DETERMINATION OF THE LINEAR ATTENUATION COEFFICIENTS AND BUILDUP FACTORS OF MCP-96 ALLOY FOR USE IN TISSUE COMPENSATION AND RADIATION PROTECTION

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DEDICATION

I want to dedicate my thesis to my two biggest supporters: my mom and dad Vicky and Howard Hopkins. They are the two most amazing people I know. I wouldn't be where I am now without their infinite amount of love, encouragement and support. I was lucky enough to be raised by parents who nurtured my dreams and led me to believe I could accomplish anything I was willing to work hard for. I am grateful for the knowledge, work ethic and values they continue to instill in me. I love you both more than words can express.

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Namaste

ABSTRACT

THESIS: Determination of the Linear Attenuation Coefficients and Buildup Factors of MCP-96 Alloy for Use in Tissue Compensation and Radiation Protection

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The linear attenuation coefficient and buildup factor are a few of the important characteristics that need to be studied and determined prior to using a material clinically in radiation treatment and protection. The linear attenuation coefficient and buildup factor, as well as several other properties, will be determined for MCP-96 alloy to assess its use in radiation therapy. A narrow collimated beam of γ -rays from sources with varying energies will pass through various thicknesses of MCP-96 alloy. The attenuation in the intensity of the beam will be determined for each varying thickness of the alloy. Plotting the thickness of the alloy versus the corresponding logarithmic intensity of the beam will allow calculation of the linear attenuation coefficient.

The narrow beam geometry will then be replaced by the broad beam geometry to determine the buildup factor. Additional radiation is obtained through the broad beam geometry as a result of scattering and secondary radiation. Comparing the broad beam geometry to the narrow beam geometry allows determination of the buildup factor. Since the buildup factor depends upon the thickness of the MCP-96 attenuator, the energy of the beam, and the source-to-attenuator (STA) distance, it will be calculated using three parameters. It will be calculated as a function of the energy of the incident radiation beam by using several sources with different beam energies; and finally, as a function of the source-to-attenuators.

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Chapter 1

Introduction

1.1 Cancer Statistics

Cancer is the leading cause of death worldwide. In 2009, there were an estimated 1,479,350 new cases of cancer and an estimated 562,340 deaths due to cancer in the United States alone. That is equivalent to more than 1,500 deaths a day. The 5-year relative survival rate in the US for all cancers diagnosed between 1996 and 2004 is 66%, up from 50% in 1975-1977. This improvement in survival is indicative of advancements in treatment options and earlier diagnosis. [1]

Radiation therapy, along with surgery and chemotherapy, make up the top three cancer treatments currently used in the United States [2]. Half of all people with cancer have radiation therapy as part of their treatment. Radiation effectively treats cancer by targeting the tumor or cancerous area and mutating the proliferating malignant cells so they can no longer reproduce and, therefore, die. Unfortunately, healthy cells within the patient's body can be destroyed by radiation as well. This is why there is a need for tissue compensators as well as radiation protection in radiation therapy treatment.

1.2 MCP Alloys

MCP alloys are low melting point alloys, or fusible alloys, specially built for radiation shielding protection and tissue compensating purposes [3]. The alloy is manufactured by Bismuth specialists: Mining and Chemical Products Limited. The low melting point allows for easier shaping of the metal for specific shapes often needed in radiation therapy. They can reproduce very fine details and are reusable [4]. Unlike alloys with similar characteristics, MCP-96 is free of the known carcinogen cadmium. Cadmium exposure most often occurs in the manufacturing of products containing cadmium and causes lung damage, cancer, and fragile bones [5]. MCP-96 is not classified as dangerous for either transportation or storage. It is easy to handle, and quite safe when the rules for normal handling are observed.

1.3 Statement of the Problem

There are no perfect treatments for cancer, and radiation therapy is one of those imperfect treatments. Improvements are constantly being made in the field to bring it a little closer to perfection. Healthy cells can be killed along with cancerous cells causing severe damage to the patient. Delivery of correct and precise dose plays a very important role in achieving those goals [6-10]. Methods have been devised to spare the healthy cells while providing enough radiation damage to kill the cancerous cells in the patient. If the cancerous cells are not totally obliterated, the cells may continue to proliferate and the tumor may regrow [10-14]. Safe and effective treatments require an optimum personalized dosage level so that the healthy cells receive minimum possible damage [10, 11, 15, 16]. Dische et al. [17] showed that a dose difference as small as $\pm 5\%$ may lead to

a real impairment or enhancement of tumor response as well as a change in the risk of morbidity to the normal tissues. An issue in radiation therapy is the non-uniform shape of the body. Tissue compensators are used to strategically adjust the radiation beam to provide the best therapeutic outcome [10, 18, 19]. Radiation shielding is employed to protect the healthy cells of the patient, as well as the healthy cells of those working in and around the treatment area [20,21]. MCP-96 alloy is a new material that may be suitable for radiation protection and therapy. Several properties, including the linear attenuation coefficient and buildup factor, must be determined for an attenuator before it can be used clinically in radiation treatment or shielding.

1.4 Significance of the Study

This research determines linear attenuation coefficients and buildup factors, among other radiological properties, making MCP-96 readily available for use in tissue compensation and radiation protection. These values are used to determine the shape and thickness of the alloy needed, as well as the radiation dose to use along with the alloy tissue compensator. The availability of appropriate and varying materials for radiation therapy and shielding only increases efficiency of treatment and protection from radiation.

1.5 Summary

Cancer rates are increasing every year and more advanced treatments are being developed to accommodate this fact. Higher efficiency, fewer side effects, and patient safety are all taken into account when developing new treatment options. MCP-96 alloy is safe and has many desirable characteristics that may make it suitable for radiation therapy. This research will determine the radiological properties of MCP-96 alloy allowing it to be used for radiation therapy and radiation shielding purposes.

Chapter 2

Interaction of Gamma and X-Rays

2.1 Introduction

Gamma rays are electromagnetic radiation either emitted from a nucleus or an annihilation reaction between matter and antimatter. X-rays are electromagnetic radiation emitted by charged particles (usually electrons) in changing atomic energy levels or in slowing down in a Coulomb force field [22]. X-rays and gamma rays have identical properties, only differing in origin. The two used to be distinguishable by energy of the particle, but now linear accelerators are able to produce high energy X-rays that have the same energy as gamma rays. The practical range of photon energies emitted by radioactive atoms extends from 2.6 keV to 7.1 MeV. The energy ranges of x-rays are in terms of generating voltage.

Energy range of X-rays

| Energy | X-rays |
|---------------|---------------------|
| 0.1 - 20 kV | Soft, low energy |
| 20 - 120 kV | Diagnostic range |
| 120 - 300 kV | Orthovoltage |
| 300 kV – 1 MV | Intermediate-energy |
| 1 MV + | Megavoltage |

Table 2.1 Energy range and corresponding categories of X-rays [23].

As a γ -ray or x-ray passes through a medium, an interaction occurs between the photons and matter resulting in energy transfer to the medium [23]. The interaction can result in a large energy transfer or even complete absorption of the photon [24]. However, a photon can be scattered rather than absorbed and retain most of the initial energy while only changing direction. Some of the important ways photons interact with matter will be examined more closely.

2.2 Types of Interactions with Matter

There are five major types of interactions causing attenuation of a photon beam by matter: Compton effect, photoelectric effect, pair production, coherent scattering, and photo disintegration [23]. The first three are the most important, as they result in the transfer of energy to electrons [22]. The electrons then transfer this energy to matter in many small Coulomb-force interactions. Coherent scattering is elastic and photo disintegrations are only significant for photon energies above a few MeV, where they may create radiation protection problems with the production of neutrons and consequent radioactivity. The relative importance of Compton effect, photoelectric effect, and pair production depend on both the photon quantum energy and the atomic number of the absorbing medium [22].

2.2.1 Compton scattering

Compton's scattering is the most important interaction in radiation therapy [22]. For low Z materials such as air, water, and human tissues, Compton scattering is dominant from approximately 20 keV to 30 MeV photon energies.

Compton Effect occurs when photons interact with free or weakly bound electrons in the γ -ray incident beam. In this incoherent scattering, all atomic electrons act independently of one another. [24]

The initial photon of wavelength λ_i has a much higher energy than the binding energy of the electron. When the photon collides with the electron, the electron receives energy from the photon and is emitted at an angle φ . The reduced-energy photon of wavelength λ_f is scattered at an angle θ [23]. Compton scattering diminishes at low photon energies due to atomic binding.



Figure 2.1 Compton effect

The initial photon has energy hv, and collides with the electron with a momentum of hv/c, where c is the speed of light in a vacuum as seen in Figure 2.1. The electron is initially stationary and has no initial kinetic energy or momentum. The scattered photon has a lower quantum energy of hv'.

The resulting equation follows:

$$hv' = hv/[1 + (hv/mc^2)(1 - \cos \varphi)]....(1)$$

2.2.2 Photoelectric effect

The photoelectric effect is the most important interaction of low-energy photons with matter [22]. This is dominant in energy levels from a few eV to a few keV, corresponding to visible light through soft x-rays. Photoelectric effect occurs when a bound electron from the atom is ejected after interaction with a photon of energy hv [24]. Energy contained within the photon is transferred to the electron after contact [25]. The photon provides the energy to overcome the binding energy and the remainder is the kinetic energy, *T*, of the electron.

As the energy of a photon decreases, the photoelectric cross-section ' τ ' increases rapidly [25]. The energy dependence of the photoelectric effect cross section is between E^{-2} and E^{-4} . The Z and E dependence on the photoelectric cross section near 100 keV is approximated:

2.2.3 Pair Production

Pair production is dominant in interactions of higher-energy photons with matter [22]. Pair Production refers to the conversion of electromagnetic radiation in an atomic nucleus into matter [24]. A negative electron (e^-) and a positive electron (e^+) are created from a photon interacting with the electromagnetic field while energy and momentum are

conserved. The rest mass m_0 of an electron is 511 keV, so for a photon-nucleus pair production to occur, a minimum energy of twice the rest mass (1.022 MeV) is required to create two electrons. Each electron carries off any excess energy as kinetic energy. Two particles must be emitted to conserve charge, so there is no pair production at energy levels below the threshold. The probability of this effect increases with increasing atomic number due to the fact that pair production is caused by an interaction with the electromagnetic field of the nucleus. [23]

2.2.4 Coherent Scattering

Coherent scattering, often called Rayleigh scattering, involves the scattering of a photon with no energy transfer (elastic scattering) [23]. The electron is oscillated by the electromagnetic wave from the photon. The electron, in turn, reradiates the energy at the same frequency as the incident wave. The scattered photon has the same wavelength as the incident photon. The only effect is the scattering of the photon at a small angle. This scattering occurs in high atomic number materials and with low energy photons. This effect can only be detected in narrow beam geometry. Kerma and dose are not affected by coherent scattering, since no energy is given to charged particles, and no ionization or excitation is produced. [22]

2.2.5 Photo Disintegration

Photo Disintegration occurs in high energy γ -rays at energies over 10 MeV. The γ -ray interacts with the nucleus of an atom, therefore, exciting it. The excited nucleus immediately decays into two or more daughter nuclei. A neutron or proton is emitted

contributes directly to kerma [22]. This is seen in nuclear fission, or the breakdown of an atomic nucleus. The neutrons produced in this reaction can cause radiation protection problems if they are not accounted for. This type of reaction does not occur in this experiment because the energy levels used are much lower than the threshold for this type of reaction. [23]

2.3 Radioactive Decay

The attenuation of a gamma ray photon beam is governed by the decay law given below [6, 23, 26-28]

$$N = N_0 e^{-\mu x} \qquad (4)$$

Where μ represents the linear attenuation coefficient and x gives the thickness of the attenuator. However, practically, more photons are received at a reference point than what one can calculate using the decay law. The additional dose is received due to the buildup factor *B* of the attenuator and the simple mathematical form of the decay law is modified to a new form given in equation (5).

$$N = B \cdot N_0 e^{-\mu x} \quad \dots \quad (5)$$

Equation (4) is valid only in narrow beam geometry when a collimated beam of radiation is used. However, in a broad beam geometry, we must take into account the buildup factor and hence equation (5) will give us a more correct and precise value.

2.4 Linear Attenuation Coefficient

When a photon passes through an attenuator material, the probability that an interaction will occur depends on its energy and the composition and thickness of the

attenuator [29]. The thicker the attenuator material, the more likely an interaction will occur. The dependence on photon energy and attenuator composition is more complicated.

Consider an incident beam of photons of intensity *I*. This beam is directed onto an attenuator of thickness Δx . Assume for now the attenuator consists of a single element of atomic number *Z*, and that the beam is monoenergetic with energy E. A photon detector records the transmitted beam intensity. Only the photons passing through the attenuator without interaction are detected. For a thin attenuator, it is found that the fractional decrease in beam intensity is related to absorber thickness:

$$\Delta I/I \approx -\mu \cdot \Delta x \dots \dots \dots \dots (6)$$

The quantity μ is the linear attenuation coefficient of the attenuator material. The minus sign indicates the beam intensity decreases with increasing attenuator thickness. The linear attenuation coefficient represents the 'absorptivity of the attenuating material. The quantity μ is found to increase with linearly with attenuator density ρ . This value can be used to calculate values such as the intensity of the energy transmitted through an attenuating material, intensity of the incident beam or the thickness of the material. It may also be used to identify the material if the previously mentioned values are already known. [29]

2.4.1 Mass attenuation coefficient

Mass attenuation coefficient can be a useful coefficient because only the atomic composition of the attenuator is taken into account and not the individual density of the material. To obtain the mass attenuation coefficient, the linear attenuation coefficient, μ ,

is divided by the density, ρ . It has units of cm²/g because μ/ρ has unit cm⁻¹/(g/cm³). The attenuator thickness can also be expressed in units of electrons/cm² and atoms/cm². The associated coefficients are the electronic attenuation coefficient, $_{e}\mu$, and the atomic attenuation coefficient, $_{a}\mu$. [23]

2.4.2 Half-value thickness (HVT)

The half-value thickness, or half-value layer, is the thickness of the material that reduces the intensity of the beam to half its original value [31]. When the attenuator thickness is equivalent to the HVT, N/N_0 is equal to $\frac{1}{2}$. Thus, it can be shown that

$$HVT = \ln 2/\mu \dots \dots (7)$$

This value is used clinically quite often in place of the linear attenuation coefficient. The mean free path is related to the HVT according to

$$X_m = HVT/\ln 2 \dots (8)$$

2.4.3 Mean free path

The mean free path, or relaxation length, is the quantity

$$X_m = 1/\mu \dots (9)$$

This is the average distance a single particle travels through a given attenuating medium before interacting. It is also the depth to which a fraction 1/e (~37%) of a large homogeneous population of particles in a beam can penetrate. For example, a distance of three free mean paths, $3/\mu$, reduces the primary beam intensity to 5%. [22]

2.5 Buildup Factor

The buildup factor B is a unitless value used to account for secondary and scattered radiation in a broad-beam system. It can be applied with respect to any specified geometry, attenuator, or physical quantity in radiological physics [22]. Buildup factor is one of those important properties of a material used for beam collimation, tissue compensation or radiation shielding and protection. It directly affects the dose quantity and, hence, correct and precise dose delivery to a tumor is not possible without knowing the buildup factor of a material for tissue compensation purpose [10, 20, 21, 30].

$$B = \frac{\text{Total Broad Beam Counts}}{\text{Total Narrow Beam Counts}}$$
......(10)

For narrow beam geometry B = 1 exactly, and for broad beam geometry B > 1. In narrow beam geometry, ideally there is no scattered or secondary radiation to detect, so there is no buildup factor contributing to the detection of radiation. However, the broad beam geometry will usually produce some type of secondary scattered radiation depending on the geometry and components involved. The value of *B* is a function of radiation type and energy, attenuating medium and thickness, geometry, and measured quantity [22].

Chapter 3

Radiation Biology and Radiation Therapy

3.1 Introduction

In external beam radiation therapy, ionizing radiation is used to effectively damage cancerous cells so that they cannot reproduce and eventually die. Gamma rays or electrons are emitted in a beam from a machine called a linear accelerator. Skilled medical physicists and radiation oncology teams use linear accelerators and treatment planning software to develop the best course of treatment for patients. The goal is to maximize cancerous cell death, while minimizing healthy cell death.

3.2 Radiation Biology

Radiation biology is the study of the biological effects of radiation on living cells and tissues. In order to treat cancer through radiation therapy, one must have a great understanding of radiation and its effects on the body. The primary action of radiation is to damage DNA [32]. This damage is most lethal to cells that are rapidly dividing, so radiation selectively kills actively proliferating cells. Cancerous cells are cells that are uncontrollably proliferating, so radiation treatment targets these cells.

The cell proliferation cycle is divided into two separate time periods: mitosis (M) and DNA synthesis (S). Mitosis is the stage in which cell division takes place. The M and

S stages are separated by two time periods, or gaps, G1 and G2. Metabolic processes necessary for successful cell division occur during these two gaps. In general, cells are the most sensitive to radiation in the M and G2 phase, and are most resistant in late S phase. The cell cycle of cancerous cells is shorter than that of normal tissues. However, normal tissue cells can rejuvenate faster than cancerous cells after being injured. [33, 34]

Cells exposed to radiation receive the physical effects to the atoms and molecules first, and the biological effects later. The biological effects are mainly due to DNA damage as DNA is the most critical target in the cell. However, there are other sites in the cell that may cause cell death if irradiated. Ionizing radiation may cause damage to the cell in two ways: direct and indirect [33].

Direct interaction occurs when the radiation interacts directly with the critical target in the cell. In indirect interaction, the radiation interacts with other molecules and atoms within the cell to produce free radicals. Through diffusion in the cell, these free radicals can damage critical targets. Indirect interactions with water produce extremely reactive free radicals such as H2O+ (water ion) and OH• (hydroxyl radical). These free radicals break chemical bonds and produce chemical changes that cause biological damage to the cells because they have an unpaired valence electron. [33]

When cells are irradiated there are nine consequences that may occur. First, nothing may occur to the cell at all; the free radicals or photons may not damage any critical structures. The cell may be delayed in going through division which in turn slows reproduction. The cell may die through different pathways to apoptosis. The cell may have reproductive failure and die during the subsequent mitosis. The cell may survive, but contain a mutation. The radiation may induce genomic instability, slowing reproduction. The cell may survive a mutation, resulting in a transformed genotype and possibly carcinogenesis. Bystander effect may occur in which the cell signals to neighboring cells and induces genetic damage in them. Finally, the cell may have an adaptive response; the cell is stimulated to react and becomes more resistant to subsequent irradiation. [35]

3.3 Radiation Therapy

Radiation therapy is a treatment that utilizes radiation to damage living cells [32]. It is often used along with surgery and/or chemotherapy to treat localized tumors and other cancers. However, radiation is not limited to only damaging cancer cells. Typically, a specific plan is devised to damage certain cancerous cells, while protecting healthy cells. A certain treatment area, determined by a physician specializing in oncology, will be selected depending on the patient. This range usually extends a certain distance past the known tumor or malignant cell mass in order to attempt to prevent any malignant cells from reproducing. If even one malignant cell is left undamaged, the cell may reproduce and the tumor may appear again.

A variety of types of radiation are used in cancer therapy. Some examples are Xrays, radiation produced by the decay of radioactive elements and beams of particles produced by linear accelerators. The radiation source is usually located outside the body and a beam of radiation is directed at the tumor inside the body. However, in some instances, radiation is administered from an internal source, such as an implant of radioactive material placed directly in the vicinity of a tumor. An implant is commonly used to treat cervical cancer and prostate cancer, although it is used for other types of cancer as well. [32]

Treatment is limited by the sensitivity of normal cells to the radiation. Some cells in the body naturally proliferate actively, such as the cells of the mouth and digestive system. Side effects from irradiation of healthy cells include anemia, nausea, vomiting, diarrhea, skin damage, hair loss and sterility [32].

3.3.1 Clinical Treatment energy

Clinical radiotherapy has been extended to obtain better spatial distribution and more direct cell-killing action with less dependence on oxygen. Electrons and X-rays have been extended up to 50 MeV in clinical treatments [22].

The ICRU (International Commission on Radiation Units and Measurements) classifies ionizing radiation interactions with matter in two ways: directly and indirectly. Charged particles deliver their energy to matter directly through many small Coulomb-force interactions along the particle's track [22]. Examples of directly ionizing radiation are α -particles and β -particles. Uncharged particles can travel through the large percentage of empty space in matter before interacting with an electron or the nucleus. When the particles interact, a charged particle is produced. The resulting charged particle delivers the energy to the matter as directly ionized radiation. X- and γ -rays are examples of indirectly ionizing radiation.

Chapter 4

Radiation Protection and Tissue Compensation

4.1 Introduction

In an effort to improve radiation therapy treatment, tissue compensators are used to accommodate for inhomogenities of the patient's body. Tissue compensators are used in radiation therapy for shaping the beam profiles to generate a desired, often uniform, dose distribution in the tumor and maintain the skin-sparing advantage of high-energy photons [23]. Possible applications as a bolus material, compensator and partial or total shielding material in clinical radiation therapy are discussed.

Humans are exposed to ionizing radiation on a daily basis. Radiation protection is constantly improving to protect individuals from the hazardous long-term effects of radiation. The goal is to keep exposure levels as low as possible, so the exploration of new, possibly more effective, materials for shielding is desirable.

4.2 Radiation Protection

Biological systems are especially susceptible to damage by ionizing radiation. A trivial amount of energy (~4 J/kg) from ionizing radiation will most likely cause death, even though this amount of energy will only raise the body temperature by 0.001°C [22].

Radiation exposure occurs from natural background radiation, non-medical synthetic radiation, and medical radiation [36]. Natural radiation comes from cosmic rays and elements in the soil. This is the major contributor to worldwide radiation exposure. Non-medical synthetic radiation is due to nuclear weapons testing prior to 1962 as well as commercial and industrial sources. Medical radiation comes from diagnostic imaging equipment and radiation therapy.

Ionizing radiation has been proven to induce cancer in many different species of animals, including humans, and in almost all parts of the body [36]. It is one of the few scientifically proven carcinogens in humans. Ionizing radiation is relatively weak compared to other chemical agents. Usually, many years pass between the radiation exposure and the associated cancer. This is why effective radiation protection is so important. There are no immediate signs of sufficient radiation exposure to cause cancer. There must always be adequate radiation protection in industry and medicine to protect workers, guests, and patients at all times.

The general radiation protection policy set forth by the government is radiation exposure should be ALARA (As low as reasonably achievable). The NRC annual dose limit for the general public is 0.1 rems with the exception of pregnant women and children. The occupational exposure limits are higher as seen below.



Figure 4.1 Annual dose limits for occupationally exposed adults. The SI unit for dose is sievert (Sv) and is equivalent to joules per kilogram. (NRC)

4.2.1 Radiation dose

Absorbed dose is defined as the quantity of radiation for all types of ionizing radiation [23]. It is a measure of the biological effect of ionizing radiation. The SI unit for absorbed dose is the gray (Gy). The gray is equivalent to joules per kilogram. The rad (radiation absorbed dose) is another unit used in dose measurement. One rad is equivalent to 100 ergs per second. The centigray (cGy) is equivalent to the rad and is used quite often in therapeutic radiation.

The quality factors, as defined by the National Council on Radiation Protection and Measurements (NCRP), are:

| Quality Fa | actors |
|-------------------------------|---------|
| Radiation | Factors |
| Electrons, x-rays, and γ-rays | 1 |
| Thermal neutrons | 5 |
| Neutrons, heavy particles | 20 |

Table 4.1 The biological effects of radiation depend not only on the dose, but also on the type of radiation [23]. The relevant quantity used in radiation protection is the dose equivalent (H). It is defined as the absorbed dose multiplied by the quality factor.

H = Quality factor (Q) * Absorbed dose (D)

The SI unit for dose and dose equivalent is joules per kilogram, or sievert (Sv). If the dose is expressed in rad, then the dose equivalent is given in rem. One rem is equivalent to 10^{-2} J/kg. [23]

4.3 Radiation Therapy Geometry

High energy photon beams cause a rapid increase of dose in the first few millimeters of depth. This is called dose buildup effect [23]. Benefits of using high atomic mass material as tissue compensators are the smaller size and retention of the skin-sparing effect of Megavoltage beams. Due to this skin-sparing phenomenon, the bolus is used sparingly, as the patient's skin can be damaged by the buildup. [37]

4.3.1 Bolus

A bolus is placed directly on the skin to even out irregular contours of the patient to present a flat surface orthogonal to the incident beam. It is made of a tissue-equivalent material: material similar to the tissue it is being used to compensate. However, when a bolus is used at an energy higher than orthovoltage radiation energy, the skin-sparring advantage is lost. A bolus is used when the skin needs to be irradiated as well as the underlying tissue, such as a mastectomy scar in breast cancer treatment [37]. For higher energy radiation, such as megavoltage, a compensating filter should be used to prevent the loss of skin-sparring properties. In this case, the compensating filter is placed at least 15 cm away from the patient's skin. These filters are designed to give the same benefits as a bolus while preserving the skin-sparring effects of higher energy radiation. [23]

4.3.2 Tissue compensators

A compensator is designed to provide attenuation where the body surface is irregular or curved. The lack of dose uniformity can be damaging to the irradiated tissues. In some areas, the dose will be higher and in other areas, it will be lower. This not only makes planning more complicated, but can lead to the development of 'hot spots' or areas that are receiving too much radiation. The compensator mimics the attenuation that would be present if the body surface was flat and not contoured. It is designed to be at a certain distance from the patient, usually over 20 cm from the skin, so the shape of the compensator must be adjusted to accommodate this. The dimensions and shape depend on beam divergence, relative linear attenuation coefficient of the filter material and soft tissues, and the reduction in scatter at various depths when the compensator is placed at a distance from the skin instead of directly on the skin. [23]

A compensator may be made from any material in which the attenuation coefficient is known. When materials with high density are used, a small error in thickness will cause a large error in dose. It is very important that the compensators are made as accurately as possible to avoid dosing the patient incorrectly. The advantage to using denser materials is their compact size. The design and fabrication of tissue compensators takes this into account. The needed reduction can be determined by SDD (source-to-detector distance), or target-to-compensator distance. [31]

Compensators have also been used to compensate for tissue inhomogenities, or differences in attenuation between different types of tissues.

| | | | | 0 | |
|---------------------------------|--------|--------|--------|--------|--------|
| Material | Air | Water | Muscle | Fat | Bone |
| Density (kg/m ²) | 1.205 | 1000 | 1040 | 920 | 1046 |
| Energy | | | | | |
| 5 MeV | .02225 | .02429 | .02403 | .02371 | .02440 |
| 8 MeV | .02045 | .02219 | .02195 | .02147 | .02263 |
| 10 MeV | .01810 | .01941 | .01918 | .01840 | .02040 |

Mass attenuation coefficient (μ/ρ) in cm²/g

Table 4.2 The difference in mass attenuation coefficients at varying energy levels in different mediums [23].

4.4 Shielding

When X-rays or γ -rays traverse matter, some are absorbed, some pass through without interaction, and some scatter as low energy photons in directions that are different from those in the primary beam. Values of μ generally increase as the atomic number, Z, of the absorbing material increases because photoelectric interactions are increased in high-Z materials, especially for low-energy photons. High-Z materials yield increases in pair production interactions for high-energy photons. The most effective gamma shields are materials which have a high density and high Z, such as lead, tungsten, and uranium. These materials can be rather expensive so, in situations where space is not a constraint and where structural strength is required, concrete can be used. However, it is not as effective as the higher-*Z* materials. Lead shields are frequently used where space is limited or where only a small area of absorber is required. The effectiveness of gamma ray shielding is often described as the half-value thickness (HVT). [38]

4.5 Materials

Safety is a concern when dealing with tissue compensators. Some materials may be hazardous during manufacturing of the compensator. A material must also have the ability to be finely milled to the accuracy of a fraction of a millimeter. An accuracy of ± 1 mm in filter thickness is acceptable when using medium density materials. However, this is not accurate enough with high-density alloy filters since this precision can lead to a variation of the order of 5% in the transmitted fluence, and consequently causes a large error in the dose to the patient. [3, 10, 39]

Tissue compensators have been made of many materials. Different materials have different attenuation coefficients and properties that may make them more desirable than other compensators for particular cases.

The attenuators used in this experiment are manufactured of a high-density MCP-96 alloy consisting of bismuth (52.5%), lead (32%), and tin (15.5%). [3] Table 4.3 contains several more properties of the alloy.

| MCP-96 | alloy |
|---------------|-------|
| | |

| Property | Value |
|----------------------------------|-------|
| Melting Point (°C) | 96 |
| Density (g/cm^3) | 9.72 |
| Specific Heat (J/kg•K) | 151 |
| Thermal Conductivity (W/m•K) | 12.5 |
| Electrical Resistivity (µOhm•cm) | 71.4 |

Table 4.3 Basic physical and chemical properties of the attenuator MCP-96 alloy.

Chapter 5

Experimental Work

5.1 Equipment Setup

The experiment was conducted using various gamma-ray sources and MCP-96 alloy blocks of varying thicknesses. The sources' activity and energy range from 1-2 mCi and 0.662 to 1.33 MeV, respectively.

A NaI(Tl) scintillator is used to detect the gamma rays from the sources. An advantage to using an inorganic crystal scintillator, like NaI, versus other types is its greater stopping power due to higher density and atomic number [40]. Among all the scintillators, it also has one of the highest efficiencies. This makes it very appropriate for the detection of gamma rays, as well as high energy electrons and positrons.

The NaI(Tl) detector has a built-in preamplifier and is connected to a power source supplying 1000 Volts using a BNC connector. The preamplifier adds voltage gain to the signal from the detector counts so that the signal is at sufficient amplitude to move to the amplifier. The detector is also connected to an amplifier with another BNC connector. The amplifier takes the signal coming from the preamplifier and outputs a signal with much higher amplitude. The amplifier has two output sources: the bipolar output to the timing single channel analyzer (SCA) and the unipolar output to the delay amplifier. The delays the output for an adjusted time $(1.5 \ \mu s \ in \ this \ case)$.

The linear signal from the delay amplifier goes to the input in the linear gate. The timing SCA only outputs a logic signal if the peak amplitude of the input signal falls within the pulse-height window that is preset with two threshold levels. There is an upper and lower level discriminator that only allows certain signals, or selected γ -ray energies, to output a pulse. The signal from the timing SCA goes to the gate input of the linear gate. The linear gate allows the passage of a linear signal, provided that a logic signal arrives at the unit simultaneously to "open" the gate. This instrument performs a coincidence function, its resolving time being limited by the gate width (the length of time the gate is open after the arrival of the enable signal) [40]. Of course, the linear and logic signals must arrive simultaneously, so the linear signal will have to be delayed since the logic signal has passed through more electronics taking a longer time.



Figure 5.1 Equipment set-up diagram

The ⁶⁰Co source decays to ⁶⁰Ni by β ⁻ decay. Two separate γ -ray decays occur at 1.17 and 1.33 MeV from ⁶⁰Ni. To collect counts for each of these energies separately, an upper and lower discriminator in the timing SCA were used to block the adjacent γ -ray energy. Each pulse coming from the system represents a certain energy photon detected by the NaI scintillator. Different energies are represented by two different pulse heights. Each pulse height corresponds to particular photon energy. The gamma spectrum for the Cobalt source is shown below. Both gamma peaks are shown in (a), while (b) and (c) show the 1.17 MeV and 1.33 MeV peaks separately using the linear gate 'window.'





Figure 5.2 (a) Gamma spectrum of ⁶⁰Ni from Cobalt source from 1 MeV to 1.5 MeV (b) Gamma spectrum of ⁶⁰Ni 1.17 MeV peak using the linear gate discriminator (c) Gamma spectrum of ⁶⁰Ni 1.33 MeV peak using the linear gate discriminator

(b).

The data were collected on a PC using multi-channel analyzer (MCA) software Maestro. The software analyzes the pulses coming from the electronic setup and sorts them according to their height. The program tallies the number of pulses for each voltage, producing a histogram. The MCA was setup for 1024 channels (conversion gain). Each channel represents a different energy depending on the individual setup. The time was set on the program to collect data for 25 minutes for ⁶⁰Co and ⁵⁴Mn, and 10 minutes for ¹³⁷Cs. This time difference is due to the high activity level of ¹³⁷Cs and the relatively low activity level of the other two sources.

5.2 Narrow Beam Geometry

The narrow beam geometry prevents any scattered radiation or secondary particles from reaching the detector [22]. A collimated beam of gamma radiation penetrates varying thicknesses of the MCP-96 alloy to produce the narrow beam geometry. A one centimeter diameter hole was drilled into a five centimeter thick lead block to create the collimator. This produces a narrow beam of γ -rays approaching the attenuator. This set-up allows the linear attenuation coefficient to be determined. The lead collimator was tested for any γ -ray leakage. No radiation was detected, proving that the 5cm thickness of the lead block is enough to shield radiation from the current sources.

The source was placed 16 cm away from the detector and the attenuators were placed in between the detector and the source at 5 cm from the source. A count with no attenuator is represented by N_0 . The collimated beam was then attenuated with 5 x 5 cm² plates of MCP-96 alloy and allowed to fall on the detector after passing through the attenuator. The intensity of the attenuated beam was measured by the detector and is represented by *N*. Background counts were collected and the net count was obtained by subtracting time-normalized background counts.



Figure 5.3 Narrow beam geometry

5.3 Broad Beam Geometry

In broad beam geometry, there is no collimator used as in the narrow beam geometry. Broad beam geometry allows scattered and secondary particles to reach the detector in addition to the primary beam. Ideally, every scattered and secondary particle generated in the attenuator by a primary particle will strike the detector [22]. Figure 5.5 shows a comparison of a narrow beam and broad beam geometry gamma spectrum.









(b.) Narrow



Figure 5.5 (a) 0.662 MeV gamma spectrum of 137 Cs using broad beam geometry (b) 0.662 MeV gamma spectrum of 137 Cs using narrow beam geometry

5.3.1 Buildup factor as a function of thickness and source energy

A broad beam geometry set-up was used to collect data for the buildup factors of MCP-96 alloy. The source was again placed 16 cm from the NaI detector, but this time no lead collimator was used. The γ -rays approach the attenuator in a broad beam unlike the narrow beam geometry. A count was collected with no attenuator to obtain N_0 . Then, the source was attenuated with varying thicknesses of 5 x 5 cm² MCP-96 alloy blocks. The attenuators were placed 5 cm from the source when determining the buildup factor for various energies. These data were compared to the corresponding narrow beam geometry data to obtain the buildup factor. After collecting data for all energies, the buildup factors were compared for the same attenuator thickness, but varying energy levels.

5.3.2 Buildup factor as a function of source-to-attenuator (STA) distance

The broad beam geometry set-up was used again in this experiment. However, this time the distance between the source and the attenuator varied while the source energy and attenuator thickness remained the same. A 1-cm thick MCP-96 alloy block was moved 5-15 cm from the ¹³⁷Cs source in increments of 1 cm. This experiment was also repeated with the narrow beam geometry set-up as a comparison for calculating the buildup factor.

5.4 Results

The linear attenuation data for each source is represented by the slope of the linear curve on $Ln(N_0/N)$ versus thickness graph. The following graph shows the linear attenuations for energies 0.662 MeV, 0.835 MeV, 1.17 MeV, and 1.33 MeV.



Figure 5.6 Graph of linear attenuation data as a function of MCP-96 attenuator thickness

The following table compares the source, linear attenuation coefficients, and photon energy.

| Linear Attenuation Coefficients | | |
|---------------------------------|--------------|--|
| Source | Energy (MeV) | Linear Attenuation Coefficient (cm ⁻¹) |
| ¹³⁷ Cs | 0.662 | 0.9129±0.0064 |
| ⁵⁴ Mn | 0.835 | 0.7622 ± 0.0077 |
| ⁶⁰ Co | 1.17 | 0.5961±0.0089 |
| ⁶⁰ Co | 1.33 | 0.5613±0.0081 |

Linear Attenuation Coefficients

Table 5.1 Linear attenuation coefficients per source and energy level for MCP-96 alloy attenuator.

The linear attenuation coefficient decreases with increasing beam energy. As the beam energy increases, the MCP-96 attenuates fewer photons than the lower energy photons.

The half-value thickness increases linearly with beam energy as seen in Figure 5.7. The lower energy photons are more easily stopped in the attenuating material, so there are fewer photons reaching the detector. As the energy of the photons increase, they are able to penetrate the attenuator materials more deeply, resulting in a higher HVT.



Figure 5.7 Graph of half-value thickness (HVT) for MCP-96 alloy

The buildup factors for various thicknesses are shown in Figure 5.8 for the following photon beam energies: 0.662, 0.835, 1.17 and 1.33 MeV.

The effect of attenuator thickness on the buildup factor is shown for all four gamma ray sources obtained from ⁶⁰Co, ⁵⁴Mn and ¹³⁷Cs. The thickness of the attenuator varied between 1 cm, 2 cm, 3 cm and 4 cm. The experiments were performed for a fixed energy of the beam while the attenuator thickness was varied so that no effect comes

from energy variation. The figure shows that buildup factor increases with increasing thickness of the attenuator. The data closely matched a 2^{nd} order polynomial trendline.

This result can be discussed and explained on the basis of the available scattering sites in the attenuator. Since buildup develops due to the scattering of gamma rays therefore, it is important to have an idea of the available sites for scattering. The probability of Compton scattering increases as the number of electrons or scattering sites increases inside a material.

These scattering sites are proportional to the volume of the attenuating material. As the volume increases, more scattering sites are available and, hence, more contribution to the buildup factor. In our experimental setup, the surface area of the MCP-96 alloy, used as attenuator, was fixed. Thus, we can express our results in mathematical form. The buildup factor is proportional to the scattering sites of an attenuator, and scattering sites are proportional to the volume of the attenuator; therefore, buildup factor is also proportional to the volume of the attenuator. Mathematically;

$$B = C \times V + 1 \quad \dots \quad (3)$$

where B is the buildup factor, C is the proportionality constant and V is the volume of the attenuator. Also

$$V = A \times x \quad \dots \quad (4)$$

where A is the area of that face of the attenuator which received radiation and x is the thickness of attenuator. Thus equation (3) becomes

Thus, for a constant surface area of the attenuator, the buildup factor should be directly proportional to the thickness of the attenuator. These results can be clearly observed in Figure 5.8 below.



Figure 5.8 Graph of buildup factors for various beam energies as a function of MCP-96 attenuator thickness

Figure 5.9 describes the buildup factor of MCP-96 alloy as a function of the energy of gamma ray beam. Four different beams of energy: 1.33 MeV, 1.17 MeV, 0.835 MeV and 0.662 MeV were used for this plot. The thickness of the attenuator is kept

constant while beam energy is varied so that the attenuator thickness effect may be excluded. The figure shows that for low energies of the beam, the buildup factor increases linearly and slowly since the slope of the graph is small. However as the beam energy increases, the buildup factor appears to increase more rapidly. This data fits a 2^{nd} order polynomial trendline as before in Figure 5.8



Figure 5.9 Graph of buildup factors for various MCP-96 attenuator thicknesses as a function of beam energy

As discussed in chapter 2, the buildup results are affected by the interaction of γ rays with matter. A γ -ray photon can interact with a material (attenuator) in three different ways: photoelectric absorption, Compton scattering and pair production. Since the buildup factor is mainly due to the contribution of Compton scattering (and secondary photons produced), one needs to understand how Compton scattering is related to the beam energy. Chapter 2 shows that photoelectric effect is dominant in the γ -ray energy range 0 – 0.5 MeV, Compton scattering is dominant in the 0.5 – 10 MeV while pair production is the main phenomenon of interaction when the photon energy is above 10 MeV. Energy around 0.5 MeV is the interface between photoelectric absorption and Compton scattering. If the beam energy is close to 0.5 MeV, both phenomena have equal chances of occurrence. In this experiment, the γ -ray energy varies from 0.662 MeV to 1.33 MeV.

Clearly, looking at the energy range, more Compton scattering would have occurred with 1.33 MeV versus 0.662 MeV energy. Thus, within our beam energy range, the higher γ -ray energy results in increased Compton scattering. The buildup factor appears mainly from Compton scattering of the beam in the attenuator, therefore, it is observed that the buildup factors of MCP-96 alloy increase with increasing energy.



Figure 5.10 Graph of buildup factors as a function of source-to-attenuator (STA) distances

Figure 5.10 shows the effect of geometry of the experimental setup on the buildup factor. The distance between the radiation source and attenuator, or STA distance, is varied from 5 cm to 15 cm in intervals of 1 cm. A beam of 1.17 MeV energy γ -rays from ⁶⁰Co is used in this whole process to avoid any effects on the buildup factor from beam energy changes. Initially, the graph shows the buildup factor increasing with increasing distance. When the attenuating material is very close to the source, the scattered photons are scattering too short and not reaching the detector to be counted. At a certain intermediate distance, the buildup factor reaches a maximum value and then begins decreasing. This maximum buildup value occurs when the greatest amount of photon scattering is reaching the detector. The buildup decreases as the attenuating material is

moved farther from the source due to the inverse square law and scattering angle of the photons. The results suggest a strong dependency of buildup factor on STA distance.

The smaller value of buildup factors at larger STA distances satisfies the inverse square law. According to inverse square law, the intensity of the particles of a beam moving in all directions, or the intensity of waves or the the strength of a field is inversely proportional to the square of the distance [23, 41]. Mathematically

where I is the intensity, r is the distance from the source producing radiation or field and C is the proportionality constant.

This law applies to the STA results and explains a decrease in the number of scattered photons received by the detector. The rate of scattered and secondary photon production might still be the same, but due to the large STA distance of the scattering sites, most of these photons will not fall in the solid angle of the detector. Therefore, we obtain relatively smaller buildup factor values.

The small buildup factor when the STA distance is short is due to the geometry of the arrangement from a scattered photon's point of view [22]. When the STA distance is small, a smaller area is in the view of the photon beam. This results in only those photons, which fall in a solid angle small enough to hit photons on the detector of the beam, hitting the detector. Photons with larger solid angle have an area that is larger than the detector. These photons miss the detector, and hence, the intensity of the scattered dose received by the detector will decrease. This decrease in the intensity of the scattered beam received by the detector drops the value of buildup factor. In the intermediate distance range, the solid angle is at the optimum size for the detector. Therefore, maximum scattering and initial dose is received by the detector at such intermediate distance, assigning a high value to the buildup factor.

Chapter 6

Discussion

6.1 Summary

As the γ -ray energy increases, the linear attenuation decreases. Higher energy photons are able to penetrate a material more deeply compared to a lower energy photon. The HVT increases linearly with increasing beam energy.

The buildup factors were measured for varying attenuator thickness, beam energy, and STA distance. The buildup factors increase with increasing attenuator thickness and increasing beam energy due to Compton scattering and photoelectric effect. The buildup factors have an optimum STA distance where the value reaches a maximum. This is due to the increasing and decreasing solid angle of photons from movement of the attenuator.

Below is a table of several different materials used as compensators and their respective densities. The mass attenuation coefficient (μ/ρ) was calculated for each material using the material's composition and data [42]. The materials have varying densities and mass attenuation coefficients making a particular material more suitable in certain situations.

| Tissue Compensator | $\mu/\rho \ (cm^2/g)$ | Density (g/cm ²) |
|---------------------------|-----------------------|------------------------------|
| Siemens Ezcut 20 | 0.0299 | 7.81 |
| Elekta | 0.0493 | 11.16 |
| Varian Lead | 0.0494 | 11.2 |
| Varian Steel | 0.0299 | 7.86 |
| Varian Muntsmetal (Brass) | 0.0313 | 8.39 |
| GE Saturne Denal | 0.0461 | 17.5 |
| GE Saturne Inox | 0.0298 | 7.9 |
| Mitsubishi Iron | 0.0299 | 7.8 |
| Mitsubishi Lead | 0.0497 | 11.1 |
| MCP-96 | 0.0483 | 9.72 |

Mass Attenuation Coefficients

Table 6.1 Calculated photon mass attenuation coefficient for commercial tissue compensators and MCP-96 alloy at photon beam energy 10 MeV.

6.2 Conclusion

Thus, the linear attenuation coefficient and buildup factor have been measured and calculated for MCP-96 alloy for its use in radiation shielding, protection and cancer treatment. Four different energies of the γ -ray photon beam were used in the experimental work. It is found that buildup factor increases linearly with attenuator thickness and faster than linearly with the increasing beam energy. The geometry of the experimental set-up also brings variation in the buildup factor. Therefore, we conclude that, along with the attenuator thickness and beam energy, care should also be taken for variations in buildup factor due to the geometry of the experimental set-up in a radiation working and treatment area.

6.3 Future Work

Subsequent testing with higher energy gamma rays should be performed. Ideally, larger MCP-96 blocks would be used with a clinical linear accelerator on a phantom. This would allow for the most accurate results for use at the clinical level. Testing the alloy with various clinical levels of electrons, x-rays, and γ -rays would be ideal. Also, testing the properties of various known molded shapes on a phantom would result in greater certainty and usability.

To obtain a more ideal broad beam geometry set-up, a spherical shell detector would be used. This spherical shell detector would have a small opening to allow the γ ray beam to strike the attenuators within the detector. This would allow the detection of scattered particles at almost 360° [22]. This would collect substantially more scattered particles than the present set-up.

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Appendix I

Error Analysis

Counting statistics were used in the error analysis of this project. The σ of each count, *N*, was determined by calculating the square root of each count.

 $\sigma = \sqrt{N}$

Then, the following error analysis rules were used to determine the error after adding, subtracting, multiplying, dividing, and taking the natural log. Assume the errors in the function are uncorrelated. [43, 44]

| Function | Error |
|-----------|---|
| A + B = C | $(\Delta C)^2 = (\Delta A)^2 + (\Delta B)^2$ |
| A - B = D | $(\Delta D)^2 = (\Delta A)^2 + (\Delta B)^2$ |
| A * B = E | $\left(\frac{\Delta E}{E}\right)^2 = \left(\frac{\Delta A}{A}\right)^2 + \left(\frac{\Delta B}{B}\right)^2$ |
| A/B = F | $\left(\frac{\Delta F}{F}\right)^2 = \left(\frac{\Delta A}{A}\right)^2 + \left(\frac{\Delta B}{B}\right)^2$ |

| $\ln A = G$ | $\Delta G = \frac{\Delta A}{\Delta A}$ |
|-------------|--|
| | A |