Validation of Monte Carlo simulation of 6 MV photon beam delivered by the Vero4DRT linear accelerator using BEAMnrc and DOSXYZnrc

by

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The following individuals certify that they have read, and recommend to the Faculty of Graduate and Postdoctoral Studies for acceptance, a thesis/dissertation entitled:

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Abstract

The goal of this project is to develop a Monte Carlo (MC) simulation model for the 6MV Vero4DRT platform which is a stereotactic ablative radiation therapy (SABR) medical linear accelerator. This MC model can simulate a variety of radiotherapy delivery techniques, including three-dimensional conformal radiotherapy (3DCRT), dynamic conformal arc (DCA) therapy, as well as complex treatments such as static-field step-and-shoot intensity modulated radiotherapy (sIMRT), volumetric modulated arc therapy (VMAT) and non-coplanar trajectory-based Dynamic Wave Arc (DWA) in a single MC run.

Open-source Monte Carlo applications based on EGSnrc particle transport codes are used to simulate the medical linear accelerator head components (BEAMnrc). The VERO4DRT has a maximum radiation field size of 15×15 cm² at 100 cm from the source. All beam shaping is achieved with a low-transmission multileaf collimator (MLC). The moving MLC leaves are modelled using the SYNCVMLC component module in BEAMnrc. Radiation doses to patients and phantoms are simulated using DOSXYZnrc codes. Electron energy tuning is achieved by comparing measured vs simulated percentage depth doses (PDDs) for a variety of field sizes in a water phantom. Electron spot size tuning is achieved by comparing beam profiles at depths 1.5 and 10 cm for the same field sizes. MC simulations with electron beam energy of 5.9 MeV and spot size FWHM=1.9mm demonstrated the closest agreement with measurement. The differences between measurement and calculation are <2.0% for the descending part of PDDs. The differences in beam profiles are <2.1% in all low dose gradient regions.

Treatment plans are generated using the RayStation (RaySearch) treatment planning system (TPS). MC simulations of dose are performed on voxelized (2.5 mm³) patient CT datasets. Planning-target-volume (PTV) and organs-at-risk (OAR) dose-volume histograms (DVH) are
compared to TPS calculated doses. Radiotherapy beam deliveries for 3DCRT, DCA, sIMRT, VMAT and DWA simulated on patient CT datasets result in dose distributions having acceptable DVH agreement with TPS calculated doses.

The Vero4DRT accelerator was successfully modelled using MC simulations. This MC model can be used as an independent dose calculation and quality assurance tool for complex radiotherapy treatment plans generated by the RayStation TPS.
Lay Summary

High-precision radiation therapy devices are used to treat cancer. The Brainlab Vero4DRT, installed at BC Cancer-Vancouver, is the first of its kind in Canada. Commercial treatment planning systems calculate radiation dose in patients prior to treatment. Safety protocols for radiation therapy require an independent method to verify this calculation. This project involved building a secondary dose calculator to meet this need. The Monte Carlo (MC) simulation code was used to model the Vero4DRT in detail. MC can simulate radiation passing through components of the Vero4DRT as well as radiation dose in patients. MC simulated doses are compared to measured results from a real radiation beam. MC doses simulated in patients are compared to the clinical treatment planning system calculated doses. MC simulations matched measurement AND treatment planning systems; thus, we have a functional Vero4DRT MC model. This provides a valuable quality/safety tool for clinical medical physicists working in radiation therapy.
Preface

This thesis was completed by the author, Maryam Rostamzadeh, at BC Cancer, Vancouver Centre. The project was identified when the brand-new Vero4DRT radiotherapy machine was installed at Vancouver Centre.

Chapter 1 and 2 introduce basic concepts of radiation therapy physics and the Monte Carlo method.

In Chapter 3, the manufacturer specifications for the Vero4DRT medical accelerator was shared with us after contacting, Dr. Yoshitomo Ishihara, a medical physicist at Kyoto University, Japan. The files received from Japan provided a good start but required major work in order to make them functional with BEAMnrc. I deconstructed and re-constructed the input files in order to be able to run Monte Carlo simulations successfully. I replaced the multileaf collimator (MLC) component module called “VARMLC” with a new model called “SYNCVMLC” which enabled the simulation of complex, dynamic MLC patterns. I replaced the virtual radiation source used in the Ishihara DOSXYZnrc model with “ISOURCE 20” to enable dynamic, multi-field and non-coplanar beam deliveries. I determined the calibration factors that were needed to report the Monte Carlo dose distributions in units of Gy.

In Chapter 4, the Monte Carlo model was tested against simple irradiation geometries. I performed all simulations. The simulations were compared to ion chamber measurement data which was acquired by clinical medical physicists as part of the standard commissioning process in 2017.

In Chapter 5, the clinical plans representing different treatment deliveries were generated by medical physicists and dosimetrist in the RayStation treatment planning system. I acquired the machine parameters from these plans and performed all Monte Carlo dose simulations on patient
geometries created from CT data. I performed all plan comparisons and reported the dosimetric metrics. The results of this thesis were summarized in an abstract and accepted to the Canadian Organization of Medical Physicists Annual Scientific Meeting (Kelowna, 2019).
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3DCRT  Three-Dimensional Conformal Radiation Therapy
CCC   Collapsed Cone Convolution
CM    Component Modules
CS    Convolution/Superposition
CSDA  Continuous Slowing Down Approximation
CT    Computed tomography
DCA   Dynamic Conformal Arcs
DICOM Digital Imaging and Communications in Medicine
Dm    Dose to Medium
DTT   Dynamic Tumor Tracking
DVH   Dose Volume Histograms
Dw    Dose to Water
DWA   Dynamic Wave Arc
EGSnrc Electron Gamma Shower–National Research Council of Canada
ETAR  Equivalent Tissue Air Ratio
FWHM  Full Width at Half Maximum
GEANT4 GEometry ANd Tracking
GTV   Gross Tumor Volume
HU    Hounsfield Unit
IBRI  Institute of Biomedical Research and Innovation
ICRU  International Commission on Radiation Units
KERMA Kinetic Energy Released per unit Mass
MC  Monte Carlo
MCNP  Monte Carlo N-Particle Transport Code
mfp  Mean Free Path
MHI  Mitsubishi Heavy Industries
MLC  MultiLeaf Collimator
MRI  Magnetic Resonance Images
MU  Monitor Unit
OAR  Organs At Risk
PBC  Pencil Beam Convolution
PDD  Percentage Depth Dose
pdf  Probability Distribution Function
PENELOPE  Penetration and ENergy LOss of Positrons and Electrons
PTV  Planning Target Volume
SABR  Stereotactic ABlative Radiotherapy
SAD  Source to Axis Distance
sIMRT  Static-field Step-and-shoot Intensity Modulated Radiation Therapy
SLAC  Stanford Linear Accelerator Center
SSD  Source to Surface Distances
TERMA  Total Energy Released per unit MAss
TG  Task Group
TMR  Tissue Maximum Ratio
TPS  Treatment Planning Systems
VMAT  Volumetric Modulated Arc Therapy
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Chapter 1 : Introduction

1.1 Thesis overview

Cancer is the second cause of death in the modern world. Radiation therapy is one of the main treatments for cancer patients. It uses ionizing radiation to kill cancerous cells or control their growth by damaging their DNA. Radiation is normally delivered by a medical linear accelerator (Linac). A medical linear accelerator generates megavoltage electrons. These high energy electrons produce electron and photon beams with an energy ranging from 4 to 25 MeV. The goal of radiotherapy is to deliver as much dose to the tumour whilst sparing nearby normal tissue and organs. There are different radiation therapy delivery techniques and specialized equipment used to achieve that goal. Stereotactic ablative radiotherapy (SABR) is a highly focused radiation treatment that gives an intense dose of radiation concentrated on a tumor, while limiting the dose to the surrounding organs.

A fully functional Monte Carlo (MC) model of a novel, SABR-dedicated unit is presented. The validated model can provide independent dose calculation verification for clinical treatment plans generated by the RayStation (RaySearch) commercial treatment planning environment. Complex, dynamic, non-coplanar, modulated arc therapy trajectories can be calculated accurately within a single Monte Carlo simulation.

The Brainlab Vero4DRT (Brainlab AG, Germany) is an innovative, ring gantry design 6 MV SABR unit capable of delivering conventional three-dimensional conformal radiation therapy (3DCRT) beams, dynamic conformal arcs (DCA), static field step-and-shoot intensity modulated radiotherapy (sIMRT) beams, and volumetric modulated arc therapy (VMAT) arcs. It also has a
unique delivery mode called Dynamic Wave Arc (DWA)\textsuperscript{2} which employs VMAT deliveries combined with simultaneous motion of the gantry and ring about two axes of rotation (imagine a baseball stitch). Using advanced, integrated image guidance systems (dual orthogonal kV imagers (Exactrac, Brainlab AB, Germany)) in combination with a 2D-gimbal mounted wave guide design, the Vero4DRT is capable of real time\textsuperscript{3} respiratory motion correlated Dynamic Tumor Tracking (DTT). This functionality has been used to treat pancreas, liver and lung tumours\textsuperscript{4,5}.

It is essential to offer a secondary dose calculation method to verify dosimetry from the clinically used commercial treatment planning systems (TPS). The flexibility of the MC model will provide useful information on dose calculation algorithm accuracy for smaller volume, SABR type beam deliveries. It will also prove to be a useful and efficient platform for modelling the dosimetry from 4 dimensional (4D) dynamic tracking deliveries for tumours undergoing respiratory induced motion. Ishihara et al. modelled\textsuperscript{6} the Vero4DRT in Monte Carlo using a static multileaf collimator (MLC) component module “VARMLC”. The goal of this study was to rebuild the Ishihara model and introduce the dynamic MLC MC component module “SYNCVMLC”.\textsuperscript{7} This will enable simultaneous dynamic MLC, gantry and ring motion simulations in a single MC run.

In the following chapter, interaction of photons and charged particles with matter, medical linear accelerators and different techniques for delivering external beam radiation therapy is described. In Chapter 2, The Monte Carlo codes and their application for treatment planning is discussed. In Chapter 3, the Vero4DRT medical accelerator and the associated treatment planning system (RayStation) is introduced. The detailed Monte Carlo model built as part of this project is described. In Chapter 4, this newly built MC model is tested by comparing simulated radiation
doses to measured doses using simple, well-controlled, static geometries. This process is called commissioning. In Chapter 5, more complex radiation beam deliveries generated by the TPS are simulated to create complex dose distributions in real patient geometries. This is the ultimate goal of the project – to create a Monte Carlo-based secondary dose calculator to verify treatment plans.

1.2 Interactions of photons with matter

Attenuation of a photon beam can occur as it passes through matter by a variety of physical interactions. There are six ways in which photons may interact with matter:

- Rayleigh (coherent) scattering
- Photoelectric Effect
- Compton scattering
- Pair production
- Triplet production
- Photonuclear interaction

The probability of a given interaction occurring in a particular material depends on the incident energy of the photon. The relevant range of photon energies for general radiotherapy applications is 150 kV to 18 MV. A typical medical linear accelerator generates beams in the 4 MV to 18 MV range. The Vero4DRT accelerator, which is the focus of this thesis, has a single 6 MV photon beam.

At this energy, one of the interactions listed above, the photonuclear interaction, has a negligible probability of occurring. This interaction between a photon and a nucleus is only significant at photon energies (>10 MeV). The probability of photonuclear interactions happening in the
radiotherapy energy region generated by the Vero4DRT 6 MV beam will not be considered in future discussions. For linear accelerators with energy higher than 10 MeV the photonuclear interaction creates neutrons in linear accelerators, which must be accounted for in radiation shielding calculations and patient dose estimates.

Each of the photon interaction processes can be represented by its own linear attenuation coefficient.

- Rayleigh (coherent) scattering - $\sigma_{coh}$
- Photoelectric effect - $\tau$
- Compton scattering - $\sigma_c$
- Pair production - $\kappa_p$
- Triplet production - $\kappa_t$

The linear attenuation coefficient value is a constant that specifies the number of photons that interact per unit thickness of a given material. The linear attenuation coefficient value is determined by the energy of the photon and the atomic number and density of the absorbing material being traversed. For a specific material, the total linear attenuation coefficient is the sum of attenuation coefficients from each of these processes.

$$\mu = \sigma_{coh} + \tau + \sigma_c + \kappa_p + \kappa_t$$  \hspace{1cm} (1.1)

The mass attenuation coefficient specifies the probability of an interaction for a given element or compound, independent of the density. It is found by dividing the linear attenuation coefficient by the density of the medium ($\rho$).
1.2.1 Rayleigh (coherent) scattering

Rayleigh scattering, also recognized as coherent scattering, is characterized schematically in Figure 1.1. This interaction occurs when a photon passing near an electron, sets the electron into a momentary oscillating state. The vibrating electron releases a scattered photon having the same wavelength and energy as the incident photon. No energy is transferred into kinetic energy of the electron and no energy is absorbed in the material; the photon is completely scattered with just a small angle change. With increasing incident photon energy, the cross section for Rayleigh scattering ($\sigma_{\text{coh}}$) decreases quickly. Rayleigh scattering will occur with incident photons at energies lower than 100 keV and in high atomic number (Z) materials. For the typical radiotherapy energy region (e.g. 6 MV), the probability of Rayleigh scattering is typically very small compared to other interactions.

![Figure 1-1 Schematic representation of Rayleigh scattering. The photon interacts with an atom, but it does not have enough energy to liberate the electron from its bound state. No energy transfer or deposition occurs. There is a small scatter angle change relative to the incident photon](image)

1.2.2 Photoelectric effect

The photoelectric effect is an interaction between a photon and a bound electron. In this process one of the atomic bound electrons will absorb the energy of the incident photon and gain enough
energy to overcome its binding energy and is thus ejected from the atom. The ejected electron, called the photoelectron, has a kinetic energy (KE) equal to:

\[ KE = h\nu - E_B, \]  

(1.2)

Where, \( h\nu \) is energy of the incident photon and \( E_B \) is the binding energy (work function) of the shell from which the electron is ejected.

In the photoelectric effect, an atomic bound electron will be ejected from the atom (photoelectron), which will create a vacancy or “hole” in one of the atomic shells (often the “K” shell). The atom will be left in an excited state. The shell vacancy can be filled by an electron from the outer shells (usually from the “L” or “M” shell) and the energy difference between shells can be released by creating characteristic x-rays or Auger electrons.

1.2.2.1 Characteristic x-rays

After a photoelectron is ejected, leaving a “hole” in the atomic shell, the electrons in outer shells may fall into the vacant inner shell. The transition from outer to inner shell will produce an x-ray photon with energy equal to energy difference between the higher and lower energy levels. The set of energy levels for each element are unique, so the x-rays produced after a photoelectric effect interaction have energies that are characteristic to that specific element (Figure 1.2). For materials with higher atomic numbers and incident radiation with higher frequencies (energies), the characteristic x-ray produced has a higher energy. The characteristic x-ray may travel away from the point of interaction and deposit dose in medium at a farther distance compared to the photoelectron. In such circumstances, the total local energy absorption in medium is reduced as the characteristic x-ray will be absorbed away from the initial site of the interaction.
Schematic representation of the photoelectric effect with characteristic radiation. The vacancy left by the photoelectron (having binding energy, $E_B$), is filled by another bound electron dropping down from a higher-level shell (L-shell to K-shell, in this example). The energy difference between two different shell states is released from the atom in form of a characteristic x-ray.

### 1.2.2.2 Auger electrons

The Auger effect is a phenomenon in which a monoenergetic electron might be emitted after a “hole” is created in an electron shell due to a photoelectric interaction (Figure 1.3). The atom is left in an excited state. Again, an electron from a higher energy level makes the transition to a lower energy level. The energy differences between these two levels will be given to one of the electrons in an outer shell. This Auger electron is then ejected from the atom with energy equal to the energy lost by the electron which made the downward transition to the vacant shell, minus the binding energy of the Auger electron. This process competes with the generation of characteristic radiation photons.
1.2.2.3 Photoelectric effect: probability of interaction

The probability of a photoelectric interaction occurring depends on the energy of the photon (E); it is inversely proportional to the cube of the photon energy (1/E^3). The following relationship between the photoelectric mass attenuation coefficient, τ/ρ, and photon energy is described as:

\[ \frac{\tau}{\rho} \propto \frac{1}{E^3} \]  

(1.3)

The probability of a photoelectric interaction will rise with increasing the atomic number, Z, of the absorber. The general relationship is that the probability of photoelectric interactions (mass attenuation coefficient value) is proportional to Z^3. The following approximate relationship between photoelectric mass attenuation coefficient and atomic number of the absorbing material holds:

\[ \frac{\tau}{\rho} \propto Z^3 \]  

(1.4)

By combining previous equations, we have:

\[ \frac{\tau}{\rho} \propto \frac{Z^3}{E^3} \]  

(1.5)
### 1.2.3 Compton scattering

In Compton scattering, an incident photon will transfer a portion of its energy to a free electron (if the energy of incident photon is higher than the binding energy of electron, the electron will be considered as a free electron). The photon is scattered at a reduced energy and at a different angle relative to the original photon as shown in the figure 1.4.

In this interaction, the electron gains some of the incident photon energy and is emitted at an angle $\theta$, relative to the incident photon direction. The photon will be scattered at an angle $\varphi$ and will have a higher wavelength (lower energy, $h\nu'$).

![Schematic representation of Compton scattering](image)

Figure 1-4 Schematic representation of Compton scattering. The incident photon interacts with a very loosely bound (i.e. “free”) electron. The electron will gain kinetic energy from the incident photon and the scattered photon will have a new energy ($h\nu'$).

The change in a photon's wavelength before ($\lambda_i$) and after ($\lambda_f$) the Compton interaction is, $\Delta\lambda$, and is called the “Compton shift”.

$$\Delta\lambda = \lambda_f - \lambda_i = \frac{h}{m_0 c} (1 - \cos\varphi)$$  \hspace{1cm} (1.6)

Where $\varphi$, is the scattering angle of the photon.
Compton scattering is an interaction between a photon and a free electron. By applying the conservation of the total energy and momentum of photons and electrons before and after the interaction, one can derive the following relations:

\[ E = \hbar \nu \frac{\alpha(1 - \cos \varphi)}{1 + \alpha(1 - \cos \varphi)} \quad (1.7) \]

\[ \hbar \nu' = \hbar \nu \frac{1}{1 + \alpha(1 - \cos \varphi)} \quad (1.8) \]

\[ \cos \theta = (1 + \alpha) \frac{\tan \varphi}{2} \quad (1.9) \]

where \( \hbar \nu, \hbar \nu', \) and \( E \) are the energies of the incident photon, scattered photon, and the emitted electron, respectively, and \( \alpha = \hbar \nu / m_0 c^2 \) (where \( m_0 c^2 \) is the rest energy of the electron (0.511 MeV)).

### 1.2.3.1 Compton scattering: probability of interaction

The probability for Compton scattering depends on the energy of photon. For a given material, with increasing incident photon energy the probability of a Compton interaction occurring decreases.

The probability of Compton scattering is almost independent of the atomic number of material but is proportional to the electron density. Therefore, hydrogen, with the highest electron density, has the highest probability of undergoing Compton scattering. In soft tissues and water (\( Z_{\text{eff}} = 7.42 \)), the Compton interaction is the dominant interaction for radiotherapy photon energies.
1.2.4 Pair production

Pair production is an interaction between a photon and the atomic nucleus. In this interaction, the photon’s energy is completely transformed into matter, as described by Einstein's equation $E = mc^2$. The interaction produces a pair of particles, a negatively charged electron and a positively charged positron (Figure 1.5). The electron and positron have the same rest mass equal to 0.511 MeV. As a result, the minimum energy (threshold) required to create the electron-positron pair is 1.022 MeV. If the photon energy is higher than 1.022 MeV, the excess energy is shared as kinetic energy between the pair particles. The kinetic energy transfer to the electron and positron in pair production interactions is given by $(h\nu - 1.022)$ MeV, where $h\nu$ is the energy of the original photon. The most probable kinetic energy distribution is half to each particle; however, any distribution of energy is possible.

![Figure 1-5 Schematic representation of pair production. Minimum photon energy ($h\nu$) for this interaction is $2m_0c^2(1.022$ MeV)](image)

1.2.4.1 Pair production: probability of interaction

The probability for pair production interaction depends on the energy of the incident photon and the atomic number of the interacting material.
Pair production occurs in the force field of an atomic nucleus, so the probability of interaction is higher with larger nuclear mass. The nuclear mass is roughly $Z^2$ consequently; the attenuation coefficient, $\kappa_p$, is roughly proportional to $Z^2$.

For a given material, the likelihood of this interaction occurring starts at a threshold energy of 1.022 MeV and will increase with increasing photon energy.

### 1.2.5 Triplet production

The last kind of interaction that a photon might have with matter is triplet production. Triplet production occurs in the vicinity of an atomic electron rather than a nucleus, as described for pair production. In triplet production, an electron positron pair is also created from the photon. In addition, the atomic electron that interacts with the incident photon is ejected from the atom. By applying the conservation of energy and momentum, one can derive that the threshold energy for triplet production is $4m_e c^2$ (2.044 MeV). 1.022 MeV is used to create the electron-positron pair, and 1.022 MeV is to conserve momentum and energy (Figure 1.6). Triplet production happens much less often compared to pair production. Generally, we are not going to worry so much about triplet production when we look at these interactions. We are going to focus mainly on pair production at the higher energies.
1.2.6 Relative importance of various types of interactions

The total mass attenuation coefficient ($\mu/\rho$) is the sum of the five individual mass attenuation coefficients:

$$\mu/\rho = \sigma_{coh}/\rho + \tau/\rho + \sigma_c/\rho + \kappa_p/\rho + \kappa_t/\rho$$  \hspace{1cm} (1.10)

As noted, coherent scattering is significant only for photon energies lower than 10 keV and materials with high atomic number. At the radiation therapy energy range relevant to this project (6 MV), it is often omitted from the sum.

Figure 1.7 is the plot of total coefficient ($\mu/\rho$) corresponding values of photon energy for two different materials; water ($Z_{eff} = 7.42$), representing a low atomic number material, and lead ($Z = 82$), representing a high $Z$ material. The mass attenuation coefficient is large for low energies and high atomic number materials because the probability of photoelectric effect is dominant in this range.
With increasing photon energy, up to 1 MeV, the mass attenuation coefficient declines quickly. The Compton scattering becomes the major type of interaction between 0.1 MeV to 10 MeV. In the Compton range of energies, the mass attenuation coefficient is not different for lead (high Z) and water (low Z), because Compton scattering is independent of atomic number. For energies, higher than 10 MeV, pair production becomes the predominant interaction, and as such the mass atomic coefficient, increases again with increasing energy. The domination of pair production interaction occurs at energies much higher than the threshold energy for the interaction (1.022 MeV). The relative importance of the three main photon interactions with matter (Photoelectric effect, Compton scattering and pair production interactions) is shown in Figure 1.8.
1.3 Interaction of charged particles in matter

In the last section we explained how charged particles (electrons and positrons) are created through different interactions of photons with matter such as the photoelectric effect, Compton scattering, pair production or triplet production events. These secondary charged particles will come to rest by depleting all their kinetic energy while passing through and interacting with matter. Electron interactions are categorised by two significant mechanisms; energy loss and scattering. Energy loss can happen by collisional energy loss or by radiative energy loss. In collisional energy loss, electrons will interact with atomic electrons. In radiative energy loss, electrons will interact with the nucleus’ electromagnetic field. During the interaction, the electron will transfer its energy to the material. This transferred energy is described numerically by the stopping power (S) which is defined as the energy loss per unit path length, \( S = \frac{dE}{dx} \).
The total mass stopping power \( \left( \frac{\Sigma}{\rho} \right)_{tot} \) is the sum of the mass-collisional stopping power \( \left( \frac{\Sigma}{\rho} \right)_{col} \) and mass-radiative stopping power \( \left( \frac{\Sigma}{\rho} \right)_{rad} \):

\[
\left( \frac{\Sigma}{\rho} \right)_{tot} = \left( \frac{\Sigma}{\rho} \right)_{col} + \left( \frac{\Sigma}{\rho} \right)_{rad}
\]  

(1.11)

By using the mass stopping power which has dimensions of MeV cm\(^2\)/g, the dependence on physical density has been factored out.

1.3.1 Collisional energy loss

Electrons may deposit their kinetic energy to the absorber through collisional energy losses. Collisional energy loss (sometimes called ionization energy loss) is an interaction between an incident charged particle and an atomic electron. The charged particle can collide with an atomic electron and hit it out of its shell. If the ejected electron is removed completely the atom becomes ionized. If the atomic electron is raised to an upper energy level the atom is in an excited state.

The charged particle will have several collisional energy loss interactions as it passes through the matter. The amount of energy that is transferred through this kind of interaction is very small. Sometimes the charged particle will transfer a huge amount of its energy to the ejected atomic electron. The high energy ejected electrons are called delta rays (\( \delta \)-rays). Møller scattering is an interaction event through which delta rays have been produced by electron collisions and Bhabha scattering is an interaction event through which delta rays have been produced by positron collisions. The mass collision stopping power for collisional energy loss was calculated by Berger and Seltzer\(^8\) and is defined as:

\[
\frac{1}{\rho} \left( \frac{dE}{dx} \right)_{col} = \frac{2 \pi r^2 m_e c^2 N_A}{\beta^2} \left( \frac{Z}{A} \right) \ln \left( \frac{r^2 (\tau + 2)}{2 \left( \frac{m_e c^2}{Z} \right)^2} \right) + F(\tau) - \delta
\]  

(1.12)
\[ F(\tau) = 1 - \beta^2 + \frac{r^2}{\beta^2} \left( \frac{2\tau + 1}{\tau + 1} \right) \ln 2 \]  

(1.13)

Where,

- \( m_e c^2 \) is the rest mass of the electron
- \( \beta = \frac{v}{c} \) is the velocity relative to the speed of light
- \( \tau \) is the ratio of electron kinetic energy to rest mass
- \( I \) is the mean excitation energy
- \( \delta \) is the density effect correction
- \( F(\tau) \) accounts for Møller or Bhabha scattering (depends on whether an electron or positron is involved in the interaction)

1.3.2 Radiative energy loss (Bremsstrahlung)

In the last section we have described the interaction of charged particles with nearby atomic electrons. The charged particles may also interact with an atomic nucleus’ electromagnetic field. The electron passing near the electric field produced by a nucleus may undergo a deceleration and will radiate energy in the form of a photon. This process is called *bremsstrahlung* which means “decelerating radiation” in German. The bremsstrahlung photon transmits through the matter and will interact through one of the photon interaction processes described in Section 1.2. Bremsstrahlung photons may deposit their energy in a location far from where it was been created. Another process which can create bremsstrahlung photons is electron-electron interactions in which the Coulomb field of one electron decelerates the other. The probability of this interaction is minor compared to the electron-nucleus interaction.
1.3.3 Elastic electron scattering

Elastic electron scattering is the process in which the electron interacts with the electric field of a distant nucleus. The interaction is elastic (no energy transfer) and results in the deflection of the electron. The electron can go through a large number of elastic scattering interactions.

1.4 KERMA and absorbed dose

The photon beam will transfer its energy to matter in two steps: In first step, the photon will have an interaction with the medium (through photoelectric effect, Compton scattering, pair or triplet production) resulting in one or more secondary electrons. In the next step these high energy secondary charged particles will have excitation and ionization interactions with matter and transfer their kinetic energy to the medium. The initial kinetic energy transfer from the photon to secondary electrons is called KERMA (Kinetic Energy Released per unit MAss), $K$, and is given by:

$$ K = \frac{dE_{tr}}{dm} $$

(1.15)

where $dE_{tr}$ is the initial kinetic energy transferred by the photons to the electrons to a mass $dm$ of matter. The quantity of KERMA is used when describing indirectly ionizing particles such as photon radiation and neutron radiation. The KERMA is energy per mass and its unit is Joules per kilogram (J/kg) or Gray (Gy) where,

$$ 1 \text{ Gy} = 1 \frac{\text{J}}{\text{kg}} $$

(1.16)

The second step is defined by absorbed dose, $D$, which is given by:

$$ D = \frac{dE_{ab}}{dm} $$

(1.17)
where $dE_{ab}$ is the energy deposited by ionizing radiation such as electrons to a mass $dm$ of matter.

KERMA will drop progressively with depth in medium because of the attenuation of the photon radiation. The charged particles produced by the photons are set in motion right from the surface and will start to impart their energy along their path. The quantity of these electrons which are set in motion will rise with depth so it will cause an increase in the absorbed dose with depth in medium. The number of electrons in motion will reach its maximum at a depth specific to the energy of the photon beam and the absorbed dose will be at its maximum at this depth. This depth is called depth of maximum dose ($d_{max}$). At $d_{max}$ a state of electronic equilibrium is reached, which means the number of electrons exiting a small volume is same as the number of electrons incoming to this volume. The section from the incident surface of the absorbing material to the maximum dose in material is called the buildup region. Due to attenuation effects, after depth $d_{max}$ the absorbed dose will decrease as well. Figure 1.9 illustrates KERMA and absorbed dose as a function of depth in medium.

![Figure 1-9 Relationship between KERMA and absorbed dose as function of depth in medium](image-url)
1.5 Medical linear accelerators & x-ray beam generation

A linear particle accelerator (shortened to “linac”) is a device most commonly used for external radiation therapy. It is designed to accelerate charged particles, usually electrons, to a high velocity and direct them to a high atomic number target along a beamline causing a massive collision. For radiation therapy purposes, a medical linac will accelerate electrons to high kinetic energy ranging from 4 to 25 MeV. Figure 1.10 shows a schematic diagram of a conventional linear accelerator.

Figure 1-10 Schematic of a conventional x-ray medical linear accelerator. Electron gun = generates electrons and directs them into the accelerator wave guide. Accelerating wave guide = in which the electrons are accelerated to much higher kinetic energies in the range from 4 MeV to 25 MeV. Bending magnet = changes e- beam direction and focuses beam on target, Primary collimator = defines a maximum circular field. Flattening filter = makes the forward-peaked photon intensity more uniform, symmetric, and flat across the field of radiation. Ion chambers = monitors and controls the output of the linear accelerator. Secondary collimator = to adjust the boundaries of the beam in 2 dimensions to create rectangular symmetric and asymmetric field shapes. MLC = to provide complex x-ray beam shapes
The electron beam is produced in an electron gun or electron emitter which is an electrical component that can create an electron cloud. Using electromagnetic fields to repulse/attract the electron cloud, a narrow electron beam is generated (∼3mm width). A radiofrequency (RF) waveguide directs this narrow ‘bunch’ of electrons along the linear accelerator portion of a waveguide device, increasing the electron energy and velocity as they travel. The electrons exit the waveguide with an almost mono-energetic kinetic energy.

The high energy electron beam is directed onto a high Z material such as tungsten. When the electrons collide with the target, radiative energy loss interactions occur, and high energy bremsstrahlung x-rays will be produced. Bremsstrahlung photons are generated having a spectrum of energies, with maximum energy equal to the energy of the incident electron beam. The average energy of the spectrum is around one third of the maximum energy.

The x-ray medical linear accelerator produces photons with energy ranging from 4 MV to 25 MeV. The x-ray beam intensity is distributed such that it is peaked more in forward direction (∼0° relative to incident electron beam).

Immediately after the target, there is primary collimator to shape the beam of radiation and limit the maximum field size.

A conical shaped absorber called flattening filter will be used to make the forward-peaked photon intensity more uniform, symmetric, and flat across the field of radiation. The flattening filter is often made of steel, copper, brass and/or aluminum.

After passing through the flattening filter, the flattened x-ray beam is directed to the dose monitoring chambers. The monitor chamber is typically a single chamber with various plates. The purpose of the ion chamber is to measure dose rate, integrated dose, and flatness/symmetry of the radiation field.
Additional collimation is achieved with the secondary collimator, or jaws. This collimator can be fixed (one square field size) or movable and often contains of two pairs of tungsten blocks (positioned 90° to each other). Asymmetric jaws can adjust the boundaries of the beam to form rectangular symmetric and asymmetric field sizes typically ranging from 1×1 cm² to a maximum field size up to 40×40 cm² (projected at a standard distance such as 100 cm from the x-ray source).

A multileaf collimator (MLC) is a tertiary beam shaping device made up of individual leaves of tungsten. The MLC may be comprised up anywhere from 60 to 120 independently moving leaves and are used to create simple or complex x-ray beam shapes customized for each patient’s treatment requirements.

The linear accelerators are built so that the source of radiation can rotate about a horizontal axis on gantry. The collimators can also rotate about an axis. The point of intersection of the collimator axis and the axis of rotation of the gantry is known as the isocenter. The patient lies on a couch and machine rotates around to deliver multiple beams from multiple angles, all focussed on the tumour.

1.5.1 The Vero4DRT linear accelerator

There are different linac vendors and designs. The basic physics for all systems is very similar. This project is dedicated to studying one particular linac design. It is a novel, four-dimensional (4D) image guided radiation therapy system, the Vero4DRT. Vero4DRT is a product of BrainLAB (BrainLAB AG, Feldkirchen, Germany) and MHI (Mitsubishi Heavy Industries, Tokyo, Japan). A picture of the Vero4DRT is shown in figure 1.11. The Vero4DRT has a compact C-band 6 MV waveguide mounted in-line with the x-ray treatment head. The treatment
head is mounted on an O-ring type gantry. The gantry can rotate around the patient about 2 axes of rotation. Once the patient is in position, they are stationary and, the entire linac system will move around the patient. This is different from other conventional C-arm style of linacs which relies on a moving the gantry and moving the patient while on the couch to achieve 2 axes of rotation. The Vero4DRT is capable of delivering conventional 3DCRT beams, DCA, sIMRT beams and VMAT arcs. It also has a unique delivery mode called Dynamic Wave Arc (DWA)\(^4\) which employs VMAT gantry-arcs combined with simultaneous motion of the O-ring assembly. A gantry mounted orthogonal kV x-ray imaging “ExacTrac” system allows for imaging at any treatment angle “on-the-fly”. It is capable of 2D ‘one-shot’ and fluoroscopy imaging, plus 3D cone-beam CT imaging. The Vero4DRT is also capable of real time respiratory motion correlated Dynamic Tumor Tracking (DTT)\(^{10}\). The entire linac-head/waveguide assembly is mounted on a 2D gimbal which can tilt and pan to follow moving tumours (e.g. lung, liver). The x-ray fluoroscopy function + a ceiling-mounted infrared (IR) camera system allows for building and monitoring DTT respiratory correlation models.

Figure 1-11 Vero4DRT radiotherapy platform

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The Vero4DRT only has one static secondary collimator (jaw) field size which is set to 15×15 cm² at source to axis distance (SAD) of 100 cm. All beam shaping is achieved with a low transmission (~0.15%) multileaf collimator (MLC). This MLC has 30 pairs of 5 mm width (at isocentre) tungsten leaves.

1.6 Techniques for delivering external beam radiation therapy

There are different methods to deliver external radiation therapy. Patients can receive treatments that range from quick and simple palliation of painful symptoms, to highly complex and exquisitely conformal treatments with the intent to cure a patient of their foci of cancer.

Computed tomography (CT) scans or magnetic resonance images (MRI) of the region of interest in the patient are taken. These images are transferred into the treatment planning system and are used to find the exact location and shape of tumor. A treatment plan is generated using several options for radiation beam delivery. In this project, I have been able to simulate the following treatment deliveries:

1.6.1 Three-dimensional conformal radiation therapy (3DCRT)

Three-dimensional conformal radiation therapy (3DCRT) is one of the simpler methods to deliver focused radiation to the patient. For 3DCRT, multiple static radiation beams with static MLC shapes are directed to the treatment area from different angles (Figure 1.12). High radiation doses can be delivered to the tumor while sparing nearby healthy organs. In the most basic of treatments, 1 to 4 static beams, shaped by MLC from four cardinal angles, (anterior, posterior, left lateral and right lateral) will be used to treat the patients. The limited number of beams restricts how conformal the radiation dose is around the target. For more conformal treatments,
7 to 10 3DCRT beams may be used, often in a non-coplanar geometry (Figure 1.12). All conventional linacs can deliver these beams.

Figure 1-12 Three-dimensional conformal radiotherapy (3DCRT) technique.

1.6.2 Dynamic conformal arc (DCA) therapy

Dynamic conformal arc (DCA) therapy is an external beam radiation therapy technique that conforms the beam to the target while the linac head is rotating around the patient. Each arc is coplanar and multiple arcs may be used (Figure 1.13). Most conventional linacs can deliver these beams.

Figure 1-13 Dynamic conformal arc (DCA) therapy
1.6.3 Intensity modulated radiation therapy (IMRT)

Another method to give the external radiation therapy is intensity modulated radiation therapy (IMRT) in which the intensity (strength) of radiation beam is modified by a moving multi-leaf collimator while the beam is ‘on’. In this technique called static field step-and-shoot IMRT (sIMRT), multiple static-gantry beams are directed at the target from multiple treatment angles (Figure 1.14). The goal is to create very conformal high doses to the target, while limiting the radiation dose to surrounding healthy tissue. This is an inverse planned technique in that the computer will optimize the MLC delivery to achieve the complex planning goals. For tumors that are of a difficult or complex shape and are close to sensitive healthy organs, IMRT is a very useful technique. The disadvantage of IMRT is that each complex MLC beam delivery is not very efficient to deliver, thus increasing the treatment time. Treatment times can be up to 7 times longer than the simpler, but less conformal 3DCRT techniques (e.g. 15 min delivery for IMRT instead of 2-3 min for 3DCRT). Most conventional linacs can deliver these types of beams.

Figure 1-14 Intensity-modulated radiation therapy (IMRT) technique. MLC is moving while radiation beam is “on”.

1.6.4 Volumetric-modulated arc therapy (VMAT)

Volumetric-modulated arc therapy (VMAT) is an extended form of IMRT whereby a 360-degree rotation arc (Figure 1.15) is used to deliver the radiation continuously while the MLC is moving in an IMRT-type of delivery. The delivery time for conventional treatments on conventional linacs is usually less than two minutes. VMAT can help us to achieve very conformal 3D dose distributions to target volume while sparing nearby healthy tissues and is much faster to deliver compared to static-field IMRT.

Figure 1-15 Volumetric-modulated arc therapy (VMAT). MLC is moving (modulating intensity) while the arc is rotating.

1.6.5 Dynamic wave arc (DWA)

The concept of Dynamic Wave Arc (DWA) was presented in 2013\textsuperscript{12} and has been supported commercially in 2015 by the Vero4DRT radiotherapy linear accelerator\textsuperscript{9}. Vero4DRT has this unique delivery mode called Dynamic Wave Arc which employs VMAT combined with simultaneous motion of gantry and ring around two axes of rotation. DWA is an advanced form of VMAT in which a 2\textsuperscript{nd} axis of rotation is added onto the traditional gantry rotation (Figure
1.16). The entire floor-gantry assembly can rotate during the arc which allows for a complex, synchronized, non-coplanar VMAT beam delivery trajectories.

Figure 1-16 Dynamic wave arc (DWA). Non-coplanar VMAT arcs stitched together in a seamless delivery
Chapter 2: Monte Carlo simulation of photon and particle radiation transport through medium

2.1 The Monte Carlo method

Monte Carlo simulation is a numerical method for calculating a solution to a range of analytical problems where an accurate answer is challenging or difficult due to the inputs being described by random sampling of a probability distribution\(^\text{13}\). Monte Carlo has applications in physics, finance, biology and even road traffic. In radiotherapy physics simulations, each possible particle (photon or charged particle) interaction in matter is governed by random selection, weighted by a probability distribution function for that particular interaction. The outcome of each random decision becomes the input to the next step in a particle’s journey through matter. Ultimately, the absorbed energy in matter (matter = patient undergoing radiotherapy or physics phantom) is recorded as absorbed dose. Monte Carlo simulates millions to billions of microscopic photon/electron interactions in order to describe a macroscopic property (e.g. absorbed dose). For this reason, Monte Carlo methods are computationally intense.

To simulate the transport of photons, a sampling answer for the exponential distribution is required. The typical method for sampling from the exponential distribution starts with estimating the inverse transform of the original probability distribution function (pdf). The following equation defines the basic probability distribution, \( p(x) \), for a photon reaching a distance between \( x \) and \( x + dx \) in a medium having an attenuation coefficient, \( \mu \):

\[
p(x) = e^{-\mu x} \mu
\]

(2.1)
The Monte Carlo sampling distribution is found by computing the cumulative probability distribution of equation 2.1 (the integral of equation 2.1) and putting this distribution equivalent to a random number distribution, $\xi$:

$$\int_0^x e^{-\mu x} \, dx = \xi$$  \hspace{1cm} (2.2)

This calculation can be solved and rearranged to sample $X$, the distance to collision for a photon:

$$X = -\frac{\ln \xi}{\mu}$$  \hspace{1cm} (2.3)

It has been shown as the number of the histories increases; the Monte Carlo solution meets to the precise logical solution. The Monte Carlo might need millions of histories, depend on the difficulty of the problem and the numerical accuracy needed to achieve a statistically meaningful answer\textsuperscript{14}. This process of calculating the distance to next interaction and comparing its value to the next geometrical boundary crossing represents the fundamental calculation for neutral particles such as photon. After the particle collides, the appropriate sampling is performed to determine the type of interaction and energy and angle of the particle after the interaction.

### 2.2 Photon interaction processes

The physics of photon interactions used in MC simulations were described in the previous chapter. A photon can interact within the medium through one of the following processes:

- Rayleigh scattering (generally omitted at radiotherapy energies)
- Photoelectric effect
- Compton scattering
• Pair production
• Triplet production
• Photonuclear disintegration (often omitted at radiotherapy energies)

The average distance traveled by a photon before undergoing one of the above interactions is called mean free path (mfp). The magnitude of the mean free path depends on the particle energy and the medium type. In Monte Carlo modelling, an initial particle may undergo many interactions and scattering processes, possibly producing several daughter particles along the way. The data collected while following the path of a particle throughout its journey is called a “history”. Information such as particle energy, position, direction, and weight are saved in a stack of variables.

Once the photon reaches a point of interaction, one of the interaction types above is randomly selected. The random selection is weighted by the probability of a particular interaction occurring (cross section) for a given photon energy and absorbing material. The energy, position and direction of any scattered or secondary particles will be calculated and saved in the stack of particles to be simulated. The particle’s history will be recorded until it exits the region of interest or until the energy falls below a cut-off described by the operator.

Once the history of a primary photon has completed, the system will move on to processing the history of the secondary particles. The secondary particles can also have interactions with matter and create new particles. Any new particles arising will also be added to the stack of variables to be recorded. The process of creating multiple new particles from a single primary photon is called an electron gamma shower (Figure 2.1).
2.3 Electron interaction processes

As described in previous chapter electrons or positrons can interact with the surrounding medium through the following events:

- Møller scattering
- Bhabha scattering
- Bremsstrahlung
- Positron annihilation
- Elastic scattering

A complete explanation of these interactions can be found in the literature.\textsuperscript{15,16,17}

Any electrons generated during the simulation will transfer their energy along their path. Often this process is modelled by what is called the Continuous Slowing Down Approximation (CSDA). Some of the electrons will lose energy radiatively (e.g. bremsstrahlung).
Electrons and positrons might have hundreds of interactions before coming to the rest. All of these interactions are simulated, and it may require a very large computation time. To make the simulation more efficient, often electron cut-off energy is applied by the user. For electrons below this energy, the remaining energy is assumed to be deposited locally and the path of the electron is terminated.

2.4 Monte Carlo codes

Monte Carlo radiation transport code-bases are available for radiation therapy applications. The most common Monte Carlo codes for medical radiation physics include:

- EGSnrc (Electron Gamma Shower–National Research Council of Canada)\textsuperscript{18}
- MCNP (Monte Carlo N-Particle Transport Code)\textsuperscript{19}
- PENELOPE (Penetration and ENergy LOss of Positrons and Electrons)\textsuperscript{20}
- GEANT4 (GEometry ANd Tracking)\textsuperscript{21}

These codes have differences in particle transport calculations and performance comparisons for both photon and electron radiotherapy applications. The principal differences can be located based on the interaction models used and the type of input/output data. The application of MC simulations now ranges from high energy physics to electron microscopy and medical physics. Fast improvements have allowed these codes to develop; from originally general programs used only for high-energy detector modelling, to user-friendly codes that are devoted to a specific application, such as dose calculations in patient geometry. MC simulations are used in both diagnostic imaging simulations and medical linear accelerator modeling.

A good tool for radiotherapy applications should include:

- Dose calculations for a broad range of tissues such as soft tissue, lung, bone, muscle, etc.
• An ability to model a broad range of radiation types such as photons, neutrons, electrons, positrons, heavy charged particles, etc.

• An ability to model a broad range of photon and particle energies

In this study, the EGSnrc Monte Carlo code is used for all simulations.

2.5 EGSnrc (Electron Gamma Shower – National Research Council)

The EGS (Electron gamma shower) Monte Carlo simulation code was first developed by the Stanford Linear Accelerator Center (SLAC) in the 1980’s and is named for its ability to simulate the shower (or electromagnetic cascade) of electrons and photons generated when a photon radiation beam is incident on matter.

The original EGS code has undergone much development with recent versions and dedicated gui-based applications are now offered by the National Research Council of Canada (EGSnrc). The EGSnrc code used in this project is a general-purpose, particle transport code that can be helpful to build Monte Carlo simulations for electron and photon transport, for a wide-range of energies (from 1 keV to 10 GeV). The EGSnrc application has a detailed library of atomic scattering cross-section tables. It also includes a C++ geometry library for describing the geometry of complicated simulation settings and particle sources.

Microscopic and macroscopic quantities such as absorbed dose, KERMA, energy spectra and particle fluence etc. can be calculated with EGSnrc. A more comprehensive description of the EGS code can be found in literature.22
2.6 BEAMnrc simulation code

BEAMnrc\textsuperscript{23} is an EGSnrc based Monte Carlo code for modelling radiation sources and was developed as part of the OMEGA project which was a partnership between the National Research Council Canada and the University of Wisconsin in Madison. The goal of the OMEGA project was to create a 3D treatment planning tool for radiotherapy. It takes user input in the form of a text file describing the comprehensive geometry and composition of each component of a radiation-generating device and models the transport electrons and photons through the device. BEAMnrc allows one to create a ‘virtual’ radiation beam for a ‘virtual’ medical linear accelerator or diagnostic imaging system. Predefined geometrical structures (called Component Modules or CM) are used to model specific parts of a medical linear accelerator or linac. Their dimensions and configurations can be reconstructed from manufacturer specifications.

2.6.1 Linear accelerator source modeling

BEAMnrc allows for the development of detailed 3D models of a linear accelerator treatment head. BEAMnrc simulations start at the target (it does not explicitly model the electron gun or accelerating waveguide). The electron beam incident on the target is described by adjustable parameters within the simulation interface. Then BEAMnrc model the primary collimator, flattening filter, ionization chamber, secondary collimator (jaws). Then, the virtual x-ray beam created is transported through the multileaf collimator MLC. The MLC may contain either static shapes, as used for 3DCRT, or dynamic shapes, as used for IMRT, VMAT and DWA. These MLC shapes and motions are patient-specific. The transport through the MLC leaves can be modelled in BEAMnrc (Figure 2.2).
To do this, one requires the detailed treatment head specifications which have been provided by the manufacturer. A virtual x-ray source called a phase space creates just below the static components of linac. (see Figure 2.2).

The phase space file contains information about every photon and electron crossing a virtual plane perpendicular to the beam central axis. In the phase space file, each photon or electron crossing the plane is defined by parameters: Energy (E), the 3D position (x, y, z), and the 3D direction cosines (u, v, w).

Figure 2-2 Schema of the x-ray head and MLC components of Vero4DRT

Phase Space File
2.7 DOSXYZnrc simulation code

DOSXYZnrc\textsuperscript{24} is another EGSnrc-based Monte Carlo code for computing dose distributions in a three-dimensional Cartesian volume of a rectilinear voxel phantom or patient. DOSXYZnrc was also developed as part of the OMEGA project. A variety of radiation sources (or radiation beams) may be directed onto the phantom, including full phase space files created by BEAMnrc. The phantom is used by the DOSXYZnrc program to calculate the transport of photons and particles into a patient or phantom geometry. Radiation beams (generated by BEAMnrc) can be directed onto the voxelized phantom or patient data from patient-specific gantry and ring angles and radiation doses are simulated using DOSXYZnrc codes. The features of the incident beam geometry are controlled in a module called “ISOURCE” which offers several options to the operator. There are several different source types have been established for DOSXYZnrc dose calculation code. In this study ISOURCE 20 was used which is developed by Lobo and Popescu\textsuperscript{25} to let the user to simulate dynamic motion of the virtual phase space source relative to the patient geometry or phantom for different incident directions, source to surface distances (SSDs) and isocentres locations.

The dose distributions simulated by DOSXYZnrc dose calculation code can be found in the output file, “.3ddose”. The dose output file “.3ddose” includes the data about the simulation geometry (dimensions and number of voxels) of the 3D matrix dose values in “dose per particle hitting the target” and associated statistical uncertainty with each voxel. The goal is to simulate enough histories to achieve < 2% uncertainty in the voxel dose value.
2.7.1 Anatomy modeling

The geometry of patient-specific, Monte Carlo treatment simulation is typically based upon an initial CT scan. This initial scan is composed of multiple transverse cuts with each having an axial pixel resolution of 1-2 mm$^2$ and a slice thickness of 1.25 - 2.5 mm$^2$. During the Monte Carlo phantom-building process, the 512×512 CT matrix is typically re-sampled to produce a lower resolution matrix so that the Monte Carlo phantom pixel size is representative of typical 3D treatment planning calculation dose grid resolution (2.5 mm$^2$).

For a CT scan of the human body, a Hounsfield Unit (HU) of -1000 is equivalent to air, zero equivalent to soft tissue and Hounsfield Unit value of approximately 2000 - 3000 is equivalent to cortical bone. For the Monte Carlo calculation, a thresholding algorithm is used to convert each pixel Hounsfield unit into a corresponding anatomical tissue type defined by its elemental composition and mass density.

The major tissue types typically are accounted for are: AIR, LUNG, TISSUE and four types of BONE.

The output of cctcreate is written into a file with extension. egsphant, which contains all the information necessary for DOSXYZnrc to simulate the CT phantom (number of voxels, tissue types, geometric bounds, physical density). The tissue types will have corresponding atomic interaction cross-section tables which are a part of the EGSnrc simulation package.

2.8 Variance reduction

Monte Carlo simulation results have a calculated value and an uncertainty (variance) associate with the value. Monte Carlo relies on random, stochastic processes. Uncertainty in a simulated
value can be reduced by increasing the sampled population, at the cost of increased computing time. This is why computer power and variance reduction techniques are important.

Variance reduction techniques in Monte Carlo simulations decrease the time of computation necessary to get results of appropriate accuracy. Note that although accuracy is important, simulation/calculation time is also important. To allow for more efficient calculations, a variety of variance reduction techniques have been included with the EGSnrc codes.

### 2.8.1 Electron range rejection

Range rejection is a variance reduction technique applied to electrons and is used to reduce the computing time required for a simulation. The range of charged particles (e.g. electrons, positrons) will be computed based on its energy. If the particle is not able to leave the bounds of the calculation voxel, the simulation will stop tracking its history and assume all remaining energy of that particle is deposited at that point (voxel) (Figure 2.3). In terminating a charged particle’s history and depositing all of its energy in the current region (voxel), it is also assumed that any bremsstrahlung photons that might have been produced by the charged particle will also not leave the present voxel. The energy transported by these possible future bremsstrahlung photons away from the voxel affected by range rejection is deposited locally. Since this is an approximation method, the energy threshold of the range rejection must be kept low to avoid increasing the uncertainty of the simulation of dose in the medium.
2.8.2 Photon forcing

Another variance reduction technique applied to photons is called the photon forcing technique. This technique is beneficial improving the statistics of scattered photons when photon interactions are sparse; either due to a low density of the medium or to a thin geometric region. These sparse interactions result in long simulation times to get meaningful data. Photons are forced to interact within a given component module of the linac (e.g. the primary collimator). This variance reduction technique is often used to improve the number of electrons created in the air. This is an efficient technique to examine electron contamination from a component of the linear accelerator.

2.8.3 Bremsstrahlung photon splitting

Bremsstrahlung splitting method has three different types available in the EGS system,
• Uniform Bremsstrahlung Splitting (UBS)
• Selective Bremsstrahlung Splitting (SBS)
• Directional Bremsstrahlung Splitting (DBS): the most efficient one.

With this method, an electron is forced to create not one, but several bremsstrahlung photons; these secondary photons have suitably adjusted weights (i.e. total weight divided by the photon splitting number) to maintain particle weights. Bremsstrahlung splitting can significantly reduce the uncertainty in all photon quantities for a given number of electrons incidents on the target.

2.8.4 Charged particle “Russian roulette”

Another charged particle variance reduction technique is Russian roulette. This feature is often combined with bremsstrahlung photon splitting.

This technique is used when there is little interest in a particular particle generated in an interaction (mostly electrons in the linac head). The low interest particles are removed from the simulation with a given probability and the remaining particles weight are increased by the inverse of that probability. This removal happens until one, highly weighted, low interest particle remains.

2.8.5 Energy cut-offs

The tracking of photons and electron histories is ended if their energies drop lower than a user defined energy cut-off. This technique disregards any possible future interactions of the photon and electron at this low energy so must be applied with care.
2.9 Application of Monte Carlo method for treatment planning

The application of Monte Carlo simulations to radiation therapy dose calculations has become a standard method for benchmarking other dose calculation algorithms. The commercial algorithms, which exist in several commercial treatment planning systems can vary in accuracy and the user must be aware of any approximations and the limitations in accuracy under certain irradiation conditions. It is well known that Monte Carlo simulations of radiation transport is currently the most accurate means of calculating dose in complex geometries (e.g. in heterogenous geometries). However, it has been traditionally associated with long computation times. With the improvement of computer speed and the availability of statistical variance reduction techniques, the Monte Carlo method is becoming a clinically practical method for photon and particle (e.g. electron, proton) dose calculations. The development a Monte Carlo virtual beam model for the physical linear accelerator is the first step towards calculation of dose distributions in phantom or patient. An accurate dose calculation needs an accurate description of the radiation source.

For the application of Monte Carlo calculations to external beam radiotherapy, a 2-step method is followed. The first step needs an accurate simulation of the linear accelerator treatment head to model the energy, angular, and positional distribution of photons and electrons in the virtual radiation beam. The second step directs this virtual radiation beam towards the patient or phantom and DOSXYZnrc is used to calculate absorbed dose.

Medical physicists will commission a real linear accelerator and treatment planning system prior to treating patients. This process involves measuring a standard set of simple beam properties under standard reference conditions and comparing the measurement to calculated dose distributions.
Standard measurements include percentage depth doses measured parallel to the beam axis (a measure of the beam energy and penetration power) and in/cross-beam profiles measured perpendicular to the beam axis. These measurements are typically acquired in a water or water-equivalent phantom using ion chambers or diodes. A variety of field sizes are measured (e.g. 1×1 cm² to 15×15 cm² square fields, for this study). The Monte Carlo model will simulate the same geometry as the measurements and will generate calculated percentage depth dose and dose profiles curves. The measured and Monte Carlo simulated doses are compared and metrics are used to assess how well they agree with each other.
Chapter 3: Development and evaluation of a Monte Carlo photon beam model for Vero4DRT linear accelerator

3.1 Introduction

In this study a beam model for a novel stereotactic ablative radiotherapy (SABR)-dedicated medical linear accelerator, called Vero4DRT, was developed and evaluated. Various parameters that affect the properties of the beam were tested and optimized in order to match clinically measured data. For example, the effect of incident electron beam energy and spot-size on percentage depth dose and dose profile curves is assessed. The agreement between MC and measurement results are presented and limitations discussed. This study was performed at the Medical Physics department of BC Cancer - Vancouver Centre.

3.2 Vero4DRT: features

The Brainlab Vero4DRT (Figure 3.1) is an innovative, ring-gantry design 6 MV stereotactic ablative radiotherapy (SABR) dedicated unit capable of delivering conventional treatment beams such as three-dimensional conformal radiation therapy (3DCRT), dynamic conformal arcs (DCA) therapy, static-field step-and-shoot IMRT (sIMRT), volumetric modulated arc therapy (VMAT), as well as advanced dynamic wave arc (DWA) trajectory beams which employs VMAT deliveries combined with simultaneous motion of the gantry-ring about 2-axes of rotation.
Using advanced, integrated image guidance systems (dual orthogonal kV “ExacTrac” system) in combination with a gimbal mounted wave guide design; the Vero4DRT is capable of real-time, respiratory motion correlated Dynamic Tumor Tracking (DTT)\(^5\). The Vero4DRT only has one static secondary collimator (jaw) field size which is set to \(15\times15\ \text{cm}^2\) at source to axis distance (SAD) of 100 cm.

The linac head components including target, primary collimator, flattening filter, ionization chambers and secondary collimator jaws were simulated based on manufacturer provided information. The composition of materials and alloys; their mass densities; the position, dimensions, and shape of defining surfaces of the components; and their motion must all be known in great detail to build an accurate Monte Carlo model.
3.3 Vero4DRT linear accelerator source modeling using BEAMnrc

The components of the Vero4DRT medical accelerator are shown in figure 3.2. The EGSnrc/BEAMnrc code used to create the virtual beam. This code generates 3D models of a linear accelerator treatment head.

![Geometric schematic of the x-ray head and MLC for Vero4DRT. Note the in-line waveguide and the static/fixed secondary collimator (jaws).](image)

In BEAMnrc, ISOURC =19 was used to describe the electron beam incident on the target. This is an “Elliptical Beam with Gaussian Distributions in X and Y”. The beam has parallel edges, with direction cosines specified by the user. Both the incident energy and the FWHM values for the incident electron beam width can be varied until the simulated dose to water matches experimental measurements.

Photons and electrons produced in the tungsten target (caused by colliding the electron beam with the target) are transported through each component module of the linac head. The
BEAMnrc code models a linac using a wide-ranging selection of simple geometrical figures such as slabs, cones, parallelepipeds, trapezoids. The code requires that the components do not overlap with each other as one moves along the beam direction.

The following component modules are used:

1) Target = SLABS. Composition = Tungsten.

2) Primary collimator = CONS3R. Composition = Tungsten. The CONS3R component module is a stack of truncated cones. It can be used for any case if there is cylindrical symmetry and if there are only two radial regions.

3) Flattening filter = FLATFILT. Composition = Aluminum. FLATFILT component module consists of a stacked set of truncated coaxial cones. The x-ray beam intensity profile produced by the target is highly forward peaked. The flattening filter is designed to attenuate the peak to get an almost uniform and symmetric x-ray intensity profile throughout the field size.

4) Ionization chamber = SLABS. Composition = Aluminum. This component module can be used for several planes of different thickness and material.

5) Secondary collimator (jaws) = JAWS. Composition = Tungsten. The role of the secondary jaws (two orthogonal pairs) is to define a fixed shape rectangular radiation field of 15×15 cm² at source to axis distance (SAD) of 100 cm from the target (or at isocentre).

6) Multileaf collimator (MLC) = SYNCVMLC. Composition = Tungsten. All beam shaping is achieved with a low-transmission (~0.10 - 0.21%) MLC. This MLC has 30 pairs of 5 mm width (at isocentre) tungsten leaves with a maximum field size of 15×15 cm² (Figure 3.3)²⁶. SYNCVMLC component module allows synchronization of leaf opening with any
other synchronized component in linac\textsuperscript{27} and with the motion of SOURCE20 (synchronized phase space source used in DOSXYZnrc). The simulation of multiple fields, multiple MLC shapes can be done with a single MC run using SYNCVMLC.

![Diagram of MLC with dimensions](image)

Figure 3-3 Schematic presentation of Vero4DRT MLC

### 3.4 Generating of phase space files for Vero4DRT

Each photon or electron to be followed in the simulation process needs to be described by its charge, direction of motion, energy, and location of last interaction (optional) (the phase space). The phase space can be located anywhere along the linac head component module chain during the BEAMnrc simulation. It can be considered as a virtual x-ray source which will eventually be directed towards the phantom or patient volume in which a dose distribution is to be calculated.

In this study a plane just above the multileaf collimator (MLC) and below the secondary collimator (jaw) at 32.5 cm distance from the target component was defined (see Figure 2.2). This phase space file can effectively replace the linac components above it and is essentially becomes the virtual linac. By capturing a phase space file below the static components, the user can re-use it to transport particles though the moving (patient-specific) parts of the linac. This
saves time as the static components do not need to be re-simulated every time a dose calculation is required.

3.5 Dose measurement

Dose measurements in water phantom (Blue Phantom, IBA, USA) was performed by the clinical medical physicist team for the 6 MV energy on the Vero4DRT. The CC13 0.13 cm$^3$ volume ion chamber (IBA, USA) was used for field sizes larger than $4 \times 4$ cm$^2$. The CC13 is a standard ion chamber which can be used for absolute and relative dosimetry of photon, electron, and proton beams in air, water phantoms, or solid phantoms. For smaller field sizes ($< 4 \times 4$ cm$^2$), the CC01 0.01 cm$^3$ volume (IBA, USA) ion chamber and high-resolution diamond detector (PTW, Germany) have been used.

3.6 Determining electron beam energy

The tuning of the electron beam energy incident on the target is performed by comparing calculated and measured percentage depth dose (PDD) curves for $15 \times 15$ cm$^2$ field size. For the purpose of comparison between calculation and measurements, the value of each voxel is normalized to the maximum value of the dose deposited along the central axis ($D_{d_{\text{max}}}$). In this study the incident electron beam energy ranged between 5.5 to 6.5 MeV in steps of 0.1 MeV. The dosimetry results of measurement vs calculation were compared, and best match (found by minimizing the percent difference between the descending part of the PDD) determined the optimum energy of electron beam. Local percent differences between simulation and measurements results are reported.
3.7 Effect of electron beam width on beam profile

A systematic study was performed to assess the effect of the electron beam width on the dose profiles calculated by Monte Carlo compared to measurement. The beam profile sensitivity to electron beam width was also assessed. The beam profile of a 6 MV Vero4DRT beam for a 15×15 cm² field size at $d_{max}$ (1.5 cm) and 10 cm was simulated. The dosimetry results of measurement vs calculation were compared, and best match (found by minimizing the percent difference in low dose gradient regions of the profiles) determined the optimum FWHM of electron beam.
Chapter 4: Commissioning of a Monte Carlo model for Vero4DRT

4.1 Benchmarking the Monte Carlo system

For Monte Carlo models to be beneficial in a radiotherapy environment, the complete structure must be benchmarked and validated against measured data from the actual linear accelerator modeled. The goal is to achieve precise dose deposition information inside a patient or phantom. The actual measured commissioning data used to benchmark the Monte Carlo simulation is acquired by clinical medical physicists. The linear accelerator (linac) simulated in this project is a Vero4DRT developed by Mitsubishi Heavy Industries, Ltd., Japan (MHI) in collaboration with Kyoto University and the Institute of Biomedical Research and Innovation (IBRI). The linac generates a single 6 MV x-ray photon beam energy. The exact geometry and material of each component is specified by manufacturer. First, an initial Monte Carlo simulation of the accelerator head is performed to produce the phase space file at 35.2 cm distance from the target (just above the multileaf collimator (MLC)). The phase space file is generated using $1.5 \times 10^9$ initial electrons incident on the target. This phase space file contains ~50,000,000 particles which are used for subsequent simulations through the MLC. The MLC is programmed to define various field sizes ranging from $1 \times 1$ cm$^2$ to $15 \times 15$ cm$^2$. A water phantom with dimension of $30 \times 30 \times 30$ cm$^3$ is generated. Percentage depth doses and beam profiles in this water phantom are simulated using $9 \times 10^8$ particles. The photon cut off energy (PCUT) is set to 0.01 MeV, and the electron cut off energy (ECUT) is set to 0.521 MeV for all simulations. The resulting dose distribution is compared to measured dose data.
4.2 Open field verification

In this study some initial tuning of the virtual linac was required, specifically, the electron beam energy and electron beam width. PDDs have been used to adjust the electron beam energy. The dose distributions were calculated in water phantom placed at SSD = 90 cm and SSD = 100 cm from the target.

Our simulations started by running the nominal energy 6 MeV and then other energies from 5.5 MeV to 6.5 MeV were tested using a 0.1 MeV step size. In each simulation, the obtained results were compared to actual measured commissioning data acquired by the medical physics team by calculating the percentage dose difference between values.

To investigate the influence of electron beam width on the dose distribution, the lateral dose profile must be considered. Beam profiles are calculated at depth of 1.5 and 10 cm for doses simulated within a water phantom. The field size is set to $15 \times 15$ cm$^2$ at SSD = 90 cm. Each beam profile is normalized to its value in central axis and has been scaled for inclusion on the same graph. Several electron beam full width at half maximum (FWHM) values, from 1 to 4 mm were simulated. In each simulation, the results obtained are compared with experimental data by calculating the percentage difference.

Photon energy spectra for 6 MV photon beams were calculated by using the “beamdp_gui” application (Figure 4.1) to analyze the phase space file located at 32.5 cm from target. There is a peak on the spectrum, which belongs to positron annihilation photons of 0.511 MeV. Positron particles generated through the pair production effect, interact with electrons and undergo the positron-annihilation interaction. This interaction results in the creation of two 0.511 MeV photons.
4.3 Effect of electron beam energy on depth dose

The effect of electron beam energy (incident on the target) on percentage depth dose curves was studied. The incident electron energy was found to affect depth dose curves. (see Figure 4.2). The closest agreement between MC and measurement was found to occur for an electron energy of 5.9 MeV (Figure 4.3 and 4.4). The resulting dose differences with respect to measurement is < 2.0 % in the descending part of the PDD curve.
Figure 4.2 Percentage depth dose curves as the incident electron energy is varied between 5.6 MeV and 6.2 MeV. For clarity reasons, only the calculated curves for 5.6, 6.0 and 6.2 MeV are shown.

Figure 4.3 Comparison of simulated and measured PDDs for field size of 15×15 cm² at SSD = 100 cm. Electron beam nominal energy at the target was 5.9 MeV. Doses are normalized relative to the d_{max} dose.
4.4 Effect of electron beam width on beam profile.

It is seen from results that beam profiles are sensitive to electron beam width (Figure 4.5). The beam profile of $15 \times 15$ cm$^2$ field size for 6 MV photon beam were compared at depth of maximum dose. There are horns on beam profiles close to beam edge for electron beam width less than 1.9 mm. These horns disappear for beam profiles with electron beam width more than 1.9 mm. For quantitative evaluation of electron beam width on beam profile, we have shown the relative dose of each profile for different FWHM in figure 4.5. For our 6 MV beam, the calculated beam profile with electron beam width of 1.9 mm had the best agreement with measurements (Figure 4.6 and 4.7). The resulting dose differences with respect to measurements is $< 2.1\%$ for all low dose gradient regions of the profiles curve.

Figure 4-4 Comparison of simulated and measured PDDs for field size of $15 \times 15$ cm$^2$ at SSD = 90 cm. Electron beam nominal energy at the target was 5.9 MeV. Doses are normalized relative to the $d_{\text{max}}$ dose.
Figure 4-5 In-plane dose-profile curves as a function of FWHM of the electron beam. The energy of the incident electron beam is set to 5.9 MeV. The measured profile and three simulated in-plane dose profile curves are illustrated.

Figure 4-6 In-Plane profile of simulated dose and measured dose at depth of 1.5 and 10 cm at SSD = 90 cm, field size 15×15 cm². Electron beam FWHM is 1.9mm. Doses are normalized to its value in central axis and has been scaled for inclusion on the same graph.
4.5 Variation of MLC-defined field sizes

The secondary collimator in the Vero4DRT system is of a fixed type and the field sizes and field shapes are formed using only the MLC. The MLC for the Vero4DRT was fully modeled using the BEAMnrc SYNCVMLC component module. For verification of the MLC model, I repeated the MC vs measurement comparison for a wide range of MLC-defined field sizes, ranging from 1×1 to 10×10 cm², using the optimized electron beam energy and FWHM spot size.

Figure 4.8 shows the simulated and measured PDDs for field sizes of 1×1, 3×3 and 10×10 cm², respectively.
Figure 4-8 PDDs of the simulated and measured doses for field sizes 1×1, 3×3 and 10×10 cm² at SSD=100 cm. Doses are normalized relative to the d_{max} dose.

Figure 4.9 and 4.10 shows the comparisons between the simulated and measured in-plane dose profiles at a depth of 1.5 and 10 cm in water phantom with SSD of 90 cm for field size of 1×1, 3×3, 5×5, 7×7 and 10×10 cm², respectively. Each beam profile is normalized to its value in central axis.

Figure 4.11 and 4.12 shows the comparisons between the simulated and measured cross-plane dose profiles at a depth of 1.5 and 10 cm in water phantom with SSD of 90 cm for field size of 1×1, 3×3, 5×5, 7×7 and 10×10 cm², respectively. Each beam profile is normalized to its value in central axis.
Figure 4-9 In-Plane profiles of the simulated and measured doses for various field sizes at SSD=90 cm at depth=1.5 cm

Figure 4-10 In-Plane profiles of the simulated and measured doses for various field sizes at SSD=90 cm at depth=10 cm
Figure 4-11 Cross-Plane profiles of the simulated and measured doses for various field sizes at SSD=90 cm at depth=1.5 cm

Figure 4-12 Cross-Plane profiles of the simulated and measured doses for various field sizes at SSD=90 cm at depth=10 cm
The above results have demonstrated that our MLC model using the SYNCVMLC component module is accurate for simulating 3D dose distributions using MLC-defined fields shapes.

4.6 MLC Bar Pattern

Figure 4.13 shows a standard commissioning MLC bar pattern. A MC simulation through this aperture was performed for SSD = 100 cm in water phantom with $9 \times 10^8$ histories simulated in DOSXYZnrc. Calculated dose profiles in x and y direction were acquired and compared to measured water tank data. Figures 4.14 to 4.17 illustrate the comparison of cross-plane and in-plane profiles measured at depth of 1.5 and 10 cm. Each beam profile is normalized to its value along the central axis.

![Figure 4-13 MLC field setup for in-plane and cross-plane profile measurement](image)
Figure 4-14 Cross-Plane profiles of the simulated and measured doses for MLC Bar Pattern at SSD=100 cm at depth=1.5 cm

Figure 4-15 In-Plane profiles of the simulated and measured doses for MLC Bar Pattern at SSD=100 cm at depth=1.5 cm
Figure 4-16 Cross-Plane profiles of the simulated and measured doses for MLC Bar Pattern at SSD=100 cm at depth=10 cm

Figure 4-17 In-Plane profiles of the simulated and measured doses for MLC Bar Pattern at SSD=100 cm at depth=10 cm
4.7 MLC Transmission (inter/intra-leaf leakage)

MLCs contain a connecting tongue and groove geometry between nearby leaves to minimize leakage between adjacent leaves. The inter/intra-leaf leakage test was done to estimate the transmission properties of the MLC of the Vero4DRT. The leakage dose along the path perpendicular to MLC travel was calculated as the ratio of absolute dose with the MLC fully closed to absolute dose with the MLC fully opened (Figure 4.18). The voxel size in the uniform phantom was $0.2 \times 0.2 \times 0.2 \text{ cm}^3$. This test shows the radiation leakage between adjacent leaves in Vero4DRT MLCs is around 0.6 % which is higher than what Nakamura et al.$^{11}$ reported (0.2 %).

![Graph showing relative dose vs distance to central axis](image)

Figure 4-18  Leaf leakage ratio to estimate the transmission properties of the MLC of the Vero4DRT
4.8 Absolute dose calibration

The final goal is to define the amount of absorbed dose in the monitor chamber per monitor unit (MU) such that 1 MU will deliver 1 cGy to the patient at $d = 1.5$ cm for a $10 \times 10$ cm$^2$ field size. The beam with known Monitor Units is directed onto a water phantom and dose distribution calculated using DOSXYZnrc. The calibration geometry for the virtual linac is a $10 \times 10$cm$^2$ field, SAD = 100 cm and an isocentre depth of 10 cm. The raw dose from the DOSXYZnrc Monte Carlo simulation is in units of “dose per particle incident on the target”. The Monte Carlo “dose per particle hitting the target” is converted to absolute doses by assigning a known dose to a point at a depth of 10 cm the above calibration conditions. This dose point at depth of $d = 10$ cm is converted to a dose at $d = d_{\text{max}}$ (1.5 cm) by applying a tissue maximum ratio (TMR) factor of 0.772. Once this calibration has been established, the “Gy/Monte Carlo dose” can be applied to all voxels.

This equation has been applied to our Monte Carlo simulation to convert “dose per particle hitting the target” to “absolute dose”:\(^{28}\)

$$
D_{xyz,\text{abs}} = D_{xyz} \left( \frac{D_{\text{forward}} + D_{\text{back}}(10 \times 10)}{D_{\text{ch}} + D_{\text{back}}(10 \times 10)} \right) \frac{D_{\text{cal,abs}}}{D_{xyz}} \ U
$$

(4.1)

where, $D_{xyz}$ is the dose simulated in a voxel of phantom or patient in “Gy per incident electron on the target”

$D_{xyz,\text{abs}}$ is the absolute dose in the voxel of phantom or patient where $D_{xyz}$ was scored in “Gy”

$D_{\text{forward}}^{\text{ch}}$ is the normalized dose in monitor chamber related to the beam entering the ionization chamber from above in “Gy per incident electron on the target”
\(D_{ch}^{\text{back}}\) is the normalized dose in monitor chamber related to the beam entering the ionization chamber from the particles backscattered from the secondary collimator in “Gy per incident electron on the target”

\(D_{xyz,abs}^{\text{cal}}\) is the amount of absorbed dose in the monitor chamber per monitor unit (MU) at \(d = 10\) cm for a \(10 \times 10\ cm^2\) field size.

\(D_{xyz}^{\text{cal}}\) is the amount of absorbed dose per incident electron on the target to be deposited at \(d = 10\) cm for a \(10 \times 10\ cm^2\) field size.

As the secondary collimator for Vero4DRT is of a fixed type, the backscatter factor doesn’t depend on the field size

\[
\frac{(D_{ch}^{\text{forward}} + D_{ch}^{\text{back}(10 \times 10)})}{D_{ch}^{\text{forward}} + D_{ch}^{\text{back}}} = 1
\]  

(4.2)

With applying equation 4.2 in the equation 4.1 we have:

\[
D_{xyz,abs} = D_{xyz} \frac{D_{xyz,abs}^{\text{cal}}}{D_{xyz}^{\text{cal}}} U
\]  

(4.3)

\(D_{xyz,abs}^{\text{cal}}\) is equal to 0.772 MU per cGy and \(D_{xyz}^{\text{cal}}\) is equal to \(12.4 \times 10^{-17}\) Gy per incident electron on the target to be deposited at a depth of 10 cm and calculated by DOSXYZnrc for \(10 \times 10\ cm^2\) field size, using \(9 \times 10^8\) histories in a water phantom.

The above equation identifies that for a \(10 \times 10\ cm^2\) field of our Monte Carlo simulation of 6 MV photon beam of Vero4DRT, 1 MU corresponds to \(6.2258 \times 10^{15}\) electrons incident on the target.
Chapter 5 : Monte Carlo modelling of complex, clinical, radiotherapy treatment plans

In this chapter, we evaluate the new Monte Carlo model of the Vero4DRT against treatment plans calculated in RayStation (RaySearch, USA) for realistic patient geometries. RaySearch implements the Collapsed Cone Convolution (CCC) algorithm. Monte Carlo (MC) methods can offer highly accurate dose calculations in inhomogeneous media and for complicated geometries. A well commissioned MC model can potentially help reveal limitations of model-based dose calculation algorithms. It can also provide a valuable secondary dose calculation to verify dose distributions from commercial treatment planning systems (quality assurance).

5.1 RayStation (RaySearch) treatment planning system and the Collapsed Cone Convolution (CCC) algorithm

In radiation therapy, several treatment planning systems (TPS) are commercially available to deliver a prescribed dose to the target while sparing the surrounded healthy tissues. Several kinds of dose calculation algorithms are used in TPSs. The ICRU (International Commission on Radiation Units) recommends that the dose to the target volume should be delivered to within ±5%\(^{29-30}\). This is related to overall uncertainty on the dose received by the patient at the end of all steps contributing to radiotherapy dosimetry (including uncertainty due to 1) identifying the tumour/target in planning images, 2) radiotherapy machine performance, 3) set-up and immobilization of the patient/target prior to treatment, and 4) tumour/target localization during radiotherapy). The absolute accuracy of dosimetry calculations in radiotherapy range should be less than 2 to 3%\(^{31}\). This uncertainty in the dose calculation for a given TPS depends on the type of algorithm that has been used. Traditionally, a patient dose distribution was calculated by
correcting dose measurements obtained in a water phantom. Correction-based algorithms are the simplest to calculate but also the least accurate. Batho Power Law Method\textsuperscript{32}, Equivalent Tissue Air Ratio (ETAR) methods\textsuperscript{30} are some example of correction-based algorithms. They calculate scatter poorly (if at all) and are the least accurate of the dose correction algorithms, particularly in heterogeneous media (e.g. lung, tissue, bone). On the other hand, the newer model-based algorithms give a more accurate representation of the absorbed dose in heterogeneous media.

The simplest form of the model-based algorithms is the pencil-beam convolution (PBC) algorithm. For the PBC algorithm, when the beam is passing through the patient to deposit the dose, it will break up the open beam into numerous pencil-like beamlets. A main approximation of the PB algorithm is that it will ignore the nearest neighbor interaction (side-scatter) between nearby beamlets. This will cause lower accuracy in the dose calculation for low-density media (for lung, pencil beam (PB) models considerably over-predict the dose). The PBC algorithm cannot calculate doses accurately for smaller field-sizes, in low density medium and when depth is increasing within heterogeneity\textsuperscript{33}. The Collapsed Cone Calculation (CCC) is another example of a model-based algorithm which uses a 3D Convolution/Superposition (CS) model to compute dose in the patient. This algorithm is able to more accurately calculate dose distributions in regions of electronic disequilibrium, such as tissue-air and tissue-bone interfaces, particularly when compared to the PBC algorithm\textsuperscript{34}.

In this study, RayStation TPS was used and it employs the CCC algorithm. The CCC dose calculation algorithm calculates dose distribution in three steps: 1) the computation of the energy fluence, 2) a TERMA (total energy released per unit mass) component, which represents the energy transferred to the medium from the primary photons and, 3) a kernel component, which describes the energy deposited about a primary photon interaction site\textsuperscript{35}.
For most calculation algorithms, dose calculation accuracy in homogeneous medium (such as water) for conventional fields sizes (e.g. > 3x3cm$^2$) is almost same. In heterogeneous medium (e.g. a mixture of tissue, bone, lung, air), or for small field sizes (< 3x3cm$^2$) the differences in the accuracy of the dose calculation will depend on how well the kernels of these algorithms can calculate the radiation scattering in the different media. The kernels characterise the transport of energy or dose from an interaction point.

Dose can be calculated as dose-to-water, $D_w$ (per Bragg-Gray cavity theory), or dose-to-medium, $D_m$. Dose-to-water can be described as the dose imparted to a certain point if an infinitesimal volume of tissue (muscle, fat, lung, cortical bone etc.) is switched with an infinitesimal volume of water. In radiation therapy it is traditional to report dose as dose-to-water and so, treat all materials as water with different densities. This is also considered for Task Group (TG) 21$^{36}$ or TG 51$^{37,38}$ calibration guidance documents.

Monte Carlo based treatment planning systems report energy deposition in patient in dose to medium, $D_m$. This is a beneficial aspect of Monte Carlo simulation for treatment planning and many physicists consider that $D_m$ is more accurate for treatment planning, and it should replace $D_w$ in the TPS$^{39}$. Converting $D_m$ to $D_w$ requires stopping power ratios, and some physicists argue that this may add more uncertainty to the dose calculation, is more complicated and will increase the time of computation$^{39}$. In addition, clinical radiotherapy trials and their outcomes data have historically been linked to a $D_w$ formulism for calculating dose.

The reason for using Monte Carlo simulation in treatment planning is to achieve the highest accuracy in dose calculation. This conversion from Monte Carlo simulated $D_m$ to $D_w$ could be viewed as a setback, thus defeating the motivation for using Monte Carlo dose calculations.
The RayStation CCC dose algorithm calculates dose as dose-to-medium, $D_m$, and later converts it to dose-to-water, $D_w$. There are two reasons for this:

- It has been a tradition and a benchmark.
- It is said to be clinically relevant since the radiation sensitive parts of the cell are surrounded by water\textsuperscript{40-41}.

The differences between dose-to-water and dose-to-medium for soft tissues are often considered to be not clinically important (the difference between $D_m$ and $D_w$ is approximately 1.0%), while for cortical bone the dose difference is approximately 10\%\textsuperscript{42}. Some photon dose calculation algorithms, including the collapsed cone convolution calculates dose-to-water but assigns different electron densities to account for different heterogeneous tissues. This can be done since the dose deposited in medium depends mostly on the electron density and for photon beams in the energy range used for radiotherapy, the main type of interaction is Compton scattering.

### 5.2 Workflow and Scripting for Monte Carlo

An important component of any treatment plan validation process is a ability to extract all patient geometry and treatment field information from the treatment planning system. Many modern TPS use a standard DICOM file format for importing and exporting plans, CT data and dose distributions.

There are four different DICOM files that must be extracted in order to run MC calculations:

- **CT Images**: multiple CT image files corresponding to Hounsfield Unit data representing the patient geometry.
- **RT Structures**: contains structure/contour information representing patient anatomy and planning targets.
- **RT Plan:** contains all treatment plan parameters, and in particular the parameters required to model radiation beams with BEAMnrc (number of MUs per treatment field, gantry angles, ring angle, and isocentre position, number of fractions planned, number of beams, beam energy, the dynamic MLC position and number of control points)

- **RT Dose:** contains RayStation CCC-calculated 3D dose data. Used as a dummy file to wipe and replace with Monte Carlo dose distributions.

In-house scripts, codes and supplementary programs were designed by the medical physics team to read the DICOM files data and create Monte Carlo-friendly files for BEAMnrc and DOSXYZnrc simulations. The entire Monte Carlo simulation is scripted\(^\text{43}\) such that the user only has to provide the initial DICOM CT, structure, plan and dose files from RayStation. The script automatically generates the patient-geometry MC phantom, creates all input files required by the BEAMnrc and DOSXYZnrc simulations, and launches the simulation. Once a Monte Carlo 3D dose distribution is created on the patient geometry, additional processing is also automatically applied such as 3D smoothing (denoising) filters and the conversion to absolute dose. A 3D denoising filter based on the Savitzky-Golay formalism is applied to Monte Carlo 3D dose data\(^\text{44}\). Savitzky and Golay’s goal was to remove noise from 1D data\(^\text{45}\).

### 5.3 Monte Carlo Verification of RayStation Patient Plans

Radiation delivery plans for each patient in this study were recomputed with MC to compare with the collapsed cone convolution (CCC)\(^\text{46}\) algorithm utilized by the RayStation treatment planning system. The Brainlab Vero4DRT is capable of delivering conventional three-dimensional conformal radiation therapy (3DCRT) beams, dynamic conformal arcs (DCA), static field step-and-shoot IMRT (sIMRT) beams, and VMAT arcs. It also has a unique delivery mode...
called Dynamic Wave Arc (DWA) which employs VMAT deliveries combined with simultaneous motion of the gantry and ring about two axes of rotation. The following sections demonstrate examples of treatment plans generated for various beam delivery options available on Vero4DRT.

For the MC simulations, the dose calculation grid for each patient included the entire patient CT data set and the voxel size was 2.5 mm in each x, y, and z Cartesian coordinates. The analytical nature of Monte Carlo simulations is subject to a statistical uncertainty. My goal was to reduce the statistical uncertainty less than 2% which achieved by $9 \times 10^8$ numbers of histories.

### 5.3.1 Three-dimensional conformal radiation therapy (3DCRT)

A seven-field, non-coplanar 3DCRT plan for Vero4DRT was created on a patient CT dataset using the RayStation TPS (Figure 5.1). Stereotactic ablative radiation therapy (SABR) was planned to the liver. The prescribed dose was 35 Gy in 5 fractions (7 Gy per fraction). The planning goals were to cover > 95% of the planning target volume (PTV) with 100% of the prescription (35 Gy) while respecting maximum and mean dose constraints to nearby organs-at-risk (OAR), for example the stomach and great vessels. Figure 5.2 shows the dose distribution calculated by RayStation (collapsed cone convolution (CCC) algorithm) and Monte Carlo simulation for this static gantry, static ring and static conformal MLC plan. PTV and OAR dose volume histograms (DVH) for these algorithms are shown in figure 5.3.
Figure 5-1 a seven-field, non-coplanar 3DCRT plan

Figure 5-2 Comparison of dose distribution calculated by a) MC and b) CCC for 3DCRT plan. Planning target volume (PTV) shown in red, liver in blue, great vessels in pink, stomach in brown.
Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS is listed in table 5.1. The MC simulation dose shows some under coverage of the $V_{35Gy}$ compared to RS (4.7% difference). The mean dose to the PTV for the MC compared to RS agrees within 0.3 Gy. The PTV and OAR max doses (to 0.035 cc) were matching within 1.14 Gy.

<table>
<thead>
<tr>
<th></th>
<th>RS TPS (CCC algorithm)</th>
<th>MC simulation</th>
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<tbody>
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<td>$V_{35Gy}$ for the PTV</td>
<td>94.5%</td>
<td>90%</td>
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<tr>
<td>The mean dose of PTV</td>
<td>37.7 Gy</td>
<td>37.4 Gy</td>
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<td>The mean dose to Liver (minus GTV)</td>
<td>6.69 Gy</td>
<td>6.89 Gy</td>
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<tr>
<td>$D_{0.035 \text{ cc}}$ of PTV</td>
<td>40.58 Gy</td>
<td>41.72 Gy</td>
</tr>
<tr>
<td>$D_{0.035 \text{ cc}}$ of Great Vessels</td>
<td>14.65 Gy</td>
<td>15.67 Gy</td>
</tr>
<tr>
<td>$D_{0.035 \text{ cc}}$ of Stomach</td>
<td>10.42 Gy</td>
<td>10.71 Gy</td>
</tr>
</tbody>
</table>

Table 5-1 Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS for 7-fields 3DCRT SABR plan
5.3.2 Intensity modulated radiation therapy (IMRT)

A static field step-and-shoot intensity modulated radiation therapy (sIMRT) plan was created on the RayStation TPS for a patient with left iliac bone cancer (figure 5.4). The prescribed dose was 35 Gy in 5 fractions. Figure 5.5 shows the dose distribution calculated by RayStation treatment planning system (CCC algorithm) and Monte Carlo simulation for this static gantry, static ring, and dynamic MLC plan. PTV and OAR DVH for these algorithms are shown in figure 5.6.

Figure 5-4 Intensity-modulated radiation therapy (IMRT) technique. MLC is moving while radiation beam is “on”.

Figure 5-5 Comparison of dose distribution calculated by a) MC and b) CCC for sIMRT plan. PTV shown in red, sacral canal in green.
Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS is listed in table 5.2. The MC simulation dose shows some over coverage of the $V_{35\text{Gy}}$ compared to RS (~1% difference). The mean dose to the PTV for the MC compared to RS agrees within 0.3 Gy. The PTV and OAR max doses (to 0.035 cc) were matching within 1.5 Gy.

<table>
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<th>RS TPS (CCC algorithm)</th>
<th>MC simulation</th>
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<tr>
<td>$V_{35\text{Gy}}$ for the PTV</td>
<td>94.8%</td>
<td>95.8%</td>
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<td>The mean dose of PTV</td>
<td>38.2 Gy</td>
<td>38.5 Gy</td>
</tr>
<tr>
<td>$D_{0.035 \text{cc}}$ of PTV</td>
<td>42.2 Gy</td>
<td>43.7 Gy</td>
</tr>
<tr>
<td>$D_{0.035 \text{cc}}$ of Sacral Canal</td>
<td>14.58 Gy</td>
<td>15.42 Gy</td>
</tr>
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</table>

Table 5-2 Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS for a static field step-and-shoot intensity modulated radiation therapy (sIMRT) plan
5.3.3 Dynamic conformal arc (DCA)

A DCA plan was created on the RayStation TPS for a patient with left lung cancer. The prescribed dose was 48 Gy in 4 fractions (figure 5.7). Figure 5.8 shows the dose distribution calculated by RayStation treatment planning system (CCC algorithm) and Monte Carlo simulation for dynamic gantry, static ring, and dynamic conformal MLC plan. PTV and OAR DVH for these algorithms are shown in figure 5.9.

![Figure 5-7 Dynamic conformal arc (DCA) therapy](image)

![Figure 5-8 Comparison of dose distribution calculated by a) MC and b) CCC for DCA Lung SABR plan. PTV shown in red, chest-wall OAR in yellow.](image)
Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS is listed in table 5.3. The MC simulation dose shows good agreement of the V_{48Gy} compared to RS (~0.1% difference). The mean dose to the PTV for the MC compared to RS agrees within 0.3 Gy. The PTV and OAR max doses (to 0.035 cc) were matching within 1.28 Gy.

<table>
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<tr>
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<th>RS TPS (CCC algorithm)</th>
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<tr>
<td>V_{48Gy} for the PTV</td>
<td>96.64%</td>
<td>96.74%</td>
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<td>The mean dose of PTV</td>
<td>55.78 Gy</td>
<td>56.07 Gy</td>
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<tr>
<td>D_{0.035 cc} of PTV</td>
<td>67.4 Gy</td>
<td>68.1 Gy</td>
</tr>
<tr>
<td>D_{0.035 cc} of Left Chest-wall</td>
<td>42.17 Gy</td>
<td>40.89 Gy</td>
</tr>
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</table>

Table 5-3 Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS for DCA Lung SABR plan

Figure 5-9 Comparison of PTV and OAR dose-volume histograms (DVHs) calculated by RS (triangle) and MC (square) for DCA plan.
5.3.4 Volumetric modulated arc therapy (VMAT)

A VMAT plan was created on the RayStation TPS for a patient with liver cancer (figure 5.10). The prescribed dose was 45 Gy in 3 fractions. Figure 5.11 shows the dose distribution calculated by RayStation treatment planning system (CCC algorithm) and Monte Carlo simulation for this dynamic gantry, static ring, and dynamic MLC plan. PTV and OAR DVH for these algorithms are shown in figure 5.12.

![Figure 5-10 Volumetric-modulated arc therapy (VMAT)](image)

![Figure 5-11 Comparison of dose distribution calculated by a) MC and b) CCC for a liver SABR VMAT arc plan](image)
Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS is listed in table 5.4. The MC simulation dose shows some over coverage of the $V_{45Gy}$ compared to RS (~2.1% difference). The mean dose to the PTV for the MC compared to RS agrees within 0.25 Gy. The mean dose to the liver (minus GTV) for the MC compared to RS agrees within 0.66 Gy. The PTV and OAR max doses (to 0.035 cc) were matching within 0.97 Gy.

<table>
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<th>RS TPS (CCC algorithm)</th>
<th>MC simulation</th>
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<td>$V_{45Gy}$ for the PTV</td>
<td>95%</td>
<td>97%</td>
</tr>
<tr>
<td>The mean dose of PTV</td>
<td>51.35 Gy</td>
<td>52.10 Gy</td>
</tr>
<tr>
<td>The mean dose of Liver-GTV</td>
<td>15.04 Gy</td>
<td>15.70 Gy</td>
</tr>
<tr>
<td>$D_{0.035\text{ cc}}$ of PTV</td>
<td>59.21 Gy</td>
<td>60.18 Gy</td>
</tr>
<tr>
<td>$D_{0.035\text{ cc}}$ of Left Chest-wall</td>
<td>31.11 Gy</td>
<td>31.83 Gy</td>
</tr>
</tbody>
</table>

Table 5-4 Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS for a liver SABR VMAT arc plan
5.3.5 Dynamic wave arc (DWA)

A complex, patient liver SABR treatment plan using dynamic, non-coplanar trajectories (dynamic wave VMAT arc) is simulated (figure 5.13). The prescription to the planning target volume (PTV) is 54 Gy in 3 fractions. Figure 5.14 and 5.15 shows the dose distribution calculated by RayStation treatment planning system (CCC algorithm) and Monte Carlo simulation in axial and coronal view, respectively. The non-coplanar beam entry is modelled correctly. PTV and OAR DVHs are compared to RayStation TPS calculated doses (Figure 5.16).

Figure 5-13 liver SABR treatment plan using dynamic, non-coplanar trajectories (dynamic wave VMAT arc)

Figure 5-14 Comparison of dose distribution calculated by a) MC and b) CCC for liver SABR DWA plan axial view
Figure 5-15 Comparison of dose distribution calculated by a) MC and b) CCC for liver SABR DWA plan sagittal view.

Figure 5-16 Comparison of PTV and OAR dose-volume histograms (DVHs) calculated by RS (square) and MC (triangle) for DWA plan.

Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS is listed in table 5.5. The MC simulation dose shows excellent agreement of the $V_{54Gy}$ compared to RS. The mean dose to the PTV for the MC compared to RS agrees within 0.11 Gy. The mean dose to the liver (minus GTV) for the MC compared to RS agrees within 0.38 Gy. The PTV and OAR max doses (to 0.035 cc) were matching within 2.07 Gy.
<table>
<thead>
<tr>
<th></th>
<th>RS TPS (CCC algorithm)</th>
<th>MC simulation</th>
</tr>
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<tbody>
<tr>
<td>$V_{54\text{Gy}}$ for the PTV</td>
<td>95%</td>
<td>95%</td>
</tr>
<tr>
<td>The mean dose of PTV</td>
<td>61.15 Gy</td>
<td>61.26 Gy</td>
</tr>
<tr>
<td>The mean dose to Liver-GTV</td>
<td>13.34 Gy</td>
<td>13.72 Gy</td>
</tr>
<tr>
<td>$D_{0.035\text{cc}}$ of PTV</td>
<td>67.42 Gy</td>
<td>69.49 Gy</td>
</tr>
<tr>
<td>$D_{0.035\text{cc}}$ of Left Chest-wall</td>
<td>66.92 Gy</td>
<td>67.84 Gy</td>
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</table>

Table 5-5 Comparison of dosimetry metrics for OAR and PTV calculated by MC and RS for a liver SABR treatment plan using dynamic, non-coplanar trajectories (dynamic wave VMAT arc)
Chapter 6: Conclusion and future work

6.1 Conclusion

The purpose of my study was to build a simulation model of the 6 MV linear accelerator head and multileaf collimator (MLC) for the Brainlab Vero4DRT and verify the dose calculation accuracy. A Monte Carlo simulation code was implemented using EGSnrc/BEAMnrc and EGSnrc/DOSXYZnrc codes. Working with our Japanese counterparts, the main components of the linac head were modelled. Simulated simple radiation doses in a water phantom were compared to actual measured commissioning data to allow for fine tuning of the Monte Carlo model.

The Japanese model could only model static 3DCRT beam deliveries. My extensive modifications enabled Monte Carlo dose calculations for dynamic gantry, ring, and MLC deliveries – essentially covering the entire suite of radiotherapy delivery options available.

Every example presented in previous chapters required a single, Monte Carlo run, even when employing multiple fields, multiple gantry and ring angles, and dynamic MLC motions. The whole process from CT phantom generation to simulation to final data processing takes approximately 70 minutes when employing 50 nodes on a dedicated Monte Carlo distributed computer cluster. This cluster has up to 256 nodes available, so the computing time could be reduced to 15 - 20 minutes.

Every treatment delivery available in Vero4DRT with dynamic or static gantry angle, dynamic or static ring, dynamic or static MLC leaf motion, was modelled correctly in our Monte Carlo simulation.

Monte Carlo simulations are very helpful in assessing the accuracy of different dose calculation algorithms such as the collapsed cone convolution (CCC) algorithm.
It should be noted that the CCC algorithm reports dose in the patients in terms of dose to water, which is consistent with most commercial treatment planning systems. However, the Monte Carlo method calculates the dose to different biological media and provides dose to a medium $D_m$ directly.

The couch of Vero4DRT has 2% attenuation and data related to couch is imported in RayStation TPS however, in our Monte Carlo model we haven’t simulated coach yet, which can cause some differences in dose calculation compared to CCC for the fields coming from the posterior direction. The differences in dose coverage seen in 3DCRT might be because of this.

I have built an efficient Monte Carlo model of the Vero4DRT radiotherapy linac. This model will be useful as secondary dose calculation verification for the clinic’s commercial treatment planning system. Medical physicists at this centre are eager to implement the results of this work clinically as an important part of their quality assurance processes for the Vero4DRT medical accelerator.

### 6.2 Future work

A number of pathways for future research and development have been discovered by the author. The Monte Carlo model developed in this thesis could be extended to account for 4D dynamic tumour tracking (DTT) treatments. Currently the model ignores the gimbal motion relative to the patient respiratory motion in the dose simulation. No commercial treatment planning systems (TPS) currently supports DTT dose calculations accounting for the moving, gimbaled x-ray head. Some investigators have developed approaches to account for these respiratory motion effects in the calculation of dose distributions. It may be possible to use the treatment delivery log files to simulate and synchronise the dynamic motion of machine components with patient motion and generate a dose distribution which is more representative of the actual patient DTT treatment.
It would be useful to develop a dose calculation system for the dynamic tumor-tracking irradiation.
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