Modeling of megavoltage and low energy focused x-ray beams using MCNP5

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Modeling of megavoltage and low energy focused x-ray beams using MCNP5

by

Hassan Abbas

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by

Hassan Abbas

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Dedicated

To the memories of my late father Nazar Muhammad

&

To the memories of my late niece Amra Rubab
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Hassan Abbas
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Abstract

In this thesis 6 and 18 MV photon beams for a Varian linear accelerator were simulated using a general purpose Monte Carlo code MCNP5 (Monte Carlo N-Particle Transport Code). The potential of MCNP5 is explored to calculate the dose in high dose gradient regions. In addition, MCNP5 was used to model another geometry, low energy focused x-ray beams created with polycapillary optics, to explore their potential to provide a different skin-sparing technique for radiation therapy.

Modeling was benchmarked by adjusting electron beam energy and diameter to match calculations with ionization chamber (with 0.125 cm$^3$ sensitive volume) measurements of percent depth doses (PDDs) beyond d$_{\text{max}}$ and cross dose profiles at d$_{\text{max}}$ and 10 cm depths for 5×5, 10×10 and 30×30 cm$^2$ field sizes. For comparison in the buildup region, the MCNP5 voxel was reduced to 1 mm, with extrapolation to find surface dose. In this region a plane parallel chamber (with 0.055 cm$^3$ sensitive volume) was used to measure PDDs at 0, 2 and 4 mm depths for the three field sizes, using the Khan over-response correction.

Calculations and cylindrical chamber measurements for PDDs beyond d$_{\text{max}}$ agree within 2% for all field sizes and energies. Dose profiles agree within 2% in the flat region and within 10% in the penumbra region for all field sizes and energies. In the buildup region, the maximum deviation between MCNP5 calculations and the plane parallel PDD measurements at 18 MV is 3.8% for 30×30 cm$^2$ at 2 mm depth. The deviation is within 3% for all other field sizes and depths for both beam energies. For 18
MV, the extrapolation of the MCNP5 PDD overestimates the surface dose for the 30x30 cm² field size, and is within 4.6% for all other field sizes at both beam energies.

A focused beam obtained with polycapillary optics was simulated using MCNP5. Low energy focused x-ray beams could be used to irradiate the tumors inside soft tissue within 5 cm of the surface. A better skin sparing compared to megavoltage photon beams can be achieved by the low energy focused x-ray beam inside soft tissues.
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1 Introduction

Radiation therapy (also called radiotherapy or RT) is a cancer treatment option in which ionizing radiation is used to kill cancer cells. It is often used in combination with surgery and chemotherapy, but may also be used as a primary treatment mode depending upon the type and location of disease. Radiation therapy is divided into two major types depending upon the position of the radiation source. In brachytherapy, sealed or unsealed radioactive sources are placed near the treatment area either temporarily or permanently. In external beam radiation therapy, the radiation beams used in treatment originate from outside of the patient. The radiation beam can be photons, electrons, protons or heavy charged particles. Megavoltage photon beams are the most commonly used modality in cancer treatment at present.

In radiation therapy, DNA of malignant cells is the target to be damaged by the ionizing radiation. Cell damage can be direct or indirect. In direct action, radiation hits the target atoms (DNA) directly producing ionizations or excitations which lead to the physical and chemical changes that ultimately produce biological damage to the target. This action may affect the ability of the cells to reproduce or to perform certain actions. This type of cell damage is called functional cell death. The probability of the radiation interacting with the DNA molecule directly is very small since it makes up very small part of the cell. However, each cell, just as is the case for the human body, is mostly water. In indirect interaction, the radiation first interacts with the water molecules to produce highly reactive hydroxyl ions and then these ions ultimately damage the target. As the first step, water molecules become ionized.
\[ \gamma + \text{H}_2\text{O} \rightarrow \text{H}_2\text{O}^+ + e^- . \]

The ionized water molecule has a very short life time (~10\(^{-10}\) sec) before reacting with a neutral water molecule to produce the hydroxyl radical

\[ \text{H}_2\text{O}^+ + \text{H}_2\text{O} \rightarrow \text{H}_3\text{O}^+ + \text{OH}^*. \]

The hydroxyl radical, \(\text{OH}^*\), has one unpaired electron and is very reactive. This free radical diffuses into the target DNA and can damage it. It is estimated that about two thirds of the damage produced by x rays is through indirect action.\(^1\)

Once cells are irradiated, there could be many possible outcomes including repair and cell death. The time line between irradiation and the outcomes could be hours to days or up to years depending on the type of damage produced. DNA damage produced by x rays could be a single strand break (sub-lethal) or double strand break (lethal). Since cancer cells divide rapidly compared to normal cells, they are more prone to radiation damage. Radiation therapy is designed to maximize damage to the tumor while minimizing damage to the normal tissues. Normal cells can repair the damage if the dose is not too high and if the cells are provided time to recover. Due to this reason, most patients do not receive a full radiation treatment on a single day. Dividing the radiation treatment over several days is known as fractionation. Fractionation reduces the risk of secondary cancers.\(^2\)
1.1 External Beam Radiation Therapy

X rays were discovered by Wilhelm Conrad Rontgen in 1895. Radiation therapy using x rays was performed within a few months of this discovery. Radium was discovered by Marie Curie in 1898 and the use of radioactive sources was also started to treat cancer. The process of radiation treatment delivery was not well characterized and the practice was done by few specialists in very limited number of hospitals in the world. Since very little was known about radiation initially, the practice was based on empirical methods. A few parameters (e.g. field size, dose, position) were varied depending on the experience of the physicist and the location and size of the disease. The fundamental basic principles of radiotherapy were established at the start of the 1940s. These principles are still being followed today. The goals are accurate determination of the size, shape and position of the volume to be treated, achievement of uniform dose distribution inside the tumor volume and sparing of the normal tissues from unnecessary radiation dose.\(^3\)

Machines for external beam radiation treatment were developed in the 1920s and 1930s. Two types of machines were developed. X-ray machines used accelerated electrons to produce bremsstrahlung x rays to treat cancer. Although x-ray machines up to 800 KV were developed,\(^4\) most of the centers employed x-ray machines with accelerating voltages in the range of 200-300 KV. The other type of machines, originally called Radium teletherapy machines, used radioactive sources to produce gamma rays to irradiate the tumor. The average photon energy from Radium-226 machines was \(\sim 0.8\) MeV. Photon beams from x-ray machines and Radium-226 machines deposit the maximum dose at the surface or in the vicinity of first few millimeters of the skin of the
patient. The dose decreases as the beam is attenuated along the patient depth. When treating deep-seated tumors with these machines, the skin received very high unwanted radiation doses. It was proposed that high energy beams with greater penetrating power would deposit the maximum dose at some distance from the skin and hence would be more suitable to treat the deep-seated tumors. High energy (Megavoltage) x rays produce charged particles, mostly electrons when interacting with the tissue in the near surface. These charged particles deposit most of their energy at a finite distance from their birth place, reducing the surface dose. This is referred to as “skin sparing” in radiation therapy. For this reason Radium-226 radioactive sources with average photon energy of 0.8 MeV were replaced by Cobalt-60 sources with higher average photon energy (1.25 MeV) in the 1950s. Initially machines were not capable of moving around the patient so multi-angle treatment was not possible. With the development of sophisticated teletherapy machines, multi-angle treatment became possible offering better skin sparing compared to the limited field treatments.

Cobalt-60 machines were better than x-ray and Radium-226 machines but still were not suitable for tumors with larger depths. The first medical linear accelerator was developed and introduced in 1953. The main advantage of medical linear accelerators over the older x-ray machines now called “orthovoltage” was better skin sparing because of the higher accelerating voltage. Another advantage of using the linear accelerator over the radioactive source-based machines was that there was no need to replace the radioactive source. In a medical linear accelerator, the electrons are accelerated in multiple cavities of a waveguide with the help of a microwave power tube such as Klystron. These accelerated electrons are then guided with the help of magnets to hit a
thin metallic target plate. When high speed electrons hit the target, bremsstrahlung x rays are produced. The x ray beam is then shaped to conform to the volume of the tumor with the help of independent jaws, blocks and multi-leaf collimators. Medical linear accelerators with multiple electron and photon beam energies are being used all over the world for the treatment of cancer. Since the maintenance of a medical linear accelerator is somewhat complex as compared to a Cobalt-60 machine, Cobalt-60 machines are still in use in developing countries.  

Recent advances in external beam radiation treatment modalities include intensity modulated radiation therapy (IMRT), volumetric modulated arc therapy (VMAT) and image guided radiation therapy (IGRT). These offer a high conformal dose to the tumor while sparing the normal tissues. For these latest treatment techniques, the treatment planning system has to perform calculations for complex dose distributions. Historically, treatment planning systems were based on empirical, analytical and semi-analytical approaches for dose calculation. Most advanced treatment planning systems available in the market use model-based approaches (convolution/superposition) for dose calculation. In order to verify the radiation doses to normal tissues in complex treatment plans, performance of these commercial treatment planning systems need to be verified in steep dose gradients along tissue interfaces. Several authors\textsuperscript{5,6,7,8,9,10} have reported that the Monte Carlo methods give better accuracy at tissue interfaces compared to conventional dose calculation methods. Although statistical uncertainty in Monte Carlo calculation can never reach 0%, with high numbers of input photons it can meet the desired 3% uncertainty requirement in calculations (which leads to a total uncertainty of 5%) for complicated geometry and patient composition. Because of the long run times it may not
be used as stand-alone dose calculation engine for all patients but it can be used for benchmarking the calculations performed by commercial treatment planning systems in extreme cases.

MCNP5 (Monte Carlo N-Particle Transport Code (MCNP)) is a general purpose Monte Carlo code commonly used for simulating neutron, electron, photon or coupled neutron/photon/electron transport through any media. MCNP was originally developed at Los Alamos National Laboratory to study neutron transport. It offers a combinatorial geometry package that enables complex problems to be modeled. In this thesis 6 and 18 MV photon beams are simulated using MCNP5. The potential of MCNP5 is explored to calculate the dose in high dose gradient regions. Particular attention is placed on the near surface region where skin sparing dose build up occurs. In addition, MCNP5 was used to model another geometry, low energy focused x ray beams created with polycapillary optics, to explore their potential to provide a different skin-sparing technique for radiation therapy.
References:


2 Theory and Background

2.1 Radiation Interaction with Matter

2.1.1 Photon Interactions

The intensity, $I$, of a monoenergetic photon beam incident on a material changes according to exponential attenuation

$$I = I_o e^{-\mu(E)x}, \quad \text{Equation 2-1}$$

where $I_o$ is the incident intensity, $x$ is the distance travelled through medium and $\mu$ is linear attenuation coefficient which is function of incident photon energy, $E$. This expression is valid for narrow beam geometries, for which scattered radiation does not reach the detector. The linear attenuation coefficient ($\mu$) divided by the density of medium is called the mass attenuation coefficient ($\mu/\rho$) and is the commonly used quantity in radiation therapy as it depends only on elemental composition. When a photon beam passes through some medium, different types of interactions take place depending on the energy of the photon and atomic/electronic composition of the medium. The linear attenuation coefficient is the sum of the coefficients of all these possible interactions and can be written as

$$\mu = \mu_{PE} + \mu_{CS} + \mu_{RS} + \mu_{TS} + \mu_{PP}, \quad \text{Equation 2-2}$$
where $\mu_{PE}$ is photoelectric effect coefficient, $\mu_{CS}$ is Compton scattering coefficient, $\mu_{RS}$ is the Rayleigh scattering coefficient, $\mu_{TS}$ is the Thomson scattering coefficient and $\mu_{PP}$ is the pair production coefficient.

### 2.1.1.1 Photoelectric Absorption

In this process, the incident photon ejects an electron whose binding energy is equal or less than the photon energy. The incident photon is absorbed and the ejected electron deposits energy in the medium through excitations and ionizations. The photoelectric effect coefficient is proportional\(^\text{1,2}\) to $Z^4 E^{-3}$

\[ \mu_{PE} \sim Z^4 E^{-3}, \quad \text{Equation 2-3} \]

so that

\[ \left( \frac{\mu}{\rho} \right)_{PE} \sim \frac{Z^4}{A} E^{-3}, \quad \text{Equation 2-4} \]

where $E$ is the photon energy, $Z$ is atomic number and $A$ is the atomic mass number. Since $Z/A$ is roughly constant, the mass attenuation coefficient for photoelectric absorption varies as $Z^3$. In tissue, the photoelectric effect is the dominant interaction up to about 25 keV\(^\text{1,2}\).

### 2.1.1.2 Compton (Incoherent) Scattering

Compton (incoherent) scattering is an inelastic scattering in which both the energy and direction of the emitted photon is different than the incident photon. The incident photon interacts with an approximately free and orbital electron (E $\gg$ binding energy of electron) and gives part of its energy to the ejected electron. The Compton scattering interaction coefficient can be given as
so that the mass attenuation coefficient could be written

\[
\mu_{CS} \sim Z \frac{E}{A}^{1/2}, \quad \text{Equation 2-5}
\]

\[
\left( \frac{\mu}{\rho} \right)_{CS} \sim \frac{Z}{A} E^{1/2}. \quad \text{Equation 2-6}
\]

Since \( Z/A \) is constant, the mass attenuation coefficient for Compton scattering is independent of the atomic density, however it varies with the electron density of the medium. Compton scattering dominates in most of the therapeutic energy range.\(^1\)\(^2\)

### 2.1.1.3 Coherent Scattering

In the coherent (Rayleigh, Thomson) scattering process, the scattered photon has the same energy as the incident photon, but is deflected to some angle with respect to the incident direction. In tissue equivalent materials, the importance of coherent scattering is small with respect to the other interaction types when energy deposition is considered. They are important to consider in the simulation in order to track the direction of the incident photon.

### 2.1.1.4 Pair Production

A pair production event may occur when a photon with energy greater than or equal to twice the rest mass energy of electron and positron is located in the Coulomb field of the nucleus. The incident photon is absorbed and an electron-positron pair is created. The electron and positron produced in the event lose their energy by excitations, ionizations and bremsstrahlung events. The positron, after losing its most of energy, may recombine with an electron to give annihilation photons. The pair production coefficient can be written as\(^1\)\(^2\)
2.1.2 Electron Interactions

Electrons can interact with orbital electrons and atomic nuclei through Coulomb interactions. Electron interactions can be classified as soft collisions, hard collisions and radiative interactions. This classification depends upon the impact parameter, b, and on the atomic radius, a. The impact parameter b is the perpendicular distance from the direction of the electron movement to the nuclei before the interaction. If the impact parameter is much larger than the atomic radius (b>>a), then the electron interacts with the atom as a whole and the interaction is called a soft collision. In soft collision a small part of the incident electron’s energy is taken by the orbital electrons. The incident electron direction could also change in a soft collision. If the impact parameter is comparable to the atomic radius (b~a), then the incident electron collides with an orbital electron and the interaction is called a hard collision. In this interaction a large fraction or all of the energy of the incident electron is transferred to the orbital electron. This energy transfer may lead to the ejection of the bound electron. The ejected electron, called a secondary electron or delta ray, deposits its energy in the medium causing further ionization. The filling of the vacancy created by the emission of the orbital electron can cause emission of characteristic x rays, or the energy can be carried by an Auger electron emitted from the atom. If the impact parameter is smaller than the atomic radius (b << a),
then the electron interacts with the nucleus of the atom. This type of interaction is called a radiative interaction. In a radiative interaction, a bremsstrahlung photon is emitted, having an energy between zero and the kinetic energy of the incident electron. The radiative interaction of electrons gives bremsstrahlung x rays with continuous energy spectrum. The direction and energy of the bremsstrahlung x rays depend on the energy of the incident electrons and on the impact parameter. As the energy of the incident electrons increases, more bremsstrahlung x rays are produced in the forward direction. When megavoltage electrons hit the target in medical linear accelerators, most of the bremsstrahlung x rays produced are forward directed.\textsuperscript{1,2}

2.2 Basics of Radiation Therapy

The radiation treatment delivery process consists of three steps: scan, plan and deliver. The size, location and extent of the disease are determined through imaging (PET scan, CT scan, MRI etc.). Once the disease location and volume is determined, the next step is treatment planning. Treatment planning is a process in which the possible beam orientations, field sizes and beam energies are decided. The objective of treatment planning is to achieve uniform dose distribution inside the tumor and to deliver the maximum possible dose inside the tumor while sparing normal tissue. The radiation dose to be delivered to the tumor is prescribed by the physician and the physicist assures the accurate delivery of the prescribed dose to the tumor sparing the normal tissues.

2.2.1 Inhomogeneity Effects in Dose Calculation

External beam radiation therapy is a very effective treatment modality for cancer but the efficacy of the radiation therapy depends on \textit{accurate} dose delivery to the
planning target volume while minimizing the dose to normal tissue. It has been reported\(^3\) that a 1% accuracy improvement in dose delivery results in a 2% increase in cure rate for early stage tumors whereas a 5% change in dose may result in a 10% to 20% change in tumor control probability (TCP) and a 20% to 30% impact on complication rates in normal tissues. If the human body were a single tissue with a single density then the dose calculation would have been a trivial problem. However, the human body is not a homogenous medium; it consists of tissues with different densities such as lungs, oral cavities, bones, teeth, nasal passages and sinuses. The present overall desired accuracy in the whole radiation treatment delivery process is 5%. The total uncertainty in the delivered dose at the reference point is sum of different independent sources of error. Some of these sources of error include patient movement, uncertainty in the calibration of machine output and uncertainty in the dose calculation. If we consider the contribution of all these sources of error towards overall desired accuracy of 5%, then the uncertainty in dose calculation should be less than 3% to achieve the desired accuracy. In order to achieve this desired accuracy, the tissue heterogeneities need to be considered.

In modern day treatment planning systems, a CT scan is used to obtain the tissue density information. In treatment planning systems, Hounsfield units (air ~ -1000, water ~ 0, bone ~ 800) also known as CT numbers are calibrated with the help of CT scans of known density materials. Since Compton scattering is dominant for the photon energy range used in radiation therapy, electron density mapping is used for calibration instead of the physical density of the tissues. It has been reported\(^4\) that a change of 20 Hounsfield units for soft tissue and a change of 250 Hounsfield units for bone results in less than 1% change in the calculated dose for brain and less than 2% change in
calculated dose for pelvis and lungs area. Therefore it is reasonable to believe that the small variation in Hounsfield units will have negligible effect on accuracy of dose calculation.

Radiation dose calculation techniques are verified by dose measurements performed in water phantoms. The amount of radiation delivered to the tumor to produce certain biological effects is measured in terms of dose. It is defined as energy deposited per unit mass. The SI unit of dose is Gray and 1Gy = 1 Joule/Kg. Generally the time of machine to deliver required dose at point of interest can be calculated as

\[
\text{Time to Deliver Required Dose} = \frac{\text{Prescribed Dose (Gy)}}{\text{Machine Output} \times \frac{\text{Gy}}{\text{Time}} \times \text{ICF}}. \quad \text{Equation 2-9}
\]

where ICF is the Inhomogeneity Correction Factor that accounts for tissue density variation above the calculation point. The machine output is calibrated by measurements in water at a particular depth and field size, and is adjusted for the difference if the tumor depth and the treatment field size are different than the calibration depth and field size. If accessories such as wedges and trays are used in the path of the beam, the treatment delivery time must be adjusted due to their attenuation of the beam.

There are two ways to incorporate the inhomogeneity effects of tissues in the dose calculation: indirectly and directly. In indirect methods correction factors are calculated separately and then are used in the dose calculation; whereas, in direct methods the inhomogeneity correction is inherent in the dose calculation algorithm. Dose calculation methods can be divided into three categories; classical algorithms, superposition-convolution algorithms and Monte Carlo (MC) dose calculation method. Classical
algorithms are indirect; the inhomogeneity correction factor is calculated separately and then is incorporated in dose calculations. Superposition-convolution algorithms and Monte Carlo (MC) dose calculation methods are direct, so that the inhomogeneity correction is inherent in the calculation.

### 2.2.1.1 Classical Algorithms

These older algorithms calculate inhomogeneity correction factors using empirical methods. A few of them are briefly discussed here.

#### 2.2.1.1.1 Ratio of Tissue Air Ratio (RTAR)

Tissue air ratio, TAR, is defined as the ratio of the dose in the tissue, or the tissue-equivalent material, called a phantom, $D_{\text{Phantom}}$, to the dose at the same point and depth in air, $D_{\text{air}}$,

$$TAR = \frac{D_{\text{phantom}}}{D_{\text{air}}}. \quad \text{Equation 2-10}$$

The dose ratio will be different for different beam energies, field sizes, and materials. In the RTAR method, the inhomogeneity correction factor is calculated by considering the density variation in the direction of primary x-ray beam. The correction factor is calculated as

$$ICF = \frac{TAR(d')}{TAR(d)}, \quad \text{Equation 2-11}$$
where \( d \) is the density depth of calculation point in water, \( d = \rho \cdot z \), where \( \rho \) is density of water and \( z \) is the physical depth of dose calculation point, and \( d' \) is the density depth of calculation point for various densities above the calculation point,

\[
d' = \sum \rho_i z_i. \tag{2-12}
\]

This method takes into account only the dose variation along the beam direction to the calculation point. Density variation in the lateral direction and electron transport subsequent to primary photon interaction is not considered in this method.

### 2.2.1.1.2 Batho Method

Batho proposed an empirical relation to calculate the tissue inhomogeneity correction factor in 1964. This method was later on modified by Sontage and Cunningham\(^5\). The correction factor in the modified Batho method at point \( p \) in a tissue with density \( \rho_3 \) with overlying tissue of different densities \( \rho_1 \) and \( \rho_2 \), as shown in Figure 2-1, is

\[
ICF(d, r) = \frac{\left[ TAR(d_3) \right]^{\rho_3 - \rho_2}}{\left[ TAR(d_2 + d_3) \right]^{\rho_1 - \rho_2}}. \tag{2-13}
\]

![Figure 2-1 Inhomogeneity correction factor calculated at point P with overlying different densities.](image)
2.2.1.1.3 Equivalent Tissue Air Ratio (ETAR) Method

This method was an extension of the ratio of tissue air ratio (RTAR) method. In the ETAR method, the lateral direction along the field size is also scaled with respect to tissue density variation. In this method, the inhomogeneity correction factor (ICF) is calculated as

$$ICF = \frac{TAR(d', \tilde{r})}{TAR(d, r)}$$  \hspace{1cm} \text{Equation 2-14}

where

$$\tilde{r} = r . \rho'$$  \hspace{1cm} \text{Equation 2-15}

and

$$\rho' = \frac{\sum \sum \sum \rho_{ijk} W_{ijk}}{\sum \sum \sum W_{ijk}}.$$  \hspace{1cm} \text{Equation 2-16}

Here depth and field sizes are scaled with the tissue density variation and CT data is used to calculate $d'$ and $\tilde{r}$. This was the first 3D dose calculation method which used the CT data to consider the tissue density variation in the calculation. This method was a significant improvement over the existing algorithms and was adopted by many 3D treatment planning systems.

The limitations of RTAR, Batho and ETAR methods have been studied by several authors.\textsuperscript{6,7,8,9,10} The results of these studies show that the RTAR method overestimates the lung corrections by 10% even for larger field sizes. The calculations performed by these methods for small field size irradiations in the lungs area deviate by 10% to 20% from the measurements. For more challenging cases such as simulating the patient...
maxillary sinus region, even dose calculations performed by Batho and ETAR method can deviate by 20% to 70% from the measurements.

### 2.2.1.2 Convolution/Superposition Dose Calculation Algorithms

In convolution/superposition dose calculation methods, the beam is modeled in the treatment planning system and then dose is calculated from first principals. The primary beam energy fluence distribution is modeled as a two-dimensional array. The modeled incident energy fluence $\psi(\vec{r}')$ is projected through the patient CT data and the attenuation of the beam is considered using the mass attenuation coefficients $\frac{\mu}{\rho}(\vec{r}')$. These coefficients are obtained from a stored three dimensional lookup table. The product of mass attenuation coefficients and energy fluence is called TERMA, the total energy released per unit mass. The distribution of energy subsequent to photon interaction is modeled separately in terms of the energy deposition kernels. Energy deposition kernels are the mathematical description of the scattered energy distribution subsequent to the primary photon interaction. The energy deposition kernels are convolved with the product of mass attenuation coefficients and energy fluence (TERMA) to find the dose at the calculation voxel,

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

Equation 2-17

where $\frac{\mu}{\rho}$ is mass attenuation coefficient, $\psi$ is primary energy fluence and $K$ is the convolution kernel. The convolution kernels are calculated using a Monte Carlo method. These algorithms are able to calculate the dose directly inside the
heterogeneous tissues by scaling the energy deposition kernels and the TERMA distribution according to the mean electron density between the interaction point and calculation point. If the scatter contribution is computed from the entire kernel distribution then this method becomes computationally too cumbersome to be implemented clinically. In order to reduce the computation burden, a collapsed cone convolution algorithm was proposed. In this method the scattered energy distribution kernel is replaced by angular discrete energy channels in the shape of cones and the energy released is ray traced along these channels. Density scaling is applied along these cones in the dose computation.

These methods show much better accuracy than classical algorithms and are industry standards at this time, however these algorithms have limitations when used in the situation of severe charge particle disequilibrium e.g. at tissue interfaces with high beam energy used for small field sizes.

2.2.1.3 Monte Carlo Dose Calculation

In Monte Carlo (MC) methods, probability distributions governing physical events are statistically sampled to predict the average behavior of quantities of interest. MC techniques are widely used in coupled (photon-electron) radiation transport problems such as dose calculations. The main input for MC codes are the probability distribution functions of particle interactions, and a description of what quantities of interest are to be tallied. There are four steps in analog (event-by-event) MC simulation. In the first step, distance to the first interaction is calculated based on a random number and known total interaction probability. In the second step, the particle is transported to the first interaction site by simple ray tracing through the geometric model with known density...
and constraints. In the third step, the interaction type is decided. In the fourth step, the interaction is simulated. For example, if only photoelectric (PE) and Compton scattering (CS) effects are considered then the total interaction probability can be written as

$$\mu_{total} = \mu_{PE} + \mu_{CS}.$$  \textbf{Equation 2-18}

After generating a random number $R_1$, the distance “$x$” to the first interaction is taken as

$$x = -\ln(R_1) \cdot \frac{1}{\mu_{total}}.$$ \textbf{Equation 2-19}

Now a second random number $R_2$ is chosen to see if the interaction is PE or CS. If $R_2 < \frac{\mu_{CS}}{\mu_{total}}$ then it is Compton scattering, otherwise it is photoelectric effect. If any secondary particle is produced as a result of interaction, it is banked for further simulation and the primary particle is followed until it is absorbed or its energy is reduced below a given value called cut-off energy level. Energy cut-off values are defined in the simulation to reduce the computational times. Particles with energies below these cut-off values are assumed to deposit all their energy at their location.

Photons undergo a smaller number of interactions than electrons before they are absorbed and have larger mean free paths between interactions. Direct simulation of photon interactions is easy but due to larger mean free path, photons deposit their energy over large volumes which necessitate a large number of photon histories to get acceptable statistics. Electrons undergo large number of Coulomb interactions before complete absorption of their energy. The complete simulation of all these interactions is very complex. To overcome this problem, the idea of “condensed” history scheme was
In this technique many “microscopic” interactions are classified into a “macroscopic” group representing the details of particle transport. The cumulative effect of many “micro-interactions” is taken into account by sampling energy, direction and position changes from appropriate distributions (multiple scattering theories) into a single “macroscopic” interaction. The medium is considered locally homogeneous in the microscopic events making it easy to study the transport of particles. When using condensed history schemes, the electron path is divided into small steps. The choice of optimum step size is critical so that multiple-scattering theories remain valid. Step size choice affects the accuracy of calculations. Condensed history scheme techniques made the MC method practical but also introduce this “step-size” artifact in the calculations. High accuracy condensed history schemes have been developed and implemented in most of the general purpose Monte Carlo codes.

The computational time of MC simulation can be reduced by employing different variance reduction techniques. Energy cut-offs and geometrical constraints were used in this thesis.

### 2.3 MCNP5

MCNP (Monte Carlo N-Particle Transport Code (MCNP)) is a general purpose Monte Carlo code commonly used for simulating neutron, electron, photon or coupled neutron/photon/electron transport through any media. It was originally developed at Los Alamos National Laboratory to study neutron transport. It offers a combinatorial geometry package that enables complex problems to be modeled. It has the ability to construct a variety of surfaces which can be combined to enclose unique volumes called cells. Each cell can have its own material and density. Analytical phantoms can be
constructed with the combination of planes, ellipsoids, cones and user-defined surfaces. Duplication, translation and rotation of cells can be done by using special features of MCNP to reduce the complexity of setup.\textsuperscript{19}

2.3.1 MCNP5 Geometry

The input file for MCNP5 has three parts. In the first part of the file, cells are defined in terms of different surfaces. In the second part, those surfaces are defined. In the third part quantities of interest to be output are defined (surface flux etc.). The format of the input file for MCNP5 can be summarized as\textsuperscript{19}

\textit{Title Card}

\textit{Cell Cards}

\textit{Blank Line Delimiter}

\textit{Surface Cards}

\textit{Blank Line Delimiter}

\textit{Data Cards}

\textit{Blank Line Terminator (optional)}

The input format of MCNP5 can be understood from an example problem, a 4-MeV isotropic source of photons starting from the center of a 1 cm radius sphere of oxygen that is embedded in a cube of carbon, 10 cm on each side. The desired output is photon flux across the surface of the sphere. Comments on the input file can be written following “c” in start of any line or following the “$” after any input command.
2.3.1.1 Cells

MCNP5 models the geometry of the problem by defining closed 3D volumes called “cells” with the Boolean combination of surfaces. Each cell is defined by a cell number, material number, and material density followed by combination of operators and signed surfaces that bound the cell. Surfaces defining the cell boundaries have positive (+) or negative (-) signs giving the positive or negative sense to define whether the cell volume is on the right or left, up or down side of the surface. A zero (0) can be used to define the material number and material density for a vacuum cell. The cell number must be written in columns 1−5 of the input text file. The entries defining the cell are separated by blanks. The cells are defined by the intersection, union, and/or complement of the listed surfaces.

C cell cards for sample problem

C format is cell #, material #, density, surface numbers defining the cell

1 1 -0.0014 -7

C Cell 1 contains material 1 with density 0.0014 gm/cm³. The negative sign in front of density indicates mass density. Cell 1 is inside surface 7 (sphere); the negative sign in front of surface 7 means cell 1 is inside of the sphere.

2 2 -1.60 +1 -2 -3 +4 -5 +6 +7

C Cell 2, containing material 2 with density 1.6 gm/cm³, is a cube defined by intersection of six surfaces, and the outside of the sphere (+7). A space between surface numbers means that the cell is the intersection of the surfaces.

3 0 -1:2:3:-4:5:-6
C Cell 3 is an empty cell with no material and is everything outside of the cube defined by union of six surfaces. A colon sign between surface numbers means that the cell is the union of the surfaces.

C end of cell cards for sample problem

[Blank line delimiter]

\textbf{2.3.1.2 Surfaces}

The surfaces are represented by mnemonics such as C/X for a cylinder parallel to the \textit{x}-axis. Plane, sphere, cone, cylinder, ellipsoid, hyperboloid, paraboloid, elliptical or circular torus surfaces can be defined in MCNP5. Surfaces are defined on surface cards with the help of coefficients satisfying the surface equations.

C Beginning of surfaces for cube

1 PZ -5 $ Plane perpendicular to \textit{z}-axis, passing through \textit{z} = -5

2 PZ 5 $ Plane perpendicular to \textit{z}-axis, passing through \textit{z} = 5

3 PY 5 $ Plane perpendicular to \textit{y}-axis, passing through \textit{y} = 5

4 PY -5 $ Plane perpendicular to \textit{y}-axis, passing through \textit{y} = -5

5 PX 5 $ Plane perpendicular to \textit{x}-axis, passing through \textit{x} = 5

6 PX -5 $ Plane perpendicular to \textit{x}-axis, passing through \textit{x} = -5

7 S 0 0 0 1 $ Oxygen sphere at origin (0, 0, 0) with radius 1 cm

C End of surfaces

[Blank line delimiter]
2.3.1.3 *Data Cards*

In this section, the mode of simulation, material specification for cells, source description, desired output and variance reduction techniques are defined. Several simulation modes are available in MCNP5, for example transport mode MODE: N simulates neutron transport only. MODE: P E simulating photon and electron was used in this study.

MODE: P E $\$ Photons and electrons will be considered

The importance of particle transport for different cells can also be defined in data cards with mnemonic Imp: p. For example, if Imp: p is defined as zero for any cell then the photon particle transport will be terminated in that cell. Energy cut-offs and histories to be simulated are also defined on data cards.\textsuperscript{20}

IMP:P, E 1 1 0

C Photons and electrons are tracked inside first two cells and ignored in third, outermost cell

CUT:E 0.050

CUT:P 0.010

C Electrons with energy below 0.05 MeV and photons below 0.01 MeV are assumed to be absorbed, and to deposit their energy at their current location

The source is defined using SDEF card. Different types of source definitions are possible in MCNP5, including planar, point-isotropic, etc.\textsuperscript{20}. Parameters on the SDEF card include starting position, source cell or surface, particle type, source energy, etc.
C Source

SDEF POS = 0 0 0 Par = 2 Erg = 4

C Monoenergetic isotropic photon source (Par = 2) defined at origin (Pos (0 0 0)) with energy 4 MeV.

The desired output is defined on different tally cards with the help of mnemonics such as F2: P etc. F2: P gives the photon surface flux. Several tallies are available in MCNP5 such as surface flux (F2:P, F2:E, F2:N), cell flux (F4:P, F4:E, F4:N), flux at a point or ring detector (F5a:P, F5a:N), and pulse height (F8:P, F8:E, F8:P,E, F8:N) tallies. Neutral particle flux image tallies are also available in MCNP5. These tallies are flux image radiograph (FIR, a flux image radiograph on a planar image surface), flux image on a cylinder (FIC, a flux image on a cylindrical image surface) and flux image by pinhole (FIP, a flux image by pinhole on a planar image surface) tallies.¹⁹

C Output Tally

F2:P 7 $ Desired output is the flux across surface 7

C Material Cards

M1 8016 1 $ Oxygen (100%)

M2 6000 1 $ Carbon (100%)

The material card is defined by mnemonic M followed by the number of material (used in defining the cell) and its identification. Oxygen in the sample input deck discussed above was defined as M1 8016 1. It was material 1 with 100% oxygen. Composition is written as 1000Z C, where Z is the atom number and C the fraction. For example, water could be specified as
Here 8 represents material no. 8 and then 1000 represents hydrogen and 0.66 represents its fraction, 8000 represents oxygen and 0.33 represents its fraction.

NPS 100000 $ Number of particles to be simulated

After completing the simulation, MCNP5 code automatically creates the summary of results which gives insight into the physics of the problem and the statistics of the results.

### 2.3.2 Photon Interactions in MCNP5

MCNP5 has two physics modes for photon interactions. In both physics treatments, photoelectric effect, pair production and Compton scattering from free electrons are considered. In simple physics mode, both coherent scattering (Thomson) and fluorescent photon production followed by photoelectric absorption are ignored. In both physics modes, when Mode: P E is used, all photon collisions except coherent scatter can produce electrons which are banked for later transport. When using Mode: P, the thick-target bremsstrahlung model (TTB) is used for bremsstrahlung photons production but the electrons are locally slowed to rest. In this mode, electron-induced photons are not neglected but the electron transport step is not considered in the simulation. Electron transport and electron-induced photon production can be turned off optionally.¹⁹
2.3.3 Electron Interactions in MCNP5

In the transport of electrons, MCNP5 considers angular deflection through multiple Coulomb scattering, collisional energy loss with optional straggling, and the production of secondary particles. The secondary particles considered in the transport are K x-rays, knock-on and Auger electrons and bremsstrahlung photons.

Due to the complex nature of electron interactions, considerable theoretical work has been done and multiple scattering theories were proposed to make the MC simulation feasible for electrons. These theories predict the probability distributions of the quantities of interest like energy loss and angular deflections using the fundamental interaction cross sections and the statistical nature of the transport process.

Algorithms based on condensed history schemes are divided into two classes; class I and class II. Historically energy indexing of the electron transport in MCNP was based on the class I algorithm. In this algorithm, all electron collisions are simulated in a predetermined energy grid. When using class I algorithms, interpolations are needed when the real electron steps do not equal any of the grid energies. Another disadvantage is that there is no correlation between energy loss and direction change of the primary particle with the production of secondary particles. In class II condensed history algorithms, hard and soft electron collisions are treated differently. The hard collisions are simulated event by event like an analogue simulation whereas the soft collisions are treated using class I algorithms.

To improve statistics the production of bremsstrahlung photons can be increased in MCNP5 by a parameter BNUM used on PHYS:E card. A BNUM value of 100 was used in the simulation of 6 MV photon beam in this study, whereas BNUM = 1 was used
in case of 18 MV photon beam simulation. When BNUM > 0 is used, the cross-section for bremsstrahlung photon production is multiplied by that factor. The energy and angle of these photons is sampled independently but their weight is reduced by the factor. The energy loss of the electron is reduced only by the energy of the first sampled bremsstrahlung photon.

The version of MCNP5 used on this study (MCNP5 version 1.51) offers three optional electron transport models including the class I and class II algorithms along with a newly introduced algorithms based on detailed Landau energy straggling logic. The Default algorithm, class I, was used in this work.
Reference:


3 Simulation using MCNP5

A full Monte Carlo (MC) simulation of the 6 and 18 MV photon beams from a Varian linear accelerator was performed in this dissertation. For the MC simulation physical and geometrical characteristics of the components of the linear accelerator head are required including position, dimension, shape, density, atomic number and composition of the components of the linear accelerator head. These parameters were obtained from Varian Inc. after signing a confidentiality agreement and all this information is proprietary.

In the Monte Carlo code, the simulated particle’s energy, position and velocity can be saved at any plane, in a binary file called a phase space file. In full MC simulation methods, these phase space files are used as an input for the dose calculation. Phase space files generated by the full MC simulation can further be used for beam characterization to construct histogram-based and analytical models. These methods require significantly smaller storage space compared to the full simulation approach but the full MC simulation method is the most accurate for dose calculation in regions of high dose gradients and electronic disequilibrium. The full MC simulation performed in this dissertation was used for dose calculation in the surface region where the dose gradient is high. The surface region calculations and measurements are discussed in chapter 4. Steps regard to benchmark the simulation to measurement are discussed in this chapter.
3.1 Benchmarking approach

Different benchmarking procedures can be used to simulate megavoltage photon beams. The absorbed depth profile along the beam, generally termed the percent depth doses (PDDs), i.e. the absorbed dose in a given volume in the center of the field, as a function of the depth of the volume element, and cross-dose profiles, the absorbed dose in a given volume as a function of distance along lateral direction, are the most commonly used measured quantities to benchmark the calculations. A few authors have also suggested using in-air off-axis factors and dose profile measurements in the penumbral region to benchmark the model. These methods require dedicated in-air and penumbral measurement tools. The penumbra is the region at the edges of the radiation field where dose changes rapidly as a function of distance from the central axis. Penumbra is primarily caused by the finite source size, but the phantom/patient scatter at the edges of the field also contributes. Instead of using penumbral region measurements and in-air off-axis measurements, percent depth dose measurements and dose profile measurements for 5×5, 10×10 and 30×30 cm² field sizes were used to benchmark the model in this dissertation.

Percent depth dose and dose profile simulations are sensitive to input parameters such as the incident electron beam energy. The exact incident electron beam energy is unknown for a particular accelerator as a setting of 6 MV may typically range from 5.8 to 6.2 MeV. In a recent study, the sensitivity of PDDs and profiles were reinvestigated by considering four parameters (mean energy, energy spectral broadening, radius, and the angular divergence (θ) of primary electrons) of the incident electron beam for modeling 4, 6, 10, 15 and 18 MV photon beams. The angular divergence was shown to have no
effect in simulating the 6 and 18 MV photon beams. In this dissertation, three parameters (mean incident electron energy, Gaussian width of the energy spectrum, and the incident electron beam radius) of the incident electron beam were considered as fitting parameters in the simulation.

### 3.1.1 Measurements

Percent depth doses (PDDs) and dose profiles were measured using a PTW Type 31013 Semiflex ion chamber. This is a cylindrical chamber with an inner diameter of 5.5 mm and sensitive volume of 0.125 cm$^3$. A PTW 3D water tank was used for these measurements. The water tank and ion chambers are shown in Figure 3-1 and Figure 3-2.

![Figure 3-1 Water phantom used for measurements](image1)

![Figure 3-2 Ionization chambers used for measurements](image2)
The ion chamber was connected to an electrometer (Kiethley 35040) and its motion was controlled by PTW MEPHYSTO (Medical Physics Tools) software. This is a Microsoft Windows based software package, which is used for the control movement of the ion chamber and for data processing. The alignment of both the gantry and the water phantom was verified. The phantom tank was filled with water up to a marker, and then the leveling of water tank was done by matching the water surface with the help of level markers on the inside of water tank. With the help of reference markers in the phantom, the ion chamber was positioned at reference point. Then the chamber was positioned to the isocenter of the accelerator with the help of orthogonal lasers and a cross hair. The isocenter position of the chamber was called the null point. It was observed that the measurement probe ran straight and level through the water phantom during measurements. PDD and dose profiles were measured for 5×5, 10×10 and 30×30 cm² field sizes.

Measurement uncertainty for PDDs and dose profiles include uncertainty in the measured charge, the positioning inaccuracy of the chamber, short-term fluctuations of chamber, electrometer, air pressure, and temperature during measurement. The literature associates these with an overall experimental uncertainty of 2.5% which was used in the plots in this dissertation\textsuperscript{15}.

3.2 Simulation Geometry

The simulated components of the linear accelerator head are shown in Figure 3-3 and include the target, primary collimator, beam flattening filter, monitor chamber, shielding and adjustable independent x and y jaws (secondary collimators). The
geometric components of the linear accelerator head are same for 6 and 18 MV photon beams except the beam flattening filter (FF) is thicker in the center for the 18 MV beam than the 6 MV photon beam flattening filter. When the linear accelerator is switched from 6 to 18 MV, the flattening filters are replaced automatically.

![Figure 3-3 Schematic diagram of simulated components of linear accelerator](image-url)
The whole simulation process was divided into three steps. In the first step, the target, primary collimator, flattening filter and monitor chamber were simulated and the particles transported through these components to a plane after the monitor chamber and stored in the first phase space file. The first geometric component in the linear accelerator head is the target. The accelerated electron beam hits the target to produce the bremsstrahlung photons. High Z material is used in the target to increase the bremsstrahlung photon production. The target was simulated as a cylindrical slab in MCNP5. The target is surrounded by a shield known as a primary collimator. The primary collimator is conical in shape and is used to collimate the beam towards the area of the interest. The thickness and material depends on the maximum possible energy of the linear accelerator. The primary collimator was simulated in MCNP5 as a volume between pair of concentric cones truncated with planar surfaces from the top and the bottom. The bremsstrahlung photon beam obtained from the target is non-uniform in intensity and a flattening filter (FF) which is thicker in the center is used to give differential attenuation to achieve homogeneous intensity over the field widths used for treatment. The flattening filter is necessary because the edges of the field are farther from the source than the center of the field, and hence would be lower intensity. A flattening filter for an ideal point source corrects for the decrease in intensity at the edges of the field due to the inverse square law as the edges are further from the source than the center. The filter is thicker in the center and so reduces the intensity at the center of the field. However, the source is not an ideal point, and the beam size affects the beam profile. For example, if the source were very large, comparable in size to the field, the intensity without the filter would be uniform. In that extreme case, the filter would
greatly overcorrect in the center, so that the intensity would be highest at the edges and dip near the beam axis. For calculations for a range of probable real source sizes, the on-axis intensity will tend to exhibit a dip near the center as the source size is increased. The geometry of the flattening filter was provided as a curved shape specified with two dozen coordinates. This was simulated using a piecewise linear fit as series of truncated cones lying on top of each other covered with two planar surfaces perpendicular to z axis. The flattening filter for 18 MV photon beam has two parts; top and the bottom. Both parts were simulated with the help of multiple cones lying on top of each other with variable radii and a common vertex on the z axis. Again, the cones were truncated with two planar surfaces perpendicular to the z axis. A dual sealed monitor ion chamber was mounted just below the flattening filter for beam control. Sealed chambers are used for monitors as their response is independent of ambient temperature and pressure. Monitor chambers are designed such that they have a minimal effect on photon and electron radiation beams and can operate under saturation conditions. They are used as control system, turning off the beam if the radiation dose exceeds the prescribed dose. The monitor chamber measures the dose in arbitrary units called “monitor units” (MU). The monitor chamber sensitivity is adjusted such that 1 MU corresponds to a dose of 1 cGy in a phantom under reference conditions. The monitor chamber was simulated as a set of rectangular blocks. Since the target, the primary collimator, the flattening filter and the monitor chamber do not affect the treatment field size, they were simulated once and the first phase space file was used as input for all three field sizes.

Independent jaws and the shielding between the jaws and the flattening filter were simulated in step 2. Shielding was simulated similar to the primary collimator as a
volume between concentric truncated cones. After the shielding, independent jaws, also known as the secondary collimators were simulated. Independent jaws (x jaws and y jaws) are used to define the required treatment field size. Since they all can move independent of each other, they can define symmetric as well as asymmetric field sizes. Each of the four jaws was simulated as a union of six planar surfaces defined using coordinates of their positions relative to the central ray. Three field sizes were used in the simulation, with different symmetric positions of the jaws. Three phase space files were saved at the top of water phantom surface for each beam energy.

In the third step, a cubical water phantom (x =60 cm, y=60 cm and z=60 cm) was simulated and the percent depth dose and the cross dose profiles were calculated in separate runs. Circular voxels with dimension similar to the experimental detector were used for dose calculation for all three field sizes. Phase space files generated in step 2 were used as input for dose calculation in this step.

3.2.1 Simulation Efficiency

There are various tools available in MCNP5 to increase calculation efficiency. One of them is the energy cut off method. Electron and photon energy cut offs are available. When particle energy falls to less than the cut off energy level, the entire kinetic energy is deposited immediately. The range of travel before absorption in water of electrons with kinetic energy lower than 189 KeV (E ~ 700 KeV) and photons with kinetic energy lower than 10 KeV is a fraction of a millimeter so little accuracy is lost in assuming that they deposit at the location at which their energy falls below these values. These values are used to increase computational efficiency while preserving accuracy. There are two
types of energy cutoffs available in MCNP5; the global energy cutoff and the local
energy cutoff. Global energy cutoffs are followed in all geometric parts of the simulation
whereas a local energy cutoff can be employed in any geometric component of the
simulation. Electrons were followed to ECUT=189 KeV and photons to PCUT=10 KeV
in the whole simulation. The effect of a local electron energy cutoff (ELPT) was tested
using ELPT:E= 500 KeV in the independent jaws only. It increases computational
efficiency by factor of 5 with no effect on accuracy of calculations.

The computational time can further be reduced if the number of bremsstrahlung
photons generated per incident electron on the target is increased. Bremsstrahlung
photon production is controlled by a BNUM value on the PHYS card, which is used for
electron transport in the MCNP5 input file. Bremsstrahlung production can be turned
off in the simulation by using BNUM=0. A number of bremsstrahlung photons equal to
that given by the physical cross-section are sampled when BNUM=1 is used (the default
value in MCNP5). The usual bremsstrahlung photon production rate is multiplied by the
BNUM value when BNUM>0 is used in the simulation, and then the corresponding
weight reduction is applied to the photon when calculating dose. The energy loss of the
electron which generated the bremsstrahlung photon is computed from the first photon,
regardless of BNUM. BNUM has a maximum value of 100, which was used in the
simulation. For the 6 MV photon beam simulation, BNUM=100 was found to give
maximum photon production.

When using BNUM=100, number of tracks achieved at the phantom surface were
approximately 100 times more than running the simulation without BNUM for the 6 MV
photon beam. The simulation time per electron was increased by factor of 33 when
BNUM=100 was used for the same number of initial electrons resulting in a net decrease of about a factor of 3 in calculation time per photon. Approximately 52 million electrons were simulated for 6 MV photon beam using BNUM=100 which gave around 50 million photon tracks at phantom surface for 10×10 cm² field size. It took around 60 hours to run the simulation on a system with 3.40 GHz Intel processor and 8.00 GB installed memory. Photon production is increased by using BNUM value, but it also increases the simulation time. For the 18 MV photon beam, the tradeoff between increased time per electron and increased number of photons did not increase computational efficiency, so the default value (BNUM=1) was used in the simulation. For the 18 MV photon beam approximately 5 billion initial electrons were simulated which gave around 50 million number of tracks at phantom surface. It took around 80 hours to run the simulation for 18 MV photon beam on a system with 3.40 GHz Intel processor and 8.00 GB installed memory.

3.3 Modeling for the 6 MV photon beam

3.3.1 PDD Comparison for 6 MV photon beam

When megavoltage x rays hit the phantom surface they produce charged particles, mostly the electrons. These charge particles deposit their energy at a finite distance from their birth place. This process causes the initial buildup of the dose to a maximum value and the region before the depth of maximum dose ($d_{\text{max}}$) is called the buildup region. Large discrepancies can be observed in the buildup region ($d < 1.5$ cm for 6 MV and $d < 3.3$ cm for 18 MV) and are attributed to both calculation and measurement issues in this region.
Dose calculation and measurement beyond $d_{\text{max}}$ will be discussed in this chapter and the comparison in the buildup region ($< 1.5$ cm) will be discussed in the next chapter.

The electron beam energy was used as a fitting parameter to match the simulations with the measurements, first for the $10 \times 10$ cm$^2$ field size. For these calculations, cylindrical voxels of 2 mm depth and 5 mm radius were used as shown in sketch below.

![Sketch for circular voxels simulated along central axis of the beam for dose calculations.](image)

The energy deposition in each voxel was calculated using the *F8 tally of MCNP5. Large number of electrons ($5 \times 10^9$ electrons for 18 MV) were simulated to obtain the statistical uncertainty of less than 0.5% for all voxels. In order to obtain PDDs, the calculations were normalized with respect to the maximum dose voxel. For PDD matching, the default value of 1 mm for the incident electron beam radius was used in the calculations. The incident electron energy was varied from 5.8 to 6.3 MeV and the resulting dose depth profiles compared to measurements are shown in Figure 3-5 and Figure 3-6. Percent difference was calculated as

$$\text{% Difference} = \frac{(\text{Calculation} - \text{Measurement}) \times 100}{\text{Calculation}},$$  \hspace{1cm} \text{Equation 3-1}

so the positive percent difference means the calculation is systematically high compared to the measurement. The average % difference and RMS % difference for different incident electron beam energies for the $10 \times 10$ cm$^2$ field size is presented in Table 3-1.
For 5.8 and 6.0 MeV, calculations were systematically low and for 6.3 MeV, they were high. The best match was achieved at 6.2 MeV. After matching the PDD for 10×10 cm$^2$ field size, the cross-dose profile matching was performed for the 6.2 MeV beam.

![Figure 3-5 PDD variation with incident electron energy for 6 MV photon beam for 10×10 cm$^2$ field size.](image)

Figure 3-5 PDD variation with incident electron energy for 6 MV photon beam for 10×10 cm$^2$ field size.
Figure 3-6 Detailed PDD variation for range x=1.5-10 cm from Figure 3-4.

Table 3-1 Average % difference and RMS % difference for different energies of the incident electron beam.

<table>
<thead>
<tr>
<th>Energy</th>
<th>Average % Difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5.8 MeV</td>
<td>-5.3</td>
<td>5.9</td>
</tr>
<tr>
<td>6.0 MeV</td>
<td>-3.8</td>
<td>4.0</td>
</tr>
<tr>
<td>6.2 MeV</td>
<td>0.001</td>
<td>0.5</td>
</tr>
<tr>
<td>6.3 MeV</td>
<td>2.1</td>
<td>2.6</td>
</tr>
</tbody>
</table>
3.3.2 Profile Comparison for 6 MV photon beam

For benchmarking to the cross dose profile, the incident electron beam radius was considered the fitting parameter. For calculations, a circular cylinder in the lateral direction was simulated and divided into voxels of 4 mm depth in the lateral direction with 3.5 mm radius for each voxel.

Energy deposition in each voxel was calculated using the *F8 tally of MCNP5. Calculations were normalized with respect to the voxel at the center of the beam axis. The optimum energy found in the PDD comparison (6.2 MeV) was used in the dose profile calculations. For comparison, measured beam profiles were divided into three regions; flat region, field edges (penumbral region) and out-of-field region. The flat region is the region where the radiation dose is within 80% of the central dose. The penumbral region is at the edges of the field where dose falls off as a function of distance from the central axis and is between 20-80 % of the central value. The percent difference should be within 2% in the flat region, and within 10% at the field edges. It can be more than 10% for the out-of-field region.

For profile matching, the calculations were performed for the incident electron beam with Gaussian and uniform radii. Dose profile calculations for the incident electron beam with FWHM of 1.3, 1.7, 1.9 mm were performed for 30×30 cm² field size at 1.5 cm depth.

![Figure 3-7 Sketch for circular voxels in lateral direction for dose profile calculations.](image)
and comparison with measurement is shown in Figure 3-8. The local percent difference from measurements for these calculations is presented in Figure 3-9 through Figure 3-11. The calculations for the incident electron beam with FWHM of 0.91 mm were performed for the 10×10 cm² field size at 10 cm depth and the comparison with measurement is shown in Figure 3-12. The local percent difference for the incident electron beam with FWHM of 0.91 mm is shown in Figure 3-13. Calculations for the incident electron beam with uniform radii of 1, 1.3 and 1.5 mm were performed for 30×30 cm² field size at 10 cm depth and are presented in Figure 3-14 along with measurements. Table 3-2 represents the average % difference and RMS % difference for all calculations. Calculations were systematically high for the incident electron beam with FWHM of 1.3, 1.7 and 1.9 mm and were low for the incident beam radius with FWHM of 0.9 mm. When considering the incident electron beam with uniform radii, the calculations were systematically low for 1 mm and high for 1.5 mm. As expected, as the source size is increased, the intensity without the flattening filter becomes more uniform, so that the flattening filter overcorrects the profile and a dip is seen near the beam axis at the center of the field. The best match was achieved for the incident electron beam with uniform radius of 1.3 mm.
Figure 3-8 Dose-Profile comparison for the incident electron beam with different Gaussian radii for 30×30 cm² field size at 1.5 cm depth.

Figure 3-9 Local percent difference for the incident electron beam with FWHM = 1.3 mm. Average % difference was 0.75 and RMS % difference was 1.2. Calculations were systematically high.
Figure 3-10 Local percent difference for FWHM = 1.7 mm. Average % difference was 0.92 and RMS % difference was 1.19. Calculations were systematically high.

Figure 3-11 Local percent difference for FWHM = 1.9 mm. Average % difference was 1.02 and RMS % difference was 1.36. Calculations were systematically high.
Figure 3-12 Dose-Profile comparison for the incident electron beam with FWHM=0.9 mm for 10×10 cm$^2$ field size at 10 cm depth.

Figure 3-13 Local percent difference between measurement and calculation at 10 cm depth for 10×10 cm$^2$ field size for incident electron beam with FWHM=0.9 mm. Average % difference = -2.0, RMS % difference = 2.5. Calculations were systematically low.
Figure 3-14 Dose-Profile variation with incident electron beam with different values of uniform radii for 30×30 cm² field size at 10 cm depth.

Figure 3-15 Local percent difference for incident electron beam with radius 1 mm. Average % difference was -1.13 and RMS % difference was 1.40. Calculations were systematically low.
Figure 3-16 Local percent difference for incident electron beam with radius 1.3 mm. Average % difference was -0.03 and RMS % difference was 0.84. Calculations were within 2% of the measurements.

Figure 3-17 Local percent difference for incident electron beam with radius 1.5 mm. Average % difference was 0.24 and RMS % difference was 1.21. Calculations were systematically high.
Table 3-2 Average % difference and RMS % difference for Gaussian and uniform radii for the incident electron beam for the dose profile calculation for different field sizes at different calculation depths.

<table>
<thead>
<tr>
<th>Radius of the incident e-beam</th>
<th>Field Size</th>
<th>Depth</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.9 mm (Gaussian)</td>
<td>10×10 cm²</td>
<td>10 cm</td>
<td>-2.0</td>
<td>2.5</td>
</tr>
<tr>
<td>1.3 mm (Gaussian)</td>
<td>30×30 cm²</td>
<td>1.5 cm</td>
<td>0.7</td>
<td>1.7</td>
</tr>
<tr>
<td>1.7 mm (Gaussian)</td>
<td>30×30 cm²</td>
<td>1.5 cm</td>
<td>0.9</td>
<td>1.2</td>
</tr>
<tr>
<td>1.9 mm (Gaussian)</td>
<td>30×30 cm²</td>
<td>1.5 cm</td>
<td>1.0</td>
<td>1.4</td>
</tr>
<tr>
<td>1 mm (Uniform)</td>
<td>30×30 cm²</td>
<td>10 cm</td>
<td>-1.1</td>
<td>1.4</td>
</tr>
<tr>
<td>1.3 mm (Uniform)</td>
<td>30×30 cm²</td>
<td>10 cm</td>
<td>-0.03</td>
<td>0.8</td>
</tr>
<tr>
<td>1.5 mm (Uniform)</td>
<td>30×30 cm²</td>
<td>10 cm</td>
<td>0.2</td>
<td>1.2</td>
</tr>
</tbody>
</table>

### 3.3.3 Final Comparison

After matching the PDD calculations for 10×10 cm² field size and profile calculations for 30×30 cm² field size at 10 cm depth, the PDDs and dose profiles were calculated for all field sizes and at both depths (1.5 cm and 10 cm) using the optimized energy (6.2 MeV) and the optimized radius (uniform r=1.3 mm) for the incident electron beam. Comparison plots with the local percent differences next to each plot are shown in Figure 3-18 through Figure 3-29. The average % difference and RMS % difference of the comparison of PDD calculations performed for the incident electron beam with
energy 6.2 MeV is shown in Table 3-3 for all three field sizes. If we look at the Local percent difference for all three field sizes (Figure 3-19, Figure 3-21 and Figure 3-23), the spread increases beyond the depth of 20 cm due to poorer statistics as the beam is attenuated while passing through the medium. Average % difference and RMS % difference for dose profile comparisons is presented in Table 3-4 and Table 3-5.

Calculations and measurements for PDDs beyond $d_{\text{max}}$ (1.5 cm for 6 MV) agree within 2% for all field sizes and energies. Dose profiles at $d_{\text{max}}$ and at 10 cm agree within 2% in the flat region and within 10% in the penumbra. Slight disagreement between measurements and calculations up to 2% can be attributed to the uncertainty in the manufacturer provided information (material specification, dimension, composition and location of the linear accelerator head components). The optimal primary electron beam parameters for 6 MV photon beam were found to be monoenergetic energy of 6.2 MeV and uniform radius of 1.3 mm. These parameters differed from the optimal parameters reported by Chibani et al.,\textsuperscript{14} (monoenergetic energy of 6.3 MeV and Gaussian radius with FWHM of 0.7 mm) indicating that the beam parameters may be machine specific even if used for the same model from the same vendor.
Figure 3-18 Comparison of measured and calculated PDDs for the 5×5 cm² field size. The incident electron beam energy was 6.2 MeV and the radius was 1.3 mm.

Figure 3-19 Local percent difference between measured and calculated PDDs for 6.2 MeV photon beam for the 5×5 cm² field size. Note that because ~20% of the particles reach at higher depths, the spread becomes ~ 2 times larger.
Figure 3-20 Comparison of measured and calculated PDDs for the \(10 \times 10\) cm\(^2\) field size. The incident electron beam energy was 6.2 MeV and the radius was 1.3 mm.

Figure 3-21 Local percent difference between measured and calculated PDDs for the \(10 \times 10\) cm\(^2\) field size. The spread becomes larger at higher depth due to attenuation of the beam.
Figure 3-22 Comparison of measured and calculated PDD for the 30×30 cm² field size. The incident electron beam energy was 6.2 MeV and the radius was 1.3 mm.

Figure 3-23 Local percent difference between measured and calculated PDDs for the 30×30 cm² field size.
Table 3-3 Average % difference and RMS % difference for the comparison of measured and calculated PDDs for different field sizes at 6.2 MeV energy.

<table>
<thead>
<tr>
<th>Energy</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm²</td>
<td>0.4</td>
<td>0.9</td>
</tr>
<tr>
<td>10×10 cm²</td>
<td>0.005</td>
<td>0.5</td>
</tr>
<tr>
<td>30×30 cm²</td>
<td>0.2</td>
<td>0.5</td>
</tr>
</tbody>
</table>

Figure 3-24 Dose-Profile comparison for the 5×5 cm² field size at d=1.5 cm and d=10 cm depths.
Figure 3-25 Local percent difference between measurements and calculations for the 5×5 cm² field size at d=1.5 (left) and d=10 cm (right) depths.

Figure 3-26 Dose-Profile comparison for the 10×10 cm² field size at d=1.5 and d=10 cm depths.
Figure 3-27 Local percent difference between measurements and calculations for the 10×10 cm² field size at d=1.5 (left) and d=10 cm (right) depths.

Figure 3-28 Dose-Profile comparison for the 30×30 cm² field size at d=1.5 and d=10 cm depths.
Figure 3-29 Local percent difference between measurements and calculations for the 30×30 cm\(^2\) field size at d=1.5 (left) and d=10 cm (right) depths.

Table 3-4 Average % difference and RMS % difference for dose profile comparison at \(d_{\text{max}}\) for 6 MV photon beam for three field sizes.

<table>
<thead>
<tr>
<th>Depth = 1.5 cm</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm(^2)</td>
<td>0.1</td>
<td>0.7</td>
</tr>
<tr>
<td>10×10 cm(^2)</td>
<td>0.05</td>
<td>0.9</td>
</tr>
<tr>
<td>30×30 cm(^2)</td>
<td>0.03</td>
<td>1.0</td>
</tr>
</tbody>
</table>

Table 3-5 Average % difference and RMS % difference for dose profile comparison at 10 cm depth for 6 MV photon beam for three field sizes.

<table>
<thead>
<tr>
<th>Depth = 10 cm</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm(^2)</td>
<td>0.7</td>
<td>0.9</td>
</tr>
<tr>
<td>10×10 cm(^2)</td>
<td>-0.1</td>
<td>0.7</td>
</tr>
<tr>
<td>30×30 cm(^2)</td>
<td>-0.02</td>
<td>0.8</td>
</tr>
</tbody>
</table>
3.4 Modeling for the 18 MV photon beam

The only difference between modeling geometry of 6 and 18 MV photon beams was the flattening filter. The flattening filter used for the 18 MV photon beam is thicker in the center compared to flattening filter of 6 MV photon beam. Once the geometry was modeled, a benchmarking approach similar to 6 MV photon beam was used for the 18 MV photon beam.

3.4.1 PDD Comparison for the 18 MV photon beam

For PDD matching, the incident electron beam energy was initially considered monoenergetic and was varied from 17.8 to 18.2 MeV to get the desired results while keeping the uniform beam radius of 1 mm. Figure 3-30 shows the variation in PDD with incident electron beam energy for the 10×10 cm² field size. The local percent difference between measurements and calculations for all tested energies is shown in Figure 3-31 through Figure 3-32. The average % differences and the RMS % differences for all tested energies are presented in Table 3-6. Calculations were systematically low for the incident electron beam with energy 17.8 MeV. Then the simulation energy was increased to 18.0 MeV. Calculations were still systematically low so the energy of the incident electron beam was raised to 18.2 MeV. Minimum average % difference was achieved for the incident electron beam with energy 18.2 MeV. Local percent difference between calculations and measurements was less than 2% beyond $d_{\text{max}}$ ($d_{\text{max}} = 3.3$ cm for 18 MV photon beam) for the incident electron beam with energy 18.2 MeV for 10×10 cm² field size. Once the PDD agreement was achieved for 10×10 cm² field size, dose profiles were calculated for different radii of the incident electron beam for profile matching.
Figure 3-30 PDD variation for 18 MV photon beam with incident electron beam energy for the 10×10 cm² field size.

Figure 3-31 Local percent difference between measurements and calculations at E=17.8 MeV for the 10×10 cm² field size.
Figure 3-32 Local percent difference between measurements and calculations at $E=18.0$ MeV for the $10\times10$ cm$^2$ field size.

Figure 3-33 Local percent difference between measurements and calculations at $E=18.2$ MeV for the $10\times10$ cm$^2$ field size.
Table 3-6 Average % difference and RMS % difference for different energies of the incident electron beam for 18 MV photon beam

<table>
<thead>
<tr>
<th>Energy</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>17.8 MeV</td>
<td>-1.0</td>
<td>1.5</td>
</tr>
<tr>
<td>18.0 MeV</td>
<td>-0.9</td>
<td>1.4</td>
</tr>
<tr>
<td>18.2 MeV</td>
<td>-0.2</td>
<td>0.8</td>
</tr>
</tbody>
</table>

### 3.4.2 Profile Comparison for the 18 MV photon beam

For 6 MV photon beam dose profile matching, the optimum incident electron beam radius was found to be uniform (1.3 mm). It was therefore thought to consider first the uniform radius for 18 MV photon beam for dose-profile matching. Calculations were performed for the incident electron beam with uniform radii of 1 and 1.3 mm for 30×30 cm² field size at 10 cm depth. The optimum energy (18.2 MeV) for the incident electron beam found for PDD matching was used in the dose profile calculations. Results are shown in Figure 3-34. Average % difference and RMS % difference was shown in Table 3-7. Large positive Average % difference was found in both calculations, which shows that the calculations were systematically high. Figure 3-34 shows that the difference was more pronounced towards the edges of the field.

After observing the large error for calculations performed for the incident electron beam with uniform radii, we then considered a Gaussian profile for the incident e-beam. Calculations were performed for 10×10 cm² field size at depth of 10 cm for FWHM of 1.3 mm for incident electron beam. Results are shown in Figure 3-35.
Figure 3-34 Dose-Profile variation with incident electron beam with uniform radii of 1 and 1.3 mm.

Table 3-7 Average % difference and RMS % difference for different uniform radii of incident electron beam.

<table>
<thead>
<tr>
<th>Radius of e-beam</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 mm</td>
<td>3.4</td>
<td>4</td>
</tr>
<tr>
<td>1.3 mm</td>
<td>3.3</td>
<td>3.9</td>
</tr>
</tbody>
</table>
Figure 3-35 Dose-Profile comparison for the incident electron beam with 18.2 MeV energy and Gaussian radial
distribution with FWHM of 1.3 mm for 10×10 cm² field size at 10 cm depth.

Figure 3-36 Local percent difference between measurements and calculations for the 10×10 cm² field size at 10
cm depth for 18 MV beam for incident electron beam with FWHM=1.3 mm. Average % difference was 1.5 and
RMS % difference was 1.9

Large positive average % difference shows that calculations were systematically
high for the incident e-beam with FWHM of 1.3 mm. Comparison is shown in Figure 3-
41 for an increased radius with FWHM of 1.9 mm also increasing the field size to 30×30
cm$^2$ field size at 10 cm depth. Field size was increased to see the difference at the edges of the larger field size.

Figure 3-37 Dose-Profile comparison for the 30×30 cm$^2$ field size at 10 cm depth for the incident electron beam with FWHM of 1.9 mm.

Figure 3-38 Local percent difference between measurements and calculations for the 30×30 cm$^2$ field size at 10 cm depth for 18 MV beam for the incident electron beam with FWHM of 1.9 mm. Average % difference was 2.2 and RMS % difference was 2.5.
Increasing the FWHM also increased the average % difference and RMS % difference. The Local percent difference between measurements and the calculations shows that the difference increases towards the edges of the field, i.e. calculated dose profiles had more pronounced horns at the field edges as compare to measurements. These horns could be due to too much thickness of the flattening filter at the center which causes non-uniform beam hardening of the incident beam. Varying the position of the flattening filter vertically did not reduce the error as shown in Figure 3-39.

![Figure 3-39 Comparison of calculations and measurements at various positions of flattening filter.](image)

It was observed that by changing the position of flattening filter the calculations became worse, so original position of the flattening filter was maintained.

To get the desired results, the incident electron beam parameters in the next step were chosen based on a recently published study\textsuperscript{14} for a 18 MV photon beam. In this study, the incident electron beam energy was considered having a Gaussian spread and the e-beam radius was considered to be uniform. In our calculations, we input the
incident electron beam energy with mean value of 18.2 MeV and added straggle with a FWHM of 0.2 MeV. The incident electron beam radius of 1.3 mm as for the 6 MV beam was used in the simulation. Results are shown in Figure 3-40 and Figure 3-41.

![Graph showing comparison of measurements with calculation for 30×30 cm² field size at 10 cm depth.]

**Figure 3-40** Comparison of measurements with calculation performed for the 30×30 cm² field size at 10 cm depth by considering the incident beam energy as Gaussian with mean 18.2 MeV and FWHM of 0.2 MeV and uniform e-radius with 1.3 mm.

![Graph showing local percent difference for 30×30 cm² field size at 10 cm depth.]

**Figure 3-41** Local percent difference between measurements and calculations performed for the 30×30 cm² field size at 10 cm depth by considering the incident beam energy having Gaussian spread with mean 18.2 MeV and FWHM of 0.2 MeV and uniform e-radius with 1.3 mm. Average % difference was 0.8 and RMS % difference was 1.5.
Although the average % difference was reduced but the observed Local percent difference between measurements and calculations was still more than 2%. The systematic change in sign between measurements and calculations towards the field boundaries and center was hypothesized to decrease by increasing the incident electron beam radius. To investigate the hypothesis, the incident electron beam radius profile was considered to be uniform and was varied from 1.5 to 2.5 mm. The energy of the incident electron beam was considered uniform (18.2 MeV) in these calculations. Calculations were performed for $30 \times 30 \text{ cm}^2$ field size at 10 cm depth for 1.5, 2 and 2.5 mm uniform incident e-beam radii. Results are shown in the Figure 3-41. Average % difference and the RMS % difference are shown in Table 3-8. Best results were achieved for the incident electron beam with 2 mm uniform radius.

Figure 3-42 Dose-Profile variation for the incident electron beam with uniform radii of 1, 1.2 and 2 mm.
Table 3-8 Average % difference and RMS % difference for different uniform radii of incident electron beam.

<table>
<thead>
<tr>
<th>Radius of e-beam</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mm</td>
<td>-0.9</td>
<td>1.8</td>
</tr>
<tr>
<td>2 mm</td>
<td>-0.11</td>
<td>0.7</td>
</tr>
<tr>
<td>2.5 mm</td>
<td>-0.13</td>
<td>1.3</td>
</tr>
</tbody>
</table>

### 3.4.3 Final Comparison

Since the best match was achieved for 30×30 cm² field size at 10 cm depth, dose profile calculations were performed at 3.3 cm for 30×30 cm² field size and also for 5×5 and 10×10 cm² field sizes at 3.3 and 10 cm depths. Results are shown in Figure 3-43 through Figure 3-48. Average % difference and RMS % difference was shown in Table 3-9 and Table 3-10 for comparison of measurements with dose-profile calculations performed for the incident electron beam with energy 18.2 MeV and uniform radius of 2 mm. Although calculations were somewhat systematically low, they were within the acceptable limit of 2%.

After matching the profiles for all field sizes, PDDs were calculated with optimized energy (18.2 MeV) and the optimized radius (2 mm) for the incident electron beam for all field sizes. Comparison of the calculations with measurements and Local percent difference is shown in Figure 3-49 through Figure 3-54. The average % difference and the RMS % difference for three field sizes for the incident electron beam with energy 18.2 MeV is shown in Table 3-11. Calculated PDD for 30×30 cm² field size was systematically high which may be attributed to the difficulty in both measurement...
and calculation of the increased scatter contribution for larger field sizes. All PDD calculations were within 2% of the measurements for all three field sizes beyond $d_{\text{max}}$ for incident electron beam with energy 18.2 MeV.

![Figure 3-43 Dose-Profile comparison for the 5×5 cm$^2$ field size at d=3.3 and d=10 cm depths.](image)

![Figure 3-44 Local percent difference between measurements and calculations at 3.3 and 10 cm depths for the 5×5 cm$^2$ field size.](image)
Figure 3-45 Dose-Profile comparison for the $10\times10 \text{ cm}^2$ field size at $d=3.3$ and $d=10$ cm depths.

Figure 3-46 Local percent difference between measurements and calculations at 3.3 and 10 cm depths for the $10\times10 \text{ cm}^2$ field size.
Figure 3-47 Dose-Profile comparison for the 30×30 cm$^2$ field size at d=3.3 and d=10 cm depths.

Figure 3-48 Local percent difference between measurements and calculations at 3.3 (left) and 10 cm (right) depths for the 30×30 cm$^2$ field size.
Table 3-9 Average % difference and RMS % difference at 3.3 cm depth for three field sizes.

<table>
<thead>
<tr>
<th>Depth = 3.3 cm</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm²</td>
<td>-0.3</td>
<td>1.2</td>
</tr>
<tr>
<td>10×10 cm²</td>
<td>-0.3</td>
<td>0.9</td>
</tr>
<tr>
<td>30×30 cm²</td>
<td>-0.4</td>
<td>0.7</td>
</tr>
</tbody>
</table>

Table 3-10 Average % difference and RMS % difference at 10 cm depth for three field sizes.

<table>
<thead>
<tr>
<th>Depth = 10 cm</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm²</td>
<td>-0.4</td>
<td>1.1</td>
</tr>
<tr>
<td>10×10 cm²</td>
<td>-0.2</td>
<td>0.8</td>
</tr>
<tr>
<td>30×30 cm²</td>
<td>-0.1</td>
<td>0.6</td>
</tr>
</tbody>
</table>

Figure 3-49 Comparison of PDD calculations with measurements for the 5×5 cm² field size for the incident electron beam with energy 18.2 MeV.
Figure 3-50 Local percent difference between measurements and calculations for the 5×5 cm² field size for the incident electron beam with energy 18.2 MeV.

Figure 3-51 Comparison of PDD calculations with measurements for the 10×10 cm² field size for the incident electron beam with energy 18.2 MeV.
Figure 3.52 Local percent difference between measurements and calculations for the 10×10 cm² field size for the incident electron beam with energy 18.2 MeV

Figure 3.53 Comparison of PDD calculations with measurements for the 30×30 cm² field size for the incident electron beam with energy 18.2 MeV
Figure 3-54 Local percent difference between measurements and calculations for the 30×30 cm² field size for the incident electron beam with energy 18.2 MeV.

Table 3-11 Average % difference and RMS % difference for different field sizes at 18.2 MeV.

<table>
<thead>
<tr>
<th>Energy</th>
<th>Average % difference</th>
<th>RMS % difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm²</td>
<td>0.3</td>
<td>0.5</td>
</tr>
<tr>
<td>10×10 cm²</td>
<td>-0.2</td>
<td>0.8</td>
</tr>
<tr>
<td>30×30 cm²</td>
<td>1.5</td>
<td>1.2</td>
</tr>
</tbody>
</table>

3.5 Conclusion

In this work, 6 and 18 MV photon beams of Varian’s linear accelerator were modeled using MCNP5. To benchmark the modeling, measured PDDs and dose profiles for 5×5, 10×10 and 30×30 cm² field sizes were used. It was verified that the PDDs are sensitive to the incident electron beam energy. Dose profiles for 30×30 cm² field size were found to be more sensitive to the incident electron beam radius. Dose profiles were
also found to be sensitive to the position of the flattening filter. The optimal primary electron beam parameters were found to be monoenergetic spectra with energies of 6.2 and 18.2 MeV and uniform radii of 1.3 and 2 mm respectively. These parameters differed from the optimal parameters reported by Chibani et al.,\textsuperscript{14} indicating that the beam parameters may be machine specific even if used for the same model from the same vendor.

PDD calculations agree with measurements within 2% beyond 1.5 cm for the 6 MV photon beam and beyond 3.3 cm for the 18 MV photon beam for all three field sizes. Dose profile calculations agree with measurements within 2% in the flat region and within 10% for penumbral region at 3.3 and 10 cm depths for both beam energies and for all three field sizes.

The simulated photon beams can be used to verify the dose calculations performed by the commercial treatment planning systems in high dose gradients. These models were used to investigate the radiation dose deposition in the buildup region in chapter 4 of this thesis.
References:


17 Roeers D W O “Low energy electron transport with EGS,” Nuclear Instruments and Methods B, 221, 535-48


4 Dose Calculation in the Buildup Region

4.1 Introduction

For megavoltage x-ray beams, the penetrating power of the secondary charged particles created by the beam increases relative to those created by lower energy beams, leading to a deeper position of the maximum dose point. The region before the depth of maximum dose is called the buildup region. Dose deposition within the first few millimeters of the skin is important for treatment planning. The primary contribution to dose in the buildup region is from contaminant electrons produced by primary photon beam interactions with the linear accelerator head components, such as the independent jaws and beam modifiers, as well as with the air volume between the patient (or phantom) surface and the linear accelerator head. The secondary contribution to the buildup dose is from the backscattering of the photon beam inside the phantom. Dose measurement in the buildup region is prone to a high degree of uncertainty due to the high dose gradient and lack of charged particle equilibrium.

Monte Carlo simulation techniques are considered the gold standard for dose calculation in radiation therapy, however many authors\textsuperscript{1,2,3} have reported discrepancies between measurements and MC calculations in the buildup region. These were partially rectified by the previously mentioned corrections to the geometry of linear accelerator head by Chibani and Ma.\textsuperscript{4} However, to best of our knowledge, MCNP5 has not been tested to investigate the discrepancy between measurements and calculations in the buildup region. MCNP5 has enhanced electron transport as compared to its older
versions. MCNP5 was used in this thesis to investigate the discrepancy between measurements and calculations in the buildup region. The geometry corrections in the primary collimator proposed by Chibani and Ma were considered in the simulation and the shielding geometry was also modeled.

4.2 Measurements in the buildup region

Ionization chambers are the most commonly used for dose measurements in radiation therapy departments. When radiation passes through the detector, ionization is produced in the sensitive volume. An electric field is applied across the sensitive volume to measure the charge produced by the radiation. Mostly air is used as sensitive volume in the ion chambers. The ionization produced depends on the mass of the sensitive volume, so any density variation due to changes in temperature and pressure must be taken into account. The dose deposited in the sensitive air volume can be related to the charge produced as

\[ D_{\text{air}} = \frac{Q_{\text{air}}}{\rho_{\text{air}} \cdot V} \cdot \frac{W_{\text{air}}}{e}. \]

Equation 4-1

where \( Q_{\text{air}} \) is the amount of charge produced, \( \rho_{\text{air}} \) is the air density, \( V \) is the air volume and \( W_{\text{air}} \) is the mean energy needed to produce an ion pair in dry air (~33.97 eV for photon and electron beams in the therapeutic energy range).

Equation 4-1 gives the dose to the air volume inside the ion chamber. When the ion chamber is placed in water, it is intended to measure the dose to the water at the position of ion chamber. The stopping power, \( S \), is defined as the energy loss per unit path length in any medium. Using a stopping power ratio of the two mediums compensates for the
difference between the dose in the chamber and the dose in the surrounding medium so the dose can be computed as if there was water at the point of interest. Mass stopping power is defined as

\[ S' = \frac{S}{\rho}, \]  

where \( \rho \) is the mass density. The dose to the sensitive volume of air can be related to the dose to water at the same point as \(^5\,^6\)

\[ D_{\text{water}} = D_{\text{air}} \left( \frac{S'_{\text{water}}}{S'_{\text{air}}} \right). \]  

Equation 4-3

where \( \left( \frac{S'_{\text{water}}}{S'_{\text{air}}} \right) \) is the weighted mean ratio of water to air stopping powers of the electrons. The assumptions underlying equation 4-3 are that the cavity must be small compared to the range of the incident charged particles so that its presence does not perturb the fluence of charged particles in the medium, and the absorbed dose in the cavity is deposited by the charged particles only. That is, photon interactions in the cavity are negligible. These assumptions may not be fulfilled in areas of high dose gradient and charged particle disequilibrium, so the measurements made in such areas should be done very carefully. In other words, stopping power ratios do not represent the detector response accurately in areas of high dose gradients and charge particle disequilibrium.

Large volume cylindrical ion chambers are not suitable for measurements in the buildup region and when used they greatly overestimate the dose. Extrapolated ion chambers are the instruments of choice for dose measurement in the buildup region. In
extrapolation chambers, the separation between electrodes is adjustable so the measurements are performed at one position for different separation of electrodes and then the measured percentage depth ionization (PDI) values are extrapolated to obtain the dose value for zero separation of electrodes. Extrapolation chambers are recommended for dose measurement in the buildup region but they are cumbersome to use and are not widely available.

One of the possible choices for dose measurement in the buildup region is the plane parallel ion chamber, such as the PTW Markus 23343 plane parallel chamber used in this work. It has an electrode separation of 2 mm, with a small measuring volume diameter of 6 mm and sidewall to collector distance of 0.35 mm. Plane parallel chambers\(^7\) have also been found to overestimate the dose due to the primary electron fluence perturbation caused by secondary electrons emitted or scattered from the side walls of the chamber. An empirical method was proposed by Velkley et al.\(^8\) in terms of percentage change per millimeter of plate separation to correct the over response of plane parallel chamber measurements. Gerbi and Khan\(^9\) investigated the drawback of the method and they proposed a more generalized method to correct the over response of the plane parallel ion chamber. They developed an empirical formula for over response correction comparing the plane parallel ionization chamber measurements to extrapolation chamber measurements\(^9\)

\[
P'(d) = P(d) - \xi(E,0) \cdot l \cdot e^{-a \left( \frac{d}{d_{\text{max}}} \right)} \quad \text{Equation 4-4}
\]

where
\[ \xi(E, 0) = [-1.666 + (1.982IR)](C - 15.8)\left(\frac{\%}{mm}\right) \]  \hspace{1cm} \text{Equation 4-5}

\(P'(d)\) is corrected PDD, \(P(d)\) is uncorrected PDD, \(IR\) is the ratio of the ionization at a depth of 20 to the ionization measured at 10 cm at constant source to detector distance of 100 cm for a 10×10 cm² field size, \(C\) is the collector edge to side wall distance in mm, \(l\) is the plate separation in mm, \(\alpha\) is constant, \(d\) is depth of any dose and \(d_{max}\) is depth of maximum dose. For the Markus chamber used in the measurements \(C = 0.35\) mm, \(l = 2\) mm. The measured ionization ratios were 0.668 and 0.776 for the 6 and 18 MV photon beams. The constant \(\alpha\) used in the correction was 5.5, taken from Gerbi and Khan. The effective point of measurement was assumed to be located just below the top electrode of the chamber.

Measurements were performed inside the solid water phantom with a cavity suitable to place the parallel plate ion chamber. When measuring the charge produced in the plane parallel chamber due to radiation, the polarising potential is applied to the outer electrode and the signal charge is collected from the inner electrode. Photon beams interacting with the collecting electrode may produce charge particles which may affect the charge collection from the ionization produced inside the sensitive air cavity and the charge collection may vary with the polarity of the collecting electrode. This polarity effect was taken into account by measuring the charge with both positive and negative polarities and then taking the average as the measured charge. Measurements were performed at 0, 2 mm and 4 mm depths for 5×5, 10×10 and 30×30 cm² field sizes for both beam energies and were corrected using Equation 4-4. Cylindrical ion chamber measured, plane parallel ion chamber measured uncorrected and corrected PDDs are
presented in Figure 4-1 through Figure 4-3. Results show that the cylindrical ion chamber overestimates the surface dose enormously and then it underestimates the dose at 2 and 4 mm depths compared to plane parallel chamber measurements and MC calculations. This is expected because the chamber averages the dose along a vertical path, and so overestimates the dose at the start of the buildup and underestimates it when it begins to level off. The correction is reduced as the depth is increased away from the buildup region.

![Graph](image)

**Figure 4-1** Comparison of cylindrical chamber measured (Cyl. Chamber), plane parallel chamber measured (PP uncorrected) and plane parallel chamber corrected (PP corrected) PDD in the buildup region for 6 MV (left) and 18 MV (right) photon beams for 5×5 cm² field size.
Figure 4-2 Comparison of cylindrical chamber measured (Cyl. Chamber), plane parallel chamber measured (PP uncorrected) and plane parallel chamber corrected (PP corrected) PDD in the buildup region for 6 MV (left) and 18 MV (right) photon beams for 10×10 cm$^2$ field size.

Figure 4-3 Comparison of cylindrical chamber measured (Cyl. Chamber), plane parallel chamber measured (PP uncorrected) and plane parallel chamber corrected (PP corrected) PDD in the buildup region for 6 MV (left) and 18 MV (right) photon beams for 30×30 cm$^2$ field size.

4.3 Surface Dose Comparison

Calculation voxels of 1 mm depth were used for the PDD calculation in the buildup region. Two separate sets of simulations were performed. In the first set of simulations, the starting position of the first voxel was defined at 0 mm giving the dose at the 0.5 mm
position and the subsequent voxels were defined with 1 mm separation. In the second set of simulation, the starting position of the first voxel was defined at 0.5 mm giving the dose at 1 mm position. Measurements were compared with the calculations at 2 and 4 mm depths.

For comparison at the surface, MCNP5 computations were extrapolated to obtain the surface dose. Figure 4-4 shows 3\textsuperscript{rd} through 6\textsuperscript{th} order polynomials fit to the MCNP5 computations in the buildup region for a 6 MV photon beam with a 10×10 cm\textsuperscript{2} field size. It can be seen from the plot that the lower order polynomials have a systematic bias for the region where the dose is less than 80%. Lower order polynomials were overshooting and undershooting the data alternatively. The sixth order polynomial passes more cleanly through the data. Fitting for 6\textsuperscript{th} through 9\textsuperscript{th} order polynomials is shown in Figure 4-5 through 4-10. In most cases these higher order polynomials were nearly indistinguishable, however the highest order polynomial begins to curve away from the data for the 5×5 field size for the 6 MV beam and the 30×30 field size for the 18 MV beam, as shown in Figure 4-7 and 4-10. RMS error was calculated for all field sizes at both beam energies for 3\textsuperscript{rd}, 4\textsuperscript{th}, 5\textsuperscript{th}, 6\textsuperscript{th} and 7\textsuperscript{th} order polynomials shown in Table 4-1.

Table 4-1 RMS values for various order polynomial fitting for all field sizes at both beam energies.

<table>
<thead>
<tr>
<th>Polynomial</th>
<th>5×5</th>
<th>10×10</th>
<th>30×30</th>
<th>5×5</th>
<th>10×10</th>
<th>30×30</th>
</tr>
</thead>
<tbody>
<tr>
<td>6 MV</td>
<td>6 MV</td>
<td>6 MV</td>
<td>18 MV</td>
<td>18 MV</td>
<td>18 MV</td>
<td></td>
</tr>
<tr>
<td>3\textsuperscript{rd} Order</td>
<td>1.22</td>
<td>1.23</td>
<td>1.35</td>
<td>1.16</td>
<td>1.09</td>
<td>1.31</td>
</tr>
<tr>
<td>4\textsuperscript{th} Order</td>
<td>0.58</td>
<td>0.60</td>
<td>0.67</td>
<td>0.77</td>
<td>0.49</td>
<td>0.58</td>
</tr>
</tbody>
</table>
Higher order polynomials gave less RMS error but that could be attributed to the higher number of parameters, as there were only 30 data points.

To avoid bias, it was decided to average the polynomial extrapolation to the surface for the full datasets for the 6th through 9th order polynomials. Comparison of the measured and extrapolated surface dose is shown in Table 4-1 and Table 4-3. The difference from the measured values was less than the standard deviation for all of the field sizes at 6 MV and the two smaller field sizes at 18 MV.

**Table 4-2 Comparison of measured (PP corrected) and extrapolated surface dose for 6 MV beam for all field sizes.**

<table>
<thead>
<tr>
<th>6 MV</th>
<th>Measured (PP Corrected)</th>
<th>Average of 6-9th order polynomials</th>
<th>Standard Deviation</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm²</td>
<td>10.7</td>
<td>11.2</td>
<td>0.87</td>
<td>0.5</td>
</tr>
<tr>
<td>10×10 cm²</td>
<td>15.7</td>
<td>16.0</td>
<td>1.03</td>
<td>0.3</td>
</tr>
<tr>
<td>30×30 cm²</td>
<td>36.4</td>
<td>36.1</td>
<td>0.87</td>
<td>-0.3</td>
</tr>
</tbody>
</table>
Comparison of plane parallel chamber measured PDDs (PP corrected) and MCNP5 calculations at 2 and 4 mm depths and comparison of measured and extrapolated surface dose is presented in Figure 4-11 through Figure 4-16. The maximum deviation between MCNP5 calculations and the plane parallel PDD measurements at 18 MV is 3.8% for 30×30 cm² at 2 mm depth. The deviation is within 3% for all other field sizes and depths for both beam energies. For 18 MV, the extrapolation of the MCNP5 PDD overestimates the surface dose for the 30×30 cm² field size, and is within 4.6% for all other field sizes at both beam energies.

### Table 4-3 Comparison of measured (PP corrected) and extrapolated surface dose for 18 MV beam for all field sizes.

<table>
<thead>
<tr>
<th>Field Size</th>
<th>Measured (PP Corrected)</th>
<th>Average of 6-9th order polynomials</th>
<th>Standard Deviation</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>5×5 cm²</td>
<td>6.9</td>
<td>7.2</td>
<td>0.71</td>
<td>0.3</td>
</tr>
<tr>
<td>10×10 cm²</td>
<td>14.3</td>
<td>14.3</td>
<td>0.14</td>
<td>0.04</td>
</tr>
<tr>
<td>30×30 cm²</td>
<td>37.6</td>
<td>34.2</td>
<td>0.53</td>
<td>-3.4</td>
</tr>
</tbody>
</table>
Figure 4-4 Polynomials of various degrees were fitted to MCNP5 data for 6 MV photon beam for 10×10 cm² field size.
Figure 4-5 Higher order polynomial (6th, 7th, 8th and 9th) fitting to MCNP5 data in buildup region for 5×5 cm² field size for 6 MV photon beam.
Figure 4-6 Extended view of higher order polynomial (6th, 7th, 8th and 9th) fitting to MCNP5 data in buildup region for 10×10 cm² field size for 6 MV photon beam.
Figure 4-7 Higher order polynomial (6th, 7th, 8th and 9th) fitting to MCNP5 data in buildup region for 30×30 cm² field size for 6 MV photon beam.
Figure 4-8 Higher order polynomial (6th, 7th, 8th and 9th) fitting to MCNP5 data in buildup region for 5×5 cm² field size for 18 MV photon beam.
Figure 4-9 Higher order polynomial (6th, 7th, 8th and 9th) fitting to MCNP5 data in buildup region for 10×10 cm² field size for 18 MV photon beam.
Figure 4-10 Higher order polynomial (6th, 7th, 8th and 9th) fitting to MCNP5 data in buildup region for 30×30 cm² field size for 18 MV photon beam.
Figure 4-11 Comparison between measured (PP corrected) and calculated (MCNP5) PDDs in the buildup region for 6 MV photon beam for 5×5 cm$^2$ field size.

Figure 4-12 Comparison between measured (PP corrected) and calculated (MCNP5) PDDs in the buildup region for 6 MV photon beam for 10×10 cm$^2$ field size.
Figure 4-13 Comparison between measured (PP corrected) and calculated (MCNP5) PDDs in the buildup region for 6 MV photon beam for 30×30 cm$^2$ field size.

Figure 4-14 Comparison between measured (PP corrected) and calculated (MCNP5) PDDs in the buildup region for 18 MV photon beam for 5×5 cm$^2$ field size.
Figure 4-15 Comparison between measured (PP corrected) and calculated (MCNP5) PDDs in the buildup region for 18 MV photon beam for 10×10 cm² field size.

Figure 4-16 Comparison between measured (PP corrected) and calculated (MCNP5) PDDs in the buildup region for 18 MV photon beam for 30×30 cm² field size.
References:


5 Focused Beam Therapy

5.1 Introduction

In conventional external beam radiation therapy, megavoltage beams are used to reduce the skin dose, even for relatively shallow tumors (depth < 5 cm). Instead, focusing a low energy x-ray beam could result in spreading the beam across the skin surface while increasing the beam intensity at the tumor. Such a beam can be produced with polycapillary optics. MCNP5 was used to model dose profiles for a scanned focused beam, using measured beam parameters. The potential of low energy focused x-ray beams for radiation therapy was assessed. The schematic diagram of the proposed method is shown in Figure 5-1.

![Figure 5-1 Schematic diagram for focusing x-ray beam using polycapillary optics.](image)

5.2 Polycapillary Optics

Polycapillary focusing and collimating lenses were invented in the 1980s. Monolithic collimating and focusing lenses contain arrays of thousands of hollow capillary tubes fused together, as shown in Figure 5-2. These capillary tubes guide the
divergent x rays by the principle of total external reflection if the incident angle of the ray
is less than the critical angle. The index of refraction for glass in x-ray energy spectrum
is less than unity, so only the x rays falling at grazing angles will be reflected by the
polycapillary optics. The angle of incidence are kept small by using large numbers of very
small tubes, all pointed at the source at one end and at the tumor at the other.
Polycapillary optics are used in diverse applications, especially for materials analysis such as
x-ray diffraction, fluorescence, and electron-excited spectroscopy.

Figure 5-2 Cross sectional view of polycapillary fiber of 500 μm in diameter with approximately 250 channels,
each 50 μm in diameter.

5.3 Existing Polycapillary Beam Data

Earlier measurements have been made with a polycapillary lens used to collect x-
rays from a conventional divergent x-ray source and redirect them to make a focused x-
ray beam. The measurements were performed at 50 kV for optic 4280. The size of the
output focal point was determined by taking images of the lens output. The image
profiles were fitted with Gaussians to estimate the width at the focus point. The
minimum width was found to be 0.20 mm at the output focal point distance of 50 mm
with a 10 mm depth of field. PMMA blocks of varying thickness were placed between
the optic and focal point to establish that focusing of the beam was maintained through
40 mm of PMMA. The output spectrum of the optic was measured in 2 keV photon
energy bins. This data and the output focus distance were obtained and used to simulate
the focused beam in MCNP5.

5.4 Simulating a focused x-ray beam using MCNP5

MCNP5 does not have a predefined source template for a focused beam, so instead
an array of rays was generated to model the polycapillary optic. A C++ program was
written to produce hundreds of positions (x, y, z) of source particles and the
corresponding direction vectors (u, v, w) which were used in the SDEF card of MCNP5
to define a convergent beam. Starting positions of the source particles (x, y, z) were
generated as random numbers inside a circular disk of radius 0.2 cm. The corresponding
direction vectors (u, v, w) were defined such a way that the source particles were passing
through the focal point of the optic at 5 cm (u=-x, v=-y, w=f). As a polycapillary optic
does not produce a perfect focus, a small random value in the range of ±0.65θ_c was added
to the direction of the source particles. The factor of 0.65 is based on experimental
measurements of the optic divergence. Here θ_c is the critical angle which is 1.7 mrad for
the energy range used in the experiment. The offset in the direction vectors was defined as

\[ \theta = -0.65 \cdot \theta_c + 1.3 \cdot \theta_c \cdot \text{rand}( ) \]  \hspace{1cm} \text{Equation 5-1} \\
\[ \phi = 2 \cdot \pi \cdot \text{rand}( ) \]  \hspace{1cm} \text{Equation 5-2} \\
\[ \text{Offset}(u) = \sqrt{x^2 + y^2 + f^2} \cdot \sin(\theta) \cos(\phi) \]  \hspace{1cm} \text{Equation 5-3} \\
\[ \text{Offset}(v) = \sqrt{x^2 + y^2 + f^2} \cdot \sin(\theta) \sin(\phi) \]  \hspace{1cm} \text{Equation 5-4}
Once the starting positions and the corresponding vectors with offsets were defined, they were then given to MCNP5 using the SDEF card, which is used in MCNP5 for source definition. The SDEF card has many variables that are used to define all the characteristics of radiation source in the problem, such as reference vector for direction (VEC), direction of flight (DIR), energy (ERG). The first variable on the SDEF card was PAR, which designates the particle type. For the simulation of photons, PAR P was used, so that the first part of the SDEF card read

SDEF PAR P

The next variable is the starting position (POS) of the source particles, which was defined as a distribution D1, so that the SDEF card now read SDEF PAR P POS=D1. Values of SDFE card variables can be specified at three levels; explicitly e.g., POS=0 0 0, with a distribution number (e.g., POS=D1), or as a function of another variable. When variables are specified as a distribution or as a function of another variable then other source cards are required. The position distribution D1 was given by the source information card SI1 and source probability card SP1. SI1 H means the source card is followed by a histogram and SI1 L means it is followed by discrete source values. The C++ code generated the source positions in the format

SI1 L x1 y1 z1

x2 y2 z2

x3 y3 z3 ...
The SP1 describes the corresponding relative intensity for each ray, which were set to 1 for all,

\[ \text{SP1} \quad 1 \ 1 \ 1 \ ... \]

The direction of each ray was specified as a reference vector (VEC) for direction, defined as a dependent distribution, \( D_2 \), a function of the starting position of source particles, so that the SDEF card now read

\[ \text{SDEF} \quad \text{PAR P POS=D1 VEC=FPOSD2} \]

The corresponding reference vectors for all source particles were defined on the dependent source card \( \text{DS2 L} \), which were also generated by the C++ code.

\[ \text{DS2 L} \quad u_1 \ v_1 \ w_1 \]
\[ \quad u_2 \ v_2 \ w_2 \]
\[ \quad u_3 \ v_3 \ w_3 \ ... \]

The direction (DIR) of the source particles was defined as 1 which causes the particles to move in direction of their reference vectors (e.g. as opposed to -1, which moves them in the opposite direction), so the card now read

\[ \text{SDEF} \quad \text{PAR P POS=D1 VEC=FPOSD2 DIR=1} \]

The simulated beam spectrum was given by the experimental values, so the energy (ERG) of the source particles was defined as a distribution \( D_3 \),

\[ \text{SDEF} \quad \text{PAR P POS=D1 VEC=FPOSD2 DIR=1 ERG=D3}. \]
The information about the energy distribution D3 was defined on SI3 H and SI3 D cards from the reported experimental measurements. The SI3 H card defines the bin boundaries for a histogram distribution. The bin probabilities were defined on SI3 D card.

SI3 H 0.012 0.014 0.016….. $ Energy Bins

SP3 D 0 .055 .085 ……. $ Sampling Probability

Figure 5-3 shows the particle tracks of the focused x-ray beam generated by MCNP5. After passing through the focal point, x ray beam diverges again.

![Particle tracks for a focus beam generated by MCNP5.](image)

Flux and dose distribution data were calculated using an MCNP5 superimposed mesh tally (fmesh4). The mesh tally divided the tissue phantom volume into a grid of (orthogonal) voxels and the x-ray flux in each voxel was calculated. Calculated flux in each voxel was converted to dose rate using a standard flux-to-dose table (ICRP-21)
given in the MCNP5 manual. The table was entered into the input file as a tally multiplier card.

After defining the focus beam, the focal spot size of the beam was verified by calculating the flux within a grid of voxels along the x axis at the focal plane. The FWHM was calculated from the Gaussian fit to the data presented in Figure 5-4. Calculated FWHM was 0.015 cm, but the voxel size was .009 cm so there were only ~3 voxels in the focus, whereas the measured focal spot size was 0.02 cm, so the agreement shows the focus of the lens has been correctly modeled.

![Figure 5-4 Flux calculated across a grid of voxels on x-axis at focal plane.](image)

Relative dose was calculated along the central axis of the beam from a static lens. A breast tissue phantom was simulated in shape of a hemisphere in MCNP5. The optic
was positioned at 2 cm from the skin surface and the beam was focused at 3 cm inside the
tissue. For dose calculations, cubical voxels of 0.2 mm side length were simulated along
the central axis of the beam. The energy deposition in each voxel was calculated using
the *F8 tally of MCNP5. Percent depth dose calculated inside the phantom is shown in
Figure 5-5. The focused beam was found to concentrate the dose within the tumor with
sharp fall off before and after the treatment volume. The ratio of peak tumor to skin dose
for this idealized case of a tiny tumor in a spherical patient was 14.

![Figure 5-5 Percent depth dose calculated by MCNP5 in tissue phantom.](image)

Because the focal spot size of the beam is so small, a scan is necessary to irradiate
a reasonable size tumor. Since the beam was translated and rotated for scanning purpose,
TR card was used as a distribution to define source at different positions on SDEF card of
MCNP5. The information about the TR distribution (D4) was given on SI4, SP4, SB4
and *TR cards.

SI4 L 11

SP4 1
SB4 1

*TR1 x’ y’ z’ θ(x, x’) θ(y, y’) θ(z, z’) θ(x, y’) θ(y, z’) θ(z, x’) M

SL4 L defines the discrete number of transformations and gives them different names like 11, 12 etc. The SP4 card defines the corresponding relative intensities for different source positions and the SB4 card defines the directions biases. The new position of the source is given on *TR card for each transformation. Co-ordinates after the translation (x’ y’ z’) and angles (θ) of the rotation matrix are defined on the *TR card. The position of the origin of the displacement vector is also defined on *TR card represented by M and could be 1 or -1. When M=1 the displacement vector is the location of the origin of the auxiliary coordinate system, defined in the main system.

For simplicity, a symmetric case was considered first. A cylindrical tumor with volume 63 mm³ was simulated in a spherical tissue of radius 3 cm. The optic was rotated in one plane around the sphere as shown in Figure 5-6. In Figure 5-6, the focused beam is rotated at three different angles but when treating the tumor, it was rotated at forty different angle positions.
Figure 5-6 Focus beam rotated at different positions to scan the tumor.

The original lens position was \( x = 0, \; y = 0 \) and \( z = 0 \). For a rotation of 4 degree, the following *TR card was used

*TR11 0.3487 0 0.01218 4 90 86 90 0 90 94 90 4 1

Here \( x' = f \sin \theta + x = 0.3487, \; y' = y = 0, \; z' = f(1-\cos \theta) + z = 0.01218, \; \theta(x, x') = 4, \; \theta(y, x') = 90, \; \theta(z, x') = 86, \; \theta(x, y') = 90, \; \theta(y, y') = 0, \; \theta(z, y') = 90, \; \theta(x, z') = 94, \; \theta(y, z') = 90, \; \theta(z, z') = 4 \) and \( M = 1 \). The calculated dose distribution for a rotation scan is presented in Figure 5-7. The peak to skin ratio for the case in which it is possible to rotate the optic around the patient was 78.
Figure 5-7 Dose distribution calculated with superimposed mesh tally. Cylindrical tumor was simulated at the focus point of the beam. Maximum dose was achieved at the tumor while sparing the skin dose.

If the patient shape prohibits a full rotation, a linear scan may be necessary, however, simply overlapping the lens position along a straight line will not be skin sparing. To demonstrate that, a single slice of rectangular shaped tumor was simulated with dimension $1 \times 0.02 \times 1 \text{ cm}^3$ in a rectangular shape tissue with dimension $6 \times 6 \times 4 \text{ cm}^3$. The thickness of the tumor slice in y-direction was chosen equal to the focal spot size of the x ray beam (0.02 cm) and the height of the tumor was chosen equal to the field depth of the focused beam. The tissue surface was at 3.6 cm from the optic and tumor was 1.4 cm deep in to the tissue. The density of the tumor and the phantom was simulated as a soft tissue. Linear scan was performed in the x-direction. For one step of a linear scan in x direction, the following *TR card was used

*TR12 0.2 0 0 0 90 90 0 90 90 0 90 0 1

The lens was translated in steps equal to the radius of the lens (0.2 cm). Here $x' = x + 0.2 = 0.2$, $y' = y = 0$, $z' = z = 0.01218$, $\theta(x, x') = 0$, $\theta(y, y') = 90$, $\theta(z, z') = 90$, $\theta(x, y') = 90$, $\theta(x, z') = 90$, $\theta(y, z') = 90$, $\theta(y, x') = 90$, $\theta(z, x') = 90$, $\theta(z, y') = 90$, $\theta(z, z') = 90$. 

The dose distribution is shown in Figure 5-7.
\( \theta(y, y') = 0, \theta(z, y') = 90, \theta(x, z') = 90, \theta(y, z') = 90, \theta(z, z') = 0 \) and \( M = 1 \). The optic was translated at fifty steps to cover a tumor with dimension 1 cm along x-axis. The dose distribution calculated with superimposed mesh tally is shown in Figure 5-8.

In a linear scan, the beam overlaps at the skin surface which increases the skin dose. Although the skin dose is smaller than the tumor dose by a ratio of 18, the dose is still significant. Rotating the beam as it is scanned will maintain the beam spreading and will give better skin sparing effect than the linear scan. The focused beam was translated and rotated in the next step for the same tumor. For translation and rotation of the lens, the following *TR card was used:

*TR13 0.420211 0 0.01218 4 90 86 90 0 90 94 90 4 1

Here \( x' = f \sin \theta + x + \text{step-size} = 0.420211, y' = y = 0, z' = f(1-\cos \theta) + z = 0.01218, \theta(x, x') = 4, \theta(y, x') = 90, \theta(x, y') = 86, \theta(x, y') = 90, \theta(y, y') = 0, \theta(z, y') = 90, \theta(x, z') = 94, \theta(y, z') = 90, \theta(z, z') = 0 \) and \( M = 1 \). The optic was translated and rotated at fifty
different positions to cover the tumor. The dose distribution calculated with the superimposed mesh tally is shown in Figure 5-9.

![Figure 5-9 Dose inside the patient geometry for a translated and rotated beam for tumor at 5 cm and tissue at 3.6 cm from the optic.](image)

Better skin sparing was observed when the beam was angled as it is translated. The peak to skin dose ratio was 45. Skin sparing is very much important in radiation therapy and is the limiting factor in dose escalation. The proposed method has potential for much better skin sparing compared to the megavoltage photon beams in treating the soft tissue.

### 5.5 Conclusion

A focused beam obtained with polycapillary optics was simulated using MCNP5. The focusing of the beam was maintained inside the tissue phantom. High conformal dose to the tumor while sparing the skin was achieved by scanning the tumor with the polycapillary optic. Low energy focused x-ray beams could be used to irradiate the tumors inside soft tissue within 5 cm of the surface. Better skin sparing compared to
megavoltage photon beams could be achieved by low energy focused x ray beam inside soft tissues.
References:


6 Thesis Summary

In this work, 6 and 18 MV photon beams of Varian’s linear accelerator were modeled using MCNP5. To benchmark the modeling, measured PDDs and dose profiles for 5×5, 10×10 and 30×30 cm² field sizes were used. It was verified that the PDDs are sensitive to the incident electron beam energy. Dose profiles for 30×30 cm² field size were found to be more sensitive to the incident electron beam radius. Dose profiles were also found to be sensitive to the position of the flattening filter. The optimal primary electron beam parameters were found to be monoenergetic spectra with energies of 6.2 and 18.2 MeV and uniform radii of 1.3 and 2 mm respectively. These parameters differed from the optimal parameters reported in literature indicating that the beam parameters may be machine specific even if used for the same model from the same vendor.

PDD calculations agree with measurements within 2% beyond 1.5 cm for the 6 MV photon beam and beyond 3.3 cm for the 18 MV photon beam for all three field sizes. Dose profile calculations agree with measurements within 2% in the flat region and within 10% for penumbral region at d_{max} and 10 cm depths for both beam energies and for all three field sizes.

The simulated photon beams were used for dose calculation in the buildup region. For comparison in the buildup region, the MCNP5 voxel was reduced to 1 mm, with extrapolation to find surface dose. In this region a plane parallel chamber (with 0.055 cm³ sensitive volume) was used to measure the PDDs at 0, 2 and 4 mm depths for the
three field sizes, using the Khan over-response correction. Plane parallel chamber corrected and uncorrected measurements were compared with the cylindrical ion chamber measurements. Results show that the cylindrical ion chamber overestimates the surface dose enormously and then it underestimates the dose at 2 and 4 mm depths compared to plane parallel chamber measurements and MC calculations. This is expected because the chamber averages the dose along a vertical path, and so overestimates the dose at the start of the buildup and underestimates it when it begins to level off. The correction is reduced as the depth is increased away from the buildup region. The plane parallel chamber corrected PDDs were compared with the MCNP5 calculations at 2 and 4 mm depths. For comparison at the surface, MCNP5 computations were extrapolated to obtain the surface dose. In the buildup region, the maximum deviation between MCNP5 calculations and the plane parallel PDD measurements at 18 MV is 3.8% for 30×30 cm^2 at 2 mm depth. The deviation is within 3% for all other field sizes and depths for both beam energies. For 18 MV, the extrapolation of the MCNP5 PDD overestimates the surface dose for the 30x30 cm^2 field size, and is within 4.6% for all other field sizes at both beam energies.

A focused beam obtained with polycapillary optics was simulated using MCNP5. The focusing of the beam was maintained inside the tissue phantom. Relative dose was calculated inside a tissue phantom along the central axis of the beam from a static lens. The focused beam was found to concentrate the dose within the tumor with sharp fall off before and after the treatment volume. The ratio of peak tumor to skin dose for this idealized case of a tiny tumor in a spherical patient was 14. Because the focal spot size of the beam is so small, a scan is necessary to irradiate a reasonable size tumor. For
simplicity, a symmetric case was considered first. A cylindrical tumor with volume 63 mm$^3$ was simulated in a spherical tissue of radius 3 cm. The optic was rotated in one plane around the tumor. The peak to skin ratio for the case in which it is possible to rotate the optic around the patient was 78. A linear scan was performed to demonstrate that the skin dose will increase by overlapping the beam at the skin surface. A single slice of rectangular shaped tumor was simulated with dimension $1 \times 0.02 \times 1$ cm$^3$ in a rectangular shape tissue with dimension $6 \times 6 \times 4$ cm$^3$. The ratio of peak tumor to skin dose for linear scan was 18. Rotating the beam as it is scanned will maintain the beam spreading and will give better skin sparing effect than the linear scan. Better skin sparing was observed when the beam was angled as it is translated. The peak to skin dose ratio was 45.

Low energy focused x-ray beams could be used to irradiate the tumors inside soft tissue within 5 cm of the surface. Better skin sparing compared to megavoltage photon beams could be achieved by low energy focused x-ray beam inside the soft tissues.