Chapter 1: Introduction

1.1 Overview

Amongst women worldwide, breast cancer is regarded as the common cause of cancer death. Porter (2009) studied that the increase in breast cancer incidence and mortality has been found to be prevalent in lower-income countries. The basis of breast cancer treatment is surgery when the tumour is localized, followed by chemotherapy, radiotherapy and adjuvant hormonal therapy. Hiraoka et al., (2002) undertook a study confirming the use of adjuvant radiation therapy following mastectomy for breast cancer to be beneficial but raised a comment that post-mastectomy radiation therapy (PMRT) still remains a controversial issue as no studies have been conducted yet. Findings have shown that approximately 10% to 15% of patients with stage I/II invasive breast cancer will develop a clinically isolated local recurrence Freedman et al., (2000).

Breast reconstruction is a type of surgery for women who have had a breast removed (mastectomy) Med-Line Plus (2012). The surgery rebuilds the breast mount so that it is about the same size and shape as it was before Resnick et al., (2002) and Brunicardi et al., (2010). Several types of operations can be performed to reconstruct the breast. The most common prosthesis is a saline-filled prosthesis, which is a silicone shell filled with salt water. Bostwick (1995) mentioned that silicone gel-filled prostheses are another option for breast reconstruction. They are not used as often as they were in the past because of concerns that silicone leakage might cause immune system diseases Djohan et al., (2008), Park et al., (1993), Barbara (1996) and Noone (1997) but it is yet to be proven that they in fact do.

Concerns were raised that insufficient information existed whether the silicone gel prosthesis have any effect on the delivery of radiation treatment in cases of breast cancer recurrence Med-Line Plus (2012). Radiation therapy, also referred to as radiotherapy, involves delivering exact amounts of high-energy radiation to eradicate malignant cancer cells. The radiation dose targets the actively dividing cancer cells and it is important to spare the surrounding healthy tissues.
Shedbalkar et al., (1980) studied the radiation effects on silicone gel and dose distribution of radiation through breast prostheses. It was observed that silicone gel behaved like tissue and that its half-value thickness and density were comparable to that of water.

1.2 Aim of the Study

The aim of this study was to observe the effect of sizes of silicone gel on photon dose distribution using the Monte Carlo Neutron Particle transport (MCNP) code.

1.3 Specific Objectives

The specific objectives required to achieve the above-mentioned aim are as follows:

i. The measurements of the percentage depth dose (PDD) curves for 6 MV and 15 MV photon beams using a field size of $10 \times 10 \text{ cm}^2$ in a water phantom at a 100 cm source-to-surface distance (SSD).

ii. The measurements of the dose profiles for 6 MV and 15 MV photon beams using a field size of $10 \times 10 \text{ cm}^2$ in a water phantom at an SSD of 100 cm.

iii. MCNP simulation of a $30 \times 30 \times 30 \text{ cm}^3$ water phantom.

iv. MCNP simulation of photon beam geometry to obtain PDD curve and dose profile data for 6 MV and 15 MV photon beams using a field size of in a water phantom in order to validate the MCNP code.

v. MCNP simulations of photon beam geometries at field sizes of $10 \times 10 \text{ cm}^2$ and $16 \times 16 \text{ cm}^2$ at a 100 cm SSD to obtain PDD curves and dose profiles for 6 MV and 15 MV photon beams in a water phantom for 4 cm, 6 cm, 8 cm, 10 cm, 12 cm, 14 cm and 16 cm gel thicknesses.

1.4 Outline of the Book

Chapter 2 discusses the type of electromagnetic waves that can be found on the electromagnetic spectrum, their wavelength as well as their frequencies. The major interaction processes at which radiation can be produced is also discussed in this chapter.

Chapter 3 discusses the different components that make up the Linear Accelerator (linac) as well as their functions. The production of photons in a linear accelerator are also discussed to show how the
beam emitted from the electron gun travels through the other linac structures until it is delivered in terms of radiation dose for therapeutic purposes.

Chapter 4 discusses how the clinical reference dosimetry of high energy photon beams is carried out. The quantity “absorbed dose” and Kerma is introduced in this chapter, and it is explained how the dose propagated in a medium results with the loss of energy per unit area during interaction. Factors that influence the percentage depth dose (PDD) such as the inverse square law, energy, field size and source-to-surface distance (SSD) are outlined in this chapter.

Chapter 5 explains the MCNP code technique as well as its origin. One of the objectives mentioned in this chapter is to determine how the MCNP code can solve the radiation transport in order to calculate the depth dose curves together with the dose profiles. The chapter elaborates further on how the geometries for the code should be defined in order to obtain the expected dose profiles and percentage depth dose curves.

Chapter 6 discusses the materials used and enlightens as to how the project was carried out. The first section explains the properties of the silicone gel and its composition. The linac geometry outlines the set-up for the physical measurements. The MCNP geometry explains how the simulations were defined in order to achieve the objectives of the study.

Chapter 7 presents the results that are obtained from the methods outlined, and the discussion of the results follows.

Chapter 8 gives the conclusion based on the results obtained and the objectives of the study.
Chapter 2: Interaction of Radiation with Matter

2.1 Introduction

The term “radiation” refers to the transfer or propagation of energy in the form of particles or waves emitted from the radioactive material. This form of energy is regarded as electromagnetic radiation. The possible range of electromagnetic radiation is represented in the form of an electromagnetic spectrum. Figure 2.1 below illustrates the electromagnetic spectrum and is expressed in terms of energy, wavelength and frequency. Jones et al., (1938) gave the relationship between them as follows:

\[ f = \frac{c}{\lambda} \quad \text{and} \quad E = hf \]  
(2.1)

where \( f \) is the frequency, measured in hertz (Hz) or sec\(^{-1}\), \( \lambda \) is the wavelength, measured in cm and \( c \) is the speed of of light (in m/s), \( h \) is Planck’s constant and \( E \) is the photon energy, measured in electron volts (eV).

According to equation 2.1 the photon energy is directly proportional to the frequency of the electromagnetic wave. The wavelength describes the behaviour of the electromagnetic radiation in a spectrum. From Figure 2.1 below, it can be observed that the higher the frequency, the shorter the wavelength.

![Figure 2.1: Electromagnetic spectrum: radio waves, visible light, x-rays, and all the other part of the electromagnetic spectrum are fundamentally the same thing, electromagnetic radiation Viewzone (2012).](image)

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Two types of radiation sources are naturally occurring radiation sources and man-made radiation sources. The three major sources of naturally-occurring radiation include cosmic radiation, terrestrial radiation as well as internal sources. Cosmic radiation comes from the sun and outer space; terrestrial radiation is found in the ground, rocks, building materials as well as drinking water supplies Nikjoo et al., (2012) and internal sources are found in a human body, for example potassium-40.

X-rays are a form of high frequency electromagnetic radiation which is described as a form of pure energy carried by waves of photons. The terms X-rays and photons are used interchangeably even though their emission characteristics differs Turner, (2007). X-rays contain more energy than ultraviolet, infrared, radio waves, microwaves or visible light Ibbott et al., (2005). When radiation interacts with matter, ions are produced and for this reason, it is called ionizing radiation.

Ionization is a process by which a neutral atom or molecule acquires a positive or negative charge when an outer shell electron/tightly bound electron is removed from the orbit of an atom (see Figure 2.2 below). Two types of radiation are commonly differentiated in the way they interact with normal matter; that is: ionization and non-ionization radiation. For interaction to take place; the type of radiation should have sufficient energy of greater than 30 electron volt (> 30eV) to cause ionization. Some ionizing radiation can be very harmful; in essence, capable of causing chemical changes to biologically important molecules, for example, deoxyribonucleic acid (DNA) Nikjoo et al., (2012).
Non-ionization radiation does not contain energy up to 30 electron volt (<30eV); instead only causes excitation which is the movement of an electron from a lower energy to a higher energy state Nenoi (2012). Examples of non-ionization radiation are microwaves, radio waves, ultrasound, cellular phones, ultraviolet, etc.

Bomford et al., (2003) describes attenuation as the removal of radiation energy from the beam either by scattering or absorption. When describing the attenuation of electromagnetic radiation with matter, radiation has to be considered as waves taking into consideration their origin and interactions.

During the ionization process, photons may either pass straight through without interacting with the atoms of the medium, or:

- they may be totally absorbed by the medium (photoelectric absorption), or
- they may be scattered from their original path and partially absorbed (Compton scattering), or
- they may be scattered and partially or completely absorbed (pair production).
Ionization radiation can either be directly or indirectly. X-rays, gamma rays, beta rays, alpha rays etc, fall under the indirectly ionizing radiation category since they are electrically neutral and do not interact with atomic electrons through Coulombic forces Turner (2007).

In this context, X-rays and gamma rays are regarded as examples of indirectly ionizing radiation; they subsequently tend to be absorbed by processes which set electrons in motion when they pass through matter and these electrons produce ionization of other atoms or molecules in the medium. The electrons have short finite ranges and their kinetic energy is rapidly dissipated first as ionization and excitation, and eventually as heat Adrovic et al., (2012).

2.2 Types of interactions

Photons (X- and gamma rays) react with matter in a very different way from charged particles. They do not experience long-range electrostatic forces and interact only in direct collision processes. Hence, the behaviour of a beam of photons differs from the behaviour of a beam of charged particles. Charged particles, for instance electrons, have a definite range in matter as compared to photons that are much more penetrative and demonstrate no definite range in exponential absorption Bomford et al., (2003). Khan (2003) noted that ionization photons interact with the atoms of a material or absorber to produce high speed electrons by three major processes namely, photoelectric effect, Compton Effect and pair production.

2.2.1 Photoelectric Absorption

The photoelectric process, illustrated in Figure 2.3 below, involves an interaction between an incident photon and a tightly bound orbital electron of an attenuator. The photon disappears, while the orbital electron is ejected from the atom as a photoelectron with a kinetic energy \( KE \) given as:

\[
KE = h\nu - BE
\]  

(2.2)

where \( h\nu \) is the incident photon energy and \( BE \) is the binding energy of the electron Podgorsak (2003).
After the electron has been ejected from the atom, a vacancy is created in the shell, thus leaving the atom in an excited state. The vacancy can be filled by an outer orbital electron with the emission of characteristics X-rays Khan (2003 & 2010).

### 2.2.2 Compton Effect

The Compton Effect, also known as Compton scattering, is the result of a high energy incident photon colliding with a target, which releases loosely bound electrons from the outer shell of the atom or molecule, as shown in Figure 2.4.
The scattering is demonstrated in the Figure 2.4 where a high-energy photon (incident gamma-ray or X-ray) collides with a target, which has loosely bound electrons on its outer shell Hendee (1979).

The incident photon has energy $h\nu$ and momentum $\frac{h\nu}{c}$ and gives part of its energy to the stationary free electron, in the form of kinetic energy whilst conserving both its energy and momentum Khan (2003).

In this interaction, the ejected electron receives energy from the photon and is emitted at an angle ($\phi$). The photon with reduced energy is scattered at an angle ($\theta$). The Compton process can be analyzed in terms of a collision between two particles, a photon and an electron. Applying the laws of conservation of energy and momentum, Khan (2003) derived the following relationships:

$$E = h\nu_0 \frac{\alpha(1 - \cos \theta)}{1 + \alpha(1 - \cos \theta)}$$  \hspace{1cm} (2.3)

$$h\nu' = h\nu_0 \frac{1}{1 + \alpha(1 - \cos \theta)}$$ \hspace{1cm} (2.4)

$$\cos \phi = (1 + \alpha) \frac{\tan \theta}{2}$$ \hspace{1cm} (2.5)
where $hv_0$, $hv'$ and $E$ are the energies of the incident photon, scattered photon, and electron, respectively and,

$$\alpha = \frac{hv_0}{m_0c^2}$$  \hspace{1cm} (2.6)

where $m_0c^2$ is the rest energy of the electron (0.511 MeV). If it is expressed in MeV, then

$$\alpha = \frac{hv_0}{0.511 \text{ MeV}}$$  \hspace{1cm} (2.7)

There are three special cases of the Compton Effect explained by Khan (2003):

i. “Direct hit” which means that if a photon hits the electron, the electron will travel forward ($\phi = 0^\circ$) and scattered photon will travel backward $\theta = 180^\circ$ after the collision. In such a collision, the electron will receive maximum energy $E_{\text{max}}$ and the scattered photon will be left with minimum energy $hv'_{\text{min}}$. $E_{\text{max}}$ and $hv'_{\text{min}}$ can be calculated by substituting $\cos \theta = \cos 180^\circ = -1$ in equations 2.3 and 2.4 (Khan, 2003).

$$E_{\text{max}} = hv_0 \frac{2\alpha}{1+2\alpha}$$  \hspace{1cm} (2.8)

$$hv'_{\text{min}} = hv_0 \frac{1}{1+2\alpha}$$  \hspace{1cm} (2.9)

ii. Khan (2003) explained “grazing hit” which states that if a photon makes a grazing hit with the electron, the electron will be emitted at right angles ($\phi = 90^\circ$) and the scattered photon will go in the forward direction ($\theta = 0^\circ$). By substituting $\cos \theta = \cos 0^\circ = 1$ in equations 2.3 and 2.4, it can also be shown that for collision $E=0$ and $hv' = hv_0$.

iii. Khan (2003) further said that $90^\circ$ photon scatter means that if a photon is scattered at right angles to its original direction ($\theta = 90^\circ$), one can calculate $E$ and $hv'$ from equations 2.3 and 2.4 by substituting $\cos \theta = \cos 90^\circ = 0$. The angle of the electron emission in this case will depend on $\alpha$, according to equation 2.6.
### 2.2.3 Pair Production

Pair production occurs when the X-ray photon energy is greater than 1.022 MeV, but only becomes significant at energies around 10 MeV. In this process (see Figure 2.5), the photon interacts strongly with the electromagnetic field of an atomic nucleus and gives up all its energy in the process creating a pair consisting of a negative electron ($e^-$) and a positive electron ($e^+$). Because the rest mass energy of the electron is equivalent to 0.511 MeV, a minimum energy of 1.022 MeV is required to create the pair of electrons. Bomford et al., (2003) states that since the electrons are opposite in charge, no creation of charge is involved. He elaborated further that if the energy of the initiating photon $h\nu$ is greater than 1.022 MeV then the excess energy will appear as kinetic energy, $E$, and be shared between the two particles as indicated in the following equation:

$$ h\nu = mc^2 + mc^2 + E^+ + E^- $$

where $h\nu$ is the photon energy, $E^+$ is the positron, $E^-$ is the electron, $m$ is the mass of an electron and $c$ is the speed of light.

![Figure 2.5: The absorption of photons by pair production Giffin (1996).](image-url)
Chapter 3: Linear Accelerator Based External Beam Radiation Therapy

3.1 Introduction

The application of X-rays in the field of medicine thus far is focused mainly for diagnosis and therapeutic purposes in the following branches of medicine: Radiotherapy, Nuclear Medicine and Radiology. X-rays and high-energy electrons as discussed in Section 2.1 are high frequency electromagnetic waves generated by the linear accelerator (linac) for medicinal purposes in radiation therapy (radiotherapy) Boudville (2008). The focus in this study will mainly be on one category of radiotherapy called the external beam radiotherapy [EBRT], where a radiation beam is produced by a linear accelerator and then targeted at the tumour.

The linac is an electrical device that accelerates subatomic particles producing a high-energy radiation beam that is delivered to tumor cells for therapeutic purposes. These devices are designed to deliver radiation dose which is accurately defined in terms of its energy, uniformity, distribution and duration.

Greene et al., (1997) explained that inside the device, a beam of electrons is generated by a heated cathode inside the electron gun and accelerated through a waveguide, a structure that increases their energy from the keV to MeV range as indicated in Figure 3.1 below. These electrons strike a tungsten target and produce X-rays Gadza et al., (2007). The dual mode linacs today offer both one or two X-ray energies and a range of electron beam energies Khan (2003 & 2010). Photon beams produced are in the range of about 6 MV to a maximum of 20 MV and are used in the treatment of the deep-seated tumors; whilst the electron beam energies in a range of about 6-20 MeV permit the treatment of the superficial tumors Bomford et al., (2003).
3.2 Linear Accelerator Components

The major components of the linac are the gantry, the stand, the control console and the treatment couch. Linacs are usually mounted isocentrically, with its head facing downward and some include a modulator cabinet which will be described below. The Figure 3.1 below illustrates a flow diagram of a typical linac.

![Figure 3.1: A flow representation of the typical linear accelerator Khan (2003).](image)

The stand of the linac is attached firmly to the floor and the gantry rotates in both the clockwise ($0^\circ - 180^\circ$) and anticlockwise ($360^\circ - 180^\circ$) directions, in order to irradiate the target volume from different directions successfully without repositioning the patient. The operational accelerator structure, housed in the gantry, rotates about a horizontal axis fixed by the stand, and the rotation occurs about an isocentric point in space to a very high degree of accuracy Wiedemann (2003). The major components in the stand are klystron or a magnetron, waveguide, circulator and a water cooling system.

The major components found in the gantry are the accelerator structure, the electron gun (or cathode), the bending magnet, the treatment head and the modulator cabinet.
The treatment head of a linac consists of a thick shell of high density shielding material such as lead, tungsten, or lead-tungsten alloy. The head provides sufficient shielding against leakage radiation. It contains the following component modules: the X-ray target, the primary collimator, the scattering foil, the flattening filter, the ion chamber, secondary collimators and the light localizer system Boudville (2004).

The treatment couch motions are controlled by a hand pendant most often operated by the radiographer. The 3-D positioning of the patient on the couch is motor-driven Wiedemann (2003). Most couches also provide couch rotation about a vertical axis passing through the isocentre and some permit the attachment of a treatment chair. The control console is the operation centre for a linac and it supplies the timing pulses that initiate each pulse of radiation. It is located outside the treatment room for personnel safety purposes. It provides visual and electronic monitors for a host of linac operating parameters including the patient’s dose prescription.

3.3 The Production of Radiation in a Linear Accelerator

The linac components that produce the photon beam consists of three components namely primary photons, secondary photons and contaminant photons Hall (2010). The components that produce these photons are indicated in Figure 3.1 Hall (2010) indicated that the primary photons are the Bremsstrahlung photons produced when the electron beam strikes the X-ray target and is responsible for the $96.96 \pm 0.08\%$ of the photon fluence (which will be discussed in chapter 4).

3.4 Functions of Components in the Production Process

The functions of major components mentioned in Section 3.2 above in the stand are as follows:

- Klystron or Magnetron - which sits at the top of an insulating oil tank and provides a source of microwave power to accelerate electrons through the waveguide;

- waveguide – which conveys this power to the accelerator in the gantry Wangler (2008);

- circulator – a device inserted in the waveguide to isolate the klystron from microwaves reflected back from the accelerator; and
- water cooling system - which cools various components that dissipate energy as heat and establishes a stable operating temperature sufficiently above room temperature to prevent condensation of moisture from the air Greene et al., (1997).

The functions of major components found in the gantry are as follows:

- the accelerator structure - which is energized by the microwave power supplied from the klystron via the waveguide;
- the electron gun (or cathode) - which provides the source of electrons injected into the structure. The electrons generated are in the range from 1 to 50 MeV for the production of photons TRS398 (2001).
- the bending magnet – which deflects the electrons emerging from the accelerator structure around a loop in order to strike the target to produce X-rays or to be used directly for electron treatments;
- the treatment head – which contains component modules (collimators and monitoring devices) Wangler (2008) and
- the modulator cabinet - which contains components that distribute and control primary electrical power to all areas of the machine from the utility connection and also supplies high voltage pulses.

In the treatment head, the X-rays are produced by high-speed electrons of various energies incident on X-ray target. Since linear accelerators produce electrons in the megavoltage range, the X-ray intensity is directed in the forward direction. The treatment beam is first collimated by a fixed primary collimator located immediately beyond the X-ray target and beam uniformity across the field is achieved by the introduction of a beam flattening filter located in the carousel. This is a metal filter, usually made of lead, although tungsten, uranium, steel, aluminium, or a combination of these has also been used Wangler (2008). The photon beam is incident on the dose monitoring chambers. The monitoring system consists of two ion chambers. The function of the ion chambers is to monitor the dose rate, integrate the dose, as well as monitoring the symmetry of the photon field. The monitor chambers in the treatment head are usually sealed so that their response is not influenced by temperature and pressure of the outside air Wangler (2008).
After passing through the ion chambers, the beam is further collimated by a continuously movable X-ray collimator. This collimator consists of two pairs of lead or tungsten blocks (jaws) which provide a rectangular opening from $0 \times 0 \, \text{cm}^2$ to the maximum field size ($40 \times 40 \, \text{cm}^2$ or a little less) projected at a standard SSD of about 100 cm from the X-ray source (focal spot on the target). The collimator blocks are controlled to move so that the block edge is always along a radial line passing through the target Wiedemann (2003). The field size definition is provided by a light localizing system in the treatment head. A combination of a mirror and light source located in the space between the chambers and the jaws projects a light beam as it is emitted from the X-ray focal spot.
Chapter 4: Photon Beam Dosimetry

4.1 Phantoms

Phantom is a common name for materials that are used for radiation measurements in studies of radiation interaction. The phantoms have been designed after some recommendations done by the AAPM protocol (1983). These include water phantoms, polystyrene phantoms, and perspex (Poly-methyl methacrylate) (PMMA).

Water is recommended as the reference medium for measurements of absorbed dose and beam quality in photon beams. Plastic phantoms should not be used for reference dosimetry; however they can be used for routine quality assurance measurements provided the transfer factor between plastic and water has been established TRS398 (2001).

The phantom material should meet the following criteria:

i. absorb photons in the same manner as tissue;
ii. scatter photons in the same manner as tissue;
iii. have the same density as tissue;
iv. contain the same number of electrons per gram as tissue; and
v. have the same effective atomic number as tissue.

The water phantom closely approximates the radiation absorption and scattering properties of tissue; and is universally available with reproducible radiation properties Khan (2003). The water phantom is designed for absolute dose measurements in radiation beams with horizontal/vertical beam incidence. It is further suitable for the calibration of ionization chambers used in radiation therapy. The phantom’s design allows cross-calibration of a field ionization chamber against a calibrated reference chamber at a user’s facility. Figures 4.1 and 4.2 give the water phantom positioned on the stand and the schematic drawing of the water phantom, respectively.
Figure 4.1: The water phantom positioned on a stand.

Figure 4.2: Schematic drawing of a water phantom on a stand.
4.2 Dosimetric Quantities and Units

Podgorsak (2005) discussed radiation dosimetry as the measurement and calculation of the absorbed dose in matter resulting from the exposure to indirect and direct ionization radiation in two stages in which the energy of photons is imparted into matter. In the first stage, the photon radiation transfers energy to the secondary charged particles (electrons) through various photon interactions discussed in Section 2.2. In the second stage, the charged particle (electrons) transfers energy to the medium through atomic excitations and ionizations.

Podgorsak (2005) further defined Kerma as the Kinetic Energy Released in the Medium. Kerma is the quantity that most directly connects the description of the radiation beam with its effects Johns et al., (1983). It is thus given by the following equation:

\[ K = \frac{d\bar{\varepsilon}_e}{dm} \]  

(4.1)

where \( \bar{\varepsilon}_e \) is the average energy that is transferred to the electrons in a volume of tissue with mass \( dm \).

The unit for Kerma is the same as for dose, i.e. gray, Gy, where 1 Gy = joule per kilogram \((J \cdot kg^{-1})\).

For a mono-energetic photon beam of energy \( h\nu \) and photon fluence \( \Phi \), Kerma is given as

\[ K = \Phi \frac{\mu}{\rho} \bar{\varepsilon}_e \]  

(4.2)

where \( \Phi \) is the fluence; which refers to the number of photons in a beam, \( \frac{\mu}{\rho} \) is the mass attenuation coefficient; this is the percent reduction in the number of photons in the beam.

For a photon beam traversing a medium, Kerma at a point is directly proportional to the photon energy fluence \( \Phi \) and is given by

\[ K = K^{col} + K^{rad} \]  

(4.3)

where \( K^{col} \) is the collision part of Kerma given by

\[ K^{col} = \Phi \left( \frac{\mu_{en}}{\rho} \right) \]  

(4.4)

and is also called the absorbed dose given away during excitation and ionization, whilst \( K^{rad} \) is the energy radiated away by bremsstrahlung given by

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The quantity “absorbed dose” or simply the term dose is referred by Khan (2003) as the amount of energy absorbed per unit mass that the photon beam may deposit in a given medium, such as air, water and or any biological material. Mathematically, the following ratio describes the absorbed dose:

\[ \frac{d\bar{\varepsilon}}{dm} \]  

where \(d\bar{\varepsilon}\) is the mean energy imparted by ionizing radiation to material of mass \(dm\).

The absorbed dose \(D(\bar{x})\) is a function of the spatial variable \(\bar{x}\), and this spatial variation is termed the dose distribution. A second important aspect refers to the temporal pattern of dose accumulation. The absorbed-dose-rate is defined as:

\[ \dot{D}(\bar{x})t = \frac{dD(\bar{x})}{dt} \]  

The absorbed-dose-rate is measured in Gy/h.

The number and energies of all photons constituting the beam is referred to as the photon beam spectrum. The following quantities can be used to describe the photon beam:

i. Photon fluence: \(\phi = \frac{dN}{dA}\),  

where

- \(dN\) is the number of photons that enter an imaginary sphere of cross-sectional area \(dA\).

ii. Energy fluence: \(\psi = \frac{d\bar{\varepsilon}}{dA}\);  

where

- \(d\bar{\varepsilon}\) is the amount of energy crossing a unit area \(dA\).

Figure 4.3 below gives the photon beam incident on the water phantom first without attenuation and giving rise to a build-up region at certain depth. Absorbed dose will start low and build up to the maximum level due to the low energy electrons liberated; therefore, the photon beam attenuation will
not be appreciable over this distance Attix (1986). The absorbed dose at this point is just the collision part of the Kerma, and if no bremsstrahlung loss occur then absorbed dose is equal to Kerma.

At a maximum depth, the electronic equilibrium is reached for both Kerma and absorbed dose. With attenuation of the photon beam the absorbed dose which will be at a higher level than Kerma, then decrease from there onwards. Kerma will now decrease by 5% in depth. The reason that the absorbed dose is higher than Kerma at any point beyond the peak is that the absorbed dose is in line to Kerma further upstream, and both the absorbed dose and Kerma will remain at the same point.

4.3 Factors that influence the Percentage Depth Dose (PDD) Distribution

The delivery of radiation therapy is characterized with PDD curves and dose profiles. The incident beam varies with depth as it is absorbed in a phantom. The depth is the distance beneath the surface where the prescribed dose is to be delivered. This variation depends on the following parameters: energy, depth, field size, and source-to-surface distance (SSD) Khan (2003).

Podgorsak (2010) described the PDD as the ratio, expressed as a percentage, of the absorbed dose \(D_d\) at a given depth to the absorbed dose at a fixed reference depth usually \(D_{d_{max}}\) will be given by:
\[
PDD(hv; SSD; FS; d) = \frac{D_d}{D_{d_{\text{max}}}} \times 100\% \quad (4.10)
\]

where \(D_d\) is the dose rate at any arbitrary depth, and \(D_{d_{\text{max}}}\) is the dose rate at a depth of maximum dose.

The PDD can also be defined in terms of the dose rate and will be given by the following equation:

\[
PDD(hv; SSD; FS; d) = \frac{\dot{D}_d}{\dot{D}_{d_{\text{max}}}} \times 100\% \quad (4.11)
\]

where is \(\dot{D}_d\) the dose rate at any arbitrary depth, and is \(\dot{D}_{d_{\text{max}}}\) the dose rate at a depth of maximum dose.

![Diagram](image)

Figure 4.4: The set-up for the measurement of the percentage depth dose.

The measurement of PDD is one of the methods that measure the attenuation of the beam as it travels through the phantom. The region between the surface \((d = 0)\) and depth \(d = d_{\text{max}}\) in megavoltage photon beams is called the dose build up region.

The surface dose for megavoltage photon beams is in the order of 10\% to 25\% Podgorsak (2010), and that accounts for the skin sparing effect.
Close to the beam exit point, the dose distribution curves slightly downwards from the dose curve obtained for an infinitely thick phantom as a result of missing scatter contribution for points beyond the dose exit point. The effect is small and generally negligible. The curve in Figure 4.5 above has added detail known as error bars, whose purpose is to show the variability in the measurements plotted in the chart Peltier Technical Services (2012). Figure 4.5 shows an example of a PDD curve as a result of the set-up in Figure 4.4.

### 4.3.1 Inverse Square Law as a Factor that Influences PDD

As the photon beam traverses through a phantom, it is affected by the inverse square law as well as attenuation and scattering of the photon beam inside the phantom Podgorsak *et al.*, (2003). When a photon beam diverges from its point of production, its intensity is reduced as the distance from the source increases. Podgorsak (2005) and Khan (2003) pointed out that this law indicates that external beam radiotherapy photon sources are often assumed to be point sources that produce divergent beams. Assume that there is a photon point source $S$ and a square field with side $a$ (area $A = a^2$) at a distance $f_a$ from the source. At a distance $f_b$, we then get a square field with side $b$ (area $B = b^2$), and the two fields are geometrically related as follows Podgorsak (2005):

$$\beta = \frac{a}{b} = \frac{f_a}{f_b}, \quad (4.12)$$
where $\beta$ is the angle between the beam central axis and the geometric beam edge.

The photon source emits photons and produces a photon fluence $\Phi_A$ at a distance $f_a$ and photon fluence $\Phi_B$ at a distance $f_b$. Since the total number of photons $N_{tot}$ crossing area equal to the total number of photons crossing area B (assuming no photons interactions take place in air between area A and area B), then

$$N_{tot} = \phi_A A = \phi_B B \quad (4.13)$$

and

$$\frac{\phi_A}{\phi_B} = \frac{B}{A} = \frac{b^2}{a^2} = \frac{f_{b}^2}{f_{a}^2} \quad (4.14)$$

Figure 4.6: The divergent photon beam originating in a photon point-source S. At a distance $f_a$ from the source S the field size is $A = a^2$, at a distance $f_b$ the field size $B = b^2$ Podgorsak (2005).

The photon fluence is thus inversely proportional to the square of the distance from the source. For example, if $f_b = 2f_a$ then the photon fluence at B will be exactly $\frac{1}{4}$ of the photon fluence at A (i.e. $\Phi_B = \frac{\Phi_A}{4}$). Since at a given point P in air the exposure in air $X$, air-Kerma in air $(K_{air})_{air}$, and “dose to small mass of medium in air” $D'_{med}$ are directly proportional to the photon fluence at point P, it is
reasonable to conclude that the three quantities: \( X \), \( (K_{air})_{air} \), and \( D'_{med} \) all follow this inverse square law behaviour, i.e.

\[
\frac{X(f_a)}{X(f_b)} = \frac{(K_{air}(f_a))_{air}}{(K_{air}(f_b))_{air}} = \frac{D'_{med}(f_a)}{D'_{med}(f_b)} = \left(\frac{f_b}{f_a}\right)^2
\]  

(4.15)

4.3.2 Energy as a Factor that Influences PDD

Energy is one of the parameters that affect the dose distribution Podgorsak (2005). The PDD increases with beam energy. Higher energy beams have greater penetrating power and thus deliver a higher PDD, but the point of maximum dose lies deeper into the tissue or phantom giving rise to the skin sparing effect Hoppe et al., (2010). However, Megavoltage beams such as cobalt-60 and higher energies, the surface dose are much smaller than the (depth of maximum) \( d_{max} \). This offers distinct advantage over lower energies where \( d_{max} \) occurs at the skin surface. The \( d_{max} \) is used for the fixed reference depth, and it varies with photon energy. High-energy photon beams can be delivered to deep seated tumors without exceeding the tolerance of the skin Khan (2010).

![Figure 4.7: Illustration of PDD (d) as a function of depth.](image)
4.3.3 The Source-to-Surface Distance (SSD) as a Factor that Influences PDD

The photon beam setup can be carried in a source-to-surface distance (SSD) technique. The SSD refers to the distance from the source of radiation in the treatment machine to the surface of the water in the form of the light intersecting the beam light Podgorsak (2005).

The SSD is normally measured using an optical distance indicator (ODI) on a phantom. This device projects a distance scale onto the phantom’s surface Washington et al., (1996) in the form of light. Photon fluence emitted by a point source of radiation varies inversely as a square of the distance from the source. Figure 4.8 shows how actual dose rate at a point decreases with increase in distance from the source; the PDD, which is a relative dose, increases with SSD.

\[
P = \frac{D_d}{D_{\text{max}}} \times 100
\]

![Figure 4.8: Illustration of the relative dose rate dependence on SSD.](image)

Figure 4.8: Illustration of the relative dose rate dependence on SSD.
Figure 4.9 above the Mayneord factor $F$ that is used for correction of PDDs from one SSD to another. The formula is given by:

$$F = \left( \frac{f_2 + d_m}{f_1 + d_m} \right)^2 \times \left( \frac{f_1 + d}{f_2 + d} \right)^2$$  \hspace{1cm} (4.16)

Where $f_1$ and $f_2$ are the SSD1 and SSD2, $d$ is the arbitrary depth and $d_m$ is the reference depth.

### 4.3.4 Effect of Field Size on Percentage Depth Dose

Field size can be defined either geometrically or dosimetrically Podgorsak (2005). According to the ICRU (1976), the geometric field size is defined as the projection of the distal end of the machine collimator on a plane perpendicular to the central axis of the radiation beam as seen from the front centre of the source. The dosimetric field size, (also called the physical field size) is defined by the intercept of a given isodose surface usually 50% up to 80 % with a plane perpendicular to the central axis of the radiation beam at a defined distance from the source Podgorsak (2005). The dosimetrical and geometrical field sizes are illustrated in Figures 4.10 and 4.11.
As the field size is increased, the contribution of scattered radiation to the absorbed dose increases. For small field sizes, the PDD is due to the primary beam only. This increase in scattered dose is greater at
larger depths than at $d_{max}$ where the PDD increases with increasing field size. PDDs are usually tabulated by equivalent square field calculated from area per perimeter. The scatter contribution is corrected by the output factors for different field sizes. PDD data for radiotherapy beams is usually tabulated for square fields. In clinical practice the rectangular and irregular shaped fields are required; therefore equations 4.17 and 4.18 are introduced to equate these fields to a square field and equivalent circle respectively.

$$\frac{A}{P} = 2 \times \frac{a \times b}{a + b} \quad \text{and} \quad r = \frac{4}{\sqrt{\pi}} \times \frac{A}{P}$$

(4.17) and (4.18)

and $r$ is the radius of the circle.

### 4.4 Beam Profiles

The accuracy of the dose distribution inside the water phantom cannot only be obtained along the central axis of the beam. It can further be characterized by measuring dose profile shown in Figure 4.10. The dose profile uniformity is usually measured by a scan horizontally along the centre major beam axes for various depths in a water phantom Podgorsak (2005). Two parameters that quantify field uniformity are beam flatness and beam symmetry. The beam flatness and symmetry are measured within the flattened area.

Beam flatness ($F$) is the ratio of dose at $D_{max}$ to dose at $D_{min}$ values on the beam profile within the central 80% of the beam width Armoogum (2004). The obtained beam flatness should be less than 3% Kouloulias et al., (2003) and is given by the following equation:

$$F = \frac{D_{max} - D_{min}}{D_{max} + D_{min}} \times 100$$

(4.19)

Beam symmetry ($S$) is usually determined at $D_{max}$ which represents the most sensitive depth for assessment of this beam uniformity parameter. Podgorsak (2005) and Armoogum (2004) define the areas under the $D_{max}$ beam profile on each side (left or right) of the central axis extending to the 50% dose level (normalized to 100% at the central axis point) as indicated in Figure 4.10 above and $S$ is calculated from the following equation Podgorsak (2005),

$$S = 100 \times \frac{(area_{left} - area_{right})}{(area_{left} + area_{right})}$$

(4.20)
Chapter 5: Monte Carlo Neutron-Particle (MCNP) Transport Techniques

5.1 Introduction

The MCNP is a type of code that is internationally recognized for analyzing the transport of neutrons and gamma rays by the Monte Carlo method Burgett (2007). The MCNP has been benchmarked for transporting photon and electron energies at a range of 1 keV to 1000 MeV Wilson (2011). The code makes use of commands and prompts during the input of files, execution of files and output of files. The origin and structure of the MCNP will be discussed in the following section.

5.2 MCNP Origin

According to Briesmeister (1997), the MCNP method was recognized from the work done at Los Alamos during World War II and further added that it is the successor to the work attributed to Fermi, von Neumann, Ulam, Metropolis and Richtmyer (machine language programs) as demonstrated in Figure 5.1. This code and its data libraries have been developed more than 25 years ago Brown et al., (2004) and McKinney et al., (2006) demonstrated the history of MCNP diagrammatically as shown in Figure 5.1. The MCNP code undergoes continuous development and has periodic new releases Briesmeister (1997).

Figure 5.1: The flowchart of the MCNP code history development Los Alamos National Laboratory (2011).
Taranenko et al., (2005) and Xu et al., (2000) reflected that the Monte Carlo concept was employed to select random nuclear processes and this type of code generates random numbers between 0 and 1. These random numbers determine which interaction will occur by comparing probabilities of each intersection. Shultis et al., (2006) said that the Monte Carlo code uses a three dimensional (3-D) Cartesian coordinate system for simulations. Yoriyaz et al., (2001) confirmed the fact the code treats the geometry of the problem primarily in terms of regions or volumes bounded by first and second degree surfaces. All dimensions used are measured in centimetres as required by the code Shultis et al., (2006).

5.3 Monte Carlo Simulations

The Monte Carlo method uses the Physics laws and probabilities during the transport of photon and electron beam through matter. The code applies minimal amount of estimates and requires large amounts of computer time due to its statistical procedures Martz (2006). The MCNP code is used to describe the geometry of the problem, specific materials and radiation sources.

The geometry that describes the problem is called the input file whereas the geometry that contains the solution to the problem is referred to as the output file. The MCNP code’s input file is divided into three sections, namely the cell card section, the surface card section and the data card section. The title card should be the first line to begin the input geometry within the first 5 columns (lines) and it gives a filename in relation to the contents of the whole input file. There is no blank line/blank line delimiter to indicate the separation between the title card and the cell card that follows it. The structure of the code is illustrated in Table 5.1. Each of the above-mentioned sections will be discussed in the following sections.
There is no restriction in terms of the letter cases used in the geometry, that is lower, upper, and mixed case may be used. The comment character “c” indicates a comment line at the beginning of the line. A dollar sign “$” sign is used to give a detailed comment about the particular line. Tabs are restricted in the geometry as they are not recognized by the code; instead the space bar should be used at all times. The output file will contain all the results calculated by the code as was instructed in the input file and the copy of the particular input file would still be included in the beginning of the output file.

5.3.1 **Cell Cards and Surface Cards**

Cell cards and surface cards are discussed together, as cells are the units of the geometry. Each (x, y, z) coordinate defined in the input geometry must belong to a cell or be on the surface of a cell and there can be no “gaps” in the geometry Shultis et al., (2006) except when the blank line delimiter is expected. Simple cells can have several surfaces which are combined using Boolean operators i.e. Intersection (logical AND) (space); Union (logical OR) (;) and Complement (logical NOT) (#) Mueller (2003). Table 5.2 below is a representation of the cell card format.
In Table 5.2 above, the cell number is denoted by “j”, which can be any integer between 0 and 99999. The material number “m” identifies the material present in a particular cell and is also an integer from 0 to 99999. The character “d” defines the cell material density and if it is given with a negative sign it indicates the density unit to be in grams per cubic centimeter (\( \frac{g}{cm^3} \)) and if it is given with a positive sign therefore the density unit is in atoms per barn-centimeter (\( \frac{atoms}{barn \cdot cm} \)). Then, “geom” lists all the surfaces that specify the cell Goorley (2010).

In the example specified above, the surface defined is a spherical and as a result, the “-1” denotes the surface inside the sphere in cell 1. The surface’s coefficient goes with the shape of material in application. The params on the cell card format are used to specify the cell parameters, for example the importance meaning of the cell gives it weighting and keeping its population constant. That implies “imp:p=1” the photons particles in cell 1 will likely be tracked, if the “imp:p=0” it means that the photons particles will not be tracked for that particular cell.

The surface card format is presented in Table 5.3 below and each parameter will be discussed.

Table 5.3: Surface card format.

<table>
<thead>
<tr>
<th>Surface #</th>
<th>surface mnemonic</th>
<th>list</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>cz</td>
<td>4.935</td>
</tr>
</tbody>
</table>

The character “j” represents the surface number (1-99999) and must start in columns 1-5 of the surface cards. The surface type, represented here by the character “a”, is the next parameter in the surface
definition. The “list” feature on the surface card is a space for the user to list numbers that describe the surface, such as dimensions and radius in cm. The mnemonic “cz” defines an infinite cylinder centered on the z-axis with a radius of 4.935 cm.

5.3.2 Data Cards

Data card which is the last section of the geometry defines the following input cards: problem type, problem materials which deal with the specification of materials filling the various cells, source specification, tally specification and cross-section specification Shultis et al., (2006).

The problem type makes use of the “MODE” card, and this card specifies which particles being are transported in the geometry. MODE P E refers to the photon particle and an electron particle respectively. Source specification uses the “SDEF” function to describe the type of source to be tracked (Wilson, 2011). The source definition uses the following command SDEF PAR ERG. The source definition differs with the problem and the following functions can be used: source information “SIn”, and source probability “SPn” where n is an arbitrary number 1-9999. The energy, time, direction, position, weight are defined as they form part of the general source definition (reference). The “par” command on the SDEF line tells the code what type of particle is the source emitting. The par = 1 refers to the neutron source, par = 2 the photon source and par = 3 refers to the electron source. If the par command is not specified, the code will assume the default particle to be the neutron.

ERG command defined on the SDEF function describes the energy of the emitted particle (in MeV). All energies are defaulted to MeV as indicated in Table 5.4, and cannot be changed.

Tally card is a tool that MCNP uses to calculate the linear functions of current or fluence, energy deposition etc. Tallies can be made at different volumes, surfaces, or points in space where the emitted source particles interact and can be counted. For the most part, a fluence tally is measured in particles/cm² as shown in Table 5.4. The Table 5.4 gives the types of tallies to be used for a particular problem.
Table 5.4: A Table showing the types of tallies, the type of particles denoted by pl. Tally type particle pl Fn Units, *Fn Units.

<table>
<thead>
<tr>
<th>Mnemonic</th>
<th>Tally type</th>
<th>Particles pl</th>
<th>Fn Units</th>
<th>*Fn Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>F1: pl</td>
<td>Surface current</td>
<td>N or P or N,P or E</td>
<td>#</td>
<td>MeV</td>
</tr>
<tr>
<td>F2: pl</td>
<td>Average surface flux</td>
<td>N or P or N,P or E</td>
<td>$\frac{#}{cm^2}$</td>
<td>$\frac{MeV}{cm^2}$</td>
</tr>
<tr>
<td>F4: pl</td>
<td>Average flux in a cell</td>
<td>N or P or N, P or E</td>
<td>$\frac{#}{cm^2}$</td>
<td>$\frac{MeV}{cm^2}$</td>
</tr>
<tr>
<td>F6: pl</td>
<td>Energy deposition</td>
<td>Nor P or N, P</td>
<td>$\frac{MeV}{g}$</td>
<td>$\frac{jerks}{g}$</td>
</tr>
<tr>
<td>F7: pl</td>
<td>Fission energy deposition in a cell</td>
<td>N</td>
<td>$\frac{MeV}{g}$</td>
<td>$\frac{jerks}{g}$</td>
</tr>
<tr>
<td>F8: pl</td>
<td>Pulse height distribution in a cell</td>
<td>P or E or P, E</td>
<td>pulses</td>
<td>MeV</td>
</tr>
</tbody>
</table>

Cross-section specification refers to the probability of the occurrence of a specific interaction by a single radiation quantum. The examples of a cross-sectional area is a rectangular circular cylinder (RCC), rectangular parallelepiped (RPP) located in a radiation field that is perpendicular to the incident beam Van Rooyen (2012). The code uses MCNP cross-section tables and nuclear models that contain (p,nX) and (p,γX) reactions, i.e. the cross-sections for nuclear reactions with ionizing photons in the exit channel. Whenever the tabular cross-section data is not available in the libraries, the code makes use of nuclear models. In this way, the production of secondary types of ionizing radiation such as ionizing photons with accurate spectra and angular distributions are accounted for Van Rooyen (2012).

A material card is simply many lines of input information structures that look like this: Mn ZAID$_1$ fraction$_1$ ZAID$_2$ fraction$_2$

where M is the letter “M” and

n is an arbitrary number known as the material number. ZAID number is a six digit element/ nuclide identifier. E.g. ZZZ = atomic number, AAA = atomic mass number. $^{235}_{92}U$ = 092235.
5.4 MCNP Geometry Commands

The geometry plotting can be done using the command line MCNP ip i=filename, where “p” character stands for plot. The running of the code is executed by the following command: MCNP ix i=filename. The geometry input file with the command lines is saved into the directory containing MCNP files. At a command window, the command cd MCNP is entered to direct the execution commands to look up in the MCNP location. The MCNP command, i=filename will process the input file and gives “mcrun is done” to indicate the end of the run. The output files outp and runtpe will be created and saved automatically into the MCNP directory. If the outp file exists then the outq file will be created instead. The same applies to the runtpe which will be runtpf instead. The plotting geometry is defined as MCNP z runtpe=runtpf, which will give the mcplot at the end. After the mcplot, the command plot is typed in order to give a plot window. The plot window can be increased by typing ex 12, ex 20 or any small number or decreased by typing ex 32, ex 45 etc. The mcplot can also plot the tally as defined in the input geometry.

Output files vary in size. However, the storage requirement for the phase space files containing millions of photons may be up to tens of gigabytes. In order to save memory space the PRDMP command is specified to direct the code as to how long it should save the output files Shultis et al., (2006). In addition, Monte Carlo simulation of a single beam can take up to several days on a desktop computer. Simulation of all the beams used in a radiotherapy clinic would represent a considerable burden on the computing resources of an average radiotherapy department Deng et al., (2000).

The command PRINT is only required to tell the code to produce tables that contains lots of useful data in the output file. The last command CTME on the input file is the particle cut off time. This tells the code how many particles to run before stopping provided that the code has no fatal errors. The particle cut off command looks like this: CTME=1000000, where 1000000 is the number of particles to be run. The particle cutoffs (low energy particles) are either of zero importance or almost zero importance Booth (1985). The results will be biased (low) if the particle cutoff (energy cutoff) is killing particles that might have contributed. The time cutoffs should only be used in time-dependent problems where the last time bin will be earlier than the cutoff. The Variance Reduction technique in Monte Carlo calculations can often reduce the computer time required to obtain results of sufficient precision and acceptable relative error Booth (1985) and Shultis et al., (2006).
The main objective for applying the variance reduction is to spend more time sampling important cells and less time sampling insignificant cells. The more particles are run, the less the amount of error produced.

The most valuable tool to assess the reliability of the results obtained is the set of 10 statistical tests performed on the tally. If any of the tests are failed, MCNP automatically produces the additional output to help the user in interpreting the urgency of the failed test(s). The ten statistical tests are summarized below in Table 5.5.

<table>
<thead>
<tr>
<th>Tally Mean, $\bar{x}$</th>
<th>1. The mean must exhibit, for the last half of the problem, only random fluctuations as $N$ increases. No up or down trends must be exhibited.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Relative Error, $R$</td>
<td>2. $R$ must be less than 0.1 (0.05 for point/ring detectors). 3. $R$ must decrease monotonically with $N$ for the last half of the problem. 4. $R$ must decrease as $\frac{1}{\sqrt{N}}$ for the last half of the problem.</td>
</tr>
<tr>
<td>Variance of the Variance, (VOV)</td>
<td>5. The magnitude of the VOV must be less than 0.1 for all types of tallies. 6. VOV must decrease monotonically for the last half of the problem. 7. VOV must decrease as $\frac{1}{N}$ for the last half of the problem.</td>
</tr>
<tr>
<td>Figure of Merit, (FOM)</td>
<td>8. FOM must remain statistically constant for the last half of the problem. 9. FOM must exhibit no monotonic up or down trends in the last half of the problem.</td>
</tr>
<tr>
<td>Tally PDF, $f(x)$:</td>
<td>10. The SLOPE determined from the 201 largest scoring events must be $&gt; 3$.</td>
</tr>
</tbody>
</table>
5.5 The Use of MCNP

MCNP code has been applied effectively industrially as well as in the medical field. The code is accessible to health physicists, medical physicists, and radiological engineers using desktop or laptop personal computers Seagraves (2012). Industrially, the code is used for nuclear criticality safety analyses Goorley et al., (2010).

In medical physics field the code has been employed in all related departments, that is radiotherapy, nuclear medicine and radiology. Siebers et al., (2000) and Bush (2009) verified that Monte Carlo dose calculation algorithms have the potential for higher dose accuracy as they have transported particles and computed the absorbed dose to the patient media such as soft tissue, lung or bone. Sood et al., (2009) reviewed the methods and new data used to include photon Doppler energy broadening in MCNP5 since this effect was not accounted for when using MCNP4C3. The comparison is made from the existing libraries of MCNP4C3 to the recent MCNP5 libraries.

Goorley et al., (2006, 2007) developed MCNP medical physics analytical and voxelized database due to the developing interest in MCNP medical physics calculations for organ–specific dose calculation, code comparison or geometric representation studies. Seven modelling problems were simulated by Redd (2003) but only three of them will be listed. The first model was for the determination of the near field angular anisotropy and dose distribution from a high dose rate $^{192}$Ir brachytherapy source in a surrounding spherical water phantom. The second model was used to obtain air Kerma backscatter profiles for 150 kVp and 120 kVp X-rays upon a water phantom. The third one was to determine indirect spectral and energy fluence upon two neutron detectors within a calibration bunker. The largest indirect contribution was found to come from low energy neutrons with an average angle $47^\circ$ where $0^\circ$ is a plane parallel to the floor.
Chapter 6: Materials and Methodology

6.1 Basic Properties of Silicone Gel-Filled Breast Prostheses

Silicones are inert, synthetic compounds with a variety of forms and uses in medical applications. The general properties of silicone as discussed by Shin-Etsu silicones (2012) include: low thermal conductivity, low chemical reactivity, low toxicity and thermal stability, their moisture and high heat resistance. Silicones do not stick to many substrates; but adheres very well to others, e.g. glass and does not support microbiological growth, flame retardancy. Silicones have excellent resistance to wind, rain and ultraviolet rays as well as radiation resistance Shin-Etsu silicones (2012).

Silicone prostheses are pre-filled with silicone gel; a thick and sticky fluid that closely mimics the feel of a human fat. Bondurant et al., (1999) mentioned that breast prosthesis made up of silicone refers to a large family of organic silicon polymer products with a main chain of alternating silicon and oxygen atoms. Typically, each silicon in the chain carries two methyl groups (CH$_3$, where C=Carbon and H=Hydrogen) and an oxygen, and the materials are called poly-dimethylsiloxane (PDMS). The density of a silicone gel ranges from $0.97 \frac{g}{cm^3}$ to $1.29 \frac{g}{cm^3}$. When silicon is mixed with oxygen, hydrogen and carbon; it becomes silicone.

6.1.1 Structure of the Silicone Gel-Filled Breast Prostheses

Figure 6.1: Silicone gel filled with “cohesive” shell-gel.
Low molecular weight silicones form oils, middle molecular weight silicones forms gels whereas high molecular weight silicones form elastomers and rubbers Sarosy (2012). Silicone gel-filled breast prostheses used medically have a silicone rubber shell made of polysiloxane(s), such as polydimethylsiloxane (PDMS) and polydiphenylsiloxane. Polydimethylsiloxane is filled with a fixed amount of silicone gel and has got a molecular composition as indicated in Figure 6.2. This prosthesis may vary in shell surface, shape, profile, volume and shell thickness.

### 6.2 Linear Accelerator Geometry

The physical measurements were carried out in a linac room. The room lights were dimmed for the alignment of the $40 \times 40 \times 40$ cm$^3$ water tank using the wall built-in lasers. The water phantom positioned on a stand was placed under the treatment head of the linear accelerator as indicated in Figure 6.3. The water tank was filled and the SSD of 100 cm was set to the surface of the water. Figure 6.3 portrays the assembly of the linear accelerator with the water phantom positioned below its gantry.
The SSD of 100 cm was observed with the help of the optical distance indicator embedded inside the gantry head. The gantry and the collimator angles were both rotated from 0°. The phantom was aligned using the positioning lasers on the wall and the center markers on the phantom. The orientation (x, y, z) of the phantom which is of great importance was considered during the alignment. RFA MCU Scanditronix Wellhöfer control unit cable was connected to the water phantom serial port for communication whilst the control unit was switched off.

The RS323 cable was connected from the control unit to the computer installed with the Omni-Pro Accept software for plotting PDD curves and the beam profile curves. The computer was started and the control unit switched ON in order to begin with the adjustments of the diodes. Collimators were adjusted to give a reference field size of $10 \times 10 \text{ cm}^2$ so that the light beam can strike on the center of the field (central axis) following the jaws orientation inside the gantry.

The Photon Field Diode (PFD) was aligned to be at the centre of the field by using the SSD of 100 cm and the cross wires. The PFD motion was enabled by the connection of the Linear Current Booster (LCB) motor cable to the local control port on the water phantom. The reference diode was also aligned in a fixed position in the field to monitor the beam intensity with time without obstructing the
beam. The PFD was adjusted to move in a water phantom to sample the dose rate at various points whereas the reference diode remained fixed throughout the measurements.

The PDD curves for energies 6 MV and 15 MV at the field size of $10 \times 10 \text{ cm}^2$ were plotted. The source surface distance was kept constant at 100 cm. The point of normalization of all the depth dose curves was 100 %. The cross plane profiles for energies 6 MV and 15 MV were also measured and the field size of $10 \times 10 \text{ cm}^2$ at the depth of 10 cm inside the water phantom. All the beam dose profiles were normalized at 100 %.

### 6.3 MCNP Geometries

The water phantom model illustrated in Figure 6.4 was simulated using the MCNP code as was explicitly explained in Section 5.3. The physical measurements in a water phantom as described in Section 6.2 was reconstructed and will be used to benchmark against the MCNP calculation in the same set-up in order to validate the MCNP code.

This code has been trustworthy for its accuracy in particle transport Bush (2009) and Wilson (2011) therefore, it has been chosen for this particular study. The water phantoms were simulated at a field size of $10 \times 10 \text{ cm}^2$ for both 6 MV and 15 MV photon beams. Only the geometry at a field size will be used for validation on both energies. The validation of the code will be calculated based on the study performed by Subbaiah et al., (2010), where the ratio between the measured data and the MCNP calculated data will be used to predict the accuracy of the MCNP code. If the ratio is less than 5% then the MCNP code is worthy of trust in generating the accuracy of the data Subbaiah et al., (2010).
Figure 6.4: The schematic 3-D diagram of a $30 \times 30 \times 30$ cm$^3$ water phantom with 6 cm silicone gel slab/water voxels as simulated using MCNP.

The geometry problem was defined on a notepad in a simple text format Burgett (2007). The title of the MCNP geometry is defined as “Linac Beam Strikes Water Phantom” and this title will be used to distinguish each geometry in the input file as well as being echoed in the output file. It is important to define the surface cards prior to the cell card definition. Numerous surface cards have been used to define a water phantom in this geometry, but the sequence does not matter. They are defined as follows:

Surface Cards

05 RPP -20 +20 -20 +20 +100 +140 Water Phantom Surface

The above surface card defines the surfaces using the macrobody format, where a rectangular parallelepiped (RPP) surface 05 was defined as $X_{\text{min}} = -20$ cm and $X_{\text{max}} = +20$ cm. This information defines the region of intersection on the surface that cuts on the X-axis starting from -20 cm up to +20 cm where the minus sign refers to the left of the plane and the positive sign refers to the right of the Cartesian plane. The next set of points indicate the $Y_{\text{min}} = -20$ cm and the $Y_{\text{max}} = +20$ cm region of the surface on the Y-axis as well as the last set of points that indicate $Z_{\text{min}} = +100$ and $Z_{\text{max}} = +140$ cm region of the surface on the z-axis. The whole surface region defines the surface numbers that make up the water phantom.
The above line defines the outer surface of the water phantom, which is referred to as the Umwelt boundary surface as indicated in Figure 6.3 above. That is also defined in terms of a rectangular parallel piped (rpp) or a box. 999 is surface that will be recalled by cell card during the simulation. The following surface cards define the planes that will be used to divide the interior part of the water phantom.

The interior part of the water phantom was defined to give out the cylindrical shape of the water phantom in order to be able to calculate the radial distribution of the dose. For the axial dose distribution the planes normal to the z-axis have been given below.

21 PZ +101 Plane in z-axis
22 PZ +102 Plane in z-axis
23 PZ +103 Plane in z-axis
.
.
.
47 PZ +127 Plane in z-axis
48 PZ +128 Plane in z-axis
49 PZ +129 Plane in z-axis

The surface 21 is used to define a plane (PZ), normal to the z-axis. 101 refer to the plane at a point of z=101 cm and the positive sign indicates above the plane. The following other planes up to surface 49 have the same parameters as surface 21 but differs with surface number and the intersection region increases by 1 cm as the z-axis increases by going down. The purpose for defining the above mentioned surface cards is to be able to track the axial dose distribution of the photon beam.
61  CZ  +0.5  cyl
62  CZ  +1.5  cyl

70  CZ  +9.5  cyl
71  CZ  +10.5  cyl

The above surface card refers to surface 61; CZ refers to the cylinder on z-axis with a radius of 0.5 cm from the centerline of the beam. The positive sign indicate that the cylinder has got a positive sense with reference to the origin/centerline of the beam. The surface cards that follow surface 64 to surface 75 are not indicated here but have the same sense parameters as the ones shown above. They have an increment of 1 cm up to the +15 cm so as to track the dose distribution radially for the required field size of 16 cm × 16 cm.

It has been explained in Section 5.3.1 that the cells are the basic unit of the MCNP geometry Mckinney (2008); therefore the subsection of the geometry called cell cards can be defined since the surface cards are known.

CELL CARDS
01  4  -1.205E-3  -999  +5  imp:e,p=1  All other Air

As clearly indicated in Section 5.3.1 the 01 in the first line refers to the cell number that the photon beam is going to interact with. Cell 01 contains material number 4 refers whose composition is defined in the data card section. The density of material, dry air is given by -1.205E-3 and the minus sign indicate that the SI unit for the material density is given by grams per cubic centimeter (g/cm³). On the cell card, surfaces are related to the cells by a surface number with either a “+” or “-“. -999 and +5 shows that cell 01 is bounded by surfaces 999 and 5 respectively. The -999 in this case is a surface 999 defined with respect to the cell 01 assigned to the inside region of the Umwelt boundary surface. Whereas the +5 refers to the outside region of the surface 5, defines as the water phantom surface.

The imp:e,p=1 is referring to the importance of the particle to be tracked in cell 01. Imp:e,p=1 means that electron and photon particles should be tracked in cell 1.
The cell 999 is bounded by surface 999 with a positive sign indicating the outside region of the surface 999, i.e. outside the Umwelt boundary surface. Cell 999 is different, its material number is zero (0), therefore it has no density. This is called a “void” cell. The imp:e,p=0 is referring to the importance of the particle to be tracked in cell 999. Imp:e,p=0 means that no electron or photon particles will be tracked in it, and MCNP will not track down any particle that enters it.

Figure 6.5: Schematic and cross-sectional water phantom.

011 3 -1.0 -5 -21 -61 vol=03.142 imp:e,p=1 Water, Z-Layer 01, R-layer 01
The above command line indicates that cell 011 consists water indicated as material 3 with a density of $1.00 \frac{g}{cm^3}$ and is bounded by surfaces 5, 21 and 61. The volume inside the cell is 3.142 cm$^3$, only electrons and photons are tracked in the cell.

012 3 -1.0 -5 -21 -62 +61 vol=09.425 imp:e,p=1 Water, Z-Layer 01, R-layer 02
013 3 -1.0 -5 -21 -63 +62 vol=15.708 imp:e,p=1 Water, Z-Layer 01, R-layer 03
014 3 -1.0 -5 -21 -64 +63 vol=21.991 imp:e,p=1 Water, Z-Layer 01, R-layer 04
015 3 -1.0 -5 -21 -65 +64 vol=28.274 imp:e,p=1 Water, Z-Layer 01, R-layer 05
016 3 -1.0 -5 -21 -66 +65 vol=34.558 imp:e,p=1 Water, Z-Layer 01, R-layer 06
017 3 -1.0 -5 -21 -67 +66 vol=40.841 imp:e,p=1 Water, Z-Layer 01, R-layer 07
The above commands give the input data for layer 1 in the z-axis of 1 cm as should be read by the code. The layer consists of a different cell 11 up to 25. The data commands are similar to cell 11 except for the surface cards. The following set of cell card commands is also similar to the cell 11 except that this is a second layer with different surface cards. The medium of preference is still water as indicated by material 3 with density of 1.00 g/cm³. The layers continue up to 30 shown cross-sectionally in Figure 6.5. The same data commands applies for the radial part of the problem, which is starting from the center of the circle. It goes up to the concentric circle of 15 cm radius, which is given in Figure 6.5.

018 3 -1.0 -5 -21 -68 +67 vol=47.124 imp:e,p=1 Water, Z-Layer 01, R-layer 08
019 3 -1.0 -5 -21 -69 +68 vol=53.407 imp:e,p=1 Water, Z-Layer 01, R-layer 09
020 3 -1.0 -5 -21 -70 +69 vol=59.690 imp:e,p=1 Water, Z-Layer 01, R-layer 10
021 3 -1.0 -5 -21 -71 +70 vol=65.973 imp:e,p=1 Water, Z-Layer 01, R-layer 11
022 3 -1.0 -5 -21 -72 +71 vol=72.257 imp:e,p=1 Water, Z-Layer 01, R-layer 12
023 3 -1.0 -5 -21 -73 +72 vol=78.540 imp:e,p=1 Water, Z-Layer 01, R-layer 13
024 3 -1.0 -5 -21 -74 +73 vol=84.823 imp:e,p=1 Water, Z-Layer 01, R-layer 14
025 3 -1.0 -5 -21 -75 +74 vol=91.106 imp:e,p=1 Water, Z-Layer 01, R-layer 15
026 3 -1.0 -5 -21 -76 +75 vol=97.389 imp:e,p=1 Water, Z-Layer 01, R-layer 16
027 3 -1.0 -5 -21 -77 +76 vol=103.773 imp:e,p=1 Water, Z-Layer 01, R-layer 17
028 3 -1.0 -5 -21 -78 +77 vol=109.157 imp:e,p=1 Water, Z-Layer 01, R-layer 18
029 3 -1.0 -5 -21 -79 +78 vol=114.540 imp:e,p=1 Water, Z-Layer 01, R-layer 19
030 3 -1.0 -5 -21 -80 +79 vol=120.924 imp:e,p=1 Water, Z-Layer 01, R-layer 20
031 3 -1.0 -5 -22 +21 -61 vol=03.142 imp:e,p=1 Water, Z-Layer 02, R-layer 01
032 3 -1.0 -5 -22 +21 -62 +61 vol=09.425 imp:e,p=1 Water, Z-Layer 02, R-layer 02
033 3 -1.0 -5 -23 +22 -61 vol=03.142 imp:e,p=1 Water, Z-Layer 03, R-layer 01
034 3 -1.0 -5 -23 +22 -62 +61 vol=09.425 imp:e,p=1 Water, Z-Layer 03, R-layer 02
035 3 -1.0 -5 +49 -73 +72 vol=78.540 imp:e,p=1 Water, Z-Layer 30, R-layer 13
036 3 -1.0 -5 +49 -74 +73 vol=84.823 imp:e,p=1 Water, Z-Layer 30, R-layer 14
037 3 -1.0 -5 +49 -75 +74 vol=91.106 imp:e,p=1 Water, Z-Layer 30, R-layer 15
900 3 -1.0 -5 +75 imp:e,p=1 Remainder of Water Phantom
The cell 900 gives the remaining part of the water phantom extending in the z-axis. It is only bounded by surfaces 5 and 75. The interaction of both electron and photon particles is expected but will not be used to calculate the dose distributions for this particular case.

Data cards
Mode p e
The type of particle to be tracked down during interaction is the photon and electron.

SDEF par = P
Erg = D4
SDEF command gives the fixed source; par tells the photon particle is emitted by the fixed source. ERG is a continuum spectrum described in terms of the distribution 4 since there is more than one cell to locate the source. D4 refers to distribution four, elaborated in SI4, SP4 card.

pos = 0 0 0, is a vector quantity and gives the position of the source from origin in terms of axis.
axs = 0 0 1, is the axs command gives the source in a cylinder that protrudes in the z-axis (0 0 1), axially.
vec = 0 0 1, is the vector is positioned in the z-axis.
dir = +1, gives the direction of the source is in the positive z-axis.
ext = 0, indicates that the source in a cylinder is not extended
rad = D1, gives the radial distribution of the radiation emitted by the source. it is defined in terms of SI1 and SP1 below.

si1 h 0 5, this command says that the source is evenly distributed along 0 and 5 cm radius, giving a field size of $10 \text{ cm} \times 10 \text{ cm}$.
sp1 -21 1 command tells the code to sample uniformly throughout the volume of the source proportionally with respect to a radial distribution; i.e. following the power law of a circle.

The PHYS card gives the physics information in relation to the energy simulated.
phys:p 20 p=20 is the maximum possible energy that can ever be found in the device.
0 no energies will go above 20 MeV.
-1 1 0 switches ON the possibility of photonuclear reactions.
The SI4 is the energy spectrum and SP4 is probability of occurrence of the energy generated from the source geometry. The spectrum generated here is for a 6 MV photon beam.

<table>
<thead>
<tr>
<th>SI4</th>
<th>SP4</th>
</tr>
</thead>
<tbody>
<tr>
<td>L</td>
<td>D</td>
</tr>
<tr>
<td>8.20000E-03</td>
<td>1.03013E-03</td>
</tr>
</tbody>
</table>

4.20000E-02  9.76766E-03
9.00000E-02  6.74180E-02
1.80000E-01  1.05669E-01
2.80000E-01  1.62836E-01
3.80000E-01  1.39217E-01
4.80000E-01  1.08599E-01
5.08000E-01  9.81731E-03
5.18000E-01  1.25354E-02
5.44000E-01  2.62002E-02
5.90000E-01  3.91823E-02
6.40000E-01  3.56575E-02
6.90000E-01  3.26292E-02
7.40000E-01  2.96877E-02
7.90000E-01  2.74661E-02
8.40000E-01  2.49715E-02
8.90000E-01  2.26133E-02
9.40000E-01  2.00938E-02
9.90000E-01  1.87782E-02
1.20000E+00  7.87370E-02
1.45000E+00  5.68808E-02
1.70000E+00  4.37000E-02
1.95000E+00  3.37214E-02
2.16800E+00  2.35317E-02
2.24200E+00  4.08330E-03
2.45000E+00  2.16204E-02
2.70000E+00     1.74254E-02
2.95000E+00     1.36400E-02
3.20000E+00     1.18403E-02
3.45000E+00     9.74284E-03
3.70000E+00     7.33505E-03
3.95000E+00     6.30492E-03
4.20000E+00     4.71628E-03
4.45000E+00     3.76061E-03
4.70000E+00     3.15246E-03
4.95000E+00     2.09750E-03
5.20000E+00     1.30318E-03
5.45000E+00     7.94321E-04
5.70000E+00     6.08152E-04
5.95000E+00     1.24113E-04

M1 defines the composition of the tungsten nuclide as evoked in the cell card.

m1     74180      +0.12
       74182      +26.50
       74183      +14.31
       74184      +30.64
       74186      +28.43

M2 defines the composition of the lead nuclide as evoked in the cell card

m2     82206     +24.1
       82207     +22.1
       82208     +52.4

M3 defines the composition of the pure water as evoked in the cell card

m3      1001     +1.9995
       1002     +0.0005
       8016     +1.0
m4  6000  -1.24E-4  Air
    7014  -0.755267  Air
    8016  -0.231781  Air
    18000 -0.012827  Air
M4 defines the composition of the dry air as evoked in the cell card.

ENERGY SPECTRUM OF IONISING PHOTONS @ centre line R = 0

F14:p  11  31  51  71  91  111  131  151  171  191  211  231  251  271  291
    311  331  351  371  391  411  431  451  471  491  511  531  551  571  591

F14 refers to the cell flux tally which is the number of particles per unit area. In this case the number of photons (p) per unit area in cells 11, 13, 51, 71, …, 551, 571, and 591.

E14    0.001  0.005  0.01  0.05  0.1  0.2  0.3  0.4  0.50  0.51  0.52  0.55  0.6  0.65 0.7  0.75 0.8  0.85
    0.9  0.95 1.0 1.25 1.5 1.75 2.0 2.21 2.25 2.5 2.75 3.0 3.25 3.5 3.75 4.0 4.25 4.5 4.75 5.0
    5.25 5.5 5.75 6.0 6.25 6.5 6.75 7.0 7.25 7.5 7.75 8.0 8.25 8.5 8.75 9.0 9.25 9.5 9.75 10
    11  12  13  14  15  16  17  18  19  20  21  22  23  24  25  26  27  28  29  30  31  32  33  34  35  36  37
    38  39 40 41 42 43 44 45 46 47 48 49 50

E14 refers to the energy bins applied to the F14 card, which gives the integral of the energy from the first energy to the maximum incident energy. The 0.001 energy bin means that all the particles with the energy of 0.001 MeV should be deposited in that particular energy bin. In this case the maximum energy that could be obtained is the 5.9 MeV since it is the maximum incident energy.

fm14 3.126E7, tally normalization factor to calibrate and correct for Dose rate * 1.04E8 only if dose rate in Gy /min particles per second

TMESH, tmesh command gives a mesh for the above defined tally.

Rmesh11: p flux, a rectangular mesh tally for a photon flux is defined.
    Cora11 -50 99i +50, coordinates a for mesh tally for photon flux
    Corb11 -50 99i +50, coordinates b for mesh tally for photon flux
    Corc11 +100 130, coordinates c for mesh tally for photon flux

Rmesh21: e flux, a rectangular mesh tally for an electron flux is defined.
    Cora21 -50 99i +50, coordinates a for mesh tally for electron flux
    Corb21 -50 99i +50, coordinates a for mesh tally for electron flux
Corc21 +100   +130 , coordinates a for mesh tally for electron flux
ENDMD, end of mesh tally data.

PRDMP -500 -500  0  5 J, this is the print and dump cycle card, which tells the code to create a
RUNTPE file every 500 minutes to avoid consumption of computer memory.
CTME 1800, ctme gives the computing cutoff time in minutes.

6.3.1 MCNP simulations with a silicone gel prosthesis in the water phantom

The simulation procedure in Section 6.3 above was repeated. The simulation geometry indicated in
Figure 6.6 where a 6 cm layer of material 3 (water) is substituted by the material 6 (silicone gel layer)
just after the first layer of 1 cm of material 3. Figure 6.6 below indicates layout of the geometry of the
silicone gel prostheses described in the input file.

![Figure 6.6: The axial view of the cylinder with a 6 cm layer of silicone gel.](image)

All geometries of 6 MV energy beam are simulated with a field size of $10 \times 10$ cm$^2$ similarly, all
geometries of 15 MV energy beam are simulated at a field size of $16 \times 16$ cm$^2$. They are both defined
at a constant SSD of 100 cm. The silicone gel prostheses thickness was varied from 4 cm to 16 cm with
an increment of 2 cm for both energies to evaluate the dose deposition in the cells.
Chapter 7: Results and Discussions

7.1 Physical measurements

7.1.1 The Measured Percentage Depth Dose Curves for 6 MV and 15 MV Photon Beams at a field size of 10 × 10 cm² in a Water Phantom

Figure 7.1: Measured PDD curves for 6 MV and 15 MV at a field size of 10 × 10 cm² in a water phantom.

Figure 7.1 gives the percentage depth dose curves for 6 MV and 15 MV. The build-up region for 6 MV photon beam energy as indicated in Figure 7.1 starts from the depth of 0 cm up to the depth of 1.8 cm where the curve reached an electronic equilibrium at the peak dose of 100%, also referred to as the maximum dose. Similarly, the 15 MV photon beam achieve the build-up region from a depth of 0 cm building up to the maximum depth of 2.6 cm giving a dose of 100%. The curve for 6 MV photon beam starts to drop from the depth of 2 cm by 5%, with every increment of 0.2 cm in depth, the dose is attenuated by 5%. These effects of inverse square law and attenuation in water is supported by Podgorsak (2003), indicating that when the photon beam traverses in a water phantom its intensity is reduced as the distance from the source is increased. The curve for 15 MV photon beam starts to drop by 2% at a depth of 2.8 cm.
7.1.2 The Measured Dose Profiles for 6 MV and 15 MV Photon Beams at a Field Size of 10 × 10 cm² in a water phantom

Figure 7.2: Measured cross-plane dose profile for 6 MV and 15 MV at a field size of 10 × 10 cm² in a water phantom.

Figure 7.2 gives the results obtained from the set-up in Section 6.2. The Figure represents the relative dose against the off axis distance. Dose distribution along the beam central axis give only part of the information required for an accurate dose distribution in a phantom Podgorsak (2005). The beam flatness was calculated using the following equation:

\[
F = \frac{D_{\text{max}} - D_{\text{min}}}{D_{\text{max}} + D_{\text{min}}} \times 100
\]  

(7.1)

where the \(D_{\text{max}}\) for 6 MV photon beam at 80% of the beam width was found to be 101.0% at 2.6 cm. The \(D_{\text{min}}\) was found to be 100% at 0 cm within the 80% of the beam width. Using the above data, the beam flatness is equal to 0.5%. For 15 MV the flatness was calculated to give 0.5%. The obtained beam flatness for the above curves fall within the 3% range as recommended by Kouloulias et al., (2003).
The beam symmetry was calculated using the following equation:

\[ S = 100 \times \frac{\text{area}_{\text{left}} - \text{area}_{\text{right}}}{\text{area}_{\text{left}} + \text{area}_{\text{right}}} \]  \hspace{1cm} (7.2)

The data calculate beam symmetry was tabulated in appendix A-2. The \( D_{\text{max}} \) for \( \text{area}_{\text{left}} \) is given as 101.0\% and the \( D_{\text{max}} \) for \( \text{area}_{\text{right}} \) is given as 102\%. When using the beam symmetry equation for 6 MV, \( S = 0.5\% \). The beam symmetry for 15 MV was obtained to be 0\%. The beam symmetry for both energies falls within the 3\% range.

### 7.2 MCNP simulations

Figure 7.3 shows the simulated 30 × 30 × 30 cm\(^3\) water phantom in absence of inhomogeneities. The large water phantom contained water as the material of interest and as the medium for transporting photon particles. The MCNP photon dose rate calculations were yielded by the generation of random distribution of dose in energy bins defined in Section 5.3 according to the treatment geometry characteristics and effective energy spectrum. The field size of 10 × 10 cm\(^2\) was chosen to be at the central-axis of the beam for the calculation of absorbed dose, whilst the 16 × 16 cm\(^2\) was used with an
assumption that it conform the dose coverage of the largest size of the breast in external beam radiotherapy (EBRT).

7.2.1 Percentage Depth Dose Curves for 6 MV and 15 MV Photon Beams at Field Size of 10 × 10 cm² in a Water Phantom

Figure 7.4 shows the MCNP simulated 30 × 30 × 30 cm³ water phantom in absence of inhomogeneities. The two PDD curves are an indication of how the photon fluence behaves in a volume of water. The relative dose curves were calculated at a field size of 10 × 10 cm² in a water phantom. The 6 MV PDD curve gave a build-up of dose from a depth of 0.5 cm with a relative dose rate of 6.7E-03 Gy/initial particle and reached a maximum dose at a depth of 1.5 cm with a relative dose of 7.8E-03 Gy/initial particle. The 15 MV PDD curve obtained gave a dose build up from the depth of 0.5 cm with a relative dose rate of 8.37E-03 Gy/initial particle and reached a maximum dose of 1.25E-02 Gy/initial particle at a depth of 3.0 cm. The two calculated curves follow the trend of a typical PDD curve.
Figure 7.5: MCNP simulated dose profile curves for 6 MV and 15 MV at a field size of $10 \times 10$ cm$^2$ in a water phantom at a depth of 10 cm.

The beam flatness and beam symmetry for the 6 MV photon dose profile indicated in Figure 7.5 was calculated. The beam symmetry was obtained as 0.9%, and the beam flatness was obtained to be 5%.
7.3 Validation of the MCNP code

Table 7.1: Comparison of measured PDD data with the computed PDD data for a 6 MV photon beam in a water phantom.

<table>
<thead>
<tr>
<th>Depth (cm)</th>
<th>6 MV PDD</th>
<th>Output Factor (OF) (10 × 10 cm²) (measured)</th>
<th>Scatter Factor (SF) (10 × 10 cm²) (calculated)</th>
<th>Rel. dose factor (RDF) measured (Gy/min)</th>
<th>Rel. dose Rate (MCNP calculated) (Gy/min)</th>
<th>Ratio = ( \frac{\text{measured dose rate}}{\text{calculated dose rate}} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.5</td>
<td>0.78</td>
<td>1.0</td>
<td>1.2E+00</td>
<td>9.5E-03</td>
<td>4.0E-01</td>
<td>0.02</td>
</tr>
<tr>
<td>1.5</td>
<td>1.00</td>
<td>1.0</td>
<td>1.5E+00</td>
<td>1.5E-02</td>
<td>4.7E-01</td>
<td>0.03</td>
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<tr>
<td>2.5</td>
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<td>1.4E-02</td>
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</tr>
<tr>
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</tr>
<tr>
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</tr>
<tr>
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<td>5.3E-01</td>
<td>1.8E-03</td>
<td>1.8E-01</td>
<td>0.01</td>
</tr>
<tr>
<td>23.5</td>
<td>0.32</td>
<td>1.0</td>
<td>5.0E-01</td>
<td>1.6E-03</td>
<td>1.7E-01</td>
<td>0.01</td>
</tr>
<tr>
<td>24.5</td>
<td>0.30</td>
<td>1.0</td>
<td>4.7E-01</td>
<td>1.4E-03</td>
<td>1.6E-01</td>
<td>0.01</td>
</tr>
<tr>
<td>25.5</td>
<td>0.29</td>
<td>1.0</td>
<td>4.5E-01</td>
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<td>1.5E-01</td>
<td>0.01</td>
</tr>
<tr>
<td>26.5</td>
<td>0.27</td>
<td>1.0</td>
<td>4.2E-01</td>
<td>1.2E-03</td>
<td>1.5E-01</td>
<td>0.01</td>
</tr>
<tr>
<td>27.5</td>
<td>0.26</td>
<td>1.0</td>
<td>4.0E-01</td>
<td>1.0E-03</td>
<td>1.4E-01</td>
<td>0.01</td>
</tr>
</tbody>
</table>

Table 7.1 above indicates the different depths (cm) at which PDD’s were measured in a water phantom, indicated in the first column and the second column lists the PDD values as obtained from the measurements. The output factors as a function of field size were taken from the data compiled at the Radiation Oncology department (Polokwane Hospital) and the value is approximately 1 for standard conditions of 10 × 10 cm² field size, 10 cm depth and 100 cm SSD.
The scatter factor values were obtained by calculating the tissue-phantom ratio (TPR) from this equation:

\[ TPR(d) = \frac{D_d}{D_{d_{10}}} \]  

(7.3)

where \( D_d \) is the dose at an arbitrary depth and \( D_{d_{10}} \) is the dose at the depth of 10 cm. The percentage dose at a depth of 10 cm was 64.3 % as obtained from the PDD data.

Thus, for the 0.5 cm depth, the scatter factor was calculated as follows:

\[ \text{Scatter Factor (10×10 cm}^2) = \frac{0.784}{0.643} = 1.22 \]  

(7.4)

The relative dose factor (RDF) was used to convert the absorbed dose to the dose rate and it was calculated using the following equation:

\[ \text{RDF (FS)} = PDD(hv, FS, SSD, d) \times SF(FS) \times OF(FS) \]  

(7.5)

The calculated relative dose rate was calculated by the MCNP code. The ratio of the measured dose rate and the calculated dose rate were computed. The values are listed in the last column of Table 7.1. The average ratio of the measured dose rate to the calculated dose rate is below 0.5, which is acceptable according to Subbaiah et al., (2010). As such, the MCNP code is ready to be used, since it has passed the acceptable condition.
The measured relative dose rate factors calculated are listed in column 5 of Table 7.2: Comparison of measured data with the computed data for a 15 MV photon beam.

<table>
<thead>
<tr>
<th>Depth (cm)</th>
<th>15 MV PDD</th>
<th>Output Factor (OF) ((10 \times 10 \text{ cm}^2)) ((\text{measured}))</th>
<th>Scatter Factor (SF) ((10 \times 10 \text{ cm}^2)) ((\text{calculated}))</th>
<th>Rel. dose factor (RDF) ((\text{calculated})) ((\text{Gy/min}))</th>
<th>Rel. dose Rate ((\text{calculated})) ((\text{Gy/min}))</th>
<th>Ratio (= \frac{\text{measured}}{\text{calculated}})</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.5</td>
<td>0.63</td>
<td>1.0</td>
<td>8.2E-03</td>
<td>5.2E-05</td>
<td>5.0E-01</td>
<td>0.00010</td>
</tr>
<tr>
<td>1.5</td>
<td>0.94</td>
<td>1.0</td>
<td>1.2E-02</td>
<td>1.2E-04</td>
<td>7.0E-01</td>
<td>0.00017</td>
</tr>
<tr>
<td>2.5</td>
<td>0.99</td>
<td>1.0</td>
<td>3.1E-02</td>
<td>1.3E-04</td>
<td>7.4E-01</td>
<td>0.00018</td>
</tr>
<tr>
<td>3.5</td>
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<td>1.0</td>
<td>6.5E-03</td>
<td>3.2E-05</td>
<td>7.5E-01</td>
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</tr>
<tr>
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<td>0.96</td>
<td>1.0</td>
<td>1.3E-02</td>
<td>1.2E-04</td>
<td>7.4E-01</td>
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</tr>
<tr>
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<td>1.0</td>
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<td>1.1E-03</td>
<td>7.1E-01</td>
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</tr>
<tr>
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<td>1.2E-02</td>
<td>1.0E-04</td>
<td>6.9E-01</td>
<td>0.00015</td>
</tr>
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<td>1.0</td>
<td>1.1E-02</td>
<td>9.5E-05</td>
<td>6.7E-01</td>
<td>0.00014</td>
</tr>
<tr>
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<td>0.81</td>
<td>1.0</td>
<td>1.1E-02</td>
<td>8.6E-05</td>
<td>6.5E-01</td>
<td>0.00013</td>
</tr>
<tr>
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<td>1.0</td>
<td>1.0E-02</td>
<td>8.1E-05</td>
<td>6.2E-01</td>
<td>0.00013</td>
</tr>
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<td>6.1E-01</td>
<td>0.00012</td>
</tr>
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<td>1.0</td>
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<td>6.8E-05</td>
<td>5.8E-01</td>
<td>0.00012</td>
</tr>
<tr>
<td>12.5</td>
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<td>1.0</td>
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<td>6.2E-05</td>
<td>5.6E-01</td>
<td>0.00011</td>
</tr>
<tr>
<td>13.5</td>
<td>0.66</td>
<td>1.0</td>
<td>8.7E-03</td>
<td>5.7E-05</td>
<td>5.4E-01</td>
<td>0.00011</td>
</tr>
<tr>
<td>14.5</td>
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<td>1.0</td>
<td>8.3E-03</td>
<td>5.2E-05</td>
<td>5.2E-01</td>
<td>0.00010</td>
</tr>
<tr>
<td>15.5</td>
<td>0.61</td>
<td>1.0</td>
<td>8.0E-03</td>
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<td>5.1E-01</td>
<td>0.00010</td>
</tr>
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<td>4.4E-05</td>
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</tr>
<tr>
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<td>4.7E-01</td>
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<td>3.7E-05</td>
<td>4.5E-01</td>
<td>0.00008</td>
</tr>
<tr>
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<td>6.7E-03</td>
<td>3.4E-05</td>
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</tr>
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<td>3.2E-05</td>
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<td>0.00008</td>
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</tr>
<tr>
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<td>1.0</td>
<td>5.9E-03</td>
<td>2.6E-05</td>
<td>3.9E-01</td>
<td>0.00007</td>
</tr>
<tr>
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<td>2.5E-05</td>
<td>3.8E-01</td>
<td>0.00006</td>
</tr>
<tr>
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<td>0.42</td>
<td>1.0</td>
<td>5.5E-03</td>
<td>2.3E-05</td>
<td>3.7E-01</td>
<td>0.00006</td>
</tr>
<tr>
<td>25.5</td>
<td>0.40</td>
<td>1.0</td>
<td>5.3E-03</td>
<td>2.1E-05</td>
<td>3.6E-01</td>
<td>0.00006</td>
</tr>
<tr>
<td>26.5</td>
<td>0.38</td>
<td>1.0</td>
<td>5.0E-03</td>
<td>1.9E-05</td>
<td>3.4E-01</td>
<td>0.00006</td>
</tr>
<tr>
<td>27.5</td>
<td>0.36</td>
<td>1.0</td>
<td>4.8E-03</td>
<td>1.7E-05</td>
<td>3.3E-01</td>
<td>0.00005</td>
</tr>
</tbody>
</table>

The data tabulated in Table 7.2 above was used to validate the MCNP code for the use of 15 MV photon beam. The data in the first column gives the different depth at which the photon beam will traverse during the simulation. The relative dose factor (RDF) is used to convert the absorbed dose to the dose rate and it was calculated using the following equation:

\[
RDF (FS) = \frac{PDD(hv, FS, SSD, d) \times SF(FS) \times OF(FS)}{7.6}
\]

The scatter factor and output factor were calculated for the 15 MV photon beam making use of the procedure in Section 7.4. The measured relative dose rate factors calculated are listed in column 5 of Table 7.2. The column 6 is listing the MCNP computed relative dose rate measured in Gy per initial particle. To validate the code the ratio of the measured dose rate to the calculated dose rate was...
calculated to give an indication of whether the code can be trusted. The last column gives the ratio of relative dose rates per depth. The average ratio is below 0.5 which is acceptable according to Subbaiah et al., (2010).

7.4 MCNP simulated 30 × 30 × 30 cm³ Water Phantom with a Silicone Gel layer

Figure 7.6: MCNP simulated 6 cm layer of silicone gel prostheses in the water phantom.

Figure 7.6 above shows the MCNP simulated water phantom with a 1 cm layer of water followed by a 6 cm layer of silicone gel indicated. Below the 6 cm layer of silicone gel follows a 22 cm layer of water for photon interactions to proceed.

The field size of choice is indicated in the defined input and output geometries at the central axis of the beam. The effect of silicone gel prosthesis on dose distribution will be established in this water phantom. The cylindrical water phantom was designed and modelled by the MCNP code in two-dimension (2-D).
The percentage depth dose curves and the dose profiles for 6 MV and 15 MV photon beams as generated by the MCNP code are shown in Figures 7.7–7.11.

7.4.1 The Percentage Depth Dose Curves for the 6 MV Photon Beam at a Field Size of 10 × 10 cm² in a Water Phantom for Different Thicknesses of the Silicone Gel Prostheses.

Figure 7.7: MCNP relative depth dose curves for a 6 MV photon beam for different thicknesses of the silicone gel prosthesis.

The PDD in water without any silicone gel layers and the PDD with silicone gel layers were generated with different thicknesses (4 cm – 16 cm) of silicone gel shown in Figure 7.7. The PDD data obtained above was simulated from the depth of 0.5 cm up to 28.5 cm. The surface dose rate for water was obtained at a depth of 0.5 cm with a relative dose rate of 6.7E-03 Gy/initial particle. The surface dose rate obtained for the silicone gel thicknesses (6 cm - 16 cm) was obtained as 6.7E-03 Gy/initial particle except for the 4 cm silicone gel thickness which read 5.5E-03 Gy/initial particle.
The readings are similar due to the photon fluence entering the cross-sectional area on the first layer of water which is the same for all geometries. The dose build up to the maximum at a depth of 1.5 cm for water as well as for all the silicone gels which is expected of the 6 MV photon beam.

The relative dose rate for the 4 cm silicone gel at a depth of 4.5 cm was obtained to be 5.8E-03 Gy/initial particle whilst at the same depth the relative dose rate in water without the gel was obtained to be 7.2E-03 Gy/initial particle. The silicone gel of 8 cm thickness gave the peaks at the depths of 8 cm, 8.5 cm, 9 cm and 12.5 cm. This could have resulted from the computation time of the MCNP code. Table 7.3 shows the relative dose rates and percentage differences for all the results with and without the silicone gel.

Table 7.3: Statistical comparison for the relative dose rates in a water phantom with and without the silicone gel thicknesses for 6 MV photon beam 0.5 cm below the prosthesis.
7.4.2 The Dose Profiles for 6 MV Photon Beam at a Field Size of 10 × 10 cm² in a Water Phantom for the Different Thicknesses of the Silicone Gel Prostheses.

For water in the phantom without the gel the beam flatness is 0.9%, and the beam symmetry was obtained to be 0%. For the calculation of the silicone gels from 4 cm to 14 cm the beam flatness is 2%; whilst the beams symmetry is 0%. The beam flatness for 16 cm silicone gel is obtained as 0% and the beam symmetry is also gave 0%.
7.4.3 The Percentage Depth Dose Curves for the 15 MV Photon Beam at a Field Size of $16 \times 16 \text{ cm}^2$ in a Water Phantom for Different Thicknesses of the Silicone Gel Prostheses.

The PDD data that was simulated from the depth of 0.5 cm up to 27.5 cm is shown in Figure 7.9. The surface dose rate for water was obtained at a depth of 0.5 cm with a relative dose rate of 7.5E-03 Gy/initial particle which is similar for all simulations; this is due to the 1 cm layer of water above the silicone gel. The photon fluence passing through that cross-sectional area is uniform throughout. The dose builds up to the maximum at a depth of about 3.5 cm for water as well as for all the silicone gels which is expected from the 15 MV photon beam behaviour.

The relative dose rate for the silicone gel at a depth of 4.5 cm was obtained to be 7.5E-03 Gy/initial particle whilst at the same depth the relative dose rate in water without the gel was also obtained to be 7.5E-03 Gy/initial particle. The relative dose rate observed 0.5 cm below the gel is equal to the relative dose rate in water without the gel. The statistical comparison of the relative dose rates and percentage difference for all the results with and without the silicone gel are tabulated below:
Table 7.4: Statistical comparison for the relative dose rates in a water phantom with and without the silicone gel thicknesses for 15 MV photon beam 0.5 cm below each prosthesis.

<table>
<thead>
<tr>
<th>Gel thickness (cm)</th>
<th>4</th>
<th>6</th>
<th>8</th>
<th>10</th>
<th>12</th>
<th>14</th>
<th>16</th>
</tr>
</thead>
<tbody>
<tr>
<td>Depth (cm)</td>
<td>4.5</td>
<td>6.5</td>
<td>8.5</td>
<td>10.5</td>
<td>12.5</td>
<td>14.5</td>
<td>16.5</td>
</tr>
<tr>
<td>Rel. dose rate</td>
<td>H$_2$O</td>
<td>Gel</td>
<td>H$_2$O</td>
<td>Gel</td>
<td>H$_2$O</td>
<td>Gel</td>
<td>H$_2$O</td>
</tr>
<tr>
<td>(Gy/initial particle)</td>
<td>7.5E-03</td>
<td>7.5E-03</td>
<td>7.1E-03</td>
<td>7.1E-03</td>
<td>6.8E-03</td>
<td>6.8E-03</td>
<td>6.4E-03</td>
</tr>
<tr>
<td>% difference</td>
<td>0.0%</td>
<td>0.0%</td>
<td>0.0%</td>
<td>1.0%</td>
<td>0.02%</td>
<td>0.02%</td>
<td>0.02%</td>
</tr>
<tr>
<td>Variance of Variance (VOV)</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
</tr>
<tr>
<td>Figure of Merit (FOM)</td>
<td>constant</td>
<td>constant</td>
<td>constant</td>
<td>constant</td>
<td>constant</td>
<td>constant</td>
<td>constant</td>
</tr>
<tr>
<td>Std. dev.</td>
<td>0.042</td>
<td>0.042</td>
<td>0.044</td>
<td>0.045</td>
<td>0.050</td>
<td>0.051</td>
<td>0.053</td>
</tr>
<tr>
<td>Abs. Rel. error</td>
<td>0.0003</td>
<td>0.0003</td>
<td>0.0003</td>
<td>0.0003</td>
<td>0.0003</td>
<td>0.0003</td>
<td>0.0003</td>
</tr>
<tr>
<td>Mean ($\bar{x}$)</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
<td>&lt;0.1</td>
</tr>
</tbody>
</table>

In Table 7.4, the first column indicates the percentage difference of 0.0% for the 4.5 cm layer of water as well as for the silicone gel. The variance of variance (VOV) for this thickness of the gel is less than 0.1. The standard deviation varies from 0.042 to 0.053. The absolute relative error of 0.0003 was also obtained from the simulation. The mean ($\bar{x}$) of less than 0.1 was achieved during the simulation. The Figure of merit (FOM) was kept constant throughout the simulations. All the statistical checks were calculated by the MCNP code.
7.4.4 The Dose Profiles for 15 MV Photon Beam at a Field Size of 16 × 16 cm² in a Water Phantom for the Different Thicknesses of the Silicone Gel Prostheses.

Figure 7.10: MCNP Relative dose profiles for a 15 MV photon beam.

Beam symmetry for Figure 7.10 above was calculated for the dose profile obtained in water without the gel and for all the silicone gels indicated was obtained to be 0%. The beam flatness was calculated to be 5% for water without the gel and for all the silicone gels indicated.
Figure 7.1: PDD curves relating doses as a function of silicone gel thickness, calculated at 0.5 cm below the prosthesis for 6 MV and 15 MV.
Figure 7.11 and Figure 7.12 show PDD curves and PDD chart relating the relative dose rates for 6 MV and 15 MV as a function of silicone gel thickness. The effect is observed at 0.5 cm below the silicone gel prosthesis.

For 6 MV PDD, the relative dose rate of 5.80E-03 Gy/initial particle is calculated at 0.5 cm below the 4 cm silicone gel thickness. As the gel thickness is increased to 6.5 cm the relative dose rate was calculated to be 6.60E-03 Gy/initial particle, which indicates a dose build-up of 12.1% as the gel thickness is increased by 2 cm. The curve goes up to give a peak at 8.0 cm gel thickness with a dose rate of 6.80E-03 Gy/initial particle. From 8.0 cm to 10 cm gel thickness the 6 MV PDD curve shows a significant decrease in the dose of 24%. As the gel thickness is further increased by 2 cm, the dose rate decreases by 10%.

For 15 MV PDD curve, the dose rate decreases from the 4.5 cm to 6.5 cm gel thickness by 5.6%. As the thickness of the gel is increased from 8 cm to 16 cm by 2 cm, the dose rate decreases by a factor of 6%.
Chapter 8: Conclusion and Recommendations

The descriptors of photon dose distributions either represented by the percentage depth dose calculated along the beam’s central axis or the dose profile calculated across the water phantom provided data to check the accuracy of computer generated dose distributions. The objective of the study was to observe the effect of the silicone gel prostheses on both 6 MV and 15 MV photon beam dose distributions.

For 6 MV photon beam, the gel built up the dose by 12.1% up to the 8.5 cm gel thickness. As the gel thickness is increased by 2 cm, the dose rate decreased by 10% each time. For 15 MV PDD curve, the dose rate showed a decrease of 6% each time when the gel thickness is increased from the lowest to the biggest gel thickness.

The increase in dose rate at lower gel thickness is influenced by the low photon energy and smaller field size as compared to the higher photon energy at a bigger field size. The slight difference observed in both scenarios irrespective of the energy or the field size can be attributed to the fact that silicone gel and water exhibit comparable densities as well as the interaction range of the low photon energy as compared to the high photon energy. The results obtained are acceptable since all the ten statistical checks run by the code were passed. The MCNPX code prediction can be safely used in the estimation of photon beam distributions in silicone gel prosthesis.

Recommendations

For future work, the author suggests that MCNP calculations for higher energies such as 18 MV to be simulated in order to observe the effect of the gel on photon dose distributions, simultaneously observing the effect in field sizes of more than $10 \times 10 \text{ cm}^2$. 
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Laboratory. LA-10363-MS.


APPENDICES

APPENDIX A: Table of Figures

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Figure 2.3: The Schematic representation of Photoelectric Effect.

Figure 2.4: Illustration of the Compton Effect.

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Figure 4.3: The relationship between the absorbed dose and Kerma at a depth.

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Figure 4.5: An example of a percentage depth dose curve.

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Figure 4.8: Illustration of the Mayneord F factor.

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Figure 7.1: Measured PDD curves for 6 MV and 15 MV at a field size of 10 × 10 cm$^2$ in a water phantom.

Figure 7.2: Measured cross-plane dose profile for 6 MV and 15 MV at a field size of in a water phantom.

Figure 7.3: MCNP simulated 30 × 30 × 30 cm$^3$ water phantom.

Figure 7.4: MCNP simulated PDD curves for 6 MV and 15 MV at a field size of 10 × 10 cm$^2$ in a water phantom.

Figure 7.5: MCNP simulated dose profile curves for 6 MV and 15 MV at a field size of 10 × 10 cm$^2$ in a water phantom.

Figure 7.6: MCNP simulated 6 cm layer of silicone gel prostheses in the water phantom.

Figure 7.7: MCNP relative depth dose curves for a 6 MV photon beam for different thicknesses of the silicone gel prosthesis.

Figure 7.8: MCNP dose profiles for a 6 MV photon beam in a silicone gel and in water.

Figure 7.9: MCNP Relative depth dose curves for a 15 MV photon beam.

Figure 7.10: MCNP Relative dose profiles for a 15 MV photon beam.

Figure 7.11: PDD curves relating doses as a function of silicone gel thickness, calculated at 0.5 cm below the prosthesis for 6 MV and 15 MV.

Figure 7.12: PDD chart relating doses as a function of silicone gel thickness, calculated at 0.5 cm below the prosthesis for 6 MV and 15 MV.
APPENDIX B: List of Tables

Table 5.1: MCNP input file structure.

Table 5.2: The cell card format.

Table 5.3: Surface card format.

Table 5.4: A Table showing the types of tallies, the type of particles denoted by pl. Tally type particle pl Fn Units, *Fn Units.

Table 5.5: The Table indicating the ten statistical checks for MCNP code.

Table 7.1: Comparison of measured PDD data with the computed PDD data for a 6 MV photon beam in a water phantom.

Table 7.2: Comparison of measured data with the computed data for a 15 MV photon beam.

Table 7.3: Statistical comparison for the relative dose rates in a water phantom with and without the silicone gel thicknesses for a 6 MV photon beam 0.5 cm below each prosthesis.

Table 7.4: Statistical comparison for the relative dose rates in a water phantom with and without the silicone gel thicknesses for a 15 MV photon beam 0.5 cm below each prosthesis.