Monte Carlo Simulation of Medical Linear Accelerator Using Primo Code

By:

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Sudan Academy of Sciences (SAS)

Atomic Energy Council

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degree of Master in Medical Physics

Supervisor:
Elhussien Hassan Mohamed Sirelkhatim

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Examination Committee

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Dedication

I dedicate this work to my family, to all the teachers who helped me, and to my friends.
Acknowledgements

This work was done at the Radiation and Isotopes Centre Khartoum. I thank my supervisor Dr. Elhussien Hassan for his support, encouragement, and advices.

I also appreciate the help from Ms. Roweyda in the study of the PRIMO code.

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Abstract

The use of Monte Carlo simulation has become very important in the medical field and especially in calculations in radiotherapy. Various Monte Carlo codes were developed for simulating interactions of particles and photons with matter.

One of these codes is PRIMO that performs simulation of radiation transport from the primary electron source of a linac to estimate the absorbed dose in a water phantom or computerized tomography (CT). PRIMO is based on PENELope Monte Carlo code. Measurements of 6 MV photon beam PDD and profile were done for Elekta precise linear accelerator at Radiation and Isotopes Center Khartoum using computerized Blue water phantom and CC13 ionization chamber. Omnipro Accept Software was used to control the phantom to measure and verify dose distribution.

Elektalinac from the list of available linacs in PRIMO was tuned to model Elekta Precise linear accelerator.

Beam parameters of 6.0 MeV initial electron energy, 0.20 MeV FWHM, and 0.20 cm focal spot FWHM were used, and an error of 4% between calculated and measured curves was found.

The buildup region $z_{\text{max}}$ was 1.40 cm and homogenous profiles in cross line and inline were acquired.

A number of studies were done to verify the model usability; one of them is the effect of the number of histories on accuracy of the simulation and the resulted profile for the same beam parameters. The effect was noticeable and inaccuracies in the profile were reduced by increasing the number of histories.

Another study was the effect of SSD setup errors on the calculated dose which was compared with the measured dose for the same settings. It was in the range of 2% for 1 cm shift, 3% to 4% for 2 cm shift, 5% to 7% for 3 cm shift, 7% to 9% for 4 cm shift, 8% to 12% for 5 cm shift, but it was higher in the calculated dose because of the small difference between the tuned model and measured dose curves.

Future developments include simulating asymmetrical fields, calculating the dose distribution in computerized tomographic (CT) volume, studying the effect of beam modifiers on beam profile for both electron and photon beams.
البحث

من خصوص البحث

استخدام المحاكاة بتقنية مونتي كارلو أصبح مهماً في المجال الطبي وخاصة في الحسابات بالنسبة للعلاج بالأشعة.

وقد طورت مجموعة من شفرات مونتي كارلو لمحاكاة تفاعل الألكترونات والفوتونات مع المادة. وتحديداً هذه الشفرات هي PRIMO المتفسة داخل فانوس من الماء أو حجم من صور مقطوعة. شفرة PRIMO من شفرة مونتي كارلو تم تشمل القياسات بالنسبة لشعاع فوتونات للفئة 6MV لتغيير الجرعة النسبية مع العمق وأيضاً نمط الجرعة بالنسبة للمعجل الخطي. والمركز القومي للعلاج بالأشعة والطب النووي الخرطوم باستخدام فانوس في التحكم في الفانوس لقياس والمتوافق Omnipro Accept وكان محذوف وغرفة التاين CC13. تم استخدام برنامج Elekta precise linear accelerator فانوسو في المراكز والчистة من PENELOPE. 

القياس تم استخدام طاقة الفوتونات 6 MV وعرض نطاق الطاقة 0.20 MeV لتغيير الطاقة ونقطة ميرة بقعة البورة ونتج عنها نسباً هي 4% بين تقديرات الجرعة المحسنة والمحاسبة.

تم الحصول على أكبر جرعة علي عمق 1.40 cm وتوزيع متجانس للجرعة على حوار الشعاع. 

تم عمل عدد من الدراسات للتكدك من فاعلياً النموذج. هذه الدراسات هي دراسة تأثير عدد الإحداثيات علي قدرة المحاكاة وعلي توزيع الجرعة الناتجة لفسد معاملات الشعاع. وقد كان النتائج لاحظاً بقلة الاخطاء الناتجة في نمط توزيع الجرعة مع زيادة عدد الإحداثيات.

اجريت دراسة أخرى لمعرفة تأثير التغيير في المساكن من المصدر للسطح علي الجرعات المحسنة والمقدمة بالنسبة لنفس الظروف تم تم مقارنتها. ووجد أن الخطأ في مدي 2% بالنسبة لتغيير 2 cm ويتراوح بين 3% إلي 4% بالنسبة لتغيير 1 cm ويتراوح بين 7% إلي 8% بالنسبة لجودة 3 cm وفي المدي من 7% إلي 9% بالنسبة لجودة 4 cm ويتراوح بين 8% إلي 12% بالنسبة لتجربة 5 cm. وقد وجد أن الخطأ أكبر بالنسبة للجرعة المحسنة بالنسبة للفرق البسيط بين النموذج المضبوط ومخططات الجرعات المحسنة.

تضمن التطويرات المستقبلية للعمل محاكاة توفير اشعة غير متجانسة وحساب توزيع الجرعات داخل حجم من صور مقطوعة. وأيضاً دراسة تأثير محددات الشعاع علي نمط الجرعة الناتج بالنسبة لشعاعي الألكترونات والفوتونات.
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Chapter One: Introduction and Literature Review
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1.1 Introduction

Monte Carlo simulation of radiation transport is considered to be the most accurate method to calculate dose distributions [1]. It produces accurate results in regions of tissue heterogeneities, such as lung and surface irregularities, providing the most convenient and accurate method for the simulation of patient treatment dose distributions [2]. The Monte Carlo (MC) method has become a widely used method for modeling linear accelerators in medical physics. In contrast to the other common techniques, the MC method starts from first principles and tracks individual particle histories, thus it takes into account the transport of secondary particles [2]. PRIMO is a program based on the codes PENELOPE 2011, PENEASY, PENEASYLINAC and a graphical user interface that encompasses all these components in a single user-friendly environment. PENELOPE is a set of subroutines for the Monte Carlo simulation of coupled electron and photon transport [3]. PRIMO performs the full Monte Carlo simulation of radiation transport from the primary electron source of a linac downstream to estimate the absorbed dose in a phantom or computerized tomography [3].

This project was carried out at Radiation and Isotopes Center Khartoum (RICK) to simulate Elekta linear accelerator photon energy using Primo Monte Carlo code.

PRIMO was used to calculate dose distributions in water phantom and to tune the beam the beam parameters of a model to match Elekta Precise Linear Accelerator at the center, then the model was used for various studies.

1.2 Literature Review:

The use of Monte Carlo simulation in the medical field has become very essential. The name Monte Carlo was coined in the 1940s by scientists working on the nuclear weapon project in Los Alamos to designate a class of numerical methods based on the use of random numbers [4]. The knowledge of random numbers is important to understand Monte Carlo simulation. A random number is a number generated for, or part of, a set exhibiting statistical randomness [5]. The ability to generate pseudorandom numbers is important for simulating events, estimating probabilities and other quantities, making randomized assignments or selections, and numerically
testing symbolic results. Such applications may require uniformly distributed numbers, non-uniformly distributed numbers, elements sampled with replacement, or elements sampled without replacement [6].

The functions RandomReal, RandomInteger, and RandomComplex generate uniformly distributed random numbers. RandomReal and RandomInteger also generate numbers for built-in distributions. RandomPrime generates primes within a range. The functions RandomChoice and RandomSample sample from a list of values with or without replacement. The elements may have equal or unequal weights. A framework is also included for defining additional methods and distributions for random number generation [6].

Number of Monte Carlo codes were written during the 1970s for application to medical physics. The codes developed by Patau [7], Nahum and Abou Mandour [8] belong to this group, all of them developed on mainframe computer systems and with a strong emphasis on the simulation of electron transport [7].

Monte Carlo method has many applications in medical radiation physics and five main categories will be considered here, namely Nuclear Medicine, Diagnostic Radiology, Radiotherapy Physics and Dosimetry, Radiation Protection calculations and transport simulation using microscopic Monte Carlo codes. Nuclear medicine is the area where most of the early Monte Carlo calculations in the field were performed [9]. Three main subgroups of applications will be considered, namely detectors, image reconstruction and dosimetry.

The Monte Carlo simulation of detector responses and efficiencies has received most attention. There are studies of detectors which include high energies (and therefore incorporate the transport of electrons), but they generally also include energies low enough to be of interest for nuclear medicine. This is the case with the tabulations of the response of 3 in x 3 in NaI detectors between 100 keV and 20 MeV by Berger and Seltzer [10], and the simulations of planar and cylindrical Ge(Li) detectors between 100 keV and 12 MeV due to Grosswendt and Waibel [11]. Rogers [12] has reported simulations for incident photons above 300 keV using cylindrical detectors of different materials which in some cases include the effect of the detector housing. Calculations below 300 keV including the same effect have been performed by Saito and Moriuchi [13] for NaI(Tl) detectors with different shapes and volumes. The design of positron emission tomography (PET) systems using the Monte Carlo method has received considerable attention and a large number of applications have been developed. Derenzo [14] has simulated
arrays of detectors of different materials (NaI(Tl), CsF, Ge, plastic detectors, etc) and sizes, as well as other parameters of influence such as the effect of inter-crystal septa. He concluded that narrow bismuth germanate (BGO) had the best detection efficiency. The Monte Carlo method has been also used in the design of multi-ring PET cameras used for three-dimensional imaging by Dahlbom et al [15], who have investigated the influence of the inter-plane septa for different source geometries, concluding that the removal of the septa can increase the efficiency of the system by a considerable amount (almost a factor of six). The geometric response of collimators in scintillation gamma-cameras and single photon emission tomography (SPECT) detection systems was determined using Monte Carlo methods [16]. Monte Carlo calculations have been found to be powerful tools to quantify and correct for photon scattering, which usually produces blurring of the image and loss of contrast in nuclear medicine imaging procedures [16]. In SPECT, the original Monte Carlo simulations of Beck et al [17] have provided a method for evaluating scatter correction methods and analyzing the contribution of different orders of scattering to the acquired images. Calculations were strongly based on variance reduction techniques, photons being restricted within a given solid angle, and forced to interact with the medium and to be detected. This Monte Carlo SPECT research program has been fruitfully continued for several years by the group at Duke University, USA [16].

The determination of absorbed dose in different organs due to internal irradiation has traditionally been based on tabulations given by the Medical Internal Radiation Dose (MIRD) Committee of the Society of Nuclear Medicine in USA through their so-called ‘MIRD Pamphlets’. Some of these Pamphlets have made extensive use of Monte Carlo calculations to derive specific absorbed fractions for photons and electrons, that is the fraction of the emitted energy per unit mass of the medium, which is absorbed at a distance from the source, and they were already in use at the time of the 1976 review article by Raeside [9]. A microscopic Monte Carlo code developed at ORNL [18] has been used by Thrner et al in applications related to B-ray dosimetry in tissue-equivalent materials, finding good agreement with experimental dose distributions. Based on the Monte Carlo results of Berger [19], Prestwich et al [20] have evaluated kernels for certain nuclides of interest in radioimmunotherapy by weighting the data for monoenergetic electrons over B-ray spectra.

The main goal of the application of Monte Carlo techniques to diagnostic radiology is the optimization of diagnostic procedures to improve the image-quality /patient-dose ratio. Many
publications in the field have dealt exclusively with radiation protection aspects of different diagnostic techniques, but the amount of scientific work investigating physical quantities or characteristics of the detection systems has been increasing since the early 1980s [16]. A very interesting and general aspect of these applications has been the improvement of sampling techniques from certain photon interactions. Electron transport has been systematically excluded from these simulations for reasons similar to those given for nuclear medicine applications (short electron ranges and negligible bremsstrahlung) [16].

The use of the Monte Carlo method to investigate basic components of the detection system in diagnostic radiology started in the early 1970s. DePalma and Gaper [21] have simulated photographic emulsion layers to obtain modulation transfer functions (MTF) of different emulsions. Morlotti [22] has performed simulations to obtain MTF and x-ray efficiency for fluorescent screens to investigate the performance of different phosphors. A Monte Carlo code which includes binding corrections to coherent and incoherent scattering has been used by Chan and Doi [23] to investigate x-ray energy absorption in different screen phosphors, including energy, angular dependence and quantum noise. The use of an air gap technique as opposed to a grid to reduce scattered radiation from small fields has been proposed by Persliden [24], who has investigated the influence of air gaps for different detectors in various conditions. Applications of the Monte Carlo technique to the study of anti-scatter grids in mammography have been developed by Dance and Day [25] where the effect of divergent beams has been included. The energy response of detectors commonly used in the measurement of x-ray spectra (Ge, Si(Li) and NaI) has been investigated by Chen et al [26], where methods for correcting the measured spectra were examined. In evaluating the potential risks of x-rays in diagnostic radiology, Chan and Doi [27] have calculated the spatial distributions of energy deposition from pencil beams in water slabs, together with 'rad/R' conversion factors and scatter-to-primary ratios for absorbed doses in a phantom. The conversion factors were used to estimate absorbed doses at different locations in the phantom for a given absorbed dose in the recording system, and to select an optimal energy. The factors necessary to obtain the energy imparted to a phantom from measurements of the collision kerma in air have been determined by Persliden and Aim-Carlsson [28]. As an application in computed tomography, a Monte Carlo code has been developed by Beck et al [29] to estimate absorbed doses in specific regions within a cylindrical phantom exposed to rotating x-ray Sources.
A large number of investigations have considered simplified configurations of external electron and photon radiotherapy sources. This is the case with the calculations of Patau et al [30] on the energy and angular spread of electron beams emerging from foils, and studies of Berger and Seltzer [31] on the effect of scattering foils in electron central-axis depth-doses, where the influence of energy and angular spread has been considered. Manfredotti et al [32] have considered a simple geometry to simulate an electron collimator and score distributions of quantities at a phantom surface which were used later in three-dimensional dose-planning simulations.

The Monte Carlo method has been demonstrated to be the most accurate method for dose calculations in radiotherapy. Clinical application of the Monte Carlo method requires detailed information on the beam characteristics including the energy, angular and spatial distributions of the particles in the beam [33]. McCall et al used MC simulations to study the effects of various targets and flattening filters on the mean energy of photon beams [34]. Petti et al investigated the electron contamination in photon beams by simulating a treatment machine head in great detail using a cylindrical geometry package to approximate various components of the linear accelerator [34]. Mohan et al used a practical approach to determine the characteristics of a clinical photon beam by simulating the transport of the particles through the treatment head using the Monte Carlo technique [33]. Rogers et al investigated the sources of electron contamination in a $^{60}$Co beam [34]. Chaney et al simulated a 6MV photon accelerator to study the origins of head scatter [34]. Lovelock et al simulated the photon beams from a Scanditronix MM50 machine to obtain the beam characteristics needed for treatment planning [34]. Sixel et al and Faddegon simulated a Therac-6 treatment head in radiosurgery mode using a cylindrically symmetric geometry [34]. To study the differential beam hardening effect of the flattening filter, Lee simulated the 6 MV beam from a Varian Clinac 2100C accelerator using the EGS4 code [34]. To determine the parameters in their photon source model used for dose calculation in the PEREGRINE system, Hartmann-Siantar et al simulated linacs using MCNP and the EGS4/BEAM code. DeMarco et al 1998 simulated photon beams from Philips SL-15/25 linear accelerators to obtain the phase space information for patient dose calculation [34]. The computer code PENELOPE performs Monte Carlo simulation of coupled electron-photon transport in arbitrary materials for a wide energy range, from a few hundred eV to about 1 GeV [4]. The code was also used by Mazurier et al to simulate photon beams from a Saturne 43 accelerator [34]. The treatment
head of a Siemens MXE accelerator was simulated to design a new flattening filter for the 6 MV photon beam for this machine by Faddegon et al [34]. Verhaegen et al applied the EGS4/BEAM code to the simulation of radiotherapy kV x-ray units [34]. Several general-purpose MC code systems have been used for radiotherapy beam modeling including ETRAN/ITS, EGS4, EGSnrc, MCNP4/MCNP5, PENELLOPE, GEANT3/ GEANT4 [34]. Kung et al studied the effect of the electron beam-on-target radial intensity distribution on dose profiles [35]. Variation of OAFs with the FWHM of the incident electron beam radial intensity distribution was studied [34]. Beatrice Jutemark used the BEAMnrc and the DOSXYZnrc Monte Carlo code system to simulate 6 and 18 MV photon beams for two different field sizes, 10 x 10 cm² and 40 x 40 cm² [1]. The Monte Carlo code PENELLOPE was used by J Sempau et al to simulate electron beams from a Siemens Mevatron KDS linac with nominal energies of 6, 12 and 18 MeV [36]. A Monte Carlo model of an Elekta Precise linear accelerator has been built and verified by measured data for a 6 MV and 10 MV photon beam running with and without a flattening filter in the beam line Marten Dalaryd et al [37]. Sempau et al introduced PENEASY and PENEASYLINAC codes to automate the Monte Carlo simulation of Varian Clinacs of the 600, 1800, 2100, and 2300 series, together with their electron applicators and multileaf collimators, and introduced a graphical user interface (GUI) named PRIMO [38].

1.3 Statement of the problem

The work load of radiotherapy machines in centers limit the access of physicists for these machines. Simulation using Monte Carlo codes to make a model for these machines can provide adequate solution to the problem. Although Monte Carlo calculations take time to produce accurate results, the time can be reduced by suitable choice of beam parameters and number of histories. Monte Carlo simulation is not the only solution to the problem but it produces accurate dose calculation method particularly in heterogeneous patient tissues where the effects of electron transport cannot be accurately handled with conventional, deterministic dose algorithms. The errors in patient setup for radiotherapy sessions result in deviation of the dose delivered. These errors were studied using the tuned model of Elekta Precise linear accelerator which can be used to study the effect of these errors on the patient delivered dose and it also can be used in dosimetry of special cases.
1.4 Methodology to solve the problem

PRIMO is a program based on PENELOPE Monte Carlo code which was used to simulate electron and photon interactions. Owing to a number of specifically developed variance-reduction techniques PRIMO simulates linac geometries efficiently. The linac model can then be used for the estimation of the dose distribution in water phantoms and computerized tomography volumes.

PRIMO was used to tune the photon percentage depth dose and profile of calculated data with the experimental measured data and to estimate the dose distribution in water phantom. The tuned model of the linear accelerator that matches measured data can be used as a powerful tool for verification of the measured data, calculating dose and dose distribution in water phantoms and computerized tomographic volume, studying the effect of setup errors in dose calculations, and can also be used for the purpose of education and training.

1.5 Outcomes

Elekta model was adjusted using PRIMO code to tune Elekta precise linear accelerator at Radiation and Isotopes Center Khartoum.

The model can be used as a verification tool of the dose measured in water in water phantoms. It can be used to study setup errors that can produce inaccuracies on patient dose.

The model can be used in further studies like the effect of beam modifiers such as blocks and wedges on both photon and electron beams and profiles.

1.6 Thesis outline

This thesis is constructed from five chapters as the following: chapter one includes introduction to the work, literature review, problem of the work, methodology to solve the problem, and outcomes of the work. Chapter two background includes the basics of Monte Carlo methods, Medical linear accelerators, Clinical dosimetry and its importance. Chapter three includes the materials used in the work and the methods used. Chapter four shows the results found from the study and their discussion. Chapter five describes the recommendations and the future work that can be done.
Chapter Two: Theoretical Background
Chapter Two: Theoretical Background

2.1 Basics of Monte Carlo Methods

The main idea behind using Monte Carlo method is to estimate the most expected value of some variable. If this variable is obeying normal distribution function then this will be equivalent to finding the mean. Numerically the expected value $E(f)$ of function $f(u)$ is calculated as the following:

$$E(f) = \int f(u) \, g(u) \, du$$  \hspace{1cm} (2.1)

With $g(u)$ being the probability density function. For normal distribution function defined in the interval $[a,b]$ we can write

$$E(f) = \frac{1}{b-a} \int_a^b f(u) \, du$$  \hspace{1cm} (2.2)

Now if we consider a true random number $\xi$, $0 \leq \xi \leq 1$. By its definition $\xi$ must be distributed homogenously in the interval $[0, 1]$. Then the sequence $\xi_i, \xi_{i+1}, ..., \xi_N$ are not distributed normally in this interval. In same time the central limit theory till us that for large $N$ the sequence $f(\xi_i), f(\xi_{i+1}), ..., f(\xi_N)$ are distributed normally in the interval $[a, b]$. Then $E(f)$ is equal to the mean of $f(\xi)$. Because $f(\xi)$ is discrete then we can write

$$E(f) = \frac{1}{N} \sum_{i=1}^{N} f(\xi_i)$$  \hspace{1cm} (2.3)

The random sequence $\xi_i, \xi_{i+1}, ..., \xi_N$ must be generated using good Random Numbers Generator (RNG). These generators use different techniques [39, 40]. One of the most usable techniques is the multiplicative linear congruential RNGs, which use the following equation:

$$\xi_i = (A\xi_{i-1} + B) \text{modulo } M$$  \hspace{1cm} (2.4)

Then a particular sequence can be defined by giving $\xi_0$, $A$, $B$ and $M$ which referred to as seed, multiplier, increment and the module respectively, all are positive integers. Depending on this mathematical base the Monte Carlo simulation has the following general scenario: at first all the possible events of the considered system must be stated and a function to score the evaluated quantity must be adopted. Then $N$ samples have to be sampled randomly and scored using the adopted function. At the end the average of the $N$ scores will be the output of the simulation. In some problems such as particle energy transport the simulation depends on using multi random sampling process.
2.2 Monte Carlo Techniques for radiation transport

In Monte Carlo simulation, the history (track) of a particle is viewed as a random sequence of free flights that ends with an interaction event where the particle change particle direction, loss energy and, occasionally, produces a secondary particle (41). To simplify the radiation transport simulation, number of assumption were adopted. At first the force between the incidents particles (electron or photon) considered to be negligible. When these particles inter the medium it refers to as primary particles and the other particles resulted from the interaction of these primary particles with the medium are called secondary particles. Both of the primary and secondary particles are supposed to move in straight lines between interactions. Each material medium is considered be homogenies, isotropic and amorphous with defined density. The atoms are assumed to be randomly distributed with uniform density. Molecules are considered as set of individual atoms and the result that each material media is considered as admixture of atomic gases. The atomic electrons are assumed to move independently in central potential field of the nucleus forming clouds of negative charges. The mass of the nucleus is considered to approach infinity regarding to mass of the interact particle. To simulate radiation transport the probability of each interaction at the end of each particle step must be known. For this purpose researchers develop the concept of interaction cross sections which represent the probability of each interaction in specific material as function of the energy of the incident radiations. The total atomic cross section \( \sigma_{\text{total}} \) is defined as the following:

\[
\sigma_{\text{total}} = \sum \sigma_i
\]

Where \( \sigma_i \) is the atomic cross section of specific interaction \( i \). Unit of the atomic cross section is barn which equal \( 10^{-28} \) m\(^2\). These cross sections must be provided as function of photon energy for all used elements and materials. The density of materials must also be given because this density will state the linear attenuation coefficient which will state the distance between the successive interactions.

2.2.1 Modeling photon transport

The important photon interactions in the range of energy suitable for different applications considered in medical physics are photoelectric effect, Compton scattering, Raleigh scattering and pair production.
**Photoelectric effect:** in this interaction an incident photon with sufficient energy interacts with electron in bounded shell of an atom of the absorbed medium. As a result of this interaction the photon is absorbed and the electron is ejected with kinetic energy $E_e$ given by

$$E_e = h\gamma - \phi_e$$  \hspace{1cm} (2.8)

Where $h$ is blank constant, $\gamma$ is the frequency of the photon and $\phi_e$ is the binding energy of the ejected electron. Due to the vacancy in the shell of the ejected electron one electron of the less binding shells will be captured. As result a fluorescence X ray will be produced and in some times one or more Auger electrons can be released. These secondary particles are always absorbed near to the position of interaction, therefore they are always ignored in simulation process. Photoelectric effect dominates at low photon energies.

**Compton scattering:** In this interaction the incident photon scatters after collision with an electron from the absorbed medium. In this collision some of photon energy will be transferred to the electron. The energy of the scattered photon is given by

$$h\gamma' = \frac{h\gamma}{1 + \frac{m_0c^2}{h\gamma}(1 - \cos \theta)}$$  \hspace{1cm} (2.9)

Where $m_0$ is the rest mass of the electron. The energy of the electron $E_e$ is given by

$$E_e = h\gamma \frac{\frac{m_0c^2(1 - \cos \theta)}{h\gamma}}{1 + \frac{m_0c^2}{h\gamma}(1 - \cos \theta)}$$  \hspace{1cm} (2.10)

The probability of Compton scatters increases linearly with atomic number $Z$ of the absorbed medium. It’s the dominate interaction in the range of energies of radiotherapy.

**Pair production:** If the photon has energy larger than 1.02MeV and pass near the nuclease then it will interact with its field the result is the disappearance of the photon and the production of negative and positive electrons –electron and positron-. The angle between the propagation directions of these two particles is $180^\circ$ and the kinetic energy of each of them is 0.511MeV. The produced positron will annihilate with an electron when it comes to rest and produce two 0.511MeV photons. The probability of pair production interaction increase with $Z^2$.

**Raleigh scattering:** In this interaction there is no energy transport and the only effect is the scattering of the incident photon. Therefore the effect of this interaction in photon transport is small comparing with other interactions.
In molding photon interaction the scenario start by considering number of photons -histories-. The distribution of these photons in the spatial and frequency dominos is stated by the description of the source. These parameters will be stored and considered one by one. The probability of any photon to travel distance \( r \) without interaction is \( e^{-\mu r} \). Where \( \mu \) is the linear attenuation coefficient which has unit of \( \text{cm}^{-1} \). This coefficient represents the total contribution of all photon interactions in attenuating the incident photons and we can write

\[
\mu = \mu_{\text{Photo electric}} + \mu_{\text{Compton scattering}} + \mu_{\text{Pair production}} \tag{2.11}
\]

Then the probability of this photon to interact at this distance is \( 1 - e^{-\mu r} \). This probability can be simulated by random number \( \xi \) then we can write

\[
\ln(1 - e^{-\mu r}) = \ln(\xi) \tag{2.12}
\]

\[
r = -\frac{\ln(1-\xi)}{\mu} \tag{2.13}
\]

If the position is within the considered volume then an interaction would be sampled using the provided cross sections of interactions. Then the position of the photon in the spatial and frequency domain, and also the value of the scored quantity will be adapted according to the sampled position and interaction. These steps would be repeated until the photon energy reaches stated level called the cut-off energy. To this end the transporting of the photon is stopped and other photon would be considered. All processes would be repeated until all stored photons are finished. One must note that secondary photons which produced during the interaction of the primary photons have to be considered also.

### 2.2.2 Modeling electron transport

The basic interactions of electrons with mater are Moller scattering of electrons from atomic electrons, Bhabha scattering of positrons from atomic electrons, bremsstrahlung photon creation in the nuclear and atomic fields, positron annihilation with atomic electrons, elastic scattering of electrons and positrons from nuclei and excitation of atoms and molecules by electrons and positrons. According to these interactions, electrons deposit there energy to the absorbed medium by two mechanisms inelastic collisions and bremsstrahlung losses. While the effect of elastic scatter is limited to change the direction of electrons, nuclear process tack place only in case of electrons with high energies therefore it always negligible in the considered range of energies in
medical applications. Electron lost small amount of energy per single interaction. Therefore it needs large number of interactions to release its energy in mater, in this case considering interactions one by one is very complex and time consuming. To treat this problem Monte Carlo codes use condensed history approximation algorithms [42] which depend on simulate the global effect of large number of interactions. The distance through which these interactions take place is represented by a single step in particle trajectory. Condensed history algorithms were categorized into two classes, class I and class II [42] depending on the processing of the secondary particles in respect to the primary particles. In class I all secondary interactions are grouped together for each step in the trajectory while in class II only the interactions with small energy losses will be treated in this manner and those with large energy losses will be considered as separated interaction. In some codes the user can stop the sampling process of the interaction of secondary particles and calculate the losses in primary particle energy using the Continues Slowing Down Approximation (CSDA). Such approximations need the transported electrons to undergo large number of interactions to avoid systematic errors.

The basic scenario of electron transport is similar to that of photon but with additional steps to count for the used condensed history techniques and there associated approximations.

2.2.3 Dosimetric quantities

In studies of radiation transport the quantities which can be evaluated using Monte Carlo simulation can be categorized into two main categories: absorbed dose and its related field quantities such as fluence, exposure, kerma, current, and track length and protection quantities used for assessing the exposure limits. The absorbed dose D can be defined as

\[ D = \frac{d\bar{\epsilon}}{dm} \]  (2.14)

With \( d\bar{\epsilon} \) being the mean energy imported to infinitesimal mass dm of a material. In case of uncharged ionizing radiations the absorbed dose can be approximated by initial kinetic energies of all charged ionizing particles liberated by uncharged ionizing particles in a material of mass dm this quantity referred to as kerma. This approximation can be used when there is charge particles equilibrium. Kerma itself can be defined as

\[ K = (\frac{d\nu}{\rho})\Phi \]  (2.15)
Where \( \frac{\mu N}{\rho} \), \( E \), and \( \phi \) are the mass energy transfer coefficient, particle energy, and particles fluence respectively. The fluence represents the number \( N \) of particles which goes through infinitesimal sphere with cross sectional area \( dA \). It can be defined as

\[
\phi = \frac{dN}{dA}
\]  

(2.16)

The fluence can also be expressed by mean of particle track \( ds \)

\[
\phi = \frac{\sum ds}{dA}
\]  

(2.17)

We have to note that dose, kerma, and fluence are related to a point in the considered medium and they are not useful for radiation protection purposes. This fact leads to the invention of protection quantities which consider the risk of the absorbed radiations on tissue or organs of human body. The first quantity is the mean absorbed dose \( D_T \) in tissue or organ \( T \) which defined as

\[
D_T = \frac{\varepsilon_T}{m_T}
\]  

(2.18)

Where \( \varepsilon_T \) is the mean total energy imported in \( T \) and \( m \) is the mass of \( T \). The second quantity is the equivalent dose \( H_T \) which was introduced to account for the variation on the biological effects due to the same mean dose from different radiations. \( H_T \) is defined as:

\[
H_T = \sum_R w_R D_{T,R}
\]  

(2.19)

Where \( w_R \) is the weighting factor of the considered type of radiation and \( D_{T,R} \) is the mean absorbed dose in organ \( T \) due to this radiation. The third quantity is the effective dose \( E \) which introduced to account for the variation of the biological effect of the same equivalent dose when absorbed in different organs

\[
E = \sum_T w_T H_T
\]  

(2.20)

Where \( w_T \) is the tissue weighting factor. Monte Carlo simulation has central role in connecting the protection quantities which are immeasurable with the different measurable quantities of radiation fields.

### 2.2.4 Variance reduction

The efficiency \( \eta \) of computational calculations is defined by the following equation:

\[
\eta = \frac{1}{\sigma^2_T}
\]  

(2.5)

15
Where $T$ is the calculation time and $\sigma^2$ is the variance. In Monte Carlo calculations, the variance is proportional to the inverse of the number of histories $N$ as can be seen from the following equation:

$$\sigma^2 = \frac{\sum_{i=1}^{N}(x_i - \bar{x})^2}{N}$$  \hspace{1cm} (2.6)

In the same time $N$ is proportional to $T$. Therefore, the equation 4.12 tells us that the efficiency is dependent of the number of histories. So to increase the efficiency one has to reduce $T$ or $\sigma^2$. For most of Monte Carlo algorithms $T$ is constant and can be reduced only by changing the algorithm itself. The other choice is to reduce $\sigma^2$ which can be achieved using what called variance reduction techniques [43, 44]. Variance reduction techniques are time consuming by itself therefore one has to balance between the final reduction in variance and the increasing in calculation time. Beside its effects on time the approximations of variance reduction techniques can also affect the accuracy of the evaluated quantity itself. Variance reduction techniques are quantity, geometry, density and energy dependable and the technique which is helpful in some case can be harmful in other.

In many cases there is no interest on the particles which go outside of the considered volume or get weight under specific threshold, therefore they discarded from calculations. This process can be achieved using Russian roulette technique. The idea of this technique is to eliminate some particles with specific probability depending on their positions, directions or importance. Due to the lost of some particles the weight of the survival particles must be increased. Other method to reduce variance is to force particles to go through the volumes of interest. One way to achieve this is to copy number of particles with suitable position and direction and using them in simulation, this is what called particle splitting technique. In this technique although of the same starting point for the original and copied particles in the spatial and frequency domain but they will propagate differently because of the random numbers used to state the next steps for each of them. The statistical weight of the original particle can be divided between the original and copied particles to keep the statistical balance of the simulation. Other technique is what called interaction forcing. In this technique the particle is forced to interact through specific probability when they pass the volume of interest. Some variance reduction techniques depend on using the true physics instead of random sampling. An example of such techniques is exponential transform techniques in which the path lengths are adopted according to the rules of exponential attenuation of the particles instead of random sampling for this length. For any simulation there
is a level of particle energy under which the interactions of this particle has no importance effect on the evaluated quantity therefore using energy cut-off will be helpful method to reduce variance.

Variance reduction can be also achieved through controlling step size used with condensed history algorithms. This size must satisfy the collection of suitable number of interactions to minimize statistical variations and to simulate the curved trajectory of electron with negligible amount of error. Some algorithms were developed to optimize the size step such as PRESTA [45] and PREASTA II [46]. In some cases the step size can be larger than the dimension of the considered volume in the direction of propagation and no step can be sampled. In such cases the only solution is to reduce the step size. In other cases one of steps can cross the boundary between two materials then the calculation of its associated quantities such as deposit energy will be interrupted because of the two different set of data related to each of the materials in the two sides. The solution in this case is to discard this step from the calculations. To reduce the discarded steps a strategy to minimize the step size near boundaries must be applied using what called boundary crossing algorithms.

2.2 Medical Linear Accelerators

Kilovoltage x-ray beams are useful for the treatment of skin lesions and shallow tumors, but for deep-seated tumors, the dose that can be delivered is limited by the high skin dose. Megavoltage beams are not only more penetrating, but they have the major benefit that the maximum dose is delivered below the skin surface [47]. In addition, because the principal interaction with tissue is through the Compton effect, the locally absorbed dose is not dependent on the atomic number of the tissue, and the dose to bone is not enhanced.

Medical linear accelerators (linacs) have become the predominant machine in treatment of cancer with ionizing radiation during the past few decades. In contrast to linacs used for high-energy physics research, medical linacs are compact machines mounted isocentrically so as to allow practical radiation treatment aiming the beam toward the patient from various directions [48]. Medical linacs are cyclic accelerators which accelerate electrons to kinetic energies from 4 MeV to 25 MeV using non-conservative microwave radiofrequency (RF) fields in the frequency range from 103MHz to 104 MHz, with the vast majority running at 2856 MHz.

In a linear accelerator the electrons are accelerated following straight trajectories in special evacuated structures called accelerating waveguides. Electrons follow a linear path
through the same, relatively low, potential difference several times; hence, linacs also fall into the class of cyclic accelerators.

2.2.1 Linac Generations:
Medical linacs have gone through five distinct generations. The first generation was Low energy photons 4–8 MV that had straight-through beam; fixed flattening filter; external wedges; symmetric jaws; single transmission ionization chamber; isocentric mounting. The second generation Medium energy photons (10–15 MV) and electrons had bent beam; movable target and flattening filter; scattering foils; dual transmission ionization chamber; electron cones. The third generation with High energy photons (18–25 MV) and electrons had dual photon energy and multiple electron energies; achromatic bending magnet; dual scattering foils or scanned electron pencil beam; motorized wedge; asymmetric or independent collimator jaws. The fourth generation used High energy photons and electrons had Computer-controlled operation; dynamic wedge; electronic portal imaging device; multi leaf collimator. The fifth generation used High energy photons and electrons with photon beam intensity modulation with multileaf collimator; full dynamic conformal dose delivery with intensity modulated beams produced with a multileaf collimator; on-board imaging for use in adaptive radiotherapy.

2.2.2 Components of Modern Linacs:
The linacs are usually mounted isocentrically and the operational systems are distributed over five major and distinct sections of the machine which are: the Gantry, Gantry stand or support, Modulator cabinet, Patient support assembly (treatment couch), Control console.
The beam-forming components of medical linacs are usually grouped into six classes which are: The Injection system, the radiofrequency power generation system, Accelerating waveguide, Auxiliary system, Beam transport system, Beam monitoring system and beam collimation.
The injection system is the source of electrons, essentially a simple electrostatic accelerator called an electron gun. Two types of electron gun are in use: diode type and triode type, both containing a heated cathode (at a negative potential of the order of−25 kV) and a perforated grounded anode. In addition, triode type gun also incorporates a grid placed between the cathode and the anode. Electrons are thermionically emitted from the heated cathode, focused into a pencil beam and accelerated toward the perforated anode through which they drift into the accelerating waveguide.
The radiofrequency (RF) power generating system produces the high power microwave radiation used for electron acceleration in the accelerating waveguide and consists of two components: the RF power source and the pulsed modulator. The RF power source is either a magnetron or a klystron in conjunction with a low power RF oscillator. Both devices use electron acceleration and deceleration in vacuum for production of the high power RF fields. The pulsed modulator produces the high voltage, high current, short duration pulses required by the RF power source and the electron injection system.

Electrons are accelerated in the accelerating waveguide by means of an energy transfer from the high power RF field which is setup in the accelerating waveguide and produced by the RF power generator. The accelerating waveguide is in principle obtained from a cylindrical uniform waveguide by adding a series of disks with circular holes at the center, positioned at equal intervals along the tube. These disks divide the waveguide into a series of cylindrical cavities that form the basic structure of the accelerating waveguide in a linac.

The auxiliary system of a linac consists of several basic systems that are not directly involved with electron acceleration, yet they make the acceleration possible and the linac viable for clinical operation. These systems are: the vacuum-pumping system, the water-cooling system, the air-pressure system, and the shielding against leakage radiation.

The electron beam transport system brings the pulsed high-energy electron beam from the accelerating waveguide onto the target in the x-ray therapy mode and onto the scattering foil in the electron therapy mode.

The beam monitoring and beam collimation system forms an essential system in a medical linac ensuring that radiation dose may be delivered to the patient as prescribed, with a high numerical and spatial accuracy.

The linac head contains several components, which influence the production, shaping, localizing, and monitoring of the clinical photon and electron beams. The important components found in a typical head of a modern linac include: Several retractable x-ray targets, Flattening filters and electron scattering foils (also referred to as scattering filters), Primary and adjustable secondary collimators, Dual transmission ionization chambers, Field defining light and range finder, Optional retractable wedges or full dynamic wedges, Multileaf collimator MLC.
Clinical photon beams are produced in medical linear accelerators with a target/flattening filter combination. The electron beam accelerated to a given kinetic energy in the accelerating waveguide is brought by the beam transport system onto an x-ray target in which a small fraction (of the order of 10%) of the electron pencil beam kinetic energy is transformed into bremsstrahlung x rays. The intensity of the x ray beam produced in the target is mainly forward peaked and a flattening filter is used to flatten the beam and make it useful for clinical applications. Each clinical photon beam produced by a given electron kinetic energy has its own specific target/flattening filter combination.

Photon beam collimation in a typical modern medical linac is achieved with three collimation devices: the primary collimator, the secondary movable beam defining collimator, and the multi leaf collimator (MLC). The primary collimator defines a maximum circular field which is further truncated with the adjustable rectangular collimator consisting of two upper and two lower independent jaws and producing rectangular or square fields with a maximum dimension of 40×40 cm$^2$ at the linac isocenter, 100 cm from the x-ray target.
The MLCs are a relatively new addition to modern linac dose delivery technology. In principle, the idea behind an MLC is simple. It allows production of irregularly shaped radiation fields with accuracy and efficiency and is based on an array of narrow collimator leaf pairs, each leaf controlled with its own miniature motor. The building of a reliable MLC system presents a substantial technological challenge and current models incorporate up to 120 leaves (60 pairs) covering radiation fields up to 40×40 cm² and requiring 120 individually computer-controlled motors and control circuits.

Clinical electron beams are produced in a medical linac by retracting the target and flattening filter from the electron pencil beam and either scattering the electron pencil beam with a scattering foil or deflecting and scanning the pencil beam magnetically to cover the field size required for electron beam treatment. Special cones (applicators) are used to collimate the clinical electron beams.

Dose monitoring systems in medical linacs are based on transmission ionization chambers permanently imbedded in the linac clinical photon and electron beams. The chambers are used to monitor the beam output (patient dose) continuously during the patient treatment. In addition to dose monitoring the chambers are also used for monitoring the radial and transverse flatness of the radiation beam as well as its symmetry and energy. For patient safety, the linac dosimetry system usually consists of two separately sealed ionization chambers with completely independent biasing power supplies and readout electrometers. If the primary chamber fails during patient treatment, the secondary chamber will terminate the irradiation, usually after an additional dose of only a few percent above the prescribed dose has been delivered.

2.3 Clinical Dosimetry Measurements for Linear Accelerators

A number of measurements have to be done to provide necessary information for models in the Treatment Planning System (TPS) for patient dose calculations. These commissioning measurements are performed after many preliminary tests of acceptance to verify that the performance of the equipment meets manufacturer’s specifications.

Data of the machine are entered to the TPS after measurements in phantoms because it is impossible to perform measurements within the patient’s body [47].

Dosimetry measurements are grouped to absolute and relative. Absolute dosimetry is the measurement of dose at reference conditions based on published codes of practice but it is
essential to measure the dose for different range of conditions that mimic clinical use. The measurements are normalized to the reference conditions to measure relative dose.

The ionization chamber is the most commonly used dosimeter for measuring the absorbed dose, which is due to the accuracy, instant readout, constant sensitivity over the detector lifespan, and good understanding of the necessary corrections [49]. The ionization chamber contains a sensitive volume filled with air, from which the ionization charge $Q$ produced by the radiation in the sensitive air mass $m_{air}$ is collected with a central electrode.

The absorbed dose to water at the reference depth $z_{ref}$ in water for a reference beam of quality $Q_0$ and in the absence of the chamber is given by

$$D_{w,Q_0} = M_{Q_0}N_{D,w,Q_0}$$

where $M_{Q_0}$ is the reading of the dosimeter under the reference conditions used in the standards laboratory and $N_{D,w,Q_0}$ is the calibration factor in terms of absorbed dose to water of the dosimeter obtained from a standards laboratory [50]. In most clinical situations the measurement conditions do not match the reference conditions used in the standards laboratory. This may affect the response of the dosimeter and it is then necessary to differentiate between the reference conditions used in the standards laboratory and the clinical measurement conditions. Reference conditions are described by a set of values of influence quantities for which the calibration factor is valid without further correction factors [50]. The reference conditions for calibrations in terms of absorbed dose to water are, for example, the geometrical arrangement (distance and depth), the field size, the material and dimensions of the irradiated phantom, polarity effect, and the ambient temperature, pressure and relative humidity.

When a dosimeter is used in a beam of quality $Q$ different from that used in its calibration, $Q_0$, the absorbed dose to water is given by

$$D_{w,Q} = M_{Q}N_{D,w,Q_0}k_{Q,Q_0}$$

where the factor $k_{Q,Q_0}$ corrects for the effects of the difference between the reference beam quality $Q_0$ and the actual user quality $Q$, and the dosimeter reading $M_{Q}$ has been corrected to the reference values of influence quantities, other than beam quality, for which the calibration factor is valid. The most common reference quality $Q_0$ used for the calibration of ionization chambers is $^{60}$Co gamma radiation [50].
Examples of beam data input required for TPS are: Machine name, modality, energy, couch, gantry, and collimator limits, SSD limits, source to collimator distance, source to wedge distance, in water output, wedge type and factor, tray factor collimator transmission, etc. After acceptance testing and commissioning of the machine, a regular quality control tests are performed to check that the machine is working properly and to avoid problems that might happen in the treatment. Results of these tests are compared with that of acceptance and commissioning tests.

Quality control tests are specified by frequency and tolerance, they are grouped to mechanical, radiation, and safety tests. Radiation tests for photon include output constancy and linearity, wedge and tray factors, beam quality, radiation leakage and survey, etc. For the weekly output check using a recommended dosimeter the tolerance is 2%, and for daily beam output constancy check the tolerance is 3%. Radiation tests for electron include output constancy and linearity, output factors for different applicators, virtual source position, beam quality, etc [51].

2.4 Clinical Dosimetric Calculations

The dose predicted by a calculation method should correspond to the real absorbed dose in the patient as accurately as possible [41]. Calculating the distribution of dose in a patient is not so straightforward. Before computerized treatment planning systems became widely available, dose distributions were calculated manually by the addition of percentages estimated from the superposition of isodose charts. The methods used involved empirically derived corrections to account for patient shape and inhomogeneities. It was time consuming and relied heavily on the experience of the planner. Modern computer technology has allowed increasingly sophisticated techniques to be routinely applied to treatment planning and has widened the aims and scope of treatment planning itself [47].

Methods based on empirical formulae [52], were the first techniques developed for dose calculation. In these methods, the primary and scattered radiation components are treated separately, since they have different physical behavior in a material. The primary component describes the distribution of the energy deposited by the first photon interaction in the material, and the scatter component describes the result of the subsequent interactions. In the method presented by Cunningham [53], the scatter is computed with the help of a scatter-air ratio (SAR),
which is derived from a measured tissue-air ratio (TAR). Handling of irregular field shapes is typically based on the integration method developed by Clarkson [54], but the method has difficulties to model generalized beam setups.

Various methods have been developed to account for the fact that the tissue density differs from the water density. Commonly, the dose distribution calculated for the homogeneous water-equivalent situation is converted into the heterogeneous situation in the same geometry by applying a point-by-point correction factor. Most methods, such as the Batho power-law method [55], determine the correction factor by a direct ray-tracing from the primary radiation source to the point of interest. More sophisticated techniques, such as the equivalent TAR method [56,57], use the electron density data from the CT image to determine the correction factors. The use of these correction factors may still lead to deviations up to 10% from the measured dose for certain type of geometries.

Kernel-based or convolution/superposition dose calculation methods are based on physical principles of the radiation behavior rather than on direct beam data measurements. Energy deposition kernels can be used to model the photon transport, since the energy deposition around the primary interaction site is independent of the position of the site in homogeneous media. Kernel-based methods are able to compute the dose directly for irregular photon beams in heterogeneous phantoms. Non-water equivalent tissues are typically taken into account by scaling the kernels with the mean electron density between the interaction point and the dose calculation point [58]. In the 3D point-spread kernel methods [59–62], the dose deposition can be viewed as a superposition of appropriately weighted responses to point irradiations. However, the point-spread kernel methods are typically still computationally expensive. In order to overcome this problem, other methods based on the superposition of 2D pencil-beam kernels have been developed [63–66]. When using the 2D pencil-beam kernels, the heterogeneities cannot be fully corrected, but the calculation times can be significantly smaller than in the 3D point-spread kernel based methods. Both the point-spread and pencil-beam kernels are usually derived from Monte Carlo (MC) simulations, although some authors have used analytical expressions to compute the first and multiple scatter kernels [60, 67].

The Monte Carlo method has been shown through many research studies to calculate accurate dose distributions for clinical radiotherapy, particularly in heterogeneous patient tissues where the effects of electron transport cannot be accurately handled with conventional, deterministic
dose algorithms [68]. Despite its proven accuracy and the potential for improved dose distributions to influence treatment outcomes, the long calculation times previously associated with MC simulation rendered this method impractical for routine clinical treatment planning. However, the development of faster codes optimized for radiotherapy calculations and improvements in computer processor technology have substantially reduced calculation times to, in some instances, within minutes on a single processor. These advances have motivated several major treatment planning system vendors to embark upon the path of MC techniques. Several commercial vendors have already released or are currently in the process of releasing MC algorithms for photon and/or electron beam treatment planning [68]. Consequently, the accessibility and use of MC treatment planning algorithms may well become widespread in the radiotherapy community. With MC simulation, dose is computed stochastically using first principles; this method is therefore quite different from conventional dose algorithms. Issues such as statistical uncertainties, the use of variance reduction techniques, the ability to account for geometric details in the accelerator treatment head simulation, and other features, are all unique components of a MC treatment planning algorithm [68].
Chapter Three: Materials and Methods
Chapter Three: Materials and Methods

3.1 Materials

3.1.1 Elekta Precise Linear Accelerator

There are two Linear accelerators and three Cobalt machines in Radiation and Isotopes Center Khartoum. Elekta Precise Linear Accelerator is one of the external beam radiotherapy machines. It was installed and accepted in 21 January 2011 and patients’ treatment began in 17 August 2011. It has 3 photon energies, 6, 10, and 15 MV respectively and 5 electron energies 4, 6, 9, 12, and 18 MeV respectively.

![Elekta Precise Linear Accelerator](image)

Figure 3.1: Image showing Elekta Precise Linear Accelerator installed in RICK and Simulated in this study

3.1.2 IBA Wellhofer Water Phantom

The computerized IBA Wellhofer water phantom was used for dose measurements. The phantom enables taking PDDs and profile by a holder that moves the chamber in Z, X, and Y directions respectively; these movements are made by connecting the phantom with the Omnipro Accept Software. The software controls the phantom to measure and verify dose distribution. It is also
used for analyzing the measured dose distribution for quality assurance, calibration of radiation devices, or in commissioning to input data to a treatment planning system TPS.

3.1.3 Slab Phantom

Solid phantoms in slab form PMMA are water equivalent plastics that were used for relative dosimetry measurements in the study. There are made of different thicknesses of 0.1 cm, 0.2 cm, 0.5 cm and 1.0 cm respectively. There is a slab of 2 cm thickness to position the chamber inside it.

3.1.4 Ionization Chambers

The ionization chamber CC13 (Compact Chamber) is a cylindrical chamber that used in the work. The chamber is the production of Scanditronix Wellhofer IBA with calibration factor $N_{DW}$: $2.63 \times 10^8$ Gy/C, voltage $V=300$ V, $T=15.101$ C, $P=97.01$ KPa.

The farmer Chamber FC-65 P is a cylindrical chamber production of Scanditronix Wellhofer IBA with serial number 577, $N_{DW}:4.823 \times 10^7$ Gy/C, $V=300$ V.
Figure 3.3: Image showing Slab phantom used in the study

Figure 3.4: Image showing CC13 ionization chamber used in the work.

Figure 3.5: Image showing FC-65 P ionization chamber used in the study of setup errors.
3.1.5 Unidos E Electrometer

UNIDOS E electrometer is PTW product that used for dose measurements for the study of the effect of SSD setup errors on point dose for different depths.

Figure 3.6: PTW UNIDOS E Electrometer used in the study of setup errors.

3.1.6 Primo Monte Carlo Code

PRIMO is the program used for the simulation and tuning of the linear accelerator. It is a program based on the codes PENELOPE 2011 [Bar+95; SFS11; Sem+97], PENEASY [SBB11], PENEASYLINAC [SBB11] and a graphical user interface that encompasses all these components in a single user-friendly environment. PENELOPE is a set of subroutines for the Monte Carlo simulation of coupled electron and photon transport. PENEASY is a general-purpose main program for PENELOPE that includes several source models, tallies, variance-reduction techniques and the possibility of combining quadric and voxelized geometries. PENEASYLINAC is a complementary tool that generates the input files required for the simulation of most Varian1 and Elekta2 linacs with PENELOPE/PENEASY [3]. PRIMO is also used for estimation of the dose distribution in water phantoms and computerized tomographies. Knowledge of the Monte Carlo method, of programming, of the peculiarities of PENELOPE and of the physics of radiation transport is not necessary in order to set up, run and analyze the simulation of a linac and the subsequent dose distribution because PRIMO has easy GUI. Users of other Monte Carlo codes can also benefit from PRIMO thanks to the possibility of importing and simulating external phase-space files written in the IAEA format.
There are different types of linacs which can be simulated using PRIMO. Table summarizes the available Linacs in PRIMO and the commercial names that they might have.

<table>
<thead>
<tr>
<th>PRIMO</th>
<th>Commercial</th>
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<tr>
<td>Elekta SL</td>
<td>SL series</td>
</tr>
<tr>
<td>Elekta MLCi</td>
<td>SLi Plus, Axesse, Affinity, Synergy, Precise</td>
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<tr>
<td>Varian Clinac 600C</td>
<td>Clinac 600 C</td>
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<tr>
<td>Varian Clinac 600CD</td>
<td>Clinac 600 C/D</td>
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<td>Varian Unique</td>
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<td>Varian Clinac 2100</td>
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<tr>
<td>Varian Clinac 2300</td>
<td>Clinac 2300 C/D</td>
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When a primary electron enters the modeled geometry, upstream of the upper part of the linac which is from the primary electron source to the flattening filter or scattering foil, an
electromagnetic shower is simulated. It may occur that the primary electron is absorbed or escapes the geometry without further consequences, or it may happen that the primary electron produces secondary particles, namely, electrons, photons or positrons. In turn, these secondary particles may produce another generation of particles, and so on. The primary particle and all its descendants are simulated until all of them have been either absorbed or escaped the geometry. When this occurs one history has been completed. Therefore, the number of simulated particles and the number of simulated histories, in general, do not coincide.

When simulating radiation transport with the Monte Carlo method it is possible to define a surface, usually a plane, at any location in the geometry. Particles traversing this plane are stopped and their state (i.e., energy, position, direction of flight, etc.) recorded on a file called phase-space file. When a phase-space file is ‘sufficiently rich’, that is, it contains a ‘large number’ of particles, it is possible to neglect the geometry upstream of the phase-space surface, and to consider the phase-space file as the radiation source for subsequent Monte Carlo simulations.

The code allows tallying a phase-space file at the downstream end of the upper part of the linac. In PRIMO the upper part of the linac is called segment 1 (s1). Similarly, a phase-space file can be tallied at the downstream end of the lower part of the linac which consist of one or two sets of jaws and possibly a multileaf collimator for photon beam or an applicator for electron beam. This region is called segment 2 (s2). The part of the simulation dedicated to the dose estimation is called segment 3 (s3).

Segments must, obviously, be simulated in sequential order, that is s1; s2; s3. However, they can be grouped according to the user’s requirements. They can be simulated individually as (s1; s2; s3); or grouped in a single simulation as (s1 + s2 + s3); or in smaller groups simulating s1 first and then s2 and s3 together (s1; s2 + s3); (s1 + s2; s3) is also possible.

A simulation can either tally a phase-space file or a dose distribution. Therefore, if simulating for example (s1 + s2 + s3) a dose distribution will be tallied. Simulation of (s1 + s2) and then a subsequent simulation of s3 will produce a phase-space file at the downstream end of the lower segment during the first simulation and a dose distribution during the simulation of s3.

The phase-space file obtained with the simulation of s1 depends on the primary beam parameters and the number of histories simulated. Once the primary beam parameters of a linac have been tuned for a given nominal energy to reproduce experimental data from that linac, it is desirable to
run, once and for all, a long simulation of s1 that can be re-used in subsequent simulations of the rest of the linac. This approach conduces to a substantial saving in simulation time, particularly in the case of photon beams. Linacs operating in electron mode have one additional segment, namely, (s1; s2; s2e; s3). s1 and s3 correspond to the segments previously described. Segment s2 simulates the movable collimators (i.e., the jaws) and the uppermost two scrapers of the electron applicator, tallying a phase-space file at the downstream end of the middle scraper. Segment s2e simulates the lowermost scraper and tallies a phase-space file at its downstream end. If the electron field is standard, as conformed by the electron applicator, then s2 and s2e should be simulated together. However, if the user is interested in adding a customized collimator at the lowermost scraper it is necessary to simulate only up to s2. In that case, segment s2e with the customized collimator must be simulated with an external program, such as, PENELOPE. When importing external phase-space files, PRIMO assumes that they have been tallied at the downstream end of s1. After importing the phase-space file, s1 will appear as already simulated and the user will be given the possibility of either simulating (s2 + s3) or (s2; s3).

To reduce this unaffordable amount of computing time the variance-reduction techniques can be used. PRIMO has three variance reduction techniques. For nominal energies below 15 MV (photon mode) it is recommended to use splitting roulette for s1. For nominal energies above 15 MV rotational splitting is usually more efficient. Regarding simulation of s3, a splitting factor of 100 usually works fine. It is advisable to check the estimated time after launching the simulation of s3. If the estimated time with a splitting factor of 100 is exceedingly long then reset the simulation, modify the splitting factor for s3 and launch it again. PRIMO reports dose in units of eV/g per primary particle. These units are equivalent to Gy/(mA s), whence the dose in Gy can be calculated knowing the current intensity at the target in mA and the irradiation time in s. When comparing with experimental profiles relative dosimetry is assumed.

3.2 Methods

3.2.1 Commissioning of Experimental Data

3.2.1.1 Measurement of Percentage depth dose for 6 MV photon beam

To measure the percentage depth dose for the photon beam the level of the phantom was adjusted and then the phantom was adjusted until its center was with the cross hair of the radiation beam.
The phantom was filled up to third quarters with distilled water to be the medium of measurement. Using the hand switch of the phantom the scan regions and level were indicated in X, Y, and Z dimensions. The ionization chamber was centered with the cross hair of the beam using the buildup cap and then fixed in the holder. Range of depth was from zero up to 20 cm, the field size was 10x10 cm² and the Source Surface Distance (SSD) was 100 cm.

3.2.1.2 Measurement of beam profile for 6 MV photon beam (cross line)
To measure the beam profile the same setting of 3.2.1.1 was used but with the chamber moving in X direction (cross line) in different depths.

3.2.1.3 Measurement of beam profile for 6 MV photon beam (In line)
To measure the beam profile the same setting of 3.2.1.1 was used but with the chamber moving in Y direction (in line) in different depths.

Figure 3.8: Positioning of the chamber inside the phantom for measurement using the buildup cap

3.2.2 Preparation of Initial files of experimental data
The files of experimental data for comparison with calculated dose were prepared as a Dat text file composed of the X, Y, Z (depth), and dose information. In the PDD file both X and Y coordinates were zero with the change in dose with depth. In the cross line and in line profile files the dose was changed according to X and Y coordinates respectively with different depths. Figure 3.9 show part of the experimental data file that used for the comparison.
installation of PRIMO Code

The PRIMO code was installed on a computer with Intel R core TM i5 2450 M CPU and 2.5 GHz processor, the operating system windows 7 version 6.1.7601 and the system type 64 bit. The installed physical memory (RAM) is 4.0 GB. PRIMO occupied less than 100 MB of disk space after installation. To make sure that the Primo code was installed correctly, Example1 from the list of examples available in the web page of Primo [69] was run. This example covers the simulation of all segments of a linac in the sequence s1; s2; s3 to tally intermediate phase-space files and the dose distribution in a water phantom. The configuration chosen for this example is the following: Linac Varian Clinac 2100 C/D, mode is Photon with nominal energy 6 MV, Field size 10x10 cm² without MLC, dose tallying in water phantom, SSD 95 cm, bin size 0.2x0.2x0.2 cm³ dose tallying volume 16.2x16.2x31.0 cm³.

3.2.4 Simulation of Elekta Precise Linear Accelerator using PRIMO code

3.2.4.1 Tuning of the existed Elekta machine model

To construct the machine, a new project was created and named LinacBCompare with the linac model Elekta and operation mode photon. From the screen of the Simulation Setup and in S1 from the simulation segments, the nominal energy used was 6MV, initial energy 6MeV, energy FWHM 0.20 MeV, Focal spot FWHM 0.20 cm. Then the segment S2 was prepared with the Gantry, couch, and collimator angles as zero, symmetrical field size of 10x10 cm² without MLC. The segment S3 was run to tally the dose in a water phantom with dimensions of 40x40x30 and a bin size of 0.20x0.20x0.20. A number of four parallel processors were used in the simulation and the splitting roulette was used for variance reduction.

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Figure 3.9: Experimental data file of the measured PDD used for tuning process.
After finishing the tuning process, the calculated PDD from the simulation was compared with the experimental data of the linac that prepared as mentioned in section 3.2.2. Lateral profiles in both X and Y axes from the tuned model were also compared with the profiles of the 6MV profile.
3.2.5 Evaluation of Histories effect on the accuracy of simulation

To study the effect of the number of histories on the accuracy of the simulation, a number of simulations were done using approximately the same parameters of initial energy, energy FWHM, Focal spot FWHM, field parameters, and phantom dimensions with the change in the number of histories. This study was so important because it also helped in the tuning process of linac model.

3.2.6 Studying the effect of SSD set up errors on the delivered dose

To study the effect of the setup errors that may result from SSD shift on delivered patient dose, the dose was measured with 100 cm as standard for 3, 5, 10, and 15 cm depths respectively using slab phantom and FC-65 P ionization chamber. Then the dose was measured at these depths with 1 cm increment from 101 cm to 105 cm. Using the same parameters the dose was calculated by adjusting the SSD on S3 segment of the tuned linear accelerator model.

Figure 3.12: Image showing the setting for dose measurements using slab phantom for the study of setup errors
Figure 3.13: Dose calculated at 15 cm depth with 101 cm SSD.
Chapter Four: Results and Discussion
Chapter Four: Results and Discussion

4.1 Results

This chapter shows the results of study from the installation of PRIMO code to the tuning of the calculated PDD and profile with the measured data.

4.1.1 Results of Installation of PRIMO code

The Primo code with version 0.1.1.3 was installed on a computer with 2.5 GHz processor, the operating system windows 7. PRIMO occupied less than 100 MB of disk space after installation. The code was in installed in the C partition and after clicking the icon of PRIMO we get the new project window. Figure 4.1 show the dose curves after running example 01 of 6MV Varian Linac.

![Figure 4.1: Dose profile and PDD after S3 segment for Varian CLinac 2100.](image)

4.1.2 Results of simulation of Elekta Precise linear accelerator using PRIMO code

A number of experiments were done to tune the calculated PDD and profile using PRIMO with Elekta precise linear accelerator measured data. The primary beam parameters were adjusted to get an acceptable difference with the measured data. The initial energy was changed from 5.90 to 6.50 MeV with an increment of 0.10 MV, the energy FWHM was modified with in steps of 0.10
from 0.00 to 0.50 MeV, the focal spot FWHM was set as 0.10, 0.20, and 0.30 cm respectively. The SSD and field size were fixed at 100 cm and 10x10 cm$^2$ respectively. The following figures show the tuning process of the 6MV Elekta Linear accelerator. Six different experiments were done to tune the accelerator. Experiment one was with beam parameters $E$=6 MeV, Energy FWHM=0.10 MeV, Focal spot FWHM=0.10 cm. The dose curve shown in Figure 4.2 resulted in buildup region $z_{\text{max}}$ (Depth of maximum dose) of 1.20 cm which is 0.20 cm lower than $z_{\text{max}}$ of the measured data; also the difference was 15% between the two curves. Experiment two was with beam parameters $E$=6.00 MeV, Energy FWHM=0.50 MeV, Focal spot FWHM=0.20 cm, the dose curve shown in Figure 4.3 resulted in buildup region $z_{\text{max}}$ of 1.20 cm and a difference of 17% was between the two curves. Experiment three was with beam parameters $E$=6.20 MeV, Energy FWHM=0.10 MeV, Focal spot FWHM=0.10 cm, the dose curve shown in figure 4.4 resulted in a buildup region $z_{\text{max}}$ of 1.40 cm which is the same as the measured data but due to the difference of about 20% between the curves it was not used as the tuned model.

Figure 4.2: The PDD curve calculated in the study with: $E$= 6.0 MeV, EFWHM=0.10 MeV, FSFWHM= 0.10 cm, $z_{\text{max}}$=1.20 cm

Experiment four was with beam parameters $E$=6.06 MeV, Energy FWHM=0.20 MeV, Focal spot FWHM=0.20 cm, the dose curve shown in figure 4.5 resulted in a buildup region $z_{\text{max}}$ of 1.60 cm which is higher than that of the measured data and with 20% difference between the curves.
Experiment five was with beam parameters $E=6.10\text{MeV}$, Energy FWHM=$0.30\text{ MeV}$, Focal spot FWHM=$0.30\text{ cm}$, the dose curve shown in figure 4.6 resulted in a buildup region $z_{\text{max}}$ of 1.20 cm which is less than that of the measured data and with more than $10\%$ difference between the curves.

Experiment six was with beam parameters $E=6.00\text{MeV}$, Energy FWHM=$0.20\text{ MeV}$, Focal spot FWHM=$0.20\text{ cm}$, the dose curve shown in figure 4.7 was with buildup region $z_{\text{max}}$ of 1.40 cm which is the same of the measured data and with less than $4\%$ difference between the curves. So Linac model in this test was taken as the tuned one because its dose curve was near the dose curve of the measured data in the buildup region and other depths.
Figure 4.5: The PDD curve calculated in the study with: $E=6.06$ MeV, $\text{EFWHM}=0.20$ MeV, $\text{FSFWHM}=0.20$ cm, $z_{\text{max}}=1.60$ cm

Figure 4.6: The PDD curve calculated in the study with: $E=6.10$ MeV, $\text{EFWHM}=0.30$ MeV, $\text{FSFWHM}=0.30$ cm, $z_{\text{max}}=1.20$ cm

Figure 4.7: The PDD curve calculated in the study with: $E=6.0$ MeV, $\text{EFWHM}=0.20$ MeV, $\text{FSFWHM}=0.20$ cm, $z_{\text{max}}=1.40$ cm
The following figures show the profiles of the calculated data in X axis (cross line) and Y axis (in line) for field size 10x10 cm$^2$ of the tuned model and the profile for field size 20x20 cm$^2$ with the same beam parameters.

Figure 4.8: Profile of the calculated data, E=6.0 MeV, EFWHM=0.20 MeV, FSFWHM=0.20 cm in X axis (Cross line), FS=10x10 cm$^2$

Figure 4.9: Profile of the calculated data E=6.0 MeV, EFWHM=0.20 MeV, FSFWHM=0.20 cm in Y axis (inline), FS=10x10 cm$^2$
4.1.3 Results of Evaluation of Histories effect on the accuracy of simulation

The number of histories was an important factor in the resulted PDDs and profiles. Next figures illustrate the effect of increasing the number of histories on the calculated dose. In all the tests the energy was 6.0 MV, SSD=100 cm, field size- 10x10 cm$^2$ with the change in the number of histories. The difference was found to be decreasing with increasing the number histories.
4.1.4 Results of the effect of SSD shift on the absorbed

Next figures show the study of the effect of SSD shift for different depths on the delivered dose. The depths are 3, 5, 10, and 15 cm respectively and SSDs are 101, 102, 103, 104, and 105 cm respectively with SSD 100 cm as standard. The comparison is between the measured and calculated doses. Figure 4.14 show the comparison between the percentage error of the measured and calculated dose for 101 cm SSD, a maximum difference of 0.80% between the two curves. Figure 4.15 show the comparison between the percentage error of the measured and calculated dose for 102 cm SSD, a maximum difference of 1.14% between the two curves.
Figure 4.16 show the comparison between the percentage error of the measured and calculated dose for 103 cm SSD, a maximum difference of 1.92% between the two curves. Figure 4.17 show the comparison between the percentage error of the measured and calculated dose for 104 cm SSD, a maximum difference of 3.05% between the two curves. Figure 4.18 show the comparison between the percentage error of the measured and calculated dose for 105 cm SSD, a maximum difference of 4.25% between the two curves.
Figure 4.16: Comparison of error between measured and calculated dose for 103 cm SSD

Figure 4.17: Comparison of error between measured and calculated dose for 104 cm SSD
4.2 Discussion

Due to the limited access of physicists for radiotherapy machines, Monte Carlo codes to make a model for these can provide adequate solution to the problem. From results we have found a good agreement between calculated dose curve and measured dose curve to tune Elekta Precise linear accelerator.

The used Code PRIMO provides a model of linac geometry and radiation transport through water phantoms or computerized tomography volumes. The code also makes use of parallel processors and variance reduction techniques for execution to reduce simulation time and produce fast and accurate results. This ability was introduced in most Monte Carlo codes due the evolution of computer technology.

The configuration of Full Width Half Maximum (FWHM) for energy and focal spot play an important role in the resulted percentage depth dose (PDD) curve and beam profile. So these parameters were adjusted in different steps to achieve the proper dose curves that match the experimental measured curves. From these parameters a good agreement of 4% was found between calculated dose curves with the same buildup region for the modeled energy and a homogenous profile was acquired. From the tuning process we found that he initial electron energy of 6.0 MeV produced an acceptable range of error over the range of FWHM and focal spot FWHM.
At some simulations this was not correct in cases of small number of histories as described in the study of Effect of histories of the accuracy of simulation. Although the choice of beam parameters was acceptable at some experiments but due to the small number of histories the buildup region and the whole curve of the calculated dose was far from that of the measured and in some examples the error exceeded 25%. In other example although the buildup region was 1.40 the same as the measured curve but the error was more than 30%. This was also verified by Elaine Conneely et al [70] in a study to tune Monte Carlo electron beam parameters to match measured data. They first used low number of histories and large energy increments initially to refine the energy range of the primary electron beam. Then they used larger number of histories to be more precise to differentiate which energy provided a better match with measured data.

For the SSD shift study as we can see from results that the shift can produce a large effect on the absorbed dose. As we know from literature that dose distribution depends on beam quality and depth, field size and shape, source surface distance, and beam collimation. For the SSD effect although the dose rate decreases with distance because of inverse square law, but the PDD increases. This was observed from this study using shifts from 1cm to 5cm from the standard 100cm SSD. The error was more pronounced in large SSD shifts and small depths. It was in the range of 2% for 1 cm shift, 3% to 4% for 2cm shift, 5% to 7% for 3 cm shift, 7% to 9% for 4 cm shift, 8% to 12% for 5 cm. It was higher with small difference in the calculated dose because of the small difference mentioned above between the calculated and measured dose curves.
Chapter Five: Conclusion and Recommendations
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In this work, PRIMO Monte Carlo code was used to adjust a model of linear accelerator that tune photon energy of Elekta Precise, one of the radiotherapy machines in Radiation and Isotopes Center Khartoum (RICK).

The tuned model resulted in 4% agreement between calculated dose curves and experimental measurements after adjusting initial electron energy, energy FWHM, and focal spot FWHM parameters. Also, the buildup region was 1.40 cm for the calculated dose curve which is the same for the measured one.

Profiles of dose distribution in X axis (cross line) and Y axis (in line) were homogenous and with good agreement with measurements.

The tuned model was used to study the effect of number of histories on the accuracy of simulation for the same beam parameters. It was found that inaccuracies and zigzags in the profile reduce with increasing the number of histories.

A study of the setup errors and their effect on patient delivered dose was done using the tuned model and measurements. The study was for the SSD setup errors. Point dose was measured for 3, 5, 10, and 15 cm depths with 100 cm SSD as standard and then with shift from 101 cm to 105 cm with 1 cm as an increment. The dose was calculated using the same settings. The error was more noticeable in large SSD shift and small depths. It was in the range of 2% for 1 cm shift, 3% to 4% for 2 cm shift, 5% to 7% for 3 cm shift, 7% to 9% for 4 cm shift, 8% to 12% for 5 cm, but it was higher in the calculated dose because of the small difference between the tuned model and measured dose curves.

The tuned model can be enhanced by reducing the error to less than 4% by using parallel processors to increase the accuracy of simulation without increasing computation time, also a wide range of field sizes can be added with and without beam modifiers.

Different photon energies can also be tuned using the same procedures referring on measurements in water phantoms.

In the future the plan is to improve the linac model by including calculation of dose in different geometries of source surface distance, different symmetrical and asymmetrical field sizes, calculating diagonal profile.
Other objective is to calculate dose distribution in a computerized tomographic (CT) volume by importing the CT volume as geometry model and indicating the tumor volumes and organs at risk.

The distribution of scatter dose outside the tumor and the risk of scattered dose on normal tissues can be studied. Electron beam PDDs and profiles also can be modelled for different energies. Also the effect of beam modifiers such as blocks and wedges on both photon and electron beams and profiles can be studied.

A comparison between calculated PDDs and profiles using PRIMO can be compared with other Monte Carlo codes to verify the accuracy of PRIMO code.
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