) but the uncertainty for dose measured at a point is usually about 3% (Khan, 2003). From this it can be seen that agreement between Monte Carlo calculations and measured doses of much less than 1% would be unnecessary. However, to achieve an adequate statistical uncertainty in the Monte Carlo simulations it can be necessary to simulate more than $10^8$ electron histories (Ma and Jiang, 1999).
2. Monte Carlo Modelling in Radiotherapy

2.3 The EGSnrc system

2.3.1 BEAMnrc code

BEAMnrc system (Rogers et al., 1995) is a general purpose EGSnrc (Electron Gamma Shower) (Kawrakow et al., 2011) user code for modelling radiotherapy units, which was developed at the NRC (National Research Council of Canada) as part of the OMEGA (Ottawa Madison Electron Gamma Algorithm) project. It is a full Monte Carlo dose calculation algorithm for radiation beams from a radiotherapy unit.

The BEAMnrc code was created for simulating radiotherapy units while the DOSXYZnrc code was created for performing phantom based and CT based dose calculations. Within the BEAMnrc GUI there is a preview option for each component module to allow the user view a graphical representation of the data which has been used in defining the component module in question. There is also the option of previewing the entire accelerator after it has been defined, to see the relative positioning of the different component modules. After the entire accelerator has been successfully defined and the input parameters have been entered, the BEAMnrc GUI allows the user to build and compile the accelerator and finally run the simulation.

The BEAMnrc and DOSXYZnrc graphical user interfaces (GUI’s) write the files required to run the code.

- BEAMnrc GUI generates a .module input file, (which contains the component modules from the accelerator) and an .egsinp (which contains all the simulation parameters). The input files must be saved before starting to compile or run the accelerator simulation.
• DOSXYZnrc GUI generates an .egsinp file also which contains a complete
description of the phantom in which the doses are to be calculated and de-
fines the source of the particles entering the phantom, (usually a phase-space
file from a BEAMnrc simulation or a BEAMnrc treatment head simulation
would be chosen as the source). Again the input file must be saved before
compiling or running the simulation.

Important features of BEAMnrc code include:

1. The use of component modules (CM’s) for building accelerators and other
radiotherapy machines.

2. The use of a phase-space file to score each particles information.

3. Using LATCH bits to track each particles history.

4. Being able to apply variance reduction techniques to speed up simulations.

5. A user friendly interface (the BEAMnrc GUI).

The first paper on the BEAM code was the work of Rogers et al. (1995). This work
discusses the different capabilities of the code including the various component
modules which were available with the original code. This code introduced the use
of LATCH technique. The benchmarking of the code showed it could reproduce
measured dose distributions well within 3% of the dose at $d_{max}$ for all depths
(Rogers et al., 1995).

2.3.1.1 Component modules

Component modules are elementary geometric entities which the user specifies
dimensions for and assembles as necessary (e.g. flattening filter, dose chamber,
jaws, and applicator etc.). Each component module acts like a building brick,
allowing an entire accelerator treatment head to be built with relative ease by simply putting a series of component modules together to match the geometrical specifications of the treatment head. This feature allows a series of different kinds of accelerators to be modelled, even by inexperienced users.

2.3.1.2 Phase space files

The BEAMnrc code generates a phase-space file which is a record of the particles:

1. Charge
2. Energy
3. Position
4. Direction
5. LATCH history tag

The LATCH history tag, for each particle crossing the scoring plane in the simulation, keeps a record of the location of each particles interactions within the accelerator model. The scoring plane is the back plane (surface furthest from the source), perpendicular to the beam axis, of any specified component module in the accelerator model.

2.3.1.3 LATCH

The history tag variable, LATCH, manipulates each bit separately to keep record of a particles history. Each bit can be mapped to or associated with a different component module or geometric region. A bit is set in the particles history if the particle passes through and interacts within a region. In the DOSXYZnrc input file, or when using BEAMDP to analyse a phase space file, the user may
choose to apply inclusive or exclusive LATCH bit filters in calculating the dose or in the analysis. Each LATCH bit is manipulated separately to keep track of each particles history, allowing the user to analyse the relative dose contributions from various accelerator components in order to evaluate the influence of each of them e.g. to determine the dose from electrons scattering off the jaws of the accelerator (Ma and Jiang, 1999).

The LATCH parameter is a 32 bit variable where bits 1 - 23 are used to record the different regions where a particle has been and/or interacted. The user employs the variable IREGION_TO_BIT for a region to define the bit set for that geometric region when defining the geometry in the input file. Therefore it is possible to have bit 5 mapped to geometric region 3 and even to have multiple geometric regions mapped to the one bit. This means that even though a component may be made up of several geometric regions they can all still correspond to the one LATCH bit. Any regions which are left unmapped to a bit are automatically assigned to bit 23, by default (Rogers et al., 2004).

### 2.3.1.4 Variance reduction techniques

Variance reduction techniques are methods to improve the efficiency of Monte Carlo calculations and reduce simulation times (Chetty et al., 2007). However, some variance reduction techniques can have minor (depending on parameters used) affects on the results, which can reduce the accuracy of the dose calculations, though these are generally not considered to be true variance reduction techniques (Verhaegen and Seuntjens, 2003). Variance reduction techniques incorporated into the BEAMnrc code include bremsstrahlung splitting, photon interaction forcing, Russian Roulette and range rejection.
2.3.2 DOSXYZnrc

DOSXYZnrc is a user code developed by the NRC for the OMEGA project, to do dose calculations in a Cartesian co-ordinate volume. The phase-space data file can be analysed by the DOSXYZnrc program to score the dose in a 3D rectilinear geometry including various phantoms or patient geometry, built from CT data (Walters and Rogers, 2004). Density and material in every voxel may vary (this allows it to be used with CT data). The geometry is a 3D rectilinear volume (X, Y and Z planes) with voxel dimensions which are completely variable in all three directions. The code allows sources such as mono-energetic diverging or parallel beam or phase-space data generated by a BEAMnrc simulation to be incident on the specified phantom (Walters and Rogers, 2004). DOSXYZnrc has a number of important features such as dose component calculations, a wide variety of source configurations and beam reconstruction techniques, CT to phantom conversion, phase-space redistribution, etc.

For the tuning done in this work the input source for the DOSXYZnrc calculations was set as ISOURCE=9: BEAM Treatment Head Simulation Incident from any Direction. This source uses particles sampled from a BEAMnrc simulation running simultaneously with the DOSXYZnrc simulation. The BEAMnrc accelerator code must first be compiled as a shared library and must have been supplied with its own input files and pegs4 material data file. Source particles for DOSXYZnrc are then sampled directly from what would have been the scoring plane during a normal run of the BEAMnrc accelerator, without the need to store a phase-space file. You can select particles from the BEAMnrc simulation to be used based on their charge and/or LATCH values (allowing dose component calculations). The “source plane” is the scoring plane in the BEAMnrc simulation, and when using source 9, particles are sampled as soon as they cross the scoring plane rather than stored in a phase-space file for later use. This source has the
obvious advantage over a phase-space source in that the intermediate phase-space data need not be stored reducing the necessary amount of computer memory.

2.3.2.1 STATDOSE

STATDOSE is a computer program for 3-dimensional dose analysis and plotting 1-dimensional dose distributions from the .3DDOSE data files generated by the program DOSXYZnrc (McGowan et al., 2010). The functions of the STATDOSE program include normalisation, rebinning, plotting and analysis of the dose distributions. Using the STATDOSE program the dose curves can be plotted along any one of the three principal axes and normalised, so as to obtain the relative central-axis depth dose and transverse dose profile graphs at specified depths in the phantom for comparison with the corresponding measurements (Kapur et al., 1998).

In order to use the STATDOSE program one of the .3DDOSE files created by DOSXYZnrc must be read in before attempting any kind of analysis. After a dose distribution is read in, the STATDOSE function, plot dose, can then be used to plot profiles of either depth doses or transverse dose profiles. The user may choose the axis which the profile plot is to be parallel to and the coordinates which the profile is to pass through i.e. for the percent depth dose plots. The plot dose function, plots the dose values in the voxels which lie parallel to the Z-axis and which pass through the X = 0 and Y = 0 (central axis) coordinates. For the transverse dose profile curves, the plot dose function plots the dose values in the voxels which lie parallel to the Y-axis and pass through the X = 0 and Z = 1 coordinates (1 cm depth in the phantom passing through the central axis). The axis of the profile must be entered as an integer from 1 - 3, corresponding to X, Y and Z axes, respectively.
2.4 The Accelerator models used

The medical linear accelerators (linacs) used for this work were the Varian CL2100 and the Siemens Oncor Avant Garde. The machines were modelled by including the vendor-supplied specifications for the different accelerator components. The linear accelerator models are comprised of different component modules or building blocks. Each part of the linacs are single component modules which take up a horizontal slab portion of the linac and sits along the Z-axis i.e. it is a block with a front surface and a back surface perpendicular to the Z-axis. Each component module is independent and reusable.

There can be gaps between the different component modules but these gaps will be filled with air by the BEAMnrc main routine. The air gap between the front surface of a module and the back surface of the previous module is considered part of this module. However, the front surface of this module is still defined as the surface of the non-air medium in the module.

Each component module can be previewed after the geometry parameters have been entered. There is also a help button beside some of the input parameters (especially for non-geometrical parameters) which displays explanatory documentation allowing for clear input formats. Each component module also has a maximum boundary, parallel to the Z-axis, beyond which particles are not followed. The photon accelerator head is comprised of a series of component modules which represent the target, primary collimator, flattening filter, dose chamber, upper y-jaws, multi-leaf collimator and an air slab.
2.5 MMCTP

The McGill Monte Carlo Treatment Planning (MMCTP) system is a radiotherapy research environment which allows the integration of patient specific treatment plans, from commercial treatment planning systems, with Monte Carlo calculated dose distributions for the same data set allowing for large-scale prospective and retrospective patient treatment studies (Alexander et al., 2007). The MMCTP code was developed in the Medical Physics Unit at McGill University, Montreal, Canada. MMCTP was designed as a flexible computational radiotherapy research environment which would allow Monte Carlo calculated dose distributions to be compared with those from commercial treatment planning systems.

The system had previously only been used at McGill, where it had only been implemented on a Macintosh computer running OSX. This work initially tested multi-platform implementation of the MMCTP code in collaboration with the group at the Medical Physics Unit in Montreal. A remote cluster was used for lengthy Monte Carlo calculations using the BEAMnrc code. The BEAMnrc input files are prepared from the beam geometry and properties and patient information from the treatment planning files. These are then uploaded and run on a cluster. It connects to the cluster through a standard secure-shell protocol using a program called “plink”. The treatment planning data is obtained from the hospital where the dose distributions are calculated using a commercial treatment planning system used in the day-to-day treatment of cancer patients.

Several publications (Ma et al., 2000 & Krieger and Sauer, 2005) have highlighted the inaccuracies in these treatment planning systems as they use a number of approximations to increase the speed of the calculations. These inaccuracies are usually particularly noticeable in areas with tissue inhomogeneities for example the femoral heads in prostate cancer treatment (Fraser et al., 2008) or in head and neck treatments where there are many soft tissue, bone and air cavity interfaces.
(Ma et al., 1999), or in lung treatments due to the low density of the air cavities in the lung (Aarup et al., 2009 & Vanderstraeten et al., 2006 & Keall et al., 2000). This is of concern as a 1% improvement in accuracy can lead to a 2% improvement in cure rate for early stage tumours (Papanikolau et al., 2004). MMCTP provides us with the opportunity to better examine the accuracy of treatment planning systems compared to Monte Carlo calculations using an unbiased platform.
Chapter 3
Implementing the MMCTP Code

This part of the work focused on implementing and testing the MMCTP system in a Windows environment and beta-testing the features of the system which was used in the subsequent stages of the research.

3.1 Background

The McGill Monte Carlo Treatment Planning (MMCTP) system (Alexander et al., 2007) is a computational research environment that allows for the comparison and analysis of dose distributions from commercial treatment planning systems with Monte Carlo calculated dose distributions using the BEAMnrc (Rogers et al., 1995, 2004) and DOSXYZnrc (Walters and Rogers, 2004) codes with EGSnrc. The MMCTP package has a GUI which runs on a laptop or PC. The program connects to a remote cluster, using an SSH protocol, for lengthy Monte Carlo calculations to be performed. The Monte Carlo calculations are...
submitted automatically to the BEAMnrc code on a remote supercomputer or cluster.

MMCTP imports hospital generated treatment plans along with CT and other patient images with structure contours. It then uses the beam geometry information from these plans and images to generate the input files required for the Monte Carlo simulations. The input files are then uploaded and run on a remote cluster. The dose distributions calculated by Monte Carlo can then be compared with those from the treatment planning system in an independent, common platform.

MMCTP was developed under REALBasic (RealSoftware Inc., Austin, Texas) which supports multi-platform environments, however, each platform environment carries unique characteristics, which require additional code in order for MMCTP to be transparent between platforms. The majority of Windows MMCTP testing and implementation was carried out in this work. Some of the problems encountered when trying to implement the MMCTP system in Windows included: SSH and FTP protocol usage, DVH calculations, window sequencing, menu bars and file paths.

In order for MMCTP to be able to connect with a remote computer or cluster, a number of steps must be carried out. Login settings such as an IP address, user name, password, and folder paths are configured within MMCTP. These settings also include specific operating system commands, which are unique to the remote computer. MMCTP uses SSH or FTP protocols for all remote login, thus, these protocols must be enabled on the remote computer.
3. Implementing the MMCTP Code

3.2 Applications of the MMCTP system

A common independent research platform is a fundamental tool for Monte Carlo treatment planning studies. In order for wide scale studies to be carried out, the research platform should be functional on multiple operating systems. In this investigation we have begun to establish the efficacy of the MMCTP system in a windows operating environment and confirmed the value of the system as a clinical study device in its previously tested Mac operating environment. This work was part of a clinical IMRT recalculation study (chapter 6) using patient data from Montreal General Hospital.

Once an accurate model has been commissioned, the user can save the BEAMnrc input file as a template input file for further simulations. These files are linked within MMCTP to a treatment machine and energy. BEAMnrc accelerator directories are also tagged with each treatment machine and energy. The DOSXYZnrc directory location is defined once for each remote computer. In addition, there is a global DOSXYZnrc template input file. This file can be modified and re-saved or tweaked for individual simulations. MMCTP has been used in many research projects as a DICOM viewer, DVH calculator, dose comparison tool, or for Monte Carlo treatment planning (Al-Yahya et al., 2005 & Soisson et al., 2011). A clinical version of MMCTP is used in the clinic at Montreal General Hospital for overnight Monte Carlo recalculation on complex patient plans.

3.3 Features of the system

1. **Multiple file format imports**: MMCTP can import treatment plans of DICOM, RTOG, and CADPlan format. This includes beam geometry properties, patient images and structure contours.
3. Implementing the MMCTP Code

To load a patient’s data and open the treatment planning window, the MMCTP program must be started after files have been imported. The folder must be changed to the location of the McGill RT folder, in which the imported files have been saved, then a patient is selected and a plan for that patient is opened.

2. Image visualisation: There are 2D and 3D visualisation methods for viewing patient images with the structure contours and dose distributions superimposed on the CT or other images. The patient images, such as CT images, can be viewed in either the sagittal, coronal, or axial planes. The images can be toggled through, slice by slice by clicking on the image and using the keyboard’s left arrow key to go back a slice and right arrow key to go up a slice.

3. Dose analysis tools: The dose analysis tools allow the dose distributions from different beams or from different plans to be added or subtracted from one another. Dose volume histograms can also be calculated within a structure and analysed for plan evaluations.

One of the features of the MMCTP system, are the dose analysis functions. These are accessed through the dose tab in the tab menu. There is a DVH (dose volume histogram) calculator and dose matrix operations. The dose matrix operations include addition, subtraction, multiplication and division of two dose distributions or a constant. The dose subtraction tool allows for easy comparison of dose distributions from the commercial treatment planning system with those calculated with Monte Carlo methods.

4. Editing of treatment plans: Treatment plans can be edited within the MMCTP GUI by adding or deleting beams, editing of beam properties as well as editing structure contours.
3. Implementing the MMCTP Code

5. Monte Carlo calculations: The input files required for Monte Carlo simulations are automatically prepared by the MMCTP program based on the external beam geometry imported from the hospital treatment plans. The phantom can be created based on the images imported. Also, the phantom materials can be set based on the structure contours imported from the hospital data.

3.4 How the MMCTP system was utilized in this work

For this work the MMCTP system will be used in tuning the Varian and Siemens linac models in a clear and efficient manner and these tuned models will then form the template input files required by the MMCTP system. The tuned Varian model and the MMCTP system will then be employed to investigate the difference between IMRT QA via a collapsed dose measurement (all gantry angles set to zero) and a rotated dose measurement (gantry angles as planned). In doing so, it will be determined if collapsed dose measurement in patient IMRT QA is a viable and reliable alternative to rotated dose measurement.

3.5 Beta testing the MMCTP code in a Windows environment

Beta testing multi-platform implementation of the MMCTP code was carried out by testing its functionality in a windows operating environment. The features discussed in section 3.3, were tested as follows.

1. File imports: Treatment plans in the DICOM format from University College Hospital Galway were successfully imported to the MMCTP system.
3. Implementing the MMCTP Code

The patient images were visible with structure contours. The beam properties matched those in the hospital treatment planning system.

2. Image visualisation: The 2D visualisation has been tested using some sample treatment plans. The patient images matched those in the treatment planning system on visual inspection. The structure contours were visible superimposed on the images and matched the structures well. The contours could be turned off or viewed as outlines or with colour fill of varying transparency. The dose distributions could also be viewed, again as contours or colour fill of varying transparency. The images could be toggled through successfully in the axial, sagittal, and coronal planes. However, this function could be slow with the dose distributions visible.

3. Dose analysis tools: The dose analysis tools were tested by adding together the dose distributions from the different beams in a plan and comparing with the total dose for the plan from the treatment planning system. The DVH’s were also successfully calculated for the different structures in the plan (see section 3.6 part 5).

4. Editing of treatment plans: Treatment plans could be edited successfully within MMCTP.

5. Monte Carlo calculations: The Monte Carlo input files were automatically prepared by the MMCTP program. These input files were sent to the remote cluster and the Monte Carlo calculations were carried out using the BEAMnrc code. This is discussed in section 3.8. One of the 3DDOSE files obtained from this test calculation case was compared with the corresponding 3DDOSE file for the same set of calculations from a MAC operating environment and the same simulation results were observed.
3.6 Using the MMCTP system

1. Importing treatment planning and patient image files: Converting patient information to the McGill RT format is the first step in using this system. MMCTP can import many different formats of files for treatment plans from the system in the hospital. Once the MMCTP GUI is open, transfer patient, is selected which opens the window in figure 3.1.

![Figure 3.1: A screen capture from MMCTP showing how hospital data is imported into the MMCTP system.](image)

2. Loading patient data: After the patient and plan data has been imported into the system, this data can be loaded and viewed by selecting openpatient. The folder must be changed to the location of the McGill RT folder, which the files have previously been imported to, then a patient is selected and a plan for that patient is opened. This opens the data for this patient’s plan in the treatment planning window (figure 3.2). This window displays three image windows for the patient images such as CT images, allowing the user to simultaneously view the dose distributions in the axial,
saggital and coronal planes, with the coordinates of the current slice are displayed at the top left of each image. The treatment planning window also initially displays a list of the plans associated with the patient, a list of the beams with their details, and a tab menu which provides access to most of the MMCTP features.

3. Setting up the login file and the template Monte Carlo input file: In order for MMCTP to function properly, the user must provide some information regarding the remote computing cluster and the linear accelerator to be simulated. In order to simulate a linear accelerator, MMCTP must be given the name and folder location of a template input file for each linac to be used. These details are stored in the BEAM.pref file. The login parameters for the remote computing cluster must also be provided in the login.txt file. This file is not encrypted, however it is saved as a hidden file as a security measure.

4. Viewing patient images: The patient images, such as CT images, can be viewed in either the saggital, coronal, or axial planes. The images can
be toggled through, slice by slice by clicking on the image and using the keyboards left arrow key to go back a slice and right arrow key to go up a slice. The structures can be seen superimposed on the images (figure 3.3), as can the dose distributions.

5. Dose Analysis: One of the features of the MMCTP system, are the dose analysis functions. These are accessed through the dose tab in the tab menu. There is a DVH calculator and dose matrix operations. In figure 3.4 this feature has been used to add together the dose distributions of 6 beams in a plan to calculate the total dose distributed within the patient for this 6 field treatment plan. The DVH calculator was then used to analyse this total dose distribution for this plan within the target volume and the organs at risk. It can be seen in figure 3.5 that 100% of the target volume receives a dose of at least 47 Gy, and less than 50% of the femoral heads receive more than 32 Gy.

6. Monte Carlo simulations: Monte Carlo simulations have been run on the McGill cluster, Clumeq, using the BEAMnrc code, which is installed on the supercomputer, using the MMCTP system. MMCTP creates the input files for the Monte Carlo simulations based on the data from the treatment
3. Implementing the MMCTP Code

Figure 3.4: The total dose from 6 beams with the dose analysis functions of the MMCTP code.

planning system but it does not include any of the patient details so the simulation can be run anonymously on the remote cluster.

3.7 Verifying the accuracy of the dose distributions in the MMCTP system in a windows environment

An initial study of dose distribution in a water phantom was carried out. The results from Monte Carlo calculations of dose distributions in this phantom were compared with measured data. This was carried out to test that the system was working correctly and to check the accuracy of the linac model in use with the BEAMnrc code. The phantom model was a 30 x 30 x 25 cm$^3$ water phantom consisting of 0.5 x 0.5 x 0.5 cm$^3$ voxels. A 10 x 10 cm$^2$ 6 MV photon beam was modeled using 100 million initial histories, which created 246.5 million particles.
Figure 3.5: Cumulative DVH’s for the target and organs at risk from a 6 field conformal prostate plan. With the bladder (orange), rectum (brown), right femoral head (purple), left femoral head (pink) and the target volume (red) shown.

in the phase space file which was recorded at 90 cm SSD. DBS was used in the simulations with a field radius of 10 cm and a splitting number of 1000. No electron splitting was used. ECUT and PCUT were set to 0.521 MeV and 0.01 MeV respectively. Approximately 245.9 million particles from this phase space file were then input to DOSXYZnrc for dose scoring in the water phantom creating a 3ddose file which was subsequently read by MMCTP. The percentage depth dose (PDD) and dose profile curves were then extracted from this 3ddose data using the dose analysis functions of MMCTP. These curves were then compared with measured data. The measurements were performed in the Clinic in Galway University Hospital, using a Linear Array Detector (LA48) with an Air-Scanner Adapter which is scanned through a 40 x 40 x 40 cm$^3$ water tank with measurements taken at increments of 0.1 cm to a depth of 2.0 cm and then 0.5 cm to a depth of 32.5 cm for the PDD and 0.1 cm increments for the dose profile in the X and Y directions going from -11 cm to +11 cm from the central axis position for the 10 x 10 cm$^2$ field.

A comparison of measurements and Monte Carlo calculated values are presented in figures 3.6 and 3.7. Profiles were taken at 10 cm depth in the water phantom.
for both the hospital measurements and the Monte Carlo simulations. The doses for the PDD were normalised to the dose at 1.4 cm ($D_{\text{max}}$) and for the dose profiles they were normalised to the isocentre dose.

![Figure 3.6: The PDD extracted from MMCTP compared with measured data from the hospital for the Siemens Oncor linac for a 6 MV 10 x 10 cm$^2$ field.](image)

The PDD plot (figure 3.6) shows good agreement between the values measured in the hospital in Galway and the Monte Carlo calculated data. All the values agree within 3%. The dose profile plot (figure 3.7) shows agreement between the measured and Monte Carlo data within 3% or 3 mm. This is slightly larger

![Figure 3.7: The dose profile extracted from MMCTP compared with measured data from the hospital for the Siemens Oncor linac for a 6 MV 10 x 10 cm$^2$ field.](image)
than the necessary accuracy of 2%-3% (Fraass et al., 2003) for dose calculation, however, this model had not undergone much tuning at this stage.

3.8 A trial Monte Carlo patient dose recalculation

In order to verify the functionality of the MMCTP system in the Windows operating environment, it was decided to carry out a test patient recalculation and to compare the resulting dose distributions to those calculated by, and imported to MMCTP from, the commercial treatment planning system from the clinic. The plan that was used was a conformal prostate treatment, which was chosen at random from the TPS. It was important to perform some preliminary clinical investigations to ensure the fields lined up as expected (i.e. the beam enters the phantom from the correct angle and the MLC’s have the correct field shape).

This preliminary clinical investigation focused on modelling a conformal prostate treatment plan, which consisted of 6 fields: a Right Anterior Oblique beam, a Right Lateral beam, a Right Posterior Oblique beam, a Left Posterior Oblique beam, a Left Lateral beam, and a Left Anterior Oblique beam as shown in figure 3.8. The DICOM files for this plan were imported into MMCTP. These provided all the necessary images, beam specifications, and dose distributions for the plan. This data was then used by the MMCTP system to generate the input files for the BEAMnrc and DOSXYZnrc Monte Carlo simulations. The patient phantom for DOSXYZnrc was generated from the CT images based on the HU to material conversions and density settings shown in table 3.1. A phantom to score the dose from the six fields was generated from the patient CT data with voxel sizes of 0.75 x 0.75 x 0.3 cm$^3$ in the X, Y, and Z directions respectively, with 64 voxels in the X Y directions and 81 in the Z direction. The CT data was converted to a phantom based on the allocation of materials to a specific range of CT numbers or Hounsfield Units using pre-assigned material and density definitions.
3. Implementing the MMCTP Code

<table>
<thead>
<tr>
<th>Material</th>
<th>HU Range</th>
<th>Density</th>
</tr>
</thead>
<tbody>
<tr>
<td>AIR521ICRU</td>
<td>-1000 : -950</td>
<td>0.001 : 0.044</td>
</tr>
<tr>
<td>LUNG521ICRU</td>
<td>-950 : -700</td>
<td>0.044 : 0.302</td>
</tr>
<tr>
<td>IRCUTISSUE521ICRU</td>
<td>-700 : 125</td>
<td>0.302 : 1.101</td>
</tr>
<tr>
<td>ICRPBONE521ICRU</td>
<td>125 : 2000</td>
<td>1.101 : 2.088</td>
</tr>
</tbody>
</table>

Table 3.1: HU to material conversions used in generating the dose scoring phantom from the patient CT data. These are the default HU and density values in DOSXYZnrc (Walters and Rogers, 2004).

Figure 3.8: A screen capture from MMCTP showing the patient CT image with contours and other screen captures from MMCTP showing the conformal beams.

The dose distributions obtained from the Monte Carlo simulations of the six-field prostate treatment plan match the isodose lines from the treatment planning system reasonably well apart from the 100% to 80% isodose lines (figure 3.9) and also in the regions around the femoral heads (figure 3.9 (a) & (b), and (e) & (f)). This shows that the beam arrangement is being modelled accurately and the discrepancies are most likely due to inaccuracies in the conversion of the CT data to a dose scoring phantom and could also be attributed to the TPS only calculating the dose to water. The fact that the Monte Carlo dose was not converted to Gy by using a calibration value, it was just converted by assuming the plan delivered 50 Gy to the isocentre as planned, is another source of error but this was considered accurate enough for this investigation as it was just intended
to test the functionality of the MMCTP system and not necessarily to investigate the accuracy of the dose distributions obtained as the BEAMnrc linac model had not been fully tuned.

For the DVH data shown in figure 3.10, the Monte Carlo calculated values are consistently lower than the doses calculated by the treatment planning system but again it is unclear how accurate the dose data was at this point in the investigation.

3.9 Dose conversion in MMCTP

The dose values imported into MMCTP from the DOSXYZnrc Monte Carlo simulations were given as dose per incident particle. In order to compare these doses to those from the clinical treatment planning system they had to be first converted to dose in Gray (Gy). This was done through the use of a calibration factor \((D_{\text{particle}}/\text{calibration})\) which was calculated as the dose per incident particle at \(d_{\text{max}}\) for a 10 x 10 cm\(^2\) field at 100 cm SSD. This was the same as the clinical calibration conditions, where the machines in McGill were calibrated to deliver 101 cGy in tissue at the calibration depth \(d_{\text{max}}\) from 100 MUs. The value obtained for \(D_{\text{particle}}/\text{calibration}\) for the tuned Varian model was \(1.43322\times10^{-16}\). MMCTP used this \(D_{\text{particle}}/\text{calibration}\) value to calculate the beam dose \(D_{\text{beam}}\) (cGy/MU) as follows:

\[
D_{\text{beam}}(\text{cGy/MU}) = \frac{D_{\text{particle}}/\text{beam}}{D_{\text{particle}}/\text{calibration}} \times \frac{1.01}{1\text{MU}} \tag{3.1}
\]

This method of dose calculation was similar to that used by Liu et al. (2001) & Ma et al. (2004). The known effects of backscattering into the monitor chamber were ignored in this equation, which for Varian machines introduces an error of approximately 1% for small field sizes (Popescu et al., 2005). This was considered
adequate for this investigation. However, MMCTP does allow for the inclusion of a MU back scatter factor which is a correction applied to the 3ddose file to take into consideration backscatter from the jaws to the ion chamber as a percentage of the total dose.

3.10 Dose-to-medium to dose-to-water conversion

MMCTP has the option of putting in a dose-to-water conversion factor. Dose to water is the stopping power ratio from material to water and is applied when importing dose values from a 3ddose file to MMCTP to convert the dose-to-medium to dose-to-water for easier comparisons with commercial treatment planning systems. However, this function was not utilised in this work as all the simulations in the research focused on the use of water phantoms. As a result of this all the dose values obtained were dose-to-water values and so no conversion was required.
3. Implementing the MMCTP Code

Figure 3.9: The dose distributions from the 6 field conformal prostate plan from the Monte Carlo simulations (left hand side) and the TPS (right hand side).
3. Implementing the MMCTP Code

Figure 3.10: DVH’s for the bladder (blue - TPS and orange - MC), rectum (brown - MC and green - TPS), right femoral head (yellow - TPS and purple - MC), left femoral head (pink - MC and dark orange - TPS) and the target volume (red - MC and light green - TPS). DVH’s shown for Monte Carlo (MC) and commercial treatment planning system (TPS).
Chapter 4

Optimization of tuning (Varian tuning)

An efficient and simple method of tuning a linear accelerator model is devised in this chapter and used to tune the Varian model based on measurements taken in the clinic in the Montreal General Hospital. The model is tuned to a clinically acceptable level (i.e. to within the accuracy of most measurements of 2% - 3%) to allow it to be used in chapter 6 of this work to investigate IMRT QA.

4.1 Modelling a clinical linear accelerator

Monte Carlo methods provide a very powerful dose calculation tool, as, unlike other dose calculation methods, they can accurately model even complex delivery techniques, involving large intensity gradients and small segments, such as those encountered in IMRT, and Monte Carlo simulation is the most accurate method for calculating the dose in heterogeneous media (Chibani et al., 2011). In these complex dose distributions, the assumptions made in conventional dose
calculation algorithms, regarding scatter equilibrium and how output ratios vary with field size, often break down. In order for Monte Carlo modelling to be implemented as the most precise way of calculating a dose distribution, an accurate model of the accelerator is needed. An accurate accelerator model needs:

1. An accurate model of the accelerator electron source

2. Accurate models of the physical accelerator components

An accurate representation of the physical accelerator components is usually the easiest part to achieve, assuming that the necessary data can be obtained from the linac manufacturer (this data can usually be obtained as part of a non-disclosure agreement with the company). It is possible for this data to be acquired via direct measurement of the components, although this is generally not practical in most clinical situations. Despite the fact that the manufacturer usually supplies data for the source also, it is usually approximate and varies from machine to machine and requires tuning. The accelerator can be modelled as a series of modular components, through the use of the BEAMnrc component modules (CM’s). Figure 4.1 shows a Monte Carlo model of a clinical accelerator in the modelling of photon beams and and its component parts.

Usually no modelling of the electron beam is done prior to it exiting the flight tube. What is used instead is an approximated primary electron beam that hits the target generating bremsstrahlung photons. These photons are then initially collimated by the primary collimator before being differentially attenuated by the flattening filter to produce an approximately flat dose distribution. The monitor chamber and mirror are often omitted as they attenuate the photon beam very little, they are however included within this model to keep as accurate a representation as possible. Finally the photon beam is shaped and modulated, in this case using X and Y jaws and an MLC.
4. Optimization of tuning (Varian Tuning)

Figure 4.1: Schematic drawing of linac components modelled in Monte Carlo simulations. Reproduced from Verhaegen and Seuntjens (2003).

4.2 The Varian accelerator model

The accelerator used for devising this optimization technique is the Varian 2100C 6 MV photon beam linear accelerator. The BEAMnrc model of this accelerator used for the purposes of this work, is based on parameters for the dimensions and the composition of the linac supplied by the manufacturer under a non-disclosure agreement. The most suitable CM for a particular accelerator component is decided upon and the parameters for it are defined to match as closely as possible, the data acquired from the manufacturer to minimize the dose discrepancies between the Monte Carlo calculations and measured data.

The component modules used and their corresponding accelerator components for this Varian linac model were as follows:
4. Optimization of tuning (Varian Tuning)

- The SLABS CM was used to create the accelerator target
- The CONS3R CM was used to create the primary collimator of the accelerator
- The SLABS CM was used to create the exit window
- The FLATFILT CM was used to create the flattening filter
- The CHAMBER CM was used to create the accelerator monitor chamber
- The SLABS CM was used to create the mirror
- The PYRAMIDS CM was used to create the accelerator shielding
- The JAWS CM was used to add a pair of Y jaws and a pair of X jaws to the model
- The DYNVMLC CM was used to model the Varian MLC so as to allow the simulation of IMRT fields (as is required later in this work)
- Finally, the SLABS CM was used to add an air slab to the end of the accelerator to bring the model to 70 cm distance from the source. This allows the phase space files to be created at a consistent location, making it easier to position the patient phantom when simulating full plans.

An image of this accelerator model as displayed in the BEAMnrc code can be seen in figure 4.2.

4.3 Tuning the primary electron beam of an accelerator model

Some assumptions must be made about the primary electron beam. In this work, the electron beam entering the linac model was assumed to be monoenergetic. The work by Sheikh-Bagheri and Rogers (2002), showed that for a
4. Optimization of tuning (Varian Tuning)

Figure 4.2: The BEAMnrc model of the Varian 2100C 6 MV photon linear accelerator.
symmetric energy distribution there is only a weak sensitivity (not large enough to be conclusive) of the depth dose values at larger depths to the energy spread of the electron beam. The angular spread of electrons resulting from scatter in air from a narrow collimated beam, for most practical applications, can be approximated by a Gaussian radial distribution (Pena et al., 2007 & Keall et al., 2003 & Verhaegen and Seuntjens, 2003). To date, tuning methods have primarily focused on two main techniques, the first being a trial and error iterative processes to determine the primary electron beam parameters (Pena et al., 2007 & Sheikh-Bagheri and Rogers, 2002). The second being the recent attempts that have been made to tune the model based on actual accelerator measurements obtained through dismantling the accelerator in question (Sawkey and Faddegon, 2009). The issue with the iterative trial and error techniques is they are unacceptably time consuming and variations that are less time consuming do not cover all possible best solutions, for example Tzedakis et al. (2004) just simulated lateral dose profile curves but only investigated FWHM’s for 1 energy and investigated the different energies at 1 FWHM. Added to this, the fact that dismantling an accelerator is not really a practical solution in most clinical situations, it was decided that there was a need for a better option. Something that focused on the ease of use of the iterative type processes that have been tried and tested (Pena et al., 2007) but in a manner that can be implemented in as short a time frame as possible and in a straight-forward manner so as to make it accessible to those with minimal Monte Carlo experience.

4.3.1 Importance of tuning

The results of Monte Carlo simulations can only be as accurate as the degree of tuning of the model being used for the simulations. Therefore, before any accelerator model can be used for looking at treatment plans or examining accelerator dose distributions, the model must first be tuned to an acceptable accuracy.
For the purposes of this work an acceptable accuracy will be considered to be 2-3% as this is in keeping with the achievable accuracy for most clinically relevant measurements (Ma and Jiang, 1999). As differences of up to 10-20% between treatment planning systems and measurements have been recorded for example in lung cases (Chetty et al., 2007), a model tuned to 2-3% accuracy would provide a useful one off verification tool for these complex situations when clinical physicists may need some assistance for atypical clinical cases. Manufacturers provide details on the geometry and composition of the components of the accelerator. However, the input beam characteristics are only known approximately, as details on these are only known approximately even to the manufacturers of the linac, and it is necessary to focus on tuning the linac model incident electron beam.

4.4 A fast Monte Carlo accelerator source optimization process - Devising a manageable, efficient tuning process on a Varian model

Monte Carlo calculations provide the possibility of a powerful treatment planning verification technique. With the increasing availability of powerful computational resources, these calculations could be used in many clinics for these purposes, providing an independent comparison tool when discrepancies arise between measured data and commercial treatment planning system dose values. This would allow physicists to determine where the error lies and perhaps point to inaccuracies in a particular procedure. However, the non-standardized and non-automated process of tuning the required accelerator model is one of the reasons for delays in the clinical implementation of the process. This work aims to establish and verify a simple and efficient tuning method that starts from standard clinical measurements and which can be carried out in a minimal time period allowing it to be easily implemented in a clinical setting by personnel with
minimal experience with Monte Carlo methods and in particular the EGSnrc and BEAMnrc Monte Carlo codes. The technique being proposed was used to establish the primary electron beam parameters (average electron beam energy and radius of the FWHM of the electron beam) for an accelerator model for the Varian Clinac 6 MV photon beam to an accuracy of 2-3%. The method is intended to provide a clear, direct and efficient process for tuning an accelerator model using readily available clinical quality assurance data. The use of this readily available data saves time as it removes the requirement to obtain extra, one-off measurements solely for the purpose of a tuning process. The accuracy of the tuned model should also be verified at both small and large field sizes to ensure there are no discrepancies in the tuned parameters. The technique proposed was used to establish the primary electron beam parameters for accelerator models for the Varian Clinac 2100 6 MV photon beam using the BEAMnrc Monte Carlo system. The method intends to provide a clear, direct and efficient process for tuning an accelerator model using readily available clinical quality assurance data. The tuning provides a refined model, which agrees with measured dose profile curves within 1.5% outside the penumbra or 3 mm in the penumbra, for square fields with sides of 3 cm up to 30 cm. These models can then be employed as the basis for Monte Carlo recalculations of dose distributions for clinical treatment plans, providing an invaluable assessment tool.

4.4.1 Initial range of tuning parameters

Work by Faddegon et al. (1999), showed that the key parameters for tuning the incident electron beam were the mean energy and focal spot size of the beam. An initial range of tuning values for these parameters had to be decided upon as the starting point for the tuning process. A review of the literature regarding previously implemented tuning methods was performed to determine these parameters. In the work by Pena et al. (2007), the initial electron energy is varied
4. Optimization of tuning (Varian Tuning)

in steps of 0.25 MeV, while Sheikh-Bagheri and Rogers (2002) varied the energy in steps of 0.1 MeV and Tzedakis et al. (2004) used increments of 0.2 MeV. Pena et al. (2004) recommended using the profiles from wide fields to tune the model as they show greater sensitivity to changes in the energy and radius of the primary electron beam. Tzedakis et al. (2004) varied the radius in steps of 0.02 cm, as did Sheikh-Bagheri and Rogers (2002), though Pena et al. (2007) used 0.05 cm. The paper by Sheikh-Bagheri and Rogers (2002) stated that they ignored the angular spread of the initial electron beam value (left it at 0 degrees) for their work, as credible divergences of up to 0.5 degrees showed no observable effect. Other studies (Sawkey and Faddegon, 2009) used alternative methods to more accurately describe the source parameters based on additional measurements of non-standard characteristics. While the shape and size of the primary beam has been found to differ from linac to linac (Munro et al., 1988), the shape of the incident electron beam is usually assumed to be circular (Verhaegen and Seuntjens, 2003). Some of these authors used wide field profiles and others used output factors but this work combines these techniques to generate a simplified and efficient tuning method which can be implemented in clinical situations by physicists with minimal experience with Monte Carlo techniques and in particular the EGSnrc and BEAMnrc systems.

4.4.2 The MMCTP code

MMCTP (Alexander et al., 2007) was used in this work to generate the input files for the Monte Carlo simulations and to submit the simulations to the remote cluster. It made the process more efficient as it automatically checked at regular intervals if the running jobs had completed and submitted the next batch of jobs once space on the cluster became available. It also downloaded and imported the results files from the cluster allowing the Monte Carlo dose distributions to be compared with the measured data. The measured data was imported to MMCTP
at the start of the process in the form of dose profiles and depth dose data in text files as recorded from the QA process in the hospital. The DOSXYZnrc part of the EGSnrc code (Walters and Rogers, 2004) was used to calculate the dose scored in the voxelised water phantom. This phantom was created using MMCTP as a user defined Cartesian phantom. Once an accurate linear accelerator model has been commissioned, the user can save the BEAMnrc input file as a template input file on which the patient-specific accelerator model is based in further simulations as is used in chapter 6 of this work. The input files are linked within MMCTP to a specific treatment machine and energy.

In this work the electron beam entering the linac model was assumed to be monoenergetic with a circular cross section and Gaussian distribution, as described above. The spread of electrons resulting from scatter in air from a narrow collimated beam can, for most practical applications, be approximated by a Gaussian distribution (Keall et al., 2003 & Pena et al., 2007 & Verhaegen and Seuntjens, 2003). In order for the BEAMnrc (Rogers et al., 1995, 2004) code to be used to accurately calculate dose distributions, the accelerator model must first be benchmarked. The energy and width of the initial electron beam values were varied to find the percentage depth dose, dose profile curves and output factors that match the hospital measured data, providing output factors that match within 1% of measured output factor values. It was decided that output factors would be the dominant factor in deciding on the tuned parameters, as it would require less computation time to calculate the output factors accurately using Monte Carlo.

4.4.3 Measurements

The percentage depth dose (PDD) and dose profile curves are extracted from the 3D dose data files, obtained from the simulations, using the dose analysis functions of MMCTP. These calculated curves were then compared with the
dose profile data obtained from measurements in the radiotherapy department at Jewish General Hospital, Montreal. The data from the Jewish General Hospital was acquired in the IBA Blue Phantom (48 x 48 x 41 cm$^3$) (figure 4.3) using a CC13 ionization chamber (figure 4.4) which can be used for both radial and axial beam incidence.

The water phantom specifications are as follows:

- Scanning volume: 480 mm (L) x 480 mm (W) x 410 mm (H)
- Scanning speed (continuous): up to 15 mm/s
- Position reproducibility: min. 0.1 mm
- Position accuracy: +/- 0.5 mm
- Approximate volume: 200 litres
- Wall thickness/material: 15 mm / acrylic (plexiglass)

The air ionisation chamber specifications are as follows:

- Cavity volume: 0.13 cm$^3$
- Cavity length: 5.8 mm
- Cavity radius: 3.0 mm
- Wall material: C552
- Central electrode material: C552
- Waterproof: Yes
4. Optimization of tuning (Varian Tuning)

Figure 4.3: The IBA Blue Phantom as used for the measurements.

Figure 4.4: The IBA CC13 air ionisation chamber as used for the measurements.

4.5 Monte Carlo simulations

The Monte Carlo simulations were run in 2 parts. First the BEAMnrc code was used to simulate the transport of the particles through the linear accelerator model to the scoring plane at 70 cm SSD from the source and were stored in a phase space file. Secondly, this phase space file was used as the input for the DOSXYZnrc code which simulates the transport of the particles through the
remaining air, through to the phantom and scores the dose in the water phantom. Directional bremsstrahlung splitting (DBS) was used with a splitting number of 2000 but no electron splitting was used (table 4.1). The EGSnrc parameters were left at the default values. The pegs4 data used were the ICRU data with production threshold for secondary electrons and cut-off energies of 189 keV and 10 keV for electrons and photons respectively.

The 6 MV photon model for the Varian linear accelerator was tuned in a number of stages, initially starting at large field (30 x 30 cm$^2$ for this model) profiles and PDD’s and progressing to examination of output factors for a range of field sizes focusing particularly on smaller field sizes for the final fine tuning steps. Small fields are more sensitive to changes in the FWHM of the radius of the incident electron beam (Pena et al., 2007) which allows tuning to be performed quickly and accurately. It was decided to focus on the larger field sizes when looking at profiles at 1.5 cm depth in the water phantom, as these are more sensitive to changes in the mean energy of the initial electron source (Pena et al., 2007) again improving the efficiency and precision of the tuning process being developed.

The tuning process was carried out in 4 phases. Phase 1 was performed to narrow the energy range estimate by looking at dose profiles for a wide field for a broad range of energy values and FWHM values. Phase 2 of the process was similar to phase 1 but used a smaller energy range with smaller increments in energy. Phase 3 used output factors for a wide field to choose the energy that provided the best agreement to measured values, followed by calculating the output factors for a range of field sizes to choose the optimum value for the FWHM of the electron beam. Finally, phase 4 was carried out to verify the accuracy of the chosen parameters by calculating PDDs and dose profiles for various field sizes and comparing these with measured data of the same. A diagram outlining the tuning process as implemented in this work is shown in figure 4.5.
4. Optimization of tuning (Varian Tuning)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Initial beam tuning</th>
<th>Energy fine-tuning</th>
<th>Output factor tuning</th>
<th>Tuned model</th>
</tr>
</thead>
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<td>No. Of Histories</td>
<td>3-7 million</td>
<td>8-11 million</td>
<td>10-13 million</td>
<td>approximately 20 million</td>
</tr>
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<td>Voxel size (cm$^3$)</td>
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<td>1 x 1 x 1</td>
<td>0.5 x 0.5 x 0.5</td>
<td>see table 4.2</td>
</tr>
<tr>
<td>Phsp data scored at</td>
<td>70 cm</td>
<td>70 cm</td>
<td>70 cm</td>
<td>70 cm</td>
</tr>
<tr>
<td>VRT’s</td>
<td>DBS (no e-split.)</td>
<td>DBS (no e-split.)</td>
<td>DBS (no e-split.)</td>
<td>DBS (with e-split.)</td>
</tr>
<tr>
<td>DBS field radius</td>
<td>27</td>
<td>27</td>
<td>27</td>
<td>27</td>
</tr>
<tr>
<td>DBS SSD (cm)</td>
<td>100</td>
<td>100</td>
<td>100</td>
<td>100</td>
</tr>
<tr>
<td>CM No. e-split.</td>
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<td>0</td>
<td>0</td>
<td>4</td>
</tr>
<tr>
<td>e-split. plane</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>12</td>
</tr>
<tr>
<td>Z of RR plane</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>12.67</td>
</tr>
<tr>
<td>ECUT (MeV)</td>
<td>0.7</td>
<td>0.7</td>
<td>0.7</td>
<td>0.7</td>
</tr>
<tr>
<td>PCUT (MeV)</td>
<td>0.01</td>
<td>0.01</td>
<td>0.01</td>
<td>0.01</td>
</tr>
</tbody>
</table>

Table 4.1: The input parameters used for the different Monte Carlo simulations.

4.5.1 Phase 1 - Energy & FWHM of the incident electron beam

For the initial phase of the tuning process the energies looked at for the 6 MV model were 5.0, 5.5, 6.0, and 6.5 MeV for the electron source with a Gaussian distribution of the radius with a full width half maximum of 0.05, 0.10, 0.15, 0.20 cm. Large energy increments were used initially to refine the energy range of the primary electron beam allowing smaller increments to be used in the subsequent tuning phases. It was decided not to investigate radii below 0.05 cm as values below this are unrealistic (Jaffray et al., 1993). The water phantom voxels were
Phase 1

Energy (MeV): 5.0, 5.5, 6.0, 6.5
FWHM (cm): 0.05, 0.10, 0.15, 0.20

Looked at wide field profiles at 1.5 cm depth.

→ Energy between 6 MeV and 6.5 MeV.

Phase 2

Energy (MeV): 6.1, 6.2, 6.3, 6.4
FWHM (cm): 0.05, 0.10, 0.15, 0.20

Looked at wide field profiles at 1.5 cm depth.

→ Difficult to ascertain which energy gave the best result other than an energy between 6 MeV and 6.5 MeV.

Phase 3

Energy (MeV): 6.1, 6.2, 6.3, 6.4
FWHM (cm): 0.05, 0.10, 0.15, 0.20

Looked at output factors for a wide field followed by output factors for a range of field sizes at one energy (6.3 MeV).

→ From wide field output factors: Energy = 6.3 MeV.
From all output factors: FWHM = 0.5 cm (radius).

Phase 4

Energy (MeV): 6.3
FWHM (cm): 0.05

Verification of parameters calculated dose profiles and PDD’s for a range of field sizes.

→ Compared to measurements to examine accuracy.

Figure 4.5: The tuning process (including the results of each step for the Varian model) as applied to the tuning of the accelerator models in this work.
set to 1 x 1 x 1 cm$^3$ for the measurement of the dose profiles at 1.5 cm depth, this somewhat large voxel size was employed to keep simulation time to a minimum. The pegs4 data used were the ICRU data with production threshold for secondary electrons and cutoff energies of 189 keV and 10 keV for electrons and photons respectively.

All doses have been normalized on the central axis and these pre-selective calculations were run at a low number of histories (3 to 7 million particles incident on the target, depending on energy) to give a more accurate estimation of the source energy in the least amount of time possible, which was in keeping with the aim of the work, to produce a fast and accurate tuning method. The results were evaluated using a root-mean-square difference value (RMSD), with:

$$RMSD = \left( \frac{\sum_{i=(-n)}^{+n} (c_i - m_i)^2}{N} \right)^{1/2}$$  \hspace{1cm} (4.1)

where, $c_i$ is the calculated normalized dose at the position $n$, in the $x$ plane (or $n$ cm distance from the central axis in the $x$ plane), $m_i$ is the measured dose at the point and $N$ is the number of points. The RMSD value was calculated for the dose profile at 1.5 cm depth (depth of maximum dose for the 6 MV beams) for each set of initial electron source parameters, from $x = -15$ cm to $+15$ cm for the 30 x 30 cm$^2$ field.

### 4.5.2 Phase 2 - Fine tuning the energy

After this initial tuning process it was decided that the best value for the initial electron energy would be found somewhere in the region between 6 MeV and 6.5 MeV for the accelerator model, as the measured dose profile curve lay between the curves obtained at these two energies. As a result, it was decided to continue the tuning process using a reduced step size of 0.1 MeV looking at 6.1, 6.2, 6.3,
and 6.4 MeV, while once again using the same values for the full width half maximum of the radius of the incident electron beam of 0.05, 0.10, 0.15, 0.20 cm. A larger number of histories were used at this stage (8 to 11 million histories striking the target, again depending on energy) as the simulations needed to be more accurate for this phase. This was necessary in order to differentiate which energy provided a better match with measured data as the energies were closer together, implying so too would the resulting profiles meaning it would be more difficult to determine which agreed best with clinical measurements.

4.5.3 Phase 3 - Output factors

After this second phase of the tuning process it was decided that it was becoming difficult to decipher the values which provided the best fit with the measured data as the differences between the graphs obtained were all within the uncertainty of the Monte Carlo calculations. In order to get a lower uncertainty and hence decide upon the optimum values for the initial electron beam, the simulations would become very long and the tuning process unacceptably time consuming for clinical implementation. As an alternative, it was decided to look at the output factors for the 30 x 30 cm² which would be used to further fine tune the energy as output factors are more sensitive to changes in the primary beam parameters (Pena et al., 2007) and as a result require less computation time than extended simulations and profile comparisons. The output factor was calculated from the dose scored in a 0.5 x 0.5 x 0.5 cm³ voxel on the central axis, at 5 cm depth for the Varian Clinac, as this was the setup under which the output factors were originally measured in the clinic.

For the Varian model, it was determined from these initial output factor investigations (using the simulation results from phase 2 for the 30 x 30 cm² output factor values) that the 6.3 MeV beam gave the closest and most consistent results
when compared with the measured output factor for the 30 x 30 cm$^2$ field at all radial distributions of the incident electron beam. Subsequently, this allowed the output factors for square fields of different sizes to be calculated at one energy only, of 6.3 MeV. The output factors for square fields of 40 x 40 cm$^2$, 30 x 30 cm$^2$, 5 x 5 cm$^2$, 4 x 4 cm$^2$, and 3 x 3 cm$^2$ were considered. The inclusion of small field sizes allowed for a more accurate determination of the FWHM of the beam radius (Pena et al., 2007) as these are the most sensitive to changes in the spatial distribution, while larger fields are more sensitive to energy variations. From this it was determined that the optimum value for the Gaussian distribution of the radius was 0.05 cm, as this radius value provided output factor values that matched measurements within 1% for all the field sizes investigated, which had been decided as the desired agreement between calculated and measured central axis output factors.

4.5.4 Phase 4 - Verification of the tuning results

Finally, once the best values had been decided upon, the dose profiles at 1.5 cm depth and the percentage depth dose curves for field sizes of 30 x 30 cm$^2$, 10 x 10 cm$^2$, 4 x 4 cm$^2$, and 3 x 3 cm$^2$ were calculated and compared with measured data to verify the accuracy of the chosen parameters of 6.3 MeV energy and 0.05 cm FWHM of the incident electron beam.

For the dose profiles, the voxel sizes were adjusted with field size in order to obtain a more accurate shape of the penumbra while using larger voxels outside this region of the dose profiles to keep the simulation time to a minimum while at the same time still obtaining smooth curves. The specific voxel sizes for the Varian dose profiles can be seen in table 4.2.
4. Optimization of tuning (Varian Tuning)

<table>
<thead>
<tr>
<th>Field Size (cm²)</th>
<th>1st Group</th>
<th>2nd Group</th>
<th>3rd Group</th>
<th>4th Group</th>
<th>5th Group</th>
</tr>
</thead>
<tbody>
<tr>
<td>30 x 30</td>
<td>1 cm x 4</td>
<td>0.2 cm x 10</td>
<td>1 cm x 28</td>
<td>0.2 cm x 10</td>
<td>1 cm x 4</td>
</tr>
<tr>
<td>10 x 10</td>
<td>1 cm x 4</td>
<td>0.2 cm x 10</td>
<td>1 cm x 8</td>
<td>0.2 cm x 10</td>
<td>1 cm x 4</td>
</tr>
<tr>
<td>4 x 4</td>
<td>0.3 cm x 14</td>
<td>0.1 cm x 15</td>
<td>0.2 cm x 13</td>
<td>0.1 cm x 15</td>
<td>0.3 cm x 14</td>
</tr>
<tr>
<td>3 x 3</td>
<td>0.3 cm x 13</td>
<td>0.1 cm x 14</td>
<td>0.3 cm x 6</td>
<td>0.1 cm x 14</td>
<td>0.3 cm x 13</td>
</tr>
</tbody>
</table>

Table 4.2: Voxel sizes used for the dose profiles at 1.5 cm depth for the verification of the tuning results simulations.

4.6 Results

The method for tuning that was developed, starts from standard commissioning data, can easily be automated and the duration of which is acceptable. The 6 MV photon model for the Varian linear accelerator was tuned in a number of stages.

4.6.1 Phase 1 - Energy & FWHM of the incident electron beam

For the initial tuning process, the RMSD value was calculated between -15 cm and +15 cm for the Varian model for the dose profile obtained at 1.5 cm depth in the water phantom from each simulation. These simulations had an average uncertainty of 1.3%. From the RMSD values (see table 4.3), it was decided that the value for the energy of the incident electron beam was located somewhere between 6 and 6.5 MeV. This is confirmed by the dose profiles (figure 4.6), as the measurements (continuous line) can be seen to lie between the 6 MeV (blue points) and 6.5 MeV (orange points) values. The RMSD values and the profiles did not point to any obvious trend in the FWHM so the same values were used again in phase 2.
### Table 4.3: The RMSD values obtained from the initial tuning process for the dose profiles at 1.5 cm depth for a 30 x 30 cm$^2$ field.

<table>
<thead>
<tr>
<th>Energy of source (MeV)</th>
<th>FWHM (cm)</th>
<th>Dose profile RMSD value</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>0.05</td>
<td>0.3352</td>
</tr>
<tr>
<td>5.5</td>
<td>0.05</td>
<td>0.1997</td>
</tr>
<tr>
<td>6</td>
<td>0.05</td>
<td>0.1126</td>
</tr>
<tr>
<td>6.5</td>
<td>0.05</td>
<td>0.1561</td>
</tr>
<tr>
<td>5</td>
<td>0.1</td>
<td>0.3077</td>
</tr>
<tr>
<td>5.5</td>
<td>0.1</td>
<td>0.1527</td>
</tr>
<tr>
<td>6</td>
<td>0.1</td>
<td>0.1183</td>
</tr>
<tr>
<td>6.5</td>
<td>0.1</td>
<td>0.1383</td>
</tr>
<tr>
<td>5</td>
<td>0.15</td>
<td>0.2586</td>
</tr>
<tr>
<td>5.5</td>
<td>0.15</td>
<td>0.1442</td>
</tr>
<tr>
<td>6</td>
<td>0.15</td>
<td>0.1256</td>
</tr>
<tr>
<td>6.5</td>
<td>0.15</td>
<td>0.1913</td>
</tr>
<tr>
<td>5</td>
<td>0.2</td>
<td>0.3194</td>
</tr>
<tr>
<td>5.5</td>
<td>0.2</td>
<td>0.2207</td>
</tr>
<tr>
<td>6</td>
<td>0.2</td>
<td>0.1094</td>
</tr>
<tr>
<td>6.5</td>
<td>0.2</td>
<td>0.2188</td>
</tr>
</tbody>
</table>

**Figure 4.6:** Dose profiles obtained at 1.5 cm depth for energies of 5.0, 5.5, 6.0, and 6.5 MeV and FWHM values of 0.05, 0.10, 0.15, and 0.20 cm from the initial tuning process (phase 1) for the Varian model.
4.6.2 Phase 2 - Fine tuning the energy:

Looking at the RMSD values obtained for the PDD’s and dose profiles from phase 2 of the tuning it was still not possible establish which values for energy and radius produce the best match to the measured data. Examination of the plots of the RMSD values deduced from the dose profiles (figure 4.7) show no clear trend in the effects of adjusting the energy or radius of the incident electron beam in phase 2 of the tuning process.

![Figure 4.7: Dose profile RMSD values as a function of energy for the data obtained running simulations with energies of 6.0, 6.1, 6.2, 6.3, 6.4, and 6.5 MeV and FWHM of 0.05, 0.10, 0.15, and 0.20 cm for the incident electron beam for phase 2 of the tuning process.](image)

4.6.3 Phase 3 - Output factors

As it was difficult to ascertain the best energy values more accurately than a rough interval determination between 6.0 MeV and 6.5 MeV, or to determine the
4. Optimization of tuning (Varian Tuning)

Figure 4.8: Dose profiles for the data obtained running simulations with energies of 6.1, 6.2, 6.3, and 6.4, and FWHM of 0.05, 0.10, 0.15, and 0.20 cm for the incident electron beam for phase 2 of the tuning process.

Gaussian radial distribution of the initial electron beam from the dose profiles or PDD’s calculated without increasing the accuracy of the simulation and hence the calculation times significantly, it was decided to change the focus of the simulations to examine the output factors obtained from the different beam energy and radius values for the 30 x 30 cm$^2$ field for this Varian model.

For the Varian linac, the dose profile simulations and initial output factor calculations indicated that the 6.3 MeV electron energy provided the best fit with least variance to the measured output factor values at all radial distributions of the incident electron beam. To determine the most appropriate value for the beam radius, we examined the output factors at different field sizes - focusing particularly on small fields as these are the most sensitive to changes in the spatial distribution in the beam.

Investigation of the output factors obtained for different square fields (40 x 40
Figure 4.9: Plot of the difference between measured and calculated output factors from phase 3 of the tuning process with the 1% difference lines drawn shown as dotted lines (this was decided as the tolerance for the accuracy of the output factors). The error bars represent the uncertainty in the Monte Carlo simulations.

cm², 30 x 30 cm², 5 x 5 cm², 4 x 4 cm², and 3 x 3 cm²) at the different Gaussian radial distributions led to the determination of the value for the beam radius as 0.05 cm as this produced a match within 1% of the measured values for the output factors at all field sizes for the Varian model (figure 4.9). The error bars show the uncertainty in the Monte Carlo calculated values. The uncertainty in the measurements is not taken into account as these values were used as the reference values. The uncertainty in measurements is usually assumed to be 2-3% for one off measurements though this can be larger for measurements for smaller fields.
4. Optimization of tuning (Varian Tuning)

4.6.4 Phase 4 - Verification of the tuning results:

Once the values were determined for the initial electron beam parameters, the PDD’s and dose profiles were simulated to a low uncertainty (average uncertainty of 0.8%), using the tuned primary electron beam energy and FWHM, for different field sizes and compared to measurements to verify the accuracy of the chosen parameters.

The tuned dose profiles can be seen in figure 4.10, while the PDDs as calculated using the tuned model are visible in figure 4.11 and the difference plots for the dose profiles and PDDs are shown in green in figures 4.10 and 4.11, respectively. The percentage difference was calculated at each position for which a Monte Carlo dose value had been obtained along the X-axis at 1.5 cm depth for the dose profile differences and along the central Z-axis for the PDD’s. The measured values were interpolated for these positions using MMCTP.

It was hoped that the inclusion of small fields in the tuning process would allow the model to be used for the simulation of IMRT fields. This application was tested in the work in chapter 6.

4.7 Timing

Every effort has been made in this work to devise a method to tune the models in the minimum amount of simulation time possible (see table 4.4), while maintaining a clinically significant level of accuracy (2-3%) in the resulting calculated doses and still keeping the complexity of the process to a minimum. This is done by using less accurate simulations in phase 1 to obtain an initial estimate of the energy of the primary electron beam. In phase 2 the number of histories is increased to obtain more accurate dose calculations but the voxels of the phantom are adjusted to allow these simulations to be subsequently used for the output
Figure 4.10: The dose profile comparisons for the tuned Varian model of Monte Carlo values (blue points) compared with measurements (red lines) for square fields of side (a) 30 cm, (b) 10 cm, (c) 4 cm, and (d) 3 cm. The dose profile differences for the tuned model compared with measurements for the fields are shown in green.

factor calculations. In phase 3, the wide field output factors are first calculated for the large field size and once a more accurate estimation of the primary electron energy is made, output factors for a range of field sizes are used to fine tune the FWHM of the radius of the electron beam. All the measured values required for this tuning method are routine QA measurements and would generally be readily available in the clinic. Consequently, this method does not require the laborious process of obtaining specialized individual measurements.
4. Optimization of tuning (Varian Tuning)

Figure 4.11: The PDD comparisons for the tuned Varian model of Monte Carlo values (blue points) compared with measurements (red lines) for square fields of side (a) 30 cm, (b) 10 cm, (c) 4 cm, and (d) 3 cm. The PDD differences for the tuned model compared with measurements for the fields are shown in green.

4.7.1 Computing resources

The computing resources used for this work is the CLUMEQ (Consortium Laval, University du Quebec, McGill and Eastern Quebec) Krylov cluster, which is located at the Montreal site in McGill University. It is a heterogeneous system comprising 21 quad-core nodes (84 cores) and 27 eight-core nodes (216 cores), for a total of 300 cores. The nodes run CENTOS 5.2 x86-64.
### 4. Optimization of tuning (Varian Tuning)

<table>
<thead>
<tr>
<th>Phase</th>
<th>Average time per calculation (Hours)</th>
<th>Number of calculations for this phase</th>
<th>Total time for this phase (CPU Hours)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5.0, 5.5, 6.0 &amp; 6.5 MeV</td>
<td>8.4</td>
<td>16</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>134.4</td>
</tr>
<tr>
<td>2</td>
<td>6.1, 6.2, 6.3 &amp; 6.4 MeV</td>
<td>25.6</td>
<td>16</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
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</tr>
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<td></td>
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</tr>
<tr>
<td></td>
<td>5 cm², 4 cm², and 3 cm²</td>
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<td>All fields</td>
<td>38.4</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>192.0</td>
</tr>
</tbody>
</table>

#### Table 4.4: Average computation hours on a 300 core cluster with a clock frequency of 2200 MHz.

#### 4.8 Discussion

For the Varian linac, the dose profile simulation and initial output factor calculations indicated that the 6.3 MeV electron energy provided the best fit with least variance to the measured values. The investigation of the output factors obtained for different square fields (40 x 40 cm\(^2\), 30 x 30 cm\(^2\), 5 x 5 cm\(^2\), 4 x 4 cm\(^2\), and 3 x 3 cm\(^2\)) at the different radial distributions determined a value for the beam radius of 0.05 cm as this produced a match within 1% of the measured values for the output factors at all field sizes. Once the values were determined for the initial electron beam parameters, the PDDs and dose profiles were simulated for different field sizes and compared to measurements to verify the results.

For this Varian model, the PDDs all agree within 1% after 1.5 cm depth apart from the 30 x 30 cm\(^2\) which is within 3.5% after 1.5 cm depth, though Pena et al. (2007) suggest the large differences for the 30 x 30 cm\(^2\) are not very clinically relevant (figure 4.11). The Varian model dose profiles agree within 1% outside of the penumbra for the smaller field sizes and within 1.5% outside of the penumbra for the 30 x 30 cm\(^2\) field and within the penumbra they agree within 3 mm, though
4. Optimization of tuning (Varian Tuning)

this is considered quite reasonable here as the detector used for the measurements had a diameter of 5 mm. Pena et al. (2007), also had worse agreement with larger fields for the Varian 2100 6 MV linac model. These verification simulations were calculated with an average uncertainty of 0.8%.

The value for the primary electron beam energy (6.3 MeV) is in keeping with the values obtained by other authors (Pena et al., 2007 & Sheikh-Bagheri and Rogers, 2002 & Keall et al., 2003 & Chibani et al., 2011) generally between 5.7 and 6.3 MeV. The beam radius value obtained (0.05 cm radius FWHM) is comparatively smaller than the values reported by Pena et al. (2007) & Sheikh-Bagheri and Rogers (2002) & Keall et al. (2003), where it has been shown to be between 0.1 cm and 0.2 cm. However this value is not unrealistic, as measurements of focal spot sizes (Jaffray et al., 1993) have shown a FWHM of 0.05 cm radius is possible as they recorded a FWHM of between 0.12 and 0.14 cm diameter for the Varian Clinac 2100. The work by Chibani et al. (2011) obtained a value of 0.07 cm for the radius of the FWHM, which is in agreement, within the accuracy of this investigation. As the values here were only correct to within 0.05 cm as this was considered sensitive enough for the purposes of this investigation.

A more rigorous and time consuming tuning process could provide a model tuned to 1% as implemented by Pena et al. (2007) or 1.5% as implemented by Sheikh-Bagheri and Rogers (2002). However, for the purposes of this investigation an accuracy of 2-3% was deemed adequate.

4.9 Conclusion

An efficient tuning method has been implemented using the MMCTP system and the BEAMnrc Monte Carlo code. The aim was to establish a model that agreed to measured data to within 2-3% so that it could be used as a verification tool for non-standard clinical treatment plans. The final model fit this requirement
apart from the $30 \times 30 \text{ cm}^2$ PDD at greater depths (approximately 20 cm depth or greater). This work shows that the goal of establishing a convenient and minimally time intensive method for linac model tuning, suitable for clinical implementation has been achieved.
Chapter 5

Tuning of the Siemens model
(Verification of the optimization technique)

The technique devised in chapter 4 is further tested here by utilising it to tune the Siemens linear accelerator from the clinic in Galway University Hospital based on measurements taken there as part of the normal QA process.

5.1 The Siemens accelerator model

The Siemens model was initially modelled based on information received from the vendor under a non-disclosure agreement. This provided data on the materials and dimensions of the primary beam components of the accelerator head in the form of data listed in excel sheets and CAD drawings. The accelerator
head was modelled using the following component modules (CM’s) to model the corresponding linac components.

- The FLATFILT CM was used to create the accelerator tungsten target
- The FLATFILT CM was used to create the primary collimator and flattening filter of the accelerator
- The CHAMBER CM was used to create the photon dose chamber
- The MIRROR CM was used to create the photon mirror
- The JAWS CM was used to add a pair of Y jaws to the model
- The MLC CM was used to model the MLC of the Siemens accelerator
- Finally, the SLABS CM was used to add an air slab to the end of the accelerator to bring the model to the desired distance from the source. This allows the phase space files to be created at a consistent location.

An image of this accelerator model as displayed in the BEAMnrc code can be seen in figure 5.1.

5.2 Siemens Monte Carlo accelerator model sources

The amount of published work on the process of tuning a Siemens accelerator model (and in particular the Oncor accelerator) is relatively limited when compared to Varian accelerators. One paper by Sawkey and Faddegon (2009) that has, investigated the tuning of the Siemens Oncor accelerator with fields of 5 x 5 cm² for depth dose curves and fields of 40 x 40 cm² for dose profiles. The use of such a small data set of field sizes (i.e. just 2 field sizes used) could produce a
Figure 5.1: The BEAMnrc model of the Siemens Oncor 6 MV photon linear accelerator.
tuned model which does not agree for all field sizes and in particular small field sizes can be especially difficult to tune (Scott et al., 2009).

This work by Sawkey and Faddegon (2009) also uses a method of dismantling a research accelerator in order to achieve a very accurate model. However, as clinically available measurements are usually only accurate to within 2%-3% (Ma and Jiang, 1999), there would be little point in trying to tune a model to measurements to less than this level of accuracy. Also, the practice of dismantling an accelerator is not a clinically viable option, not to mention incredibly time consuming and cumbersome.

5.3 Tuning the Siemens model

This section of work aimed to verify the efficacy and simplicity of the previously established tuning method by using it to tune a Siemens Oncor accelerator for 6 MV photon beams. The same procedure as was used for tuning the Varian model (figure 4.5) was followed again here as closely as possible. The process was designed to be used with already available dose data acquired during QA processes, as a result some slight alterations of the process were necessary to work with the data available.

5.4 Materials & Methods

The electron beam entering the linac model was assumed to be mono-energetic with a circular cross section and with a Gaussian spread as was done for the Varian model (section 4.3). The simulation results were compared to measurements taken in the hospital as part of the normal QA process.
The largest field with dose profile data available from the QA process was the 20 x 20 cm\(^2\) field, so this was the field used for the initial tuning process. Data from the output factors for various square fields is also used for the tuning process. The final parameters are verified for PDD’s and dose profiles for different square fields using available measurement data.

### 5.4.1 Measurements

The measurements from the radiotherapy department in GUH were performed using a linear array detector (LA48) with an air-scanner adapter (PTW, Germany) which is scanned through a 40 x 40 x 40 cm\(^3\) water tank with measurements taken at increments of 0.5 cm down to a depth of 34 cm for the PDD and 0.1 cm increments for the dose profile in the X and Y directions going from -16 cm to +16 cm from the central axis position for the 20 x 20 cm\(^2\) field. Images of the equipment used and of the setup are shown in figures 5.2 and 5.3, respectively.

![The LA48 linear array detector from PTW, Germany.](image)

#### Figure 5.2: The LA48 linear array detector from PTW, Germany.

### 5.4.2 Monte Carlo

The tuning process was carried out in 4 phases. Phase 1 started with the large field profiles (20 x 20 cm\(^2\) for this Siemens model) with a broad range of energy values and radius values for the primary electron beam. Phase 2 then aims at fine tuning the energy further. Phase 3 progresses to examination of output factors...
for a range of field sizes. Finally, phase 4 is used to verify the chosen parameters by comparison of measurements and Monte Carlo simulations of PDD’s and dose profiles for a range of field sizes.

5.4.3 Phase 1

The 20 x 20 cm$^2$ field was used for the Siemens model as this was the largest field size which data was readily available. For the initial tuning process, the dose profiles at 1.5 cm depth were calculated for the 20 x 20 cm$^2$ field for electron beam energies of 5.5 MeV, 6.0 MeV, and 6.5 MeV at each of the FWHM values of 0.05 cm, 0.10 cm, 0.15 cm, and 0.20 cm.

Dose profiles were calculated in a 30 cm x 30 cm x 30 cm water phantom, with the same X-axis and Y-axis voxel dimensions as listed in table 5.1. The Z-axis voxels were 0.2 cm from 0 cm down to 4.0 cm depth and 1 cm from 4 cm down to 28 cm depth. The resulting dose profile from each simulation was plotted against
the measured dose profile for the 20 x 20 cm² field and the RMSD value was calculated for X = -10 cm to +10 cm for the field using the equation 4.1.

<table>
<thead>
<tr>
<th>From</th>
<th>To</th>
<th>Voxel size</th>
<th>No. of Voxels</th>
</tr>
</thead>
<tbody>
<tr>
<td>-15.3 cm</td>
<td>-11.1 cm</td>
<td>0.6 cm</td>
<td>7</td>
</tr>
<tr>
<td>-11.1 cm</td>
<td>-8.5 cm</td>
<td>0.2 cm</td>
<td>13</td>
</tr>
<tr>
<td>-8.5 cm</td>
<td>8.5 cm</td>
<td>1.0 cm</td>
<td>17</td>
</tr>
<tr>
<td>8.5 cm</td>
<td>11.1 cm</td>
<td>0.2 cm</td>
<td>13</td>
</tr>
<tr>
<td>11.1 cm</td>
<td>15.3 cm</td>
<td>0.6 cm</td>
<td>7</td>
</tr>
</tbody>
</table>

Table 5.1: X-axis and Y-axis voxel dimensions used for the dose profiles at 1.5 cm depth for the initial phase of the Siemens model tuning.

5.4.4 Phase 2

From the first phase of the tuning process it was apparent that the best value for the energy of the incident electron beam lay between 6.0 and 6.5 MeV (figure 5.4). Phase 2 of the tuning process focused on energies of 6.1, 6.2, 6.3, and 6.4 MeV and FWHM values of 0.10 cm, 0.15 cm, and 0.20 cm.

The FWHM of 0.05 cm was not used as it provided the largest RMSD values in phase 1 so as a time saving measure was not included in phase 2 of the tuning process. The same phantom dimensions and voxel sizes as phase 1 were used. As was done for the Varian tuning, Phase 2 of the Siemens tuning was run with increased histories when compared to Phase 1 in an attempt to make any differences between the resulting profiles more obvious.

5.4.5 Phase 3

For Phase 3 of the tuning process the output factors were first calculated at 10 cm depth for the 20 x 20 cm² as this was the depth the clinical output factors were measured at. Following on from this, it was decided that values should be calculated for output factors for primary beam parameters of 6.3 MeV at 0.15
5. Tuning of the Siemens model

cm, 6.4 MeV at 0.10 cm, and 6.4 MeV at 0.15 cm for field sizes of 4 x 4 cm², 5 x 5 cm², 30 x 30 cm², and 40 x 40 cm².

The results from these simulations provided a match at an energy of 6.4 MeV and different radii were selected for large (10 cm and greater) fields and for the small fields (less than 10 cm). A FWHM of 0.1 cm was decided on for the small fields and a FWHM of 0.15 cm for the large fields was selected. The different values for the FWHM were necessary in order to achieve a 1% agreement between the measured and calculated output factors.

5.4.6 Phase 4

The calculations were simulated to a low uncertainty for fields of 5 x 5 cm², 10 x 10 cm², and 20 x 20 cm² for dose profiles and fields of 4 x 4 cm², 5 x 5 cm², 10 x 10 cm², and 20 x 20 cm² for PDD’s. These were plotted against measured data for comparison in order to verify the accuracy of the final result.

The voxel sizes used along the X-axis and Y-axis were varied in order to get an accurate position of the penumbra while keeping the voxel sizes outside this region reasonably large in order to have an accurate estimation of the dose in these regions, without making the simulations prohibitively time consuming. The sizes of the voxels used can be seen in tables 5.2, 5.3, and 5.4 for the 5 x 5 cm², 10 x 10 cm², and 20 x 20 cm² fields, respectively.

5.4.7 Input parameters

The EGSnrc physics parameters used in the input files for both the BEAMnrc and DOSXYZnrc parts of the simulations for tuning the Siemens model were as shown in table 5.5. The same parameters were used for all phases of the tuning process. These were the default transport options.
Table 5.2: X-axis and Y-axis voxel dimensions used for the dose profiles at 10 cm depth for the final phase of the Siemens model tuning for the 5 x 5 cm$^2$ field.

<table>
<thead>
<tr>
<th>From</th>
<th>To</th>
<th>Voxel size</th>
<th>No. of Voxels</th>
</tr>
</thead>
<tbody>
<tr>
<td>-9.25 cm</td>
<td>-4.25 cm</td>
<td>0.5 cm</td>
<td>10</td>
</tr>
<tr>
<td>-4.25 cm</td>
<td>-1.85 cm</td>
<td>0.1 cm</td>
<td>24</td>
</tr>
<tr>
<td>-1.85 cm</td>
<td>-0.25 cm</td>
<td>0.2 cm</td>
<td>8</td>
</tr>
<tr>
<td>-0.25 cm</td>
<td>0.25 cm</td>
<td>0.5 cm</td>
<td>1</td>
</tr>
<tr>
<td>0.25 cm</td>
<td>1.85 cm</td>
<td>0.2 cm</td>
<td>8</td>
</tr>
<tr>
<td>1.85 cm</td>
<td>4.25 cm</td>
<td>0.1 cm</td>
<td>24</td>
</tr>
<tr>
<td>4.25 cm</td>
<td>9.25 cm</td>
<td>0.5 cm</td>
<td>10</td>
</tr>
</tbody>
</table>

Table 5.3: X-axis and Y-axis voxel dimensions used for the dose profiles at 10 cm depth for the final phase of the Siemens model tuning for the 10 x 10 cm$^2$ field.

<table>
<thead>
<tr>
<th>From</th>
<th>To</th>
<th>Voxel size</th>
<th>No. of Voxels</th>
</tr>
</thead>
<tbody>
<tr>
<td>-10.75 cm</td>
<td>-5.75 cm</td>
<td>0.5 cm</td>
<td>10</td>
</tr>
<tr>
<td>-5.75 cm</td>
<td>-3.75 cm</td>
<td>0.2 cm</td>
<td>10</td>
</tr>
<tr>
<td>-3.75 cm</td>
<td>-3.25 cm</td>
<td>0.5 cm</td>
<td>1</td>
</tr>
<tr>
<td>-3.25 cm</td>
<td>-0.25 cm</td>
<td>1 cm</td>
<td>3</td>
</tr>
<tr>
<td>-0.25 cm</td>
<td>0.25 cm</td>
<td>0.5 cm</td>
<td>1</td>
</tr>
<tr>
<td>0.25 cm</td>
<td>3.25 cm</td>
<td>1 cm</td>
<td>3</td>
</tr>
<tr>
<td>3.25 cm</td>
<td>3.75 cm</td>
<td>0.5 cm</td>
<td>1</td>
</tr>
<tr>
<td>3.75 cm</td>
<td>5.75 cm</td>
<td>0.2 cm</td>
<td>10</td>
</tr>
<tr>
<td>5.75 cm</td>
<td>10.75 cm</td>
<td>0.5 cm</td>
<td>10</td>
</tr>
</tbody>
</table>

Table 5.4: X-axis and Y-axis voxel dimensions used for the dose profiles at 10 cm depth for the final phase of the Siemens model tuning for the 20 x 20 cm$^2$ field.

<table>
<thead>
<tr>
<th>From</th>
<th>To</th>
<th>Voxel size</th>
<th>No. of Voxels</th>
</tr>
</thead>
<tbody>
<tr>
<td>-15.85 cm</td>
<td>-10.85 cm</td>
<td>0.5 cm</td>
<td>10</td>
</tr>
<tr>
<td>-10.85 cm</td>
<td>-8.25 cm</td>
<td>0.2 cm</td>
<td>13</td>
</tr>
<tr>
<td>-8.25 cm</td>
<td>-0.25 cm</td>
<td>1 cm</td>
<td>8</td>
</tr>
<tr>
<td>-0.25 cm</td>
<td>0.25 cm</td>
<td>0.5 cm</td>
<td>1</td>
</tr>
<tr>
<td>0.25 cm</td>
<td>8.25 cm</td>
<td>1 cm</td>
<td>8</td>
</tr>
<tr>
<td>8.25 cm</td>
<td>10.85 cm</td>
<td>0.2 cm</td>
<td>13</td>
</tr>
<tr>
<td>10.85 cm</td>
<td>15.85 cm</td>
<td>0.5 cm</td>
<td>10</td>
</tr>
</tbody>
</table>

DBS was turned on in the BEAMnrc input files with a splitting number of 1000, a splitting field radius of 25 cm and a source to surface distance of 90 cm. The phase space files were all scored at 90 cm. There was no electron splitting and range rejection was not used. These same inputs were used for each phase of the
Tuning of the Siemens model

Monte Carlo transport parameters for all phases:

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Global ECUT=</td>
<td>0.521</td>
</tr>
<tr>
<td>Global PCUT=</td>
<td>0.01</td>
</tr>
<tr>
<td>Global SMAX=</td>
<td>5</td>
</tr>
<tr>
<td>ESTEPE=</td>
<td>0.25</td>
</tr>
<tr>
<td>XIMAX=</td>
<td>0.5</td>
</tr>
<tr>
<td>Boundary crossing algorithm=</td>
<td>PRESTA-I</td>
</tr>
<tr>
<td>Skin depth for BCA=</td>
<td>0</td>
</tr>
<tr>
<td>Electron-step algorithm=</td>
<td>PRESTA-II</td>
</tr>
<tr>
<td>Spin effects=</td>
<td>On</td>
</tr>
<tr>
<td>Brems angular sampling=</td>
<td>Simple</td>
</tr>
<tr>
<td>Brems cross sections=</td>
<td>BH</td>
</tr>
<tr>
<td>Bound Compton scattering=</td>
<td>Off</td>
</tr>
<tr>
<td>Pair angular sampling=</td>
<td>Simple</td>
</tr>
<tr>
<td>Photoelectron angular sampling=</td>
<td>Off</td>
</tr>
<tr>
<td>Rayleigh scattering=</td>
<td>Off</td>
</tr>
<tr>
<td>Atomic relaxations=</td>
<td>Off</td>
</tr>
<tr>
<td>Electron impact ionization=</td>
<td>Off</td>
</tr>
</tbody>
</table>

Table 5.5: The Monte Carlo transport parameters used in the .egsinp files for all phases of the Siemens tuning.

5.5 Results

5.5.1 Phase 1

Phase 1 of the tuning process was a preliminary investigation to try narrow the range of values for the energy and FWHM for the primary electron beam. Dose profiles were obtained from simulations of a 20 x 20 cm² field at 1.5 cm depth in a water phantom. The dose profiles were calculated for a combination of different energies and beam radius values of the primary electron beam.

The graph of the dose profiles obtained from the initial tuning process compared to the measured dose profile can be seen in figure 5.4. The measured values lie between 6.0 MeV (green symbols) and 6.5 MeV (blue symbols), though there is no clear trend in the results obtained with regard to the FWHM of the primary
5. Tuning of the Siemens model electron beam. The corresponding RMSD values obtained for the simulations can be seen in figure 5.5, here the RMSD values at 0.05 cm FWHM can be seen to be large for all values of beam energy relative to the RMSD values at other FWHM values.

Figure 5.5: Graph showing RMSD values for various beam energies at 0.05 cm FWHM, compared to other FWHM values.

5.5.2 Phase 2

Based on the results from phase 1, it was decided that the energy of the primary electron beam lay in the range of 6.0 MeV to 6.5 MeV. The energies used in phase 2 of the tuning process were: 6.1 MeV, 6.2 MeV, 6.3 MeV, and 6.4 MeV. The values for the FWHM used were 0.1 cm, 0.15 cm, and 0.20 cm. The FWHM of 0.05 cm was omitted as a time saving measure, as it provided the largest RMSD values in phase 1.

The dose profiles obtained can be seen in figure 5.6. The dose profiles became too close to each other to determine which matches the measured data best despite...
5. Tuning of the Siemens model

Figure 5.5: The RMSD values for the dose profiles obtained at 1.5 cm depth from the initial phase of the Siemens model tuning process.

these curves being smoother than those obtained in phase 1 due to the larger number of histories used here. The RMSD values obtained are shown in figure 5.7, while there was no obvious trend in the values, the RMSD values at 0.15 cm did seem to be consistent and to agree well for nearly all energies for this 20 x 20 cm$^2$ field.

5.5.3 Phase 3

As there was no indication to the best energy or beam radius from the results of phase 2 the output factors were first calculated for the 20 x 20 cm$^2$ field for the same energies of 6.1 MeV, 6.2 MeV, 6.3 MeV, and 6.4 MeV and same FWHM values of 0.10 cm, 0.15 cm, and 0.20 cm. The results for the output factors obtained are presented in table 5.7.

From this it was decided that energy values of 6.3 MeV and 6.4 MeV provided the best match as the output factors calculated at these energies were within 2% of the measured value for all FWHM values used (table 5.7). The output factors
5. Tuning of the Siemens model

**Figure 5.6:** Dose profiles obtained at 1.5 cm depth from the second phase of the Siemens model tuning process.

**Figure 5.7:** The RMSD values for the dose profiles obtained at 1.5 cm depth from the second phase of the Siemens model tuning process.
were then calculated for field sizes of 4 x 4 cm$^2$, 5 x 5 cm$^2$, 30 x 30 cm$^2$, and 40 x 40 cm$^2$ for 6.3 MeV at 0.15 cm, 6.4 MeV at 0.10 cm, and 6.4 MeV at 0.15 cm.

<table>
<thead>
<tr>
<th>Field Size $cm^2$</th>
<th>Output Factor</th>
</tr>
</thead>
<tbody>
<tr>
<td>4 x 4</td>
<td>0.886</td>
</tr>
<tr>
<td>5 x 5</td>
<td>0.857</td>
</tr>
<tr>
<td>20 x 20</td>
<td>1.111</td>
</tr>
<tr>
<td>30 x 30</td>
<td>1.162</td>
</tr>
<tr>
<td>40 x 40</td>
<td>1.188</td>
</tr>
</tbody>
</table>

Table 5.6: The measured output factors for the 4 x 4 cm$^2$, 5 x 5 cm$^2$, 20 x 20 cm$^2$, 30 x 30 cm$^2$, and 40 x 40 cm$^2$ fields.

<table>
<thead>
<tr>
<th>Energy (MeV)</th>
<th>FWHM (cm)</th>
<th>Output Factor</th>
<th>% Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>6.1</td>
<td>0.10</td>
<td>1.131</td>
<td>-1.77</td>
</tr>
<tr>
<td>6.1</td>
<td>0.15</td>
<td>1.069</td>
<td>3.81</td>
</tr>
<tr>
<td>6.1</td>
<td>0.20</td>
<td>1.130</td>
<td>-1.67</td>
</tr>
<tr>
<td>6.2</td>
<td>0.10</td>
<td>1.086</td>
<td>2.28</td>
</tr>
<tr>
<td>6.2</td>
<td>0.15</td>
<td>1.074</td>
<td>3.30</td>
</tr>
<tr>
<td>6.2</td>
<td>0.20</td>
<td>1.118</td>
<td>-0.61</td>
</tr>
<tr>
<td>6.3</td>
<td>0.10</td>
<td>1.129</td>
<td>-1.61</td>
</tr>
<tr>
<td>6.3</td>
<td>0.15</td>
<td>1.107</td>
<td>0.36</td>
</tr>
<tr>
<td>6.3</td>
<td>0.20</td>
<td>1.099</td>
<td>1.09</td>
</tr>
<tr>
<td>6.4</td>
<td>0.10</td>
<td>1.095</td>
<td>1.41</td>
</tr>
<tr>
<td>6.4</td>
<td>0.15</td>
<td>1.099</td>
<td>1.07</td>
</tr>
<tr>
<td>6.4</td>
<td>0.20</td>
<td>1.089</td>
<td>1.96</td>
</tr>
</tbody>
</table>

Table 5.7: The output factors calculated for the different energy and FWHM values for the 20 x 20 cm$^2$ field.

5.5.4 Phase 4

It was decided that an energy of 6.4 MeV, and radius of 0.15 cm for large fields (10 x 10 cm$^2$ fields and greater) and of 0.10 cm for the small fields (less than 10 x 10 cm$^2$) provided the best agreement. The accuracy of these parameters was verified by running simulations to low uncertainties. For the PDD’s, field sizes of 4 x 4 cm$^2$, 5 x 5 cm$^2$, 10 x 10 cm$^2$, and 20 x 20 cm$^2$ were simulated, and for dose profiles, field sizes of 5 x 5 cm$^2$, 10 x 10 cm$^2$, and 20 x 20 cm$^2$ were simulated. The resulting dose distributions were compared to the corresponding QA measurements.

The results obtained for these PDDs and dose profiles are shown in figures 5.8, and ???. The dose profile results (figure ???) agree within 2.5% outside the penumbra
### Table 5.8: The output factors calculated for the 6.3 MeV and 6.4 MeV energies and 0.10 cm and 0.15 cm FWHM values for the different field sizes.

<table>
<thead>
<tr>
<th>Field size (cm²)</th>
<th>Energy (MeV)</th>
<th>FWHM (cm)</th>
<th>Output Factor</th>
<th>Difference %</th>
</tr>
</thead>
<tbody>
<tr>
<td>4 x 4</td>
<td>6.3</td>
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For the 5 x 5 cm² and 10 x 10 cm² fields and within 3% outside the penumbra for the 20 x 20 cm². It can be seen from the green curves in figure 5.8, that the PDD’s agree within 3% after 1.5 cm depth for all field sizes calculated.

#### 5.6 Discussion

In order to maintain a 1% agreement in the output factors for the Siemens Oncor accelerator model, different tuned values were found for the FWHM of the radius of the primary electron beam for small fields (below 10 cm) and large fields (10 cm and greater). Work by Jaffray et al. (1993) has shown that the size of the incident electron beam varies with changes in the field size because of the secondary collimators blocking part of the extra-focal portion of the source.

This is also in keeping with what has been previously implemented in the commercial treatment planning system where this Siemens linac is in use clinically. The tuned values for the Siemens linac are an energy of 6.4 MeV and a FWHM
of 0.1 cm for small fields and 0.15 cm for large fields, correct to the 0.05 cm increments used. The beam radii used in the commercial treatment planning system for the Siemens model are 0.1505 cm for small fields and 0.1725 cm for large fields which is approximately in agreement with the results, within the accuracy of the values (i.e. the 0.05 cm increments used), from this work. The dose profile results agree within 2.5% outside the penumbra for the 5 x 5 cm$^2$ and 10 x 10 cm$^2$ fields and within 3% outside the penumbra for the 20 x 20 cm$^2$. The PDD’s agree within 3% after 1.5 cm depth for all field sizes calculated. These difference values are perhaps slightly high for a tuned model but this seems to be due to the smaller voxel sizes used in the penumbra of the beam for the dose profiles.
5. Tuning of the Siemens model

Figure 5.9: The dose profile comparisons for the tuned Siemens model of Monte Carlo values (blue points) compared with measurements (red lines) for square fields of side (a) 20 cm, (b) 10 cm, and (c) 5 cm. The dose profile differences for the tuned model compared with measurements for the fields are shown in green.

and in the build-up region of the beam for the PDD’s. The smaller voxel sizes were used in these areas to try to obtain a more accurate shape to the curves but it has, however, also led to a greater uncertainty in the doses in these voxels.

5.7 Conclusions

The results obtained for the initial electron beam energy of the Siemens model, 6.4 MeV, is in keeping with published data (Keall et al., 2003). The value obtained for the FWHM of the radius of the initial electron beam, of 0.1 cm and 0.15 cm is also in keeping with previously published results on Monte Carlo investigations of
Siemens linacs (Pena et al., 2007) and with published results on the measurements of the beam radius for Siemens machines (Jaffray et al., 1993).

However, no other research of Monte Carlo accelerator models suggests different values for the FWHM for the smaller and larger field sizes. This could be due to a lack of small field investigations for Siemens machines as the work by Jaffray et al. (1993) has shown with measurements that the dimensions of the incident electron beam vary with changes in the field size because of the secondary collimators blocking part of the extra-focal portion of the source. This work shows that the goal of establishing a convenient and minimally time intensive method for linac model tuning, suitable for clinical implementation has been achieved.

5.8 Further optimisation of the process

It could be possible to further optimise this tuning process by perhaps using only one FWHM for the initial tuning phases, though having multiple simulations at each energy makes it easier to more definitively identify the energy region within which the measured values lie.

It may also reduce the amount of simulation time required to by-pass phase 2 of the tuning process and skip straight to investigating the output factors, as phase 2 of the tuning process did not point to any obvious trend in the parameters for either the energy or the radius of the primary electron beam and this was the same for the tuning of the Varian model. Based on the times taken for the Varian tuning, shown in table 4.4, this would lead to a 26% reduction in the amount of time required to complete the tuning process (similar times were required for the Siemens tuning).
Chapter 6

IMRT plan verification

The work presented in this chapter focused on the novel examination of collapsed versus rotated methods for the measurement of the absolute dose for pre-treatment IMRT plan verification. A number of complexity metrics were also examined to see if any could serve as a predictor for the recommendation of a particular QA technique for a given plan.

6.1 IMRT absolute dose measurement

One of the primary issues with a more wide spread implementation of intensity modulated radiation therapy (IMRT) in clinics with the available facilities is that it can be significantly more time consuming than conventional radiotherapy (Boyer et al., 2001 & Rangel et al., 2010). Work by Miles et al. (2005) showed an average increase in physics man-hours of 4.9 hours per patient treated. One of the aspects of radiotherapy treatment that has increased the amount of time required is the dosimetric verification of each treatment plan (Chang et al., 2000
6. IMRT plan verification

& Verellen et al., 1997 & Lagares et al., 2002) to ensure an accurate estimation of patient dose. As recommended by the AAPM in their guidance document (Ezzell et al., 2003) individual patient pre-IMRT plan verification measurements on a homogeneous phantom should include the measurement of the absolute dose (at least to a point) and measurement of the dose distribution. A recent survey by Nelms and Simon (2007) showed that a significant proportion of clinics (32.8%) use the single gantry angle composite (SGAC) technique for verification of 75 - 100% of patient IMRT plans in what is assumed to be an attempt to reduce the amount of time spent carrying out IMRT plan validation for each individual patient. With such a comparatively high amount of clinics using this practice, it would seem to warrant further investigation and verification as an IMRT QA technique. This work aims to examine establish and optimize a Monte Carlo dosimetry IMRT verification of treatment plans in two different delivery modes and to compare that to the planning system dose calculations.

Intensity modulated radiotherapy (IMRT) is a significant technological advance in the field of radiotherapy which allows the dose to tumors to be maximized while concurrently minimizing the dose to the surrounding healthy tissues by sculpting a high-dose field around the disease site with hitherto unachievable precision (Webb, 2003). This increases the chances of controlling the spread of the cancer (Kuban et al., 2008 & Lee et al., 2002) while reducing the side effects of radiotherapy treatment (Zelefsky et al., 2000 & Pignol et al., 2008). IMRT is most commonly implemented in a hospital radiotherapy department using a linear accelerator with a multi-leaf collimator (MLC) to administer a series of beams, from various angles, with non-uniform fluences. The varying fluence is achieved by dividing each beam into a series of beamlets or segments of varying intensities. The primary objective of radiotherapy is to increase the dose to the cancerous tissue while at the same time keeping the dose to the surrounding normal tissues and organs to an acceptable low, safe level. The techniques being employed
to achieve this goal are becoming increasingly complex and commonplace, so the need for a manageable and feasible, yet reliable verification technique is increasing (Arnfield et al., 2001).

In order for IMRT treatments to achieve their potential it is necessary to be able to accurately verify the radiation dose that will be administered to the patient in these treatment techniques (Ramsey et al., 2003 & Niemierko, 2004). Previous studies have shown that standard treatment planning methods in certain clinical IMRT configurations are limited in the prediction of the radiation dose. Ibbott et al. (2008) reported that roughly 30% of institutions failed to deliver a dose distribution to a head and neck phantom that matched their own treatment planning system dose to within 7% for dose or 4 mm distance to agreement.

There are many reports documenting the limitations of commercial treatment planning algorithms in planning IMRT treatments especially in complicated situations such as for head and neck cancers (Ma et al., 2000 & Das et al., 2008). It has been determined that as many as 46% of patients receive a maximum dose that is more than 10% higher than the prescribed dose, and 63% of patients receive a dose that is more than 10% lower than the prescribed dose (Das et al., 2008). This finding has very significant implications for the treatment of cancer patients using IMRT as a dose 5% lower than the prescribed dose may result in clinically detectable reduction in tumour control. The International Commission on Radiation Units and Measurements (ICRU) No.50 guidelines recommends that radiation dose be delivered to within -5% and +7% of the dose prescribed to the target (ICRU, 1993, 1999). Knowledge of the precise dose, and as a result quality assurance, form a vital part of the radiotherapy treatment process (Dische et al., 1993). Each individual IMRT patient plan is verified through a two step process, as advised by the AAPM (Ezzell et al., 2003), of dose verification of the plan on a homogeneous phantom:
6. IMRT plan verification

- Measurement of the absorbed dose to a point in a high-dose, low-dose-gradient region in a clinically significant volume of the phantom with an ion chamber.

- The measurement of the two dimensional relative dose distribution, with either film or a diode array device and a gamma analysis of the resulting dose distribution when compared to the plan.

As this procedure is required for each IMRT patient plan before treatment commences, it can be seen how this would contribute to a significant increase in time and personnel requirements. This is one possible reason for the wide spread use of the faster SGAC technique (Nelms and Simon, 2007) in IMRT QA.

Alfonso et al. (2008), introduced a new approach for non-standard beam reference dosimetry. For the calculation of dose for composite fields they recommend the use of an intermediate calibration field which they call a plan-class specific reference (pcs) field. This pcs field is closer to the patient-specific clinical fields and should provide a uniform dose over a region exceeding the dimensions of the reference detector. A correction factor is provided from this to compensate for differences between the standard calibration field and the small, composite fields used in IMRT. This is done to try to compensate for the deviations from charged particle equilibrium that can have an effect when using small fields as is particularly applicable for IMRT QA. This is not accounted for in conventional dosimetry protocols which are based on the absorbed dose to water calibration at a reference field, usually a 10 x 10 cm² field.

In this work Monte Carlo methods were used to investigate the reliability of using a single gantry angle or collapsed beam configuration as opposed to measuring the doses at the planned angles or rotated beam delivery, when measuring the absorbed dose to a point in a high-dose, low-dose-gradient region. This work aimed to quantify the $Q_{clin}/Q_{ref}$ correction factor for collapsed and rotated beam
6. IMRT plan verification

Figure 6.1: Schematic overview of the dosimetry of nonstandard composite fields with reference to a plan-class specific reference field according to the formalism presented by Alfonso et al. (2008). Reproduced from Alfonso et al. (2008).

deliveries. The investigation looked at different plan complexity metrics to determine a predictor of the reliability of the QA technique. The relationship between the $k_{Q_{clin},ref}$ correction factor and the dose homogeneity index investigated, could help with determining suitable reference fields (pcsr fields) when implementing the Alfonso et al. (2008) formalism. This work was carried out due to the lack of investigations into collapsed versus rotated IMRT QA, particularly the lack of research into the effects, if any, of the QA method on the deviation of the dosimetry from the reference conditions (i.e. the use of small fields and comparing them to 10 x 10 cm$^2$).

6.2 Materials & Methods

MMCTP (Alexander et al., 2007) is a radiotherapy research platform, which enables comparison and analysis of dose distributions from treatment planning systems and quality assurance measurements with Monte Carlo calculated values.
on a common independent platform. The MMCTP system (Alexander et al., 2007) was used in this work to simplify and speed up the Monte Carlo simulation process. It automatically generates the files required for the Monte Carlo simulation process from the imported treatment planning files. The system also automatically submits the jobs to the remote cluster and downloads and imports the results once the simulation is complete.

Once an accurate linear accelerator model had been commissioned for the Varian linac (chapter 4), the BEAMnrc (Rogers et al., 1995, 2004) input file was saved as a template input file on which the patient-specific accelerator model was based in the IMRT simulations. The files were linked within MMCTP to a specific treatment machine and energy. The DOSXYZnrc part of the EGSnrc code (Walters and Rogers, 2004) was used to calculate the dose scored in the phantom. The phantom for these simulations was defined using MMCTP and was based on CT data of the measurement phantom.

The EGSnrc (Kawrakow, 2000) egs.chamber code (Wulff et al., 2008) was also used, which is an egs++ (Kawrakow et al., 2009) user code designed for chamber in phantom calculations. The user code, which defines the sources and geometries is written in C++, and is connected with the MORTRAN programmed EGSnrc backend which deals with the involved transport physics. The egs.chamber user code is similar to the old cavity (Kawrakow et al., 2009) user code but it implements three new variance reduction techniques: photon cross-section enhancement (XCSE), intermediate phase-space storage (IPSS), and correlated sampling (CS) which is the most powerful.

6.3 Measurements

Measurements were performed using a Farmer type ionization chamber inserted in a tightly fitting hole in a 30 x 30 x 17 cm$^3$ Solid Water™ phantom.
used fully rotational beam angles, as well as delivery from one angle (SGAC
technique) of the same patient plan beams with just the gantry angle being
altered. The verification plan was recalculated using the beams of the patient
plan. Dose volume statistics in the chamber volume as well as dose distributions
were recorded from the treatment planning system.

6.4 Patient treatment plans

There were 20 patient plans recalculated using the EGSnrc Monte Carlo code.
These plans were for a variety of treatment sites: 15 ENT, 3 head & neck, 1
abdominal and 1 anal canal. The number of beams per plan ranged from 7 up
to 19. The treatment plans were all dynamic IMRT plans and the number of
segments or beamlets per beam varied greatly. The 20 patients comprised of 10
patients that were the first 10 patients that QA was carried out for when the
clinic first moved to Eclipse™, Varian Medical Systems, TPS just to see what
the difference was between collapsed and rotated QA. In Corvus™, Nomos, the
collapsed QA option was not available so the collapsed option was not used until
the change was made to Eclipse. The other 10 patients were chosen as ones
where QA was first performed as collapsed and were then repeated with rotated
as they failed the collapsed QA procedure. The treatment planning system data
used was from Eclipse using Pencil Beam Convolution without corrections with
all calculations done with homogeneous calculation. As this work is based on
clinical data, there was no control over the patients chosen, as they were chosen
by the clinical staff at the hospital in Montreal.
6.5 Monte Carlo simulations: DOSXYZnrc

The plan data for the collapsed and rotated plans was imported into the MM-CTP system from the DICOM files created by the treatment planning system. This plan data provided information on the structure, beam configuration and calculated dose. This data when read in and combined with template input files creates the files necessary for the Monte Carlo simulations using the EGSnrc codes. The Monte Carlo simulations were run on a remote cluster. Previous tuning (chapter 4) had provided a refined BEAMnrc model for the Varian linac that agreed with measured dose profile curves to an accuracy of within 2% or 3 mm for square fields with sides of 3 cm up to 30 cm and within 1% for the output factors for field sizes of $40 \times 40$ cm$^2$, $30 \times 30$ cm$^2$, $5 \times 5$ cm$^2$, $4 \times 4$ cm$^2$, and $3 \times 3$ cm$^2$.

For the tuned Varian linac 6 MV photon model the electron beam entering the linac model had been chosen to be a mono-energetic 6.3 MeV beam with a circular cross section of 0.05 cm. The pegs4 data used were the ICRU data with production threshold for secondary electrons and cut-off energies of 189 keV and 10 keV for electrons and photons respectively.

Simulations were run in two steps: firstly, the BEAMnrc code modelled the photon production and particle transport in the accelerator and created a phase space file at 70 cm SSD; secondly, the DOSXYZnrc code modelled the particle transport and dose deposition in the phantom and the surrounding air or the egs_chamber code scored the average dose to the active chamber volume. The simulations were run in two parts as a time saving measure, because the same phase space files could be used as an input to the DOSXYZnrc simulations for both the rotated and collapsed simulations and later for the egs_chamber rotated and collapsed dose calculations. This is done as opposed to using isource 9 in DOSXYZnrc which is probably the more commonly used approach as it does not
require as much data storage. Also the use of isource 9 is more practical when splitting the simulation into many parallel jobs as there are not multiple sources trying to read the phase space file at the same time. In order to reduce the effect of this, the number of simultaneous jobs for the one simulation was usually kept to less than 4 jobs. Directional bremsstrahlung splitting (DBS) of 100 was used in the simulations and photon splitting of 100 was used in DOSXYZnrc to increase the calculation efficiency. ECUT and PCUT were set to 0.7 and 0.01 MeV respectively.

The dose-scoring phantom for the DOSXYZnrc simulations was generated from the CT data imported into the MMCTP program. The CT data showed a scan of a 30 x 30 x 17 cm$^3$ Solid Water$^{\text{TM}}$ phantom with the chamber centrally located within it as used for the dose measurements. The material within the phantom was set to water and the surrounding material was set to air. The dose scoring voxels were set to a size of 0.3 x 0.3 x 0.3 cm$^3$.

On completion of the simulations, the dose files output from DOSXYZnrc are downloaded from the remote cluster and read into MMCTP. MMCTP converts the dose per incident particle to Gy using a calibration factor obtained from a 10 x 10 cm$^2$ field run. This conversion allows for direct comparison between the Monte Carlo calculated doses and those obtained from the treatment planning system. This was discussed in section 3.9.

Once this is completed for each beam in the plan, MMCTP adds together the dose in each voxel from the different beams to obtain the total dose from the plan. DVH data and maximum, minimum and average doses are then derived for the chamber volume allowing for a comparison with measured values. This process was repeated for the plans for each of the 20 patients in this study.
6. IMRT plan verification

6.5.1 Input parameters for BEAMnrc and DOSXYZnrc

The EGSnrc physics parameters used in the input files for both the BEAMnrc and DOSXYZnrc parts of the simulations for tuning the Siemens model were as shown in tables 6.1 and 6.2. The same parameters were used for all the IMRT simulations.

DBS was turned on in the BEAMnrc input files with a splitting number of 1000 and a source to surface distance of 100 cm. The splitting field radius varied from field to field depending on the jaw settings and was calculated by MMCTP automatically to give the most efficient value. This is possible with the Varian model as the Varian linac has both an X and Y jaw as well as an MLC, and these jaws are set to back-up the MLC, allowing their positions to be used to calculate the most accurate splitting field radius. The phase space files were all scored at 70 cm. Electron splitting was used at the flattening filter with russian roulette. Electron range rejection was used with set ECUTRR set to 1.0 MeV. ECUT was 0.7 MeV and PCUT was 0.01 MeV. There was no photon forcing used. These same inputs were used for all the IMRT BEAMnrc simulations and the phase space files created were used as the inputs for both the DOSXYZnrc simulations and the egs.chamber simulations.

In the DOSXYZnrc simulations, photon splitting of 32 was used and range rejection was on with ESave of 2 MeV. The “HOWFARLESS” option (Walters and Rogers, 2004), was turned off and no recycling was used. The photon splitting number of 32 was chosen after a series of test simulations were run to determine the most efficient value.
6. IMRT plan verification

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Table 6.1: The Monte Carlo transport parameters used in the BEAMnrc input files for all the IMRT simulations.

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Table 6.2: The Monte Carlo transport parameters used in the DOSXYZnrc input files for all the IMRT simulations.

6.5.2 Pixelisation of the chamber

To allow for an analysis of the effects of the air perturbation of the chamber and pixelization of the chamber (figure 6.2) image when the CT data is imported into the TPS and imported into DOSXYZnrc as was done for the previous Monte
Carlo calculations, the Monte Carlo simulations were repeated, however, this time it was required that the chamber was modelled in full detail with the air cavity, wall and electrode materials and dimensions modelled as accurately as possible. In order to do this, the egs++ egs_chamber code needed to be utilized as it is the most efficient method within the EGSnrc package of calculating the dose to the active air cavity volume of a realistic ion chamber model.

While in DOSXYZnrc the chamber could be modelled more explicitly by assigning the required media to the relevant voxels in the phantom, it would be very difficult to model the chamber to any great accuracy given that the voxel sizes were 0.3 x 0.3 x 0.3 cm$^3$ and the active volume of the chamber used is only 0.65 cm$^3$ to begin with.

**Figure 6.2:** Images depicting the pixelisation of the farmer chamber in the DOSXYZnrc simulations. The XZ-plane is in the first column, the XY-plane is in the middle column, and the YZ-plane is in the third column.
6.6 Monte Carlo simulations: egs_chamber phantom

In order to investigate the effects of not explicitly modelling the chamber, the egs_chamber code was used to recalculate the dose. The same phase space files that were previously generated with the BEAMnrc code were used as the source for these simulations. The Exradin A12 chamber was modelled explicitly at 8.5 cm depth in a 30 x 30 x 17 cm$^3$ water phantom surrounded by air (i.e. the same setup as was used for the measurements). This was implemented in the egs_chamber code which is designed to accurately model chambers. It is an egs++ code that comes as part of the EGSnrc system (Kawrakow et al., 2011).

6.6.1 The egs_chamber code

One of the advantages of the egs_chamber code, over the egs++ cavity code, is the use of photon cross section enhancement (XCSE). XCSE is the implementation of photon splitting on a region by region basis. This means that photon splitting can be just implemented in the region of the chamber and the region surrounding it and time is not wasted simulating the tracks of extra photons and electrons that will most likely not end up contributing to the calculated dose which is particularly the case when the dose is being calculated to a volume which is significantly smaller than the irradiated volume as was the case here. In order to implement XCSE a special XCSE region was defined surrounding the chamber region, in which photon splitting was turned on. The optimum thickness of the XCSE shell surrounding the chamber has been found to be 1 cm (Wulff et al., 2008).

The egs_chamber simulations were carried out with esave set to 0.1 MeV, range rejection set to 256, N$_{cse}$ set to 128 for the chamber simulations and to 64 for the water sphere simulations, as these were found to be the optimum values for
efficiency as shown in figure 6.3. Range rejection based russian roulette was set to 256 and the EGS physics parameters were kept as the default parameters.

Figure 6.3 depicts the results of the tests to determine the optimum enhancement factor, \( N_{\text{cse}} \), for the cross section enhancement regions. The tests were run on the (a) water sphere and (b) chamber model, in a water phantom, and the range rejection based russian roulette factor was set to 256 for the simulations for all enhancement factors except for the simulation with an enhancement factor of 512 where it was set to 512, as the rejection must be larger or equal to the largest XCSE factor used. The efficiency was calculated using the equation:

\[
\epsilon = \frac{1}{T \cdot \sigma^2}
\]  

(6.1)

where the efficiency (\( \epsilon \)) is calculated using the CPU time required for the simulation in seconds (\( T \)) and the estimated percentage uncertainty in the resulting dose (\( \sigma \)).

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<td>31.6000</td>
<td>0.0190</td>
<td>2.7357</td>
<td>65.0000</td>
<td>5.6236e-3</td>
</tr>
<tr>
<td>128</td>
<td>1.3971</td>
<td>37.8000</td>
<td>0.0189</td>
<td>1.7503</td>
<td>99.0000</td>
<td>5.7709e-3</td>
</tr>
<tr>
<td>256</td>
<td>1.3549</td>
<td>48.1000</td>
<td>0.0153</td>
<td>1.1753</td>
<td>167.9000</td>
<td>5.0677e-3</td>
</tr>
<tr>
<td>512</td>
<td>1.2882</td>
<td>69.2000</td>
<td>0.0112</td>
<td>0.9274</td>
<td>309.0000</td>
<td>3.4897e-3</td>
</tr>
</tbody>
</table>

**Table 6.3:** The results from the test runs to examine the efficiency of the simulations for the different values for \( N_{\text{cse}} \).

The dose scoring region used for the egs_chamber simulations was the active air cavity volume of the Exradin A12 Farmer type ion chamber. The chamber was created in full detail using the egs++ geometry definitions to produce the model...
Figure 6.3: The efficiency of the egs_chamber simulations with varying $N_{cse}$ values for the photon cross section enhancement for the (a) water sphere and (b) chamber simulations.

as shown in figure 6.4. The dimensions and components were modelled based on the chamber specifications obtained from the manufacturers blueprint.
6. IMRT plan verification

6.6.2 The Exradin A12 Farmer type ion chamber model

A calibrated air-filled ionisation chamber placed in an absorbing medium is the most important method for determining the absorbed dose in absolute terms (Henkner et al., 2009). The cylindrical farmer-type ion chamber is the most popular chamber for absolute dosimetry of megavoltage photon beams in radiotherapy (Followill et al., 2003). The dose scoring region used for these simulations was the active air cavity volume of the Exradin A12 Farmer type ion chamber. The chamber was modeled in full detail using the egs++ geometry definitions. The accuracy of the chamber was verified through the calculation of the $k_Q$ value for the chamber which has been shown through measurements (Almond et al., 1999) to be 0.995 for this chamber for 6 MV photon beams. The $k_Q$ value was calculated using the equation:

$$k_Q = \frac{(D^{6\, MV}_{\text{water}}/D^{6\, MV}_{\text{chamber}})_{10\times10}}{(D^{Co}_{\text{water}}/D^{Co}_{\text{chamber}})_{10\times10}}$$

(6.2)

where, $D^{6\, MV}_{\text{water}}$ is the dose from a $10 \times 10\, \text{cm}^2$ 6 MV photon field to a water sphere, of equal volume to the active air cavity volume of the ion chamber, embedded at 10 cm depth in a water phantom of $30 \times 30 \times 20\, \text{cm}^3$ at 100 cm SSD as per the AAPM’s TG51 protocol (Almond et al., 1999). $D^{6\, MV}_{\text{chamber}}$ is the dose from a $10 \times 10\, \text{cm}^2$ 6 MV photon field to the air cavity volume of the ion chamber model.
embedded in a 30 x 30 x 20 cm³ water phantom, again at 100 cm SSD. $D_{\text{water}}^{Co}$ is the dose from a 10 x 10 cm² Cobalt-60 field to a water sphere, of equal volume to the active air cavity volume of the ion chamber, embedded at 10 cm depth in a water phantom of 30 x 30 x 20 cm³ at 100 cm SSD. $D_{\text{chamber}}^{Co}$ is the dose from a 10 x 10 cm² Cobalt-60 field to the air cavity volume of the ion chamber model embedded in a 30 x 30 x 20 cm³ water phantom at 100 cm SSD. These simulations were carried out with esave set to 0.1, $N_{\text{eic}}$ set to 128 for the chamber simulations and 64 for the water simulations, the EGS physics parameters were kept as the default parameters.

6.6.3 The egs\_chamber patient dose calculations

Once $k_Q$ had been shown to match the expected value, the chamber model was considered to be accurate and the patient dose calculations could be implemented.

6.7 $k_{Q_{\text{clin}},Q}^{f_{\text{clin}},f_{\text{ref}}}$ value

A $k_{Q_{\text{clin}},Q}^{f_{\text{clin}},f_{\text{ref}}}$ value was calculated for each plan to correct the measured values for variations in the stopping power ratio and perturbation correction factors for the small, irregular IMRT fields when compared with the reference 10 x 10 cm² field. The $k_{Q_{\text{clin}},Q}^{f_{\text{clin}},f_{\text{ref}}}$ value was calculated for each plan using the equation:

$$k_{Q_{\text{clin}},Q}^{f_{\text{clin}},f_{\text{ref}}} = \frac{(D_{\text{w},Q_{\text{clin}},f_{\text{clin}}}^{\text{clin}} / D_{\text{w},Q_{\text{ref}},f_{\text{ref}}}^{\text{clin}})}{(D_{\text{w},Q_{\text{ref}},f_{\text{ref}}}^{\text{ref}} / D_{\text{w},Q_{\text{clin}},f_{\text{ref}}}^{\text{ref}})} \quad (6.3)$$

where, $D_{\text{w},Q_{\text{clin}},f_{\text{clin}}}^{\text{clin}}$ is the dose from the clinical IMRT photon field from each plan to a water sphere, of equal volume to the active air cavity volume of the ion chamber, embedded at 8.5 cm depth in a water phantom of 30 x 30 x 17 cm³ at 100 cm SAD. $D_{\text{w},Q_{\text{ref}},f_{\text{ref}}}^{\text{ref}}$ is the dose from the clinical IMRT photon field from each plan to
the air cavity volume of the ion chamber model embedded in the water phantom, again at 100 cm SAD. $D_{w, Q_{\text{ref}}}^{f_{\text{ref}}}$ is the dose from the 10 x 10 cm$^2$ reference field to the water sphere. $D_{\text{air, } Q_{\text{ref}}}^{f_{\text{ref}}}$ is the dose from the 10 x 10 cm$^2$ reference field to the air cavity volume of the ion chamber model embedded in the water phantom at 100 cm SAD. All of these values are obtained from Monte Carlo simulations of the IMRT and 10 x 10 cm$^2$ fields using the egs_chamber code.

The $k_{Q_{\text{clin}}, Q_{\text{ref}}}^{f_{\text{clin}}, f_{\text{ref}}}$ value can then be used to convert measurements in an IMRT clinical field using a chamber calibrated in a 10 x 10 cm$^2$ field to dose to water (Alfonso et al., 2008) as follows:

$$D_{w, Q_{\text{clin}}}^{f_{\text{clin}}} = M_{Q_{\text{clin}}}^{f_{\text{clin}}} \cdot N_{D, w, Q_0} \cdot k_{Q, Q_0} \cdot k_{Q_{\text{clin}}, Q_{\text{ref}}}^{f_{\text{clin}}, f_{\text{ref}}} \quad (6.4)$$

where, $D_{w, Q_{\text{clin}}}^{f_{\text{clin}}}$ is the absorbed dose to water at a reference point in a phantom for a clinical field $f_{\text{clin}}$ of quality $Q_{\text{clin}}$ and in the absence of the chamber. $M_{Q_{\text{clin}}}^{f_{\text{clin}}}$ is the reading of the dosimeter in the field $f_{\text{clin}}$ corrected for influence quantities, such as pressure, temperature, incomplete charge collection, and polarity effects. $N_{D, w, Q_0}$ is the calibration coefficient in terms of absorbed dose to water for an ionization chamber at a reference beam quality $Q_0$ (usually $^{60}\text{Co}$). $N_{D, w, Q_0}$ is measured at the standards laboratory for a reference field of size 10 x 10 cm$^2$. $k_{Q, Q_0}$ is the beam quality correction factor, which corrects for the differences between the reference beam quality $Q_0$ at the standards laboratory and the beam quality $Q$ of the conventional field $f_{\text{ref}}$. $k_{Q_{\text{clin}}, Q_{\text{ref}}}^{f_{\text{clin}}, f_{\text{ref}}}$ is the correction factor calculated in this work to account for the difference between the responses of the ionisation chamber in the reference 10 x 10 cm$^2$ field $f_{\text{ref}}$ and the clinical IMRT field $f_{\text{clin}}$. 
6.8 Plan complexity metrics

A number of different complexity metrics were utilized to evaluate and quantify the complexity of the IMRT plans used in this study and hence examine the effect of plan complexity on dose measurement and calculation. As more complex plans deviate more from the ideal 10 x 10 cm\(^2\) field for which \(k_Q\) was originally calculated so it was believed that these more complex plans would have \(k_{Q_{\text{clin}},Q_{\text{ref}}}\) values that deviate more from the ideal value of 1.

6.8.1 Dose homogeneity index (HI)

The Homogeneity Index (HI) is a measure of the dose homogeneity within the chamber volume and is calculated as defined by Wu et al. (2003) as:

\[
\text{Homogeneity index (HI)} = \frac{D_{2\%} - D_{98\%}}{D_{\text{average}}} \tag{6.5}
\]

where, \(D_{2\%}\) is the dose to 2\% of the chamber volume as displayed on the cumulative DVH, \(D_{98\%}\) is the dose to 98\% of the chamber volume as displayed on the cumulative DVH, and \(D_{\text{average}}\) is the average dose to the chamber volume. \(D_{2\%}\) and \(D_{98\%}\) are the near maximum and near minimum doses, respectively, of the chamber volume. These concepts are defined by the ICRU (2010). The greater the HI value the more heterogeneous the dose within the chamber volume and so the more complex the plan. The values for \(D_{2\%}\) and \(D_{98\%}\) are obtained from the DVH curves for the dose to the chamber volume obtained from the DOSXYZnrc simulations. The dose to the chamber from the egs\_chamber simulations cannot be used as this only provides a single dose value and the volume is not voxelised.
6.8.2 Dose conformity index (COIN)

There are many differing definitions of the conformity index of a dose distribution as discussed by Feuvret et al. (2006). The one that has been used in this work is a slightly modified version of one of these and is defined as:

\[
\text{Conformity index (COIN)} = \frac{PTV_{\text{small, } RI}}{PTV_{\text{small}}} \times \frac{PTV_{\text{small, } RI}}{PTV_{\text{COIN, } RI}} \tag{6.6}
\]

where, \( PTV_{\text{small}} \) is the volume of the chamber contour at the center of the phantom, \( RI \) is the reference isodose which has been defined here as the average dose to the \( PTV_{\text{small}} \) volume, \( PTV_{\text{small, } RI} \) is the volume of the small target which is covered by the reference isodose, and \( PTV_{\text{COIN, } RI} \) is the volume of the added structure, of 1.5 cm border surrounding the chamber volume, which is covered by the reference isodose. While ordinarily a larger value for the conformity index would be considered better (an ideal value of 1 would indicate precise coverage of the target volume with absolutely no dose to the surrounding tissues). However, here a larger value is considered worse as it means that there is greater dependence on accurate chamber positioning (i.e. there would be a large dose gradient surrounding the chamber), and a low value would imply that a larger portion of the region surrounding the chamber is also covered by the reference isodose and so chamber positioning is less crucial.

6.8.3 Plan modulation complexity score (MCS)

The modulation complexity score (MCS) (McNiven et al., 2010), for a plan with \( J \) number of beams, is calculated by:

\[
MCS_{\text{plan}} = \sum_{j=1}^{J} MCS_{\text{beam } j} \times \frac{MU_{\text{beam } j}}{MU_{\text{plan}}} \tag{6.7}
\]
where,

$$MCS_{\text{beam}} = \sum_{i=1}^{I} AAV_{\text{segment } i} \times LSV_{\text{segment } i} \times \frac{MU_{\text{segment } i}}{MU_{\text{beam}}}$$  \hspace{1cm} (6.8)$$

for I segments per beam. The weight of each beam is taken into account by weighting each of the scores based on the number of monitor units delivered by each beam.

The aperture area variability (AAV), with A number of leaves in the leaf bank, can be calculated for each segment of the beam as follows:

$$AAV_{\text{segment}} = \frac{\sum_{a=1}^{A} ⟨\text{pos}_a⟩_{\text{left bank plan}} - ⟨\text{pos}_a⟩_{\text{right bank plan}}}{\sum_{a=1}^{A} ⟨\text{max}(\text{pos}_a)⟩_{\text{left bank beam}} - ⟨\text{max}(\text{pos}_a)⟩_{\text{right bank beam}}}$$  \hspace{1cm} (6.9)$$

The leaf sequence variability (LSV), with N number of open leaves in the leaf bank, can be calculated for each segment of the beam as follows:

$$LSV_{\text{segment}} = \left(\frac{\sum_{n=1}^{N} (\text{pos}_{\text{max}} - (\text{pos}_n - \text{pos}_{n+1}))}{N \times \text{pos}_{\text{max}}}\right)_{\text{left bank}} \times \left(\frac{\sum_{n=1}^{N} (\text{pos}_{\text{max}} - (\text{pos}_n - \text{pos}_{n+1}))}{N \times \text{pos}_{\text{max}}}\right)_{\text{right bank}}$$  \hspace{1cm} (6.10)$$

where,

$$\text{pos}_{\text{max}} = ⟨\text{max}(\text{pos}_{N \in n}) - \text{min}(\text{pos}_{N \in n})⟩_{\text{leaf bank segment}}$$  \hspace{1cm} (6.11)$$

This is calculated using the coordinates of the leaf positions (pos).
6.9 Results & Discussion

The results from DOSXYZnrc, egs_chamber and the plan complexity metrics analysis are presented and discussed here.

6.9.1 DOSXYZnrc simulations

The average dose values to the chamber from the DOSXYZnrc simulations were compared to the measured values and the percentage differences between the two sets of values were compared to other values from the simulations. There is good agreement between Monte Carlo calculated doses and measured doses once the fraction of beams passing through the chamber is >0.6 i.e. when more than 60% of the beams in a given plan pass directly through the ionization chamber as opposed to not intersecting the chamber and just passing through a part of the phantom. When this occurs, the DOSXYZnrc calculated doses for the collapsed plans agree with the corrected measured values to within 1.5% and the DOSXYZnrc calculated doses for the rotated plans agree with corrected measurements to within 2.5% (figure 6.5).

There are several possible reasons for the greater differences when the number of beams tangential to the chamber volume is >40%. Calculations in these beams are very sensitive to the accuracy of the beam model. Secondly, these are dose-to-water calculations and we are not modelling the chamber in full detail; this may influence the accuracy of the results, specifically in these beams. The egs_chamber code was used to model the chamber in full detail to investigate these effects further.

When the uncertainty in the Monte Carlo dose is approximately <0.3% then the collapsed DOSXYZnrc doses agree with the measured doses within 1% and the rotated DOSXYZnrc doses agree within 0.5% (figure 6.6). The better agreement
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Figure 6.5: How the fraction of beams passing through the chamber affects the difference between the Monte Carlo calculated dose and the measured dose.

for the rotated doses is expected to be due to the fact that the uniformity of the dose in the chamber for the rotated plans is generally better. The uncertainty value of <0.3% only occurs when 100% of the beams in the plan pass through the chamber.

From figures 6.7, 6.8, and in particular figure 6.9, it can be seen that the Monte Carlo calculated doses provide better agreement to the measured dose values than the TPS in 66% (10 out of 15) of all cases or 83% (5 out of 6) of rotated cases when the uncertainty in the Monte Carlo value is less than 0.4%.

6.9.2 egs_chamber simulations

The measured values corrected with the $k_{Q_{clin},Q_{ref}}$ factors lead to improved agreement (compared to the agreement between egs_chamber and uncorrected measurements) between the measured values and egs_chamber dose to water values in the majority of cases - 14 out of 20 for collapsed delivery and 11 out of 20 for rotated beam delivery (figure 6.10). One possible reason for the reduced effect in
Figure 6.6: Effect of the uncertainty in the dose in the chamber, as calculated with Monte Carlo, on the difference between the Monte Carlo calculated dose and the measured dose.

Figure 6.7: Difference between the Monte Carlo calculated dose and the measured dose on a patient by patient basis for the rotated plans.

The rotated beam delivery is the fact that there was already good agreement in these cases and the fact that the $k_{Q_{clin},Q_{ref}}$ values are closer to 1 in these cases so have less of an effect.
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**Figure 6.8:** Difference between the Monte Carlo calculated dose and the measured dose on a patient by patient basis for the collapsed plans.

**Figure 6.9:** Effects of the DOSXYZ Monte Carlo dose calculation uncertainty on the agreement of the dose calculation methods.

Another possible reason for the reduced effect in the rotated plans could be due to inaccuracies arising from not modelling the carbon fibre treatment couch in the Monte Carlo simulations so there could be some slight inaccuracies in the dose values for beams passing through this. Studies suggest that the couch primarily effects surface dose values, so the effect of not modelling the couch explicitly
6. IMRT plan verification

should have negligible effect in this work where the doses being investigated are at the centre of the phantom (Myint et al., 2006 & McCormack et al., 2005).

With values for $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ deviating from the ideal value of 1 by as much as 3.6% for collapsed delivery but a maximum deviation of 2% for rotated delivery (figure 6.11) would imply that IMRT QA using rotated delivery is a better option as the measured dose should be a closer match to delivered dose based on the results of this cohort of patients.

6.9.3 Effects of plan complexity

From figures 6.12, 6.13, 6.14, and 6.15, the effects of the different complexity metrics investigated can be seen on the $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ values. In figure 6.12, it can be seen that the more heterogeneous the dose within the chamber the more the $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ values deviate from the ideal value of 1. If HI is less than 0.05 then the $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ values do not deviate from unity by more than 1%. This is consistent with Chung et al. (2011). A value for HI of less than 0.05 can only be achieved with rotated delivery for the plans investigated. The $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ values for the rotated plans start to deviate more from 1 at lower HI values but overall the HI values for the rotated plans are lower than those for the collapsed plans and the $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ values for the rotated plans as a whole deviate less from 1.

In figure 6.13, it can be seen that the $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ values deviate more from 1 for larger COIN values. This is again in keeping with the results obtained by Chung et al. (2011). This effect is more pronounced in the collapsed plans which is most likely due to the more heterogeneous dose distributions in this type of beam set up.

In figure 6.14, it can be seen that the $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$ values deviate more from 1 for more complex plans (i.e. a lower MCS value). This is to be expected as more complex plans are classed as ones where the beamlets are more complex in shape.
and the leaves move most during the dose delivery resulting in a greater fluence variance (i.e. the least like the square reference field). This is also much more pronounced for the collapsed beams but this is again presumably due to the higher dose heterogeneity within the chamber in this delivery format.

In figure 6.15, it can be seen that the $k_{Q_{\text{clin}},Q_{\text{ref}}}$ values deviate more from 1 when
the fraction of beams that directly intersect the chamber volume is lower. This is in keeping with what has been previously reported by Capote et al. (2004) where they noted the largest correction values when the chamber is located outside the beamlet. Again also, a larger deviation is noticed for the collapsed plans.

Each of the 245 beams from the 20 patients were looked at individually. From this beam by beam analysis, the $k_{Q_{clin},Q_{ref}}$ values show a much greater spread in the values, particularly at the higher values for the dose homogeneity index (figure 6.16). This is due to the fact that for individual beams there are more significant deviations from CPE, which is reflected in the $k_{Q_{clin},Q_{ref}}$ values deviating more from 1.

From these results it can be seen that more complex plans (i.e. plans with lower MCS values, less beams intersecting the chamber, and less homogeneous dose both within and around the chamber) are more likely to have poor agreement
between measured data and the calculated data and are more likely to need an added correction factor \( k_{Q_{clin},Q_{ref}} \) when converting the measured charge to absolute dose. If HI is small, then the IMRT QA is easier and the difference between the measurement and calculation is smaller. This would suggest that a region of low HI (preferably less than 0.05) should be chosen to place the chamber when doing QA measurements. However, as previously mentioned this is only possible for rotated plans based on the results obtained for this cohort of patients.

### 6.10 Conclusions

This work shows the potential of using Monte Carlo techniques to assess and validate dosimetric quality assurance techniques, which is of particular importance when assessing current QA techniques to decide on a method of best practice i.e. collapsed versus rotated QA or when implementing new treatment modalities i.e.
more complex IMRT plans. Based on the outcomes of this work, it is apparent that rotated beam delivery for absolute dose measurement is more consistent as the $k_{Q_{	ext{clin}}}$ values deviate less from 1 in this delivery mode. However, if a collapsed technique must be used it would be worthwhile to take into account the homogeneity of the dose in the region in which the measurement is to be performed, the complexity of the plan and the resulting dose distribution.

**Figure 6.13:** The effect of the conformity index complexity metric on the $k_{Q_{	ext{clin}},Q_{	ext{ref}}}$ values.
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**Figure 6.14:** The effect of the modulation complexity score metric on the $k_{Q_{clin},Q_{ref}}$ values.

**Figure 6.15:** The effect of the fraction of beams intersecting the chamber on the $k_{Q_{clin},Q_{ref}}$ values.
Figure 6.16: The $k_{Q_{\text{clin}},Q_{\text{ref}}}$ values for each of the 245 beams for both the collapsed and rotated beam deliveries from all 20 patients.
The MMCTP system was implemented in a Windows environment and the features of the system applicable to this work were tested. The system successfully imported plans from the commercial treatment planning system, images were displayed correctly, input files for Monte Carlo simulations were generated correctly, and dose distributions were summed and subtracted correctly. The implementation of the code in a Windows operating system will allow greater clinical uptake of the MMCTP system as Windows is more commonly used in most clinics. This will allow the code to be used by these clinics without the need to purchase MAC computers and testing their compatibility with the hospital networks and security systems.

Once this system was up and running correctly, it was used to tune the Varian linac model firstly, followed by the Siemens model. An efficient method was devised to do this, and further speedups could be achieved using variance reduction
techniques and by reducing further the number of simulations required in the tuning process.

The Varian model was tuned to an accuracy of 1% for the output factors for field sizes of 40 x 40 cm$^2$, 30 x 30 cm$^2$, 5 x 5 cm$^2$, 4 x 4 cm$^2$, and 3 x 3 cm$^2$. This produced PDD’s which agreed to within 1% after 1.5 cm depth for all fields modelled except the 30 x 30 cm$^2$ field. The 30 x 30 cm$^2$ field agreed to within 3.5% after 1.5 cm depth but it was decided that since this investigation would focus on smaller more clinically relevant fields that this was of no major concern. The dose profiles produced agreed within 1% outside of the penumbra and within 3 mm within the penumbra for fields of 10 x 10 cm$^2$, 4 x 4 cm$^2$, and 3 x 3 cm$^2$.

The Siemens model was tuned to keep the agreement between measured and calculated output factors to within 1% for fields of 40 x 40 cm$^2$, 30 x 30 cm$^2$, 20 x 20 cm$^2$, 5 x 5 cm$^2$, and 4 x 4 cm$^2$. PDD’s were then calculated for 20 x 20 cm$^2$, 10 x 10 cm$^2$, 5 x 5 cm$^2$, and 4 x 4 cm$^2$ as these were the fields for which PDD measurements were available. These PDD’s agreed to within 3% after 1.5 cm depth. The higher percentage difference in the final simulations could be due to

1. too low a number of histories or
2. too small a voxel size

as they do appear to be somewhat noisy.

The tuned Varian model was employed in the investigation of IMRT quality assurance techniques, as used in the clinic in Montreal General Hospital. The rotated and collapsed methods for absolute dose measurement were investigated. The patient plans incident on a phantom, with a farmer ion chamber in its centre, were calculated with Monte Carlo methods, first with the DOSXYZnrc code, and then with the egs++ egs_chamber code. The results for the dose to the chamber
from DOSXYXnrc were analysed using different plan complexity metrics. These metrics showed that the collapsed method does not produce as homogeneous a dose within the chamber volume as the rotated method does. It was also shown that $k_{Q_{clin},Q_{ref}}$ value deviated by more than 1% from unity for homogeneity index values greater than 0.05.

From the work carried out on the investigation of IMRT QA techniques, it can be seen that more complex plans (i.e. plans with lower MCS values, less beams intersecting the chamber, and less homogeneous dose both within and around the chamber) are more likely to have worse agreement between measured dose values and the calculated dose values. These correlations could help with trying to devise plans that are easier to perform quality assurance on in clinical settings or provide plans that are less likely to require extensive QA measurements. From the investigation of the dose homogeneity, it can be seen that if the HI value is small, then the IMRT QA is easier and the difference between the measurement and calculation is smaller. This would suggest that a region of low HI (preferably less than 0.05) should be chosen to place the chamber when doing QA measurements.

Based on the results of this research, it is shown that rotated beam delivery for absolute dose measurement is more consistent and hence more accurate, as the $k_{Q_{clin},Q_{ref}}$ values deviate less from 1 in this delivery mode. However, if a collapsed technique must be used as a quality assurance technique, it would be worthwhile to take into account the HI value of the dose in the region in which the measurement is to be performed, the complexity of the plan and the resulting dose distribution.

The potential of using Monte Carlo techniques to assess and validate dosimetric quality assurance techniques has been shown in this work. This is of particular importance when assessing current QA techniques or to validate new techniques. This work also paves the way for using the MMCTP system as a QA tool in
itself for testing the accuracy of QA plans and may in the future lead to reducing
or even eliminating the need for clinical measurements for each patient plan for
more complex treatment methods.

7.1 Future work

The MMCTP code needs to be altered to allow to generation of input files for
linac models which use MLC component modules other than DYNVMLC. This
would allow the code to be used more readily with a greater range of linac models,
as the Siemens model which was originally used here did not use this component
module. Some of the other features of the system could be tested more extensively
in a Windows environment.

From chapters 4 and 5, the tuning process could be improved, as mentioned in
section 5.8, by leaving out “phase 2” of the tuning process. This phase did not
provide any advancement of the process when used on either the Varian model
or the Siemens model and as a result could be removed. This would speed up the
process by as much as 25%. More analysis could be done on the use of variance
reduction techniques for the process, though this was not used in this work as
access to clusters was not a limiting issue.

The dose calibration value which was used could be made more accurate by
averaging the dose across a region from the calibration run instead of just using
the dose to a point for the calibration dose.

The simulations for the ‘tuned Siemens model’ could be recalculated using a
uniform voxel size to investigate the inaccuracies in the edges of the penumbra.
It is assumed that these are due to the smaller voxel sizes in the penumbra but
recalculating the dose profiles with uniform voxel sizes would answer this question
definitively.
It could also be possible to create a database of 3ddose files with low uncertainties for the different energies and beam radii for each linac model at a range of field sizes, using a fully voxelised phantom. This would allow clinicians to extract dose profiles, PDD’s, and output factors from the 3ddose files and compare the values to measurements. This would provide a very quick method of tuning the linac for future use, though it would require a large amount of simulation time initially.

The work in chapter 6, on IMRT QA, could be added to by calculating the positioning uncertainty in the dose to the chamber for each plan and seeing how this relates to the homogeneity of the dose in the chamber. This could also be compared to the differences between measured and calculated values to determine if some of the larger discrepancies are solely due to chamber position. This work could be carried out with the egs_chamber model already developed as the egs_chamber code can vary the position of the chamber throughout the simulation to calculate the dose uncertainty due to positioning inaccuracies.

It would be a good idea to look at different trends in the plans used to see if there are any other trends which could point to the source of some of the larger differences observed. One thing which could be looked at would be to investigate if the plans with larger differences have more instances of the leaves from the negative leaf bank crossing the central axis to the positive leaf bank and vice versa. Also the angles of incidence of the beams in these plans could be examined to see if more beams in the plans with larger discrepancies pass through the treatment couch as this was another possible source of error as it was not modelled in the Monte Carlo simulations or the treatment planning system calculations.


