COMMISSIONING OF AN UPGRADED EQUINOX-100 COBALT-60 RADIATION THERAPY UNIT IN THE VARIAN ECLIPSE TREATMENT PLANNING SYSTEM

by

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Abstract

The introduction of Cobalt-60 (Co-60) units in the 1950s established megavoltage radiation therapy worldwide. Unfortunately, advances for the Co-60 radiation treatment units were not heavily pursued over the years and Co-60 units have been steadily replaced by clinical linear accelerators (linacs) x-ray systems, which can offer more precise radiation treatments. The Medical Physics Department at the Cancer Centre of Southeastern Ontario (CCCEO) has shown, however, that the Co-60 unit can be developed to enable modern radiation treatments similar to linacs. In 2015, the CCSEO collaborated with Best Theratronics (Kanata, Ontario) to upgrade a Theratron 780C Co-60 radiation therapy unit to an Equinox-100 external beam radiation therapy system. This was the first upgrade of a T780C Co-60 unit in the world, which represents an exciting opportunity to advance Co-60 radiation therapy to provide treatments equivalent to those available from modern linacs.

In this thesis, an upgraded Equinox-100 unit is commissioned in the Varian Eclipse treatment planning systems (TPS) to ascertain if the Analytical Anisotropic Algorithm (AAA) in the Eclipse could be used to generate Co-60 3D based treatment plans. Firstly, mechanical commissioning of the upgraded unit was performed according to Task Group 45 report (AAPM) recommendations. Secondly, various commissioning beam measurements were acquired using an ion chamber in standard water phantom on the upgraded unit to model the beams in the Eclipse TPS. The AAA validation was performed to test the accuracy of the Eclipse TPS dose prediction based on the local percentage differences between the measurements and the Eclipse model calculations in various simple open and blocked plans using the ion chamber in the phantom.

The mechanical acceptance tests are generally comparable to clinical linac tolerances, which suggest that upgrading conventional Co-60 units may improve treatments without compromising the mechanical accuracy. Similarly, the Eclipse model calculations of the validation plans show overall excellent
agreements with measurements within the recommended clinical criteria for large fields starting at $5 \times 5$ cm$^2$. This suggests that the Eclipse TPS is appropriate for forward planning of Co-60 3D conformal radiation therapy for large fields. Further investigations for small fields are required.
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<th>Description</th>
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<tbody>
<tr>
<td>3DCRT</td>
<td>Three Dimensional Conformal Radiation Therapy</td>
</tr>
<tr>
<td>AAA</td>
<td>Analytical Anisotropic Algorithm</td>
</tr>
<tr>
<td>CAX</td>
<td>Central axis</td>
</tr>
<tr>
<td>CCSEO</td>
<td>Cancer Center of Southeastern Ontario</td>
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<tr>
<td>CID</td>
<td>Collimator-isocentre-distance</td>
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<tr>
<td>CNS</td>
<td>Central nerve system</td>
</tr>
<tr>
<td>Co-60</td>
<td>Cobalt-60</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomography</td>
</tr>
<tr>
<td>DRR</td>
<td>Digitally reconstructed radiography</td>
</tr>
<tr>
<td>EBRT</td>
<td>External beam radiation therapy</td>
</tr>
<tr>
<td>ESS</td>
<td>Effective spot size</td>
</tr>
<tr>
<td>ETAR</td>
<td>Equivalent tissue to air ratio</td>
</tr>
<tr>
<td>ICF</td>
<td>Inhomogeneity correction factor</td>
</tr>
<tr>
<td>IGRT</td>
<td>Image-guided radiation therapy</td>
</tr>
<tr>
<td>IMRT</td>
<td>Intensity-modulated radiation therapy</td>
</tr>
<tr>
<td>Linac</td>
<td>Linear accelerator systems</td>
</tr>
<tr>
<td>LMIC</td>
<td>Lower and middle income countries</td>
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<tr>
<td>MLC</td>
<td>Multi-leaf collimators</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>OAR</td>
<td>Organs at risk</td>
</tr>
<tr>
<td>PDD</td>
<td>Percentage depth dose</td>
</tr>
<tr>
<td>POP</td>
<td>Parallel opposed pairs</td>
</tr>
<tr>
<td>PTV</td>
<td>Planning target volume</td>
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<tr>
<td>RDF</td>
<td>Relative dose factor</td>
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<tr>
<td>Acronym</td>
<td>Description</td>
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<tr>
<td>---------</td>
<td>--------------------------------------------------</td>
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<tr>
<td>RTAR</td>
<td>Ratio of tissue to air ratio</td>
</tr>
<tr>
<td>SAD</td>
<td>Source- axis-distance</td>
</tr>
<tr>
<td>SCD</td>
<td>Source- collimator-distance</td>
</tr>
<tr>
<td>SSI</td>
<td>Second source intensity</td>
</tr>
<tr>
<td>TAR</td>
<td>Tissue- air ratio</td>
</tr>
<tr>
<td>TERMA</td>
<td>Total energy released per unit mass</td>
</tr>
<tr>
<td>TMR</td>
<td>Tissue maximum ratio</td>
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<td>TPR</td>
<td>Tissue phantom ratio</td>
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<tr>
<td>TPS</td>
<td>Treatment planning system</td>
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<tr>
<td>VMAT</td>
<td>Volumetric-modulated arc therapy</td>
</tr>
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1

Introduction

1.1 Background of Radiotherapy

Cancer is currently the leading cause of death in Canada with the probability of 1 in 2 people getting the disease, while approximately 1 out of 4 is expected to die of it [Canada Statistics, 2017]. Globally, cancer is the second most common cause of death after heart diseases, with estimations showing that 9.6 million people were expected to die of cancer in 2018 [WHO, 2017]. Approximately 70% of deaths occur in the lower and middle income countries (LMICs). The high cancer mortality in LMICs is caused by population growth and ageing while their health systems remain unprepared to meet cancer care challenges [Grover, 2015].

The global options for cancer treatments include systemic therapy, radiation therapy and surgery. Of these, radiation therapy treats around 50% of all cancer patients worldwide both curatively and palliatively [Baskar, 2017]. Radiation therapy is the medical application of high energy ionizing radiation for the treatment of cancer. The goal of radiation therapy is to deliver a sufficient radiation dose to kill all tumour cells while minimizing the dose to healthy tissues. Radiation therapy can be delivered to the tumour through external beam radiation therapy (EBRT) or brachytherapy. EBRT involves targeting the tumour with radiation beams from a machine outside the body while brachytherapy involves placing a radioactive material inside the body. Over the years, radiation therapy has been changing in terms of radiation energy and dose delivery techniques. Starting with the low energy x-rays in medical applications, improvements in radiation therapy began when man-made x-rays in the range of kilovoltage x-rays (energy range
of 100 to 500 kV) were used [Dyk, 1996]. However, that energy range had poor penetration and imposed high doses on the skin when treating deep-sited tumours. The introduction of Cobalt-60 (Co-60) unit followed by linear accelerator (linac) systems unit in the early 1950s was the beginning of the megavoltage application in external beam radiation therapy [Litt, 2002; Bernier, 2004]. These two machines revolutionized the medical applications of radiation. The Co-60 unit was a major Canadian contribution to the treatment of cancer [Litt, 2002]. Characterized by operation simplicity and relatively low running costs, the Co-60 unit was the mainstay in cancer departments worldwide until in the early 1970s [Hogstrom, 2006]. Factors such as beam penetration, increased penumbra and low dose output (among others) were perceived reasons for the drop in clinical applications of Co-60 units [Suit, 1986]. The linac became the main choice for radiation therapy in the developed countries, because of the ability to provide higher energies between 4 - 25 MV for both photons and electrons on the same unit, [Bortfeld, 2006]. Currently, most of the available Co-60 units used for radiation therapy are in LMICs [Litt, 2002].

Traditionally, dose delivery was through beams with different beam blocking devices such as custom cerrobend or lead blocks, collimator jaws, and physical wedges used either to protect healthy tissues surrounding a target or provide uniform dose distribution to the target. Currently, different delivery techniques are used to optimize the spatial distribution of the radiation delivered dose. These techniques conform radiation dose to the tumour using multiple beams from different angles, which significantly lowers the dose to healthy tissues. These conformal methods are currently the standard of practice in radiation therapy. Three dimensional conformal radiation therapy (3DCRT) [Webb, 1995] uses a selected number of radiation beams and their geometry to provide a highly conformal dose according to size and shape of the tumour. The introduction of dynamic multi-leaf collimators (MLC) enhanced intensity modulated radiation therapy (IMRT) [Webb, 2001], which manipulates the number and orientation of the radiation
beams, the shape of each beam, and the in-field intensity of the radiation to provide a 3D conformal dose distribution to the tumour. The type of IMRT that delivers dose using one or more gantry arcs of continuous variation of beam aperture, gantry speed and dose rate is known as Volumetric Modulated Arc Therapy (VMAT) [Otto, 2008]. The application of VMAT in clinics is currently increasing primarily of its time efficiency. The conformal techniques depend on the application of imaging tools such as Computed Tomography (CT) and Magnetic Resonance Imaging (MRI) to provide three dimensional (3D) images that are useful in treatment planning.

1.2 Motivation

As discussed above, the advances in radiation therapy involve both the physical configuration of the treatment unit and the radiation dose delivery techniques. Unfortunately, most conventional Cobalt-60 units have remained with simple configurations that are unable to accommodate the computer driven multi-leaf collimators (MLC) to facilitate conformal deliveries. Consequently, the global advances towards modern radiation treatment for cancer patients have been limited as cancer clinics in many countries still depend on Co-60 for their radiation delivery.

Furthermore, the modern conformal treatment techniques involve complex processes that require a computerized treatment planning system to provide patient specific dose calculations and determine the optimal and practicable treatment plan to be delivered during treatment. The treatment planning system uses 3D images acquired by x-ray CT simulation for target and organ at risk delineation and also uses the CT data (the electron density) for dose calculations. Unfortunately, in many countries with Co-60 units there have been limited application of the
computerized treatment planning in Co-60 radiation therapy because modern treatment planning systems do not all have modules that can model the conventional Co-60 beams.

In order to develop a conventional Co-60 unit capable of handling conformal delivery methods, the Cancer Center of Southeastern Ontario (CCSEO) in Kingston recently upgraded their Theratron 780C (780C) (Best Theratronics, Kanata, ON, Canada) unit to an Equinox-100 Co-60 unit. This was the first such upgrade in the world of a Theratron 780C Co-60 unit and represents an exciting opportunity to advance conventional Co-60 technology so it is capable of providing treatments equivalent to those available on modern linacs. The upgrade involved the installation of a new mainframe (that holds the gantry) standing on the previous base, which increased source-to-axis distance (SAD) from 80 to 100 cm. Consequently, the collimator-to-isocentre distance (CID) increased from 35 to 49 cm, allowing the incorporation of removable multi-leaf collimator (MLC) enabling conformal radiation therapy capabilities. However, the removable MLC and a new Co-60 source were not provided by Best Theratronics at the time of the upgrade. The upgrade also included installation of a computer controlled collimator system with a

Figure 1.1: (a) The former Theratron 780C of CCSEO before the upgrade and (b) the upgraded Equinox-100 Co-60 unit.
motorized 60° wedge, where the new collimation system allows mounting of the removable multi-leaf collimators. Thus, the head of the upgraded Equinox-100 comprises the source housing from the T780C with collimation system from Equinox-100 model. The treatment unit’s analogue control system was replaced by computerized system enabling enhanced control of positioning and remote visualization of the treatment set up conditions in the treatment room from the control. A new Avanza patient positioning table (couch) was installed to enable patient positioning with improved motion accuracy and stability better than the T 780C couch that did not have integrity to handle modern deliveries. Furthermore, all emergency buttons on the unit, couch, in the treatment room, and at control console were replaced to ensure functionality with the new control system. Also, additional security interlocks, a new last person out (LPO) switch and treatment room visualization were installed to ensure radiation safety. The acceptance tests to ensure the performance of different components are within acceptable specifications were performed by Best Theratronics and CCSEO staff.

In this thesis the mechanical assessment to test the accuracy of the upgraded Equinox-100 for dose delivery is performed. The assessment includes various measurements outlines in AAPM Task Group 45 report [Nath, 1994] for linacs. Furthermore, the commissioning of the upgraded Equinox-100 in the Varian Eclipse treatment planning system (Varian Medical Systems, Palo Alto, CA) is performed. Because the removable MLC is not available at the center and the dose rate of the old Co-60 source at the isocentre has reduced to about 50 cGy/m, the commissioning of the upgraded Equinox-100 in the Eclipse TPs is only performed for open beam measurements. Various dosimetric measurements required for configuration of the Eclipse AAA algorithm were acquired, and various source modeling parameters were adapted and optimized in the Eclipse TPS to model the Cobalt-60 source. The Eclipse TPS was validated by comparing the treatment plan calculations with ion chamber, and radiochromic film dosimetry measurements in water phantom.
1.3 Thesis Outline

The successful modification of the conventional T780C Co-60 unit to an upgraded Equinox-100 at the CCSEO illustrates the potential for doing similar modifications worldwide. Also, the commissioning Co-60 in the Eclipse TPS provides an option for 3D treatment planning system that has been hindering the application of conformal techniques on Co-60 units.

Chapter 2 provides a review of the historical background of Co-60 radiation therapy. This includes its discovery, beginning of Co-60 radiation therapy and also the challenges for ongoing clinical applications. The chapter also highlights the background of treatment planning in Co-60 radiation therapy. It points out the significance of Co-60 3D treatment planning in modern EBRT. The chapter concludes by examining the historical backgrounds of different dose calculation methods that have been incorporated in TPS systems. Also, generations of different dose calculation methods are discussed to provide an understanding of the past as well as the present algorithms used in different TPSs.

Chapter 3 discusses the theoretical aspects of the dose calculations used in radiation therapy. The first section reviews the fundamentals of photon interactions with matter, energy absorption and deposition to provide an understanding of the concept of absorbed dose. In the second section, the dose algorithms used in TPS systems are described. The concepts of scatter kernels together with the particular pencil beam algorithm are described to provide an understanding of fundamentals of Analytical Anisotropic Algorithm (AAA) used in this thesis. The third section reviews the implementation of AAA in the Eclipse 3D TPS for dose calculations.

Chapter 4 describes the different materials and methods used in the thesis investigations. The chapter begins with a description of the modification carried on the previous T780C Co-60 unit to
create the upgraded Equinox-100. Then a review of the mechanical testing to check the accuracy of upgraded unit mechanical motions is provided. The third section provides the procedures and equipment used to acquire commissioning beam data. The fourth section includes a sub-section that explains the Eclipse configuration process. This helps to explain the different parameter settings that enable the Co-60 beam configuration in Eclipse. The second sub-section provides various Eclipse validation methods in which the measurements and Eclipse calculations were compared. Finally, methods for validating the delivery of simple treatment plans performed on a film are provided.

Chapter 5 provides the results of all the measurements performed in this thesis as previously outlined. Chapter 6 provides summary, conclusion and draws recommendations for future Co-60 3D treatment planning work.
2

Literature Review

This chapter reviews the literature regarding Cobalt-60 (Co-60) radiation therapy. The historical background of Co-60 radiation therapy is provided in the first section. The developments of treatment planning for Co-60 radiation therapy are then highlighted. The chapter concludes by reviewing different dose calculation algorithms that have been used in clinical treatment planning systems (TPS). The algorithms include those introduced at the beginning of megavoltage radiation therapy and those currently used in current TPS systems.

2.1 Historical Background

This section describes the development of the Co-60 isotope, the beginning of the Co-60 radiation therapy and the challenges that led to decline in use of Co-60 radiation therapy.

2.1.1 Introduction to Cobalt-60 Radiation Therapy

The discovery of the Co-60 isotope in the early 1940s came after a series of studies that began in the 1930s [Livingood, 1941]. These early investigations used Co-59 to produce Co-60 isotope through three main reactions: $^{59}\text{Co} (n, \gamma) ^{60}\text{Co}$, $^{59}\text{Co} (d, p) ^{60}\text{Co}$ and $^{62}\text{Ni} (d, \alpha) ^{60}\text{Co}$. The reactions produced two Cobalt isomers with half-lives of 10.7 minutes and 5.26 years. However, the Co-60 isomer with half-life of 5.26 years was of interest for long period applications. Currently, Co-60 isotope is produced based on the first reaction above by bombarding neutrons on a Co-59 target in a nuclear reactor. The Co-60 decays by emitting two beta particles with
Figure 2.1: (a) Co-60 isotope decay scheme and (b) the two dominant gamma ($\gamma$) photon energy peaks in megavoltage range suitable for clinical applications. From [Joshi, 2008; Traitor, 2007] with permission.

0.313 MeV and 1.486 MeV energies leading to excited Nickel-60 (Ni-60), which becomes stable by emitting two gamma photons at 1.17 MeV and 1.33 MeV. The decay process of the Co-60 isotope with half-life 5.26 years is shown in Fig 2.1 (a). The spectrum from a Co-60 T780C treatment unit is shown in Fig 2.1 (b). Note that the two high energy $\gamma$ peaks dominate the spectrum but that additional lower energy components (shown at the bottom) are also emitted during the decay.

The megavoltage range of Co-60 gamma ($\gamma$) photons prompted interest as a potential replacement to low energy radium (average energy of 0.83 MeV) for clinical applications [J. S. Mitchell, 1947]. In the early 1950s, the application of Co-60 radiation therapy began in two different independent hospitals in Canada; one in Saskatoon (Saskatchewan Cancer Institute) and the other in London (Victoria Hospital, Ontario). Since that time, different manufacturers led by Atomic Energy of Canada (Theratronics), produced Co-60 units for clinical applications worldwide. Currently, Co-60 units for external beam radiation therapy (EBRT) are available from Best
Favored by its simplicity, robustness, and low maintenance costs, Co-60 units became the standard of care worldwide. For the first few decades of megavoltage radiation therapy, Co-60 units were preferred to the alternative linear accelerator systems (linacs) that were installed around the same time in London, United Kingdom to produce 8 MV x-ray [Thwaites, 2006]. At the end of the 1960s, there were 1700 radiation therapy units worldwide, of which 90% were Co-60 units [Hogstrom, 2006]. However, large investments made to develop linac technology in 1970s marked the beginning of the downturn in the clinical use of Co-60 units. The linac can provide both photons and electrons with energies ranging from 4 - 25 MV on a single unit. The high energy photons are preferred when targeting deep-sited tumours with small number of beams, while the electrons are preferred for superficial cancers near the patient surface. A further advantage with linac treatment units is the ability to achieve high dose rates above 600 cGy/min at the isocentre.

These advantages, combined with the lack of technical improvements of the Co-60 unit, diverted clinical adoption to linacs as the treatment of choice in developed countries. As a result, new devices including multi-leaf collimators (MLCs) that support conformal dose delivery techniques such as intensity modulated radiation therapy (IMRT) were developed based on linac technologies. Furthermore, most of the commercial treatment planning systems (TPS) introduced to facilitate conformal techniques were based on linac technology as opposed to Co-60. These new devices and techniques improved patient care for patients treated with linacs, which led to the decrease and eventual discontinuation of Co-60 radiation therapy in developed countries in the 1980s [Suit, 1986]. According to the most recent data available from the International Atomic
Energy Agency (IAEA) directory of radiation therapy centers (DIRAC), there are over 13,600 radiation therapy units in member states; of those about 11,400 (84%) are linacs. The majority of the Co-60 units for clinical applications are found in low and middle income countries (LMIC), while most of those in developed countries are used for research.

2.1.2 Challenges for Cobalt-60 Radiation Therapy

The application of Co-60 units in radiation therapy has been challenged by different factors including beam penumbra, gamma photon energy and dose rate. Also, the increase of terrorist activities raises a concern of possible misuse of the Co-60 source if found not decommissioned and removed securely. However, as shown in this section, these challenges can be reduced by modifying some features of the Co-60 unit as well as undertaking required safety measures for the Co-60 source transportation and storage. Furthermore, if the new devices and dose delivery techniques are used in Co-60 radiation therapy, the effects of these challenges are potentially minimized.

2.1.2.1 Beam Penumbra

Penumbra is defined as the region at the edges of the beam at which the dose decreases rapidly as a function of distance from the beam axis [Faiz M. Khan, 2003]. The size of the penumbra is mainly associated with the finite size of the radiation source used in the treatment unit. The Co-60 beam has large penumbra because of its significant source size. The size of the penumbra is one of the reasons that led to the decline of Co-60 clinical applications. The beam penumbra has three different classifications: geometric, transmission and physical. The geometric penumbra at depth (d) in a medium depends on geometrical parameters such as the source sizes (s),
Figure 2.2: The size of the penumbra region (blue) depends on the size of the radiation source in the treatment unit: (a) the large penumbra size due to Co-60 source and (b) a smaller penumbra size for linac.

source-to-collimator distance (SCD), and source-to-axis distance (SAD) [Dyk, 1996]:

\[
Geometric \ penumbra \ (P_d) = s \times \left( \frac{SAD + d - SCD}{SCD} \right) \tag{2.1}
\]

The geometric penumbra increases with the source size and SAD, while decreasing with SCD and is independent of the field size, provided the SCD remains the same. The transmission penumbra is caused by photon transmission through the edges of the collimating jaws. This effect increases with field size caused by the divergence of the photons at the edges of the collimating jaws. The combination of the transmission and geometric penumbras create a region of dose variation at the edges of the field size known as physical penumbra. The physical penumbra (referred in this thesis as penumbra) is measured on a dose profile and defined as the distance between 20% and 80% of the central axis dose. The typical penumbra of a Co-60 unit is considered to be about 1.4
cm for a 10 x 10 cm² field size at depth of maximum dose (0.5 cm) and a source size between 1.5 cm and 2 cm in diameter [Dyk, 1996]. It increases with depth, depending not only on geometric and transmission penumbra, but also on scattered radiation produced in the medium [Faiz M. Khan, 2003]. Note, however, that the main reason for the large penumbra of the Co-60 beam is the source size which dominates over the contributions from the transmission and scattered photons, which are very minimal. Figure 2.2 illustrates the relationship between the source size and the penumbra for both Co-60 and linac units. In comparison, a linac with a focal point source between 0.1 cm and 0.3 cm in diameter has a penumbra region in the range of few millimeters, taking into consideration typical SCD and collimator to isocentre distance (CID) [Cadman, 2007].

However, the effect of the penumbra size is more pronounced when a small number of radiation beams are used to deliver the dose to the target. In such settings, if the penumbra extends outside the planning target volume (PTV), it could result into potential dose to normal tissues. Alternatively, in modern radiation therapy, the sufficient and conformal radiation dose to the target is delivered using multiple beams from different angles. The dose patterns (dose distribution) in the treated volume are determined by the inverse treatment planning in the TPS through dose constraints (specific dose distribution conditions), which takes in account the penumbra dose. Also, if the penumbra is accurately characterized in the TPS, it can be accounted for as multiple beams treatment plans are generated. A study conducted to investigate the suitability of Co-60 units for IMRT (which uses inverse treatment planning) compared to linacs concluded that there was no inferiority in terms of beam penumbra between Co-60 and linac units [Fox, 2008]. Thus, the use of the TPS and the modern dose delivery techniques potentially reduces the concern of Co-60 unit penumbra size.
2.1.2.2 Photon Energy

As previously mentioned, Co-60 decay produces $\gamma$ photons with equal probability at 1.17 MeV and 1.33 MeV which can be averaged as 1.25 MeV. Treating deep lying tumours using Co-60 energy with a small number of beams would result in high doses to tissues near the surface. However, the application of multiple beams from different angles or the use of volumetric-modulated arc therapy (VMAT) eliminates the need for high energies. For instance, the studies performed to investigate the energy required for IMRT demonstrated that there is no benefit of higher energies beyond 6 MV even for deep-sited cancers such as prostate cancer [Pirzkall, 2002; Sun, 2006]. Similar results were also obtained in a comparative study for IMRT between Co-60, 6 MV and 18 MV in the treatment of head-and-neck, prostate, central nerve system (CNS), breast and lung cancers [Fox, 2008]. This study concluded that there is little to no difference for the coverage of the PTV between the three beams, while the 18 MV demonstrated higher doses to organs at risk. These findings suggest that Co-60 energy when used in modern radiation therapy offers sufficient dose to the target while minimizing dose to normal tissues.

2.1.2.3 Dose Rate

A low dose rate offered by the Co-60 unit compared to the linac is another challenging concern for radiation therapy applications. With typical new source activity up to 527 TBq (commercially available Co-60 source, MDS Nordion), a 2 cm diameter Co-60 source can provide dose rates of about 250 cGy/min at the isocentre (100 cm source to axis distance) while a linac is typically between 400 and 600 cGy/min [Dyk, 1996]. The low dose rate translates to longer treatment times, which may results in low patient output per day. Nevertheless, various modifications of beam parameters and source dimensions and structure have been suggested to improve dose rate
to become comparable to that of the linac [Cadman, 2007; Joshi, 2008]. A study by Joshi et al., suggested that the application of a rectangular-shaped source for fan beam in combination with carefully selected SAD and SCD could enhance the output of Co-60 units and improve the penumbra compared to traditional cylindrical source geometry.

2.1.2.4 Safety Concerns

The increase of terrorism activities involving nuclear applications, inappropriate source disposal and decommissioning raise safety concerns, particularly the possibility of source theft. With a half-life of 5.26 years, the Co-60 source requires replacement after the time of one half-life or less has elapsed, to maintain a suitable clinical dose rate. This raises a safety concern, especially during transportation, operation and disposal in countries with relaxed regulations. It has even been suggested in the United States that Co-60 isotope should be phased-out from applications [Chou, 2018]. There have been reported accidents that occurred following theft or abandoning of the Co-60 source in different places around the world. In 2000, an unshielded discarded Co-60 source with activity of 15.7 Bq was left among scrap metal, which was sold to scrap dealers in Thailand [Thongpraparn, 2002]. Radiation exposure to a number of individuals in this incident caused three fatalities and ten injuries. Another example is from Mexico in 2013 [Martinez, 2013], where a Co-60 radiation therapy source that had been transported to a radiation therapy center was stolen from a cargo truck. The source was later found abandoned in a rural area with no reported injuries or deaths, although the thieves might have been exposed to high doses of radiation. The two incidents were likely caused by inadequate radiation source control regulations in those countries. If strong international and national regulations are in place, Co-60 sources can be transported and used safely [IAEA, 2018].
2.2 Treatment Planning in Radiation Therapy

Treatment planning involves several processes including patient positioning and immobilization, CT scanning, target localization and delineation, beam generation and their orientations, and dose calculations. Here the entire process of the treatment planning and its different stages are reviewed along with a general historical review of the evolution of the treatment planning over the years. The section concludes with review of the treatment planning specific for Co-60 radiation therapy.

2.2.1 Background of Treatment Planning

The beginning of megavoltage radiation therapy in the 1950s prompted the need for treatment planning to determine appropriate beam orientation and delivery time of treatment. Traditionally, radiation therapy simulation and planning was performed based on information collected through palpation and planar diagnostic x-rays [Dyk, 1999]. Beam positioning and orientation was indicated by landmarks (bony and fiducial marks) to avoid missing a target. A conventional treatment simulator, which was a combination of diagnostic x-ray unit with treatment unit geometry was introduced in the 1960s to enable the same geometrical patient position as on the treatment unit [Dyk, 1999]. This was the beginning of 2D treatment planning with manual dose calculations. The process involved obtaining a patient fluoroscopy image acquired using lead wire or plaster strips in a patient treatment position on the simulator table [Podgorsak, 2005]. The image information was used to determine anatomical shielding requirements based on the landmarks that were visible under the fluoroscopy images. Portal films acquired during treatment were also periodically compared by fluoroscopy images to ensure correct patient positioning. Figure 2.3 shows the evolution of treatment simulation methods in radiation therapy.
The introduction of x-ray computed tomography (CT) in 1971 was the beginning of 3D treatment planning [Podgorsak, 2005]. The x-ray CT provided 3D patient images of the treated volume with high resolution based on the x-ray attenuation coefficient maps in a patient. At CT scanner x-ray energies, the x-ray attenuation depends on the electron density of the tissue in the patient. The information of the electron density in the treated volume image was integrated into the computerized treatment planning system for dose calculation purposes. A special version of the CT scanner known as a CT simulator (CT-sim) was developed with the same patient couch as that available at the treatment unit to ensure the CT images were acquired at simulation with the patient in the treatment positions.

With the development of computers and related technologies, treatment planning has changed to a multiple stage process that involves several procedures that start after patient diagnosis. Figure 2.4 shows the process involved in modern treatment planning. Firstly, patient information is collected using a CT-sim to identify the location, size and extent of the tumour and its surrounding organs at risk (OAR). The positioning of the patient on the CT-sim is performed in a way that can be reproduced on the treatment unit using immobilizing devices and skin markers.
Figure 2.4: Illustration of the modern treatment planning process, starting at CT simulator, to planning unit ending at the treatment unit for plan implementation.

The images from the CT-sim are then imported in the computerized TPS station where patient contouring is performed to show the extension of the tumour and the critical organs. Next, a virtual treatment simulation is performed in the TPS to determine beam configurations based on the tumour size and position. Finally, dose distributions in the tumour and critical organs are calculated, and the optimal plan is delivered to the patient on the treatment unit.

2.2.2 Treatment Planning for Co-60 Radiation Therapy

Over the years, Co-60 radiation therapy has operated with simple treatment planning. The advances such as the application of 3D TPS observed for linac users have slowly been implemented in Co-60 radiation therapy. This is mainly due to a limited availability of the commercial 3D TPS specific for Co-60 radiation therapy. As a result, most of the cancer centers that depend primarily on Co-60 radiotherapy in most of the low and middle income countries (LMICs) are limited to the use 2D in-house TPS or even no TPS at all [Page, 2014; Mu’minah, 2016]. However, tumours and adjacent normal tissues are 3D structures with complex
relationship that makes the conventional treatment planning methods inadequate for cancer control treatments. Alternatively, the 3D TPS systems provide an advantage of clear target localization and sufficient conformal dose calculations that lead to tumour control. A study of ten prostate and seminal vesicles patients to compared dose coverage to the treated volume between 2D and 3D treatment planning indicated that the irradiated volume was reduced by 19% for the rectum and 28% for the bladder in 3D planning than observed in 2D planning [Wiegel, 1992]. This suggests that, 2D planning does not provide the tumour dose or normal tissue sparing that can be achieved with conformal 3D plans.

Co-60 units have already shown a capability to produce 3D dose distributions in target volumes that are comparable to linacs using modern dose delivery techniques. Recently the application of Co-60 magnetic resonance image guided radiation therapy (MR-IGRT) system for IMRT has been introduced [Wooten, 2015]. The MR-IGRT system combines real-time MR image guidance and IMRT methods to provide high resolution soft tissue visualization during dose delivery. To study the performance of Co-60 IMRT under the MR-IGRT system, plans were generated in ViewRay TPS (Oakwood Village, OH) with doses distribution calculated with a collapse cone convolution algorithm [Wooten, 2015]. The results showed that, the Co-60 plans achieved 100% and 95% volume coverage similar to linac. A similar study for IMRT was performed using Monte Carlo simulations to compare dose distributions for linac and Co-60 units, and suggested that Co-60 IMRT can provide similar dose distributions to those found on linacs [Fox, 2008]. Various other studies have also shown the possibility of incorporating Co-60 units and imaging tools such as x-ray CT [Schreiner, 2003], portal imaging [Van De Bunt, 2006; Marsh, 2017], EPID imaging [Dhanesar, 2013], and MRI [Kron, 2006]. All these studies indicate the feasibility for Co-60 units to provide the dose delivery required for in modern radiation therapy.
The main challenge towards the modern Co-60 radiation therapy is the availability of TPS systems for planning the complicated dose delivery techniques. Therefore, this work intends to establish Co-60 in the Eclipse TPS to allow the treatment planning for the dose delivered by the Co-60 beam. This will be a significant step that will open up the possibilities for the application of Co-60 units in the modern delivery techniques such as 3DCRT and IMRT.

2.3 Photon Dose Calculations

An important element of treatment planning involves the dose calculation, in which the dose distribution within the treated volume is determined based on the planned beam configuration (beam angle, collimator settings, etc.) as described previously. In early radiation therapy, dose calculations were performed manually as a treatment used only a limited number of beams and patient anatomy was assumed to be homogenous. However, in the current advanced dose delivery techniques such as 3DCRT and IMRT, dose calculations are more complex and require incorporation of sophisticated algorithms in the TPS systems. In fact, for successful radiation therapy it is essential for the algorithm to accurately estimate the absorbed radiation dose in various tissues during treatment.

Dose calculations were first introduced using empirical methods developed shortly after the introduction of megavoltage radiation therapy units. The dose calculations at that time only considered external body contours and assumed the rest of the body to be made of homogeneous water equivalent tissue. Thus, dose calculations ignored the effect of tissue inhomogeneity, which affected the accuracy of the calculated absorbed dose [Papanikolaou, 2009]. The advent of x-ray CT in the 1970s changed dose calculation methods significantly, as anatomical information such as tissue local electron density useful for dose calculations were then available on the images.
This enhanced the development of dose calculation algorithms that accounted for inhomogeneities, which significantly improved dose calculations [Papanikolaou, 2009]. Dose calculation algorithms used in radiation therapy are categorized into two main groups: correction-based and model-based. The basis of each category and their application period in radiation therapy treatment planning is reviewed in the following sections.

2.3.1 Correction based algorithms

Correction based dose calculation algorithms were incorporated in TPS systems in the early days of megavoltage radiation therapy and continued until after the introduction of x-ray CT in the 1970s [Dyk, 1999]. These methods were used for both Co-60 and linac beam based therapy. The correction-based algorithms were fundamentally based on the absorbed dose measurements acquired in homogeneous water phantoms. These measurements could be corrected to account for the inhomogeneity in a patient. The correction-based algorithms are categorized as one- or two-dimensions depending on whether the algorithm accounts for inhomogeneity correction in both primary and scattering photons [Purdy, 1992; Mackie, 1996; Papanikolaou, 2009]. The primary photons are radiation photons that come directly from the radiation source to the patient without prior interaction, while scattering photons are those interacting in the head of the treatment unit or in air before reaching the patient [Papanikolaou, 2004].

2.3.1.1 One-dimensional algorithms

One-dimension or effective pathlength methods such as the ratio of tissue to air ratio (RTAR) and Power-law method, were first introduced by Milan-Bentley in the 1970s [Milan, 1974]. The 1D algorithms were based on various effective pathlength approaches in which only electron density
information along the primary photon track to the dose calculation point was considered to account for the effect of inhomogeneities [Dyk, 1999]. Both methods used a central axis dose parameter namely tissue-to-air ratio (TAR) to correct for the presence of inhomogeneity in tissues such as lung, bone and air cavities in patients. TAR is the ratio of dose at a point, Z on the central axis in a phantom to dose at the same point, Z in air on the beam central axis. The TAR provided information regarding the dose fall off along the central axis according to the inverse square law. The inverse square law states that the photon fluence is inversely proportional to the square of the distance from the source. The general formula for ID methods was:

\[ D_{\text{inhom}}(x, y, z) = D_{\text{water}}(x, y, z) \times ICF, \]  

where \( D_{\text{inhom}}(x, y, z) \) was absorbed dose in inhomogeneity tissue, \( D_{\text{water}}(x, y, z) \) was absorbed dose in homogeneous water phantom, and \( ICF \) was an inhomogeneity correction factor.

The initial ICF used a ratio of tissue-air ratios (RTAR) to provide the correction:

\[ ICF(d, r) = \frac{TAR(d', r)}{TAR(d, r)}, \]  

where \( d \) was the physical depth and \( d' = d - \rho_2 \times t \) was the water equivalent depth, \( \rho_2 \) was density of the inhomogeneity, \( t \) was the thickness of the inhomogeneity, and \( r \) was the field size at \( d \).

A later ICF used a Power law correction given as:

\[ ICF(d, r) = \frac{TAR(d_1, r)^{\rho_1 - \rho_2}}{TAR(d_2, r)^{1 - \rho_2}}, \]  

where \( \rho_1 \) and \( \rho_2 \) were electron densities of material in the calculation volume and of the overlying materials respectively, and \( d_1 \) and \( d_2 \) were the depth of the point of calculation below the material and the depth below the upper surface. Figure 2.5 shows the presence of inhomogeneity in the medium for the 1D algorithms.
The 1D algorithm methods were simple, fast and could be verified through hand calculations, which was useful for that time of limited computer power. However, the 1D algorithm methods had limitations. For instance, the RTAR method only considered the absorbed dose at a point that was only affected by the variation of electron density in the path along the direction of the primary photon from the source. It did not take into account the density variation from scattering photons elsewhere in the patient. As a result, the method overestimated the dose for inhomogeneities with a lower electron density than water while underestimated the dose under a region with the electron density greater than water [Purdy, 1992; Papanikolaou, 2009]. On the other hand, the Power law method considered the location of inhomogeneity in relation to the point of calculation, thus included the scattering component. But, it did not consider the depth of the inhomogeneity material from the water surface. The Power law method also assumed the maximum build-up of dose in inhomogeneity to be the same as that of water [Purdy, 1992; Dyk, 1999]. Thus, the method broke down when $d_3$ was less than the build-up depth of that energy. Also, since the measurements used to determine the TARs were acquired in homogeneous water, the 1D methods failed to extend to situations where incoming and outgoing electrons were not in equilibrium (electron disequilibrium).
2.3.1.2 Two-dimensional algorithms

Two dimensional (2D) dose calculation methods were introduced to remedy some of the limitations of the 1D methods. An example of a 2D approach was the equivalent tissue-air ratio (ETAR) method. The ETAR was introduced by Sontag and Cunningham [Sontag, 1977] to account for the effect of lateral scattering. With the availability of x-ray CT data and enhanced computer power, this method was able to make use of the 3D CT images to include size, shape, density and position of the inhomogeneity in the patient [Purdy, 1992]. The inhomogeneity correction factor from the ETAR was given by:

\[ ICF = \frac{TAR(d', r')}{TAR(d, r')} \]  \[ \text{[2.5]} \]

where \( d' \) was the tissue equivalent pathlength between the source and the interaction point \( P \) and \( r' \) was the radius of the equivalent field size obtained by scaling the actual field \( r \) using an equivalent density given as: \( r' = r \rho' \). The equivalent density was a weighted average of voxel densities in the entire medium, where the weights depend on the relative scattering photons of each voxel.

\[ \rho' = \frac{\sum_{i,j,k} W_{i,j,k} \rho_{i,j,k}}{\sum_{i,j,k} W_{i,j,k}} \]  \[ \text{[2.6]} \]

The density scaled TAR was divided into primary and scattering components (SAR) as a result of separation of the equivalent distance and the field radius [James A Purdy, 2004].

\[ TAR(d', r') = TAR(d', 0) + SAR(d', r') \]  \[ \text{[2.7]} \]

where the first term represents TAR of the primary photons, and the second term represents the TAR from the scattering photons. The inclusion of the scattering photons to determine the dose at a point significantly improved dose calculations compared to the 1D algorithms. A study for
inhomogeneity corrections in a phantom showed that the difference between the calculated and measured dose of the Co-60 beam was within 3% using 2D methods compared to up to 10% for 1D methods [Sontag, 1977]. However, the ETAR dose calculation method also failed to account for regions with electron disequilibrium.

2.3.2 Model-Based Algorithms

A major development in dose calculation was achieved when model-based algorithms were introduced in the mid-1980s [Ahnesjö, 1987; Dyk, 1999]. The model-based algorithms were capable of providing 3D dose calculations by independently modeling the primary and scattered photons as well as accounting for tissue inhomogeneity throughout the medium. Consequently, model-based methods have higher accuracy of dose calculations compared to correction-based methods. Currently, most of the TPS systems incorporate model-based algorithms. Two types of model-based methods are reviewed in this thesis; the Monte Carlo simulations and convolution/superposition calculations.

2.3.2.1 Monte Carlo method

Monte Carlo simulations model dose distributions by simulating photon and electron transport in a medium [Rogers, 1995]. Photon simulation starts with information about the photon location, momentum and energy. This information allows the sampling of the probability whether a photon will interact when traveling between distance \( l \) and \( l+dl \) into a medium [Reynaert, 2007]. A cumulative probability for all interactions is given by Eqn.[2.8]:
Figure 2.6 Series of photon interaction histories as simulated by Monte Carlo.

\[
P(l) = \int_0^l \mu e^{-\mu s} ds = \cdots = 1 - e^{-\mu l}. \tag{2.8}
\]

The probabilities depend on the photon energy and attenuation coefficients \( \mu \) of the medium. Figure 2.6 shows the interaction of the incidence photon which results in a scattering photon and secondary electrons. The photon path lengths can be sampled from the probabilities using a uniformly distributed random number from the interval \([0, 1]\) as shown in Eqn.\([2.9]\):

\[
r = 1 - e^{-\mu l} \Rightarrow l = -\frac{1}{\mu} \ln(r). \tag{2.9}
\]

The attenuation coefficients are given as a total of linear coefficients of the photoelectric, Compton scattering and pair production photon interactions. The attenuation coefficients are different for photons with different energy. The stochastic nature of photon interaction in random cross-sections is followed until the photon leaves the medium or reaches the cut-off energy [Andreo, 1991]. In the case of a medium with different material, the photon is simulated to the nearest material boundary where distances are sampled based on the attenuation coefficient of the new material. The secondary electrons are simulated using different random numbers based on their position, energy and directions.
In radiation therapy, the Monte Carlo method for modeling dose calculations begins with simulations of millions of photon or electron transported in the head of the unit. The radiation output characterizing the beam in forward and lateral directions is stored in a space located below the head of the treatment unit known as the phase space. The beam output also carries information about the scattering component resulting from photon interaction in different materials located in the head of the unit. To increase the accuracy of the simulation, the geometry and material composition such as density of the components (radiation source and collimators) in the head of the unit must also be known. The photons stored in the phase space (virtual space containing photon information) after simulation of photon transport in the head of the treatment unit are then used as a source of simulation in the patient.

The Monte Carlo is a pure model-based method and is considered the standard for other dose calculation methods in radiation therapy. The Monte Carlo method has been shown to compute dose accurately, even in regions of different densities such as the interface between tissues, where other methods have inadequate accuracy [C. X. Yu, 1995]. In the past, this dose calculation method has been limited by the capacity of computer power to handle large numbers of photon histories required to reduce variations in dose calculations. But, with the availability of powerful computers, the Monte Carlo method is a promising tool for dose calculations in radiation therapy. The optimization of transport parameters (electron and photon energy cutoff, maximum electron energy step size, and Kerma (kinetic energy released per unit mass) cutoff) can potentially minimize time without significant loss of accuracy.

The Monte Carlo dose calculation has shown good results in the simulation of Co-60 beam characterization [McKenzie, 1996; Sichani, 2004; Joshi, 2008]. In a study to calculate a delivered dose from the Co-60 beam, the difference between Monte Carlo calculations and BJR supplement
25 [McKenzie, 1996] percentage depth doses (PDDs) and scatter factors in different field sizes was less than 1% [Sichani, 2004]. The study by Joshi et al. [2008] to validate fan beam measurement of the ionization chamber of a Theratron 780C Co-60 unit using the Monte Carlo method showed a maximum dose difference of 2% for PDD curves and profiles. Dhanesar et al. [2012, 2013] used Monte Carlo method to evaluate dose delivered using Theratron 780C Co-60 unit. In the study to use the aperture superposition Monte Carlo model to calculate dose delivered by Co-60 unit, the results showed the agreement between the aperture superposition model calculations and film measurements to be better than 1.5% in the high dose regions and 3.7% in the lower dose regions [Dhanesar, 2012]. This suggests that the Monte Carlo methods could accurately be used to estimate dose calculations from the Co-60 beam for clinical purposes in the future.

2.3.2.2 Convolution/Superposition algorithms

Unlike the corrected based methods, convolution/superposition algorithms do not use water measured data to directly calculate the dose. Instead, the dose in the patient is pre-simulated by Monte Carlo from first principles of photon transport. Then, the convolution/superposition algorithms use the homogeneous water measurements (commissioning data) to modify the pre-calculated Monte Carlo data. These modifications are used to obtain beam specific properties such as photon fluence and energy spectrum of that specific treatment unit beam [Uwe Oelfke, 2006]. This process is known as beam data configuration (beam modeling) and is performed before the algorithm is used for dose calculations. It is during beam configuration that the primary and scattering photons are modeled.
The convolution/superposition method accounts for dose contribution from both the primary and scattered photons. It computes the energy released by the primary photon at each interaction point, also referred as the total energy released per unit mass (TERMA). The TERMA at the interaction point is released locally to secondary electrons. The secondary electrons deposit the energy in a medium through assumed spatially invariant dose kernels (all dose kernels are assumed equal regardless the position of the primary photon interaction in the medium). As will be described in greater details in the next chapter, the convolution/superposition method calculates the dose deposited, \( D(\vec{r}) \) at a point in the medium as the convolution of the TERMA \( T(\vec{r}', E') \) at the primary photon interaction and dose kernels, \( k(\vec{r}', \vec{r}, E) \), as shown in Eqn. Error! Reference source not found.:

\[
D(\vec{r}) = \int dE \int d^2r' \ T(\vec{r}', E') \times k(\vec{r}', E'), \quad [2.10]
\]

For a photon beam with a spectrum of different energy components, the total dose at a point is calculated by superimposing the contributions from all dose kernels originating from the primary photon interactions of the energy components in a medium. Then, the total dose is the summation of all the points in the volume of interest, also referred as superposition method.

The convolution/Superposition method considers the effects of modifiers, patient external contours, and internal tissue inhomogeneity. This is achieved through Monte Carlo pre-computed dose kernels in water. The kernels are scaled according to electron densities of different tissues in the patient, which also facilitates dose calculations in regions of electron disequilibrium. To ensure an accurate TERMA calculation, the primary photon pathway is traced and scaled by the ratio of the electron density in water to that of the medium [Uwe Oelfke, 2006]. The convolution/Superposition methods are limited with long computation times that are irrelevant in clinical situations. Several methods including the pencil beam collapsed cone have been used to
reduce the computation time to fit clinical applications. The pencil beam methods are reviewed in
detail in Chapter 3.
3

Theory

The objective of this chapter is to provide an understanding of the main theoretical concepts used in this thesis. The first section explains the fundamentals of the radiation dose absorption in a medium. In the second section, the basics of dose calculation algorithms in the treatment planning systems (TPS) are highlighted. It provides general aspects of determining a dose distribution in a patient with a particular example of the model-based calculation engine using the pencil beam algorithm. Finally, a description of the fundamental principles of the Analytical Anisotropic Algorithm (AAA) in the Eclipse TPS is provided.

3.1 Radiation Dose

Radiations with sufficient energy to eject electrons from atoms in a medium with the resulting ionization and excitation of the atoms along their tracks are known as ionizing radiation. Ionizing radiation is either directly ionizing or indirectly ionizing. Directly ionizing occurs when charged particles (electrons, protons, alpha particles and heavy ions) directly interact and deposit energy in the medium. Indirectly ionizing radiation, involves uncharged particles (photons and neutrons) that first transfer their energy to charged particles, which in turn deposit the energy in the medium. The amount of energy deposited in the medium per unit mass is known as the absorbed dose. Absorbed dose is measured in units of Gray (Gy), defined as 1 Gy equals to 1 Joule/kg.
3.1.1 Photon Interactions

When a photon passes through a medium there is a possibility that it may go through the medium without interacting at all or it may interact. A fraction of photons attenuated in a beam as it passes through an absorbing material is expressed by the Beer Lambert law, Eqn.[3.1]:

$$\mu N(x) = -\left(\frac{dN(x)}{dx}\right),$$  \[3.1\]

where $\mu$ is the linear attenuation coefficient of the medium. Solving Eqn. [3.1], gives the following solution:

$$N(x) = N_0 e^{-\mu x},$$  \[3.2\]

where $N_0$ is initial number of photons in the beam, $N(x)$ is the number of transmitted photons after thickness $x$. Generally, $\mu$ depends on the photon energy ($E$) and atomic density of the medium ($\rho$). The probability of a photon interaction in a medium can also be given by the mass attenuation coefficient ($\frac{\mu}{\rho}$), which normalizes the attenuation coefficient to eliminate density dependence. The mass attenuation coefficient has units in $\text{m}^2/\text{kg}$.

There are three main interaction mechanisms for attenuation and dose deposition by photons from the Cobalt-60 (Co-60) beam in the medium, namely the photoelectric effect, Compton (Incoherent) scattering and pair production. These interactions enable the transfer of part or all of the incident photon energy to electrons in the medium. Note that coherent scattering, and photodisintegration interactions are not considered in this thesis. This is because coherent scattering does not contribute to any absorbed dose when interacting in the medium, while the energies for Co-60 photons are lower than the threshold for photodisintegration interaction to occur. Considering the interactions of interest there are:
• **The Photoelectric effect (τ):** The photoelectric effect occurs when a photon interacts and transfers all its energy \((h\nu)\) to an orbital bound electron of an atom. A portion of the initial energy of the incident photon is used to overcome the binding energy, \((E_B)\) of the electron. The rest of the incident photon energy is transferred as the kinetic energy of the ejected photo-electron \((E_K)\), which leaves at an angle \(\theta\), relative to the direction of the incident photon. For the photoelectric effect to occur, the photon energy has to be greater that the binding energy of the photoelectron. The electron kinetic energy is given by Eqn. [3.3]:

\[
E_K = h\nu - E_B. 
\]  

[3.3]

The mass attenuation coefficient \((\tau/\rho)\) of the photoelectric effect is approximately proportional to the third power of the atomic number of the material \((Z^3)\).

• **Compton (Incoherent) scattering \((\sigma_{inc})\):** Compton scattering involves interaction of a photon with a free electron. In the interaction, the photon transfers part of its energy as the kinetic energy to the electron while the incidence photon is deflected through an angle \(\theta\) with respect to initial track of the incident photon. The energy distribution is described by:

\[
h\nu' = \frac{h\nu}{1 + \frac{h\nu}{m_ec^2}(1 - \cos\theta)}.
\]  

[3.4]

where \(h\nu\) is the incident photon energy, \(h\nu'\) is energy of the scattering photon, \(m_e c^2\) denotes the rest mass of the electron and \(\theta\) is the angle of the scattered photon. The energy of the scattering photon depends on its scattering angle. For instance, if the collision is direct, the electron gains the maximum transferable energy from the photon
and travels straight forward at 0° while the photon scatters backward with a minimum energy at 180°. Alternatively, if the photon scatters nearly straight forward with the maximum energy at 0° then the electron would travel with minimum energy at 90°.

In the Compton interaction, the electron is most likely to be absorbed locally in the medium while the photon may escape from the medium or further interact with other electrons before losing its entire energy. The mass attenuation coefficient \((\sigma_c/\rho)\) of the Compton scattering is nearly independent of Z. Compton scattering interactions are the dominant interactions in soft tissue for the beam energy between 200 keV and 2 MeV [Fidarova, 2017].

- **Pair production** (κ): Pair production occurs when a photon at incident energy above 1.022 MeV interacts in the field of an atomic nucleus and forms a positron-electron pair. Since the rest mass of an electron is 0.511 MeV, therefore, the threshold energy is used to overcome the rest mass of the two particles. The extra energy beyond the 1.022 MeV threshold of the pair production is shared between the two particles as kinetic energies. The positron travels until it comes to rest and annihilate with another electron in the medium to produce two photons each with 0.511 MeV of energy moving in opposite directions (180°) to each other. The energy distribution of the pair production is given by Eqn. [3.5]:

\[
hv = 2m_e c^2 + T_{e^-} + T_{e^+}, \quad [3.5]
\]

where \(2m_e c^2\) is rest mass of the particles, \(T_{e^-}\) is electron energy and \(T_{e^+}\) is positron energy. The mass attenuation coefficient \((\kappa/\rho)\) of pair production is proportional to Z.
Figure 3.1: Three main interactions in radiation therapy with their regions of relative dominance. The curves represent the points with equal probability for the two adjacent interactions. Adapted from [Podgorsak, 2005].

Figure 3.1 shows each of these interactions and the different photon energies at which each dominates. With average energy of 1.25 MeV, it is clear that the most dominant interaction for the Co-60 beam is Compton scattering.

3.2 Dose Calculations in the Treatment Planning Systems

The calculation of dose for a particular photon beam from a treatment unit due to different interactions in the various tissues in a patient is a complex process. As described in Chapter 2, various dose calculation algorithms have been introduced and incorporated in treatment planning systems for clinical application. The main challenge for these dose calculation algorithms in the TPS is balancing speed and accuracy. The correction-based algorithms are fast but they have poor accuracy. Alternatively, the model-based algorithms have improved accuracy but they are slow. To improve the model-based algorithms, various methods have been implemented to achieve
acceptable speed while maintaining accuracy in clinical applications. Consequently, most of the current TPS systems have adopted model-based algorithms for the dose calculations. This section reviews the theoretical aspects of these dose calculation algorithms.

### 3.2.1 Dose Calculation Algorithms

As mentioned in Chapter 2, the dose calculations by convolution/superposition algorithms involves convolution of the total energy released per unit mass (TERMA) by primary photons at the point of interaction $\vec{r}$ in a medium and dose kernels. TERMA at interaction point is calculated using Eqn. 3.6:

$$T(\vec{r}) = \frac{\mu}{\rho}(\vec{r})\phi(\vec{r}),$$

where $\phi(\vec{r})$ and $\frac{\mu}{\rho}(\vec{r})$ are the energy fluence and the mass attenuation coefficient respectively at point of primary photon interaction $\vec{r}$ in the medium. The primary photon deposit energy in the medium through dose kernels $k(\vec{r}', r, E)$, which represent the spatial dose deposited at point $r'$ from energy released at interaction point $r$ upstream.

Dose kernels are categorized depending on the model of the beam element delivering the incidence photon at a point of interaction in the medium. For instance, a point kernel [Mackie, 1985], describes the dose deposition in an infinite medium around a primary photon interaction. A pencil beam kernel [Mohan, 1986] represents energy deposition into semi-infinite medium from a point in a mono-directional beam. A point spread kernel [Ahnesjö, 1987] defines the energy spread from a primary photon interaction in infinite medium. Figure 3.2 show different types of the dose kernels.
Figure 3.2: Dose kernels (a) the point kernel describing dose deposition around the primary photon interaction point, (b) the pencil beam representing the primary photon dose deposition in one direction of the beam, (c) the point spread function defining dose deposition from the primary photon in infinite medium. Adapted from [Papanikolaou, 2009].

Although the dose kernels can be derived from direct standard measurements, it is difficult to estimate multiple scattered photons with an empirical method [O’Connor, 1989]. Therefore, the dose kernels are obtained through Monte Carlo simulations [Mohan, 1986; Ahnesjö, 1989]. The dose kernels determined by Montel Carlo simulation are computed for monoenergetic beams in homogeneous water [Ahnesjö, 1987]. It is assumed that the dose kernels are equal regardless the position of the primary interaction (spatially invariant). For polyenergetic beams, the dose kernels are computed for each energy element in the beam and then combined with their appropriate weights in the beam [Fidarova, 2017].

3.2.2 Pencil Beam Convolution

The dose deposited in a 3D volume element (voxel) in a medium is a convolution of all dose kernels originating from different interaction voxels of the primary and scattering photons in the
surrounding medium. However, the convolution calculation using kernels from all the voxel in a patient is slow in clinical situations. One way to reduce calculation time is to reduce the dimensions of the convolutions using a pencil beam convolution [Mohan, 1986]. The pencil beam convolution uses 2D kernel convolution, which increases computation speed with acceptable accuracy.

![Figure 3.3: Dose distribution (pink) due to dose kernels around the primary interaction of the pencil beam.](image)

The pencil beam convolution divides the radiation beam from a treatment unit into small beams (pencil size). Figure 3.3 shows the dose distribution in the medium due to pencil beams. As previously mentioned, these kernels are computed in a homogeneous medium and the secondary electrons are assumed to travel on a straight line between interaction and deposition points. This allows integration of kernels accounting only for their distances and angles from the central axis of the pencil beam [Mackie, 1996]. This 2D integral of TERMA and pencil beam convolution is given by:
\[ D(\vec{r}) = \int dE \int d^2\vec{r}' T(\vec{r}, E) \times k_{pb}(\vec{r}', E'), \]  

where \( T(\vec{r}, E) \) is TERMA at interaction point \( \vec{r} \) due to photon of energy \( E \), and \( k_{pb}(\vec{r}', E') \) is a dose kernel of secondary electrons with energy \( E' \) at a distance \( \vec{r}' \) from the central axis of the pencil beam to the dose deposition point.

### 3.2.2.1 Tissue Inhomogeneity Consideration

Since the medium that comprises a patient is not homogeneous, the dose calculation assuming that dose kernels are spatially invariant does not hold. Therefore, both TERMA and dose kernel computations have to be scaled to account for tissue inhomogeneity. If the surface of the phantom is located at \( r_0 \) (Figure 3.4), then TERMA for monoenergetic beams is given by:

\[ T(\vec{r}, E) = \int \frac{\mu}{\rho}(E) \varphi_n(r_0, E) \times e^{-\mu|r-r_0|} dE, \]  

where, \( \frac{\mu}{\rho}(E) \) is the mass attenuation coefficient which determines the modification of the energy fluence as the photon beam passes through the medium. \( \varphi_n(r_0, E) \) is the energy fluence at the surface of the medium, and \( e^{-\mu|r-r_0|} \) accounts for the attenuation of energy fluence between the surface of the medium and the interaction point.

To account for the inhomogeneity during the computation of the dose kernel, a density scaling method is applied [Mohan, 1986]. Density scaling uses scaled radiological distances between the interaction and deposition voxels with the radiological distance determined by multiplying the distance between the two voxels by the effective electron density between the two voxels.
Figure 3.4: The dose distribution in homogeneous water (dashed) and the dose distribution after inhomogeneity scaling (hard).

Figure 3.4 shows the effective density calculation. Density scaling assumes that the energy loss of the secondary electrons depends on the effective (average) density between the two points,

$$k_{pb}(\vec{r}', E') = \int_{\vec{r}}^{\vec{r}'} k_{pb}(\rho_{eff} \times (\vec{r}' - \vec{r}), E') \, d\vec{r'},$$  \[3.9\]

where $\rho_{eff}$ is the effective density, and $\vec{r}' - \vec{r}$ is the distance between the interaction and deposition voxels. The total dose is calculated by convolution over all the voxels and for all energy components according to Eqn.[3.10]:

$$D(\vec{r}') = \int d^2r' \ T(\vec{r}, E) \times \frac{\rho_w}{\rho_r} \times k_{pb}(\rho_{eff} \times (\vec{r}' - \vec{r}), E'),$$  \[3.10\]

with $\rho_r$ the electron density at the interaction point, and $\rho_w$ the density of water; $\frac{\rho_w}{\rho_r}$ accounts for inhomogeneity in TERMA computation.
3.3 Analytical Anisotropic Algorithm for Photon Dose Calculations

The analytical anisotropic algorithm (AAA) [Ulmer, 1995, 1996, 1997; Sievinen, 2005] for photon dose calculation is a model-based algorithm that uses pencil beam convolution to provide a 3D absorbed dose in a patient. The AAA algorithm uses Gaussian functions to model the fundamental physical parameters of the beam, which increase dose computation speeds through analytical convolution [Ulmer, 2003]. The AAA algorithm is divided into two steps: beam modeling and dose calculation. These two steps are reviewed in the following sections. Although the AAA algorithm in this thesis is used for dose calculations from the Co-60 beam, it is primarily used for dose calculations from linac beams.

3.3.1 Beam Modeling

In contrast to the pencil beam algorithm that treats all dose deposition as a result of scattered photons [Mohan, 1986], the AAA divides a photon beam into multiple-source models including the primary photon (primary source), extra-focal photon energy (known as second source in Eclipse TPS) and electron contamination [Sievinen, 2005]. Figure 3.5 shows the typical treatment components and interactions in the head of the treatment unit and in the treated volume used to model a linear accelerator for AAA dose calculation. Prior to dose calculation, all the beam components need to be modeled to characterize their physical parameters such as fluence and energy spectrum from the beam of the treatment unit. These parameters are pre-calculated by Monte Carlo simulations and are then modified by water phantom measurements through beam modeling process (beam configuration in Eclipse TPS) to match the particular treatment unit [Sievinen, 2005].
Figure 3.5: A typical components used to model a linac in the AAA dose calculation method named right and the three radiation models named left. Bottom shows interactions leading to dose deposition in the treatment volume.

The primary source in the Eclipse AAA model represents the photons that come directly from the source to a patient without prior interaction in the head of the treatment unit. The scattering photons from the flattening filter and primary collimator are modeled by the extra-focal source, which is known as the second source in the Eclipse TPS. The second source contributes more dose in the penumbra and outside the beam field because of the lower position of the flattening filter as shown in Fig. 3.5 [Sievinen, 2005]. The electron contamination models electrons produced following photon interaction in the head. It describes the relationship between the electrons in the beam with depth. Since electrons have short range in the patient, most of the dose from the electron contamination is deposited at the beam entrance.

The modified parameters (photon energy spectrum, mean radial energy and fluence intensity) of the three radiation source models are stored in a virtual space located below the head of the treatment unit; this is referred to as the phase space and is retrieved for patient dose calculation when that particular treatment unit is used [Sievinen, 2005]. The photon energy spectrum
represents energy components in the beam. The mean radial energy describes the mean energy across the beam and is used by AAA to determine the beam hardening effect for linacs. The fluence intensity represents the distribution of photons across the beam (number of photons multiplied by energy).

3.3.2 AAA Dose Calculations

The final beam modeling for the AAA algorithm enables dose calculations through separate convolutions of the primary photon, scattered photon and electron contamination in the beam. The broad beam from the treatment unit is convoluted through small finite-sized beamlets (pencil beams). The algorithm divides the region of interest in a patient into 3D voxels depending on a selected grid size. Each voxel is aligned with the divergence of the beam and contains electron density information from the CT image of the patient. Figure 3.6 describes the beam and patient coordinates as referred to in the AAA method. The calculation point is represented in two coordinate systems: the patient coordinates ($\tilde{x}, \tilde{y}, \tilde{z}$) and beamlet coordinate system ($x, y, z$). The depth-dependent functions are defined in the $z$ direction on the central axis of the beamlet while lateral scattering is defined in the spherical shell perpendicular to the central axis of the beamlet.

3.3.2.1 Primary and Scattering Photon Calculations

In the Eclipse AAA model, both the primary and scattering photon components of the beam are calculated similarly, but with different spectral composition and position and size of the focal spot. The AAA method makes convolution in terms of energy deposited in a voxel, which allows more accurate energy conservation especially in regions of inhomogeneity compared to dose
Figure 3.6: The representation of the calculation point according to both beamlet entrance point and the patient calculation point from the center of the beamlet. *From [Sievinen, 2005] with permission.*

convolutions [Ulmer, 1995; Sievinen, 2005]. The primary photon attenuation in a patient (TERMA) is represented by the energy deposition function $T_\beta(z, \rho)$ while the lateral energy defining dose kernels are given as $K_\beta(x, y, z, \rho)$. The energy deposition function defines the energy deposition over an area of the spherical surface of the pencil beam [Ulmer, 1995; Sievinen, 2005]. To account for tissue inhomogeneity, the energy deposition function $T_\beta(z, \rho)$ is scaled by an electron density factor so that:

$$T_\beta(z, \rho) = T_\beta(z') \times \frac{\rho(x, y, z)}{\rho_{water}}$$ \hspace{1cm} [3.11]

where $\rho$ is the electron density and $z'$ is the radiological scaling depth between the entrance point and the deposition point. The scaling depth is determined by:

$$z' = \int_0^z \frac{\rho(x, y, t)}{\rho_{water}} dt.$$ \hspace{1cm} [3.12]
To account for inhomogeneity in the lateral scattering term, the dose kernel is also scaled by the electron density factor:

\[
K_{\beta}(x,y,z,\rho) = \frac{\rho_{\text{water}}}{\rho(x,y,z)} \times K_{\beta}(x,y,z',\rho)
\]  

[3.13]

In Eqn.[3.13], the kernel with \( z' \) corrects for inhomogeneity between the beamlet entry point and the depth of the calculation point. The kernel with \( z \) corrects inhomogeneity between the central axis of the beamlet and the lateral calculation point. The total energy distribution at the point of interest in a patient is given by:

\[
E_{p,\beta}(\bar{x},\bar{y},\bar{z}) = \Phi_{\beta} \times T_{\beta}(z,\rho) \times \int \int_{(u,v) \in \text{Area}(\beta)} K_{\beta}(u - x, v - y, z, \rho) \, du \, dv.
\]  

[3.14]

In the convolution, the photon fluence \( \Phi_{\beta} \) is assumed to be uniform over the cross section of the beamlet \( \beta \). The coordinate point \((u,v)\) is the point of calculation at depth \( z \) accounting for variations in photon fluence due to modification or in regions such as beam penumbra [Ulmer, 1996].

### 3.3.2.2 Electron Contamination Modeling

Electron contamination is modeled through an energy deposition function using electron contamination \( T_{c,\beta} \) and electron fluence \( \Phi_{c,\beta} \), which are assumed to be uniform over the cross-section of the beamlet, \( \beta \). The scaling for inhomogeneity in the energy deposition function of electron contamination \( T_{c,\beta} \) is similar to Eqn. [3.11]. The dose kernel of the electron contamination is modeled through a Gaussian distribution function given by;
\[ K_{c,\beta}(x, y, z) = \frac{1}{2\pi \sigma_E^2} \times \exp \left[ \frac{x^2 + y^2}{2\sigma_E^2} \right]. \]  

[3.15]

where \( \sigma_E \) is a constant derived from measured data [Ulmer, 1995, 1996]. The energy distribution due to electron contamination is given by;

\[ E_{c,\beta}(\tilde{x}, \tilde{y}, \tilde{z}) = \Phi_{c,\beta} \times T_{c,\beta}(z) \times \int \int \limits_{(u,v)\in Area(\beta)} K_{c,\beta}(u-x, v-y, z, \rho) \, du \, dv. \]  

[3.16]

3.3.2.3 Total Dose Calculation

The total absorbed energy is then simply the sum of the separate contributing energy deposition components from the primary \((E_{p,\beta})\) and second source \((E_{s,\beta})\) photons both given by Eqn.[3.14] and also from the electron contamination \((E_{c,\beta})\) given by Eqn.[3.16]. The total energy is expressed as:

\[ E(\tilde{x}, \tilde{y}, \tilde{z}) = \sum \limits_{\beta} E_{p,\beta}(\tilde{x}, \tilde{y}, \tilde{z}) + E_{s,\beta}(\tilde{x}, \tilde{y}, \tilde{z}) + E_{c,\beta}(\tilde{x}, \tilde{y}, \tilde{z}). \]  

[3.17]

The total energy is then converted to dose at the measurement point in the patient using the assumption that inhomogeneity can be modified by the relative density to water.

\[ D(\tilde{x}, \tilde{y}, \tilde{z}) = E(\tilde{x}, \tilde{y}, \tilde{z}) \times \frac{\rho_{water}}{\rho(\tilde{x}, \tilde{y}, \tilde{z})}. \]  

[3.18]
Materials and Methods

The description about the upgrade performed on the Theratron 780C (T780C) Co-60 unit teletherapy unit to an advanced Equinox-100 Co-60 unit is given in Chapter 1. This chapter describes the materials and methods used for the commissioning of the upgraded Equinox-100 unit in Eclipse treatment planning system (TPS) (Varian Medical Systems, Palo Alto, CA). First, the mechanical assessment is described. In the second section, the dose measurement procedures and materials used to acquire the dosimetric data are provided. Finally, the chapter reviews the Eclipse commissioning which involves two steps: first, the process of beam configuration to obtaining the appropriate parameters to enable Co-60 dose calculations; second, outlining the methods to validate whether Eclipse dose calculation could be used to predict Co-60 treatment delivery.

4.1 Mechanical assessments

The commissioning of the upgraded Equinox-100 unit involved the assessment of mechanical motion to establish the precision and accuracy of the treatment unit and patient support system during dose delivery. Various mechanical motions of the gantry, collimator and patient supporting system were tested according to AAPM Task Group 45 report [Nath, 1994] to ensure that their performance were within a clinically acceptable tolerances.

Mechanical axis of rotation: The mechanical axes of rotation for the gantry, collimator and couch with respect to the isocentre were checked against tolerance. For the gantry, the rigid rod pointer
Figure 4.1: Different methods used for the mechanical assessment of the upgraded Equinox-100: (a) the rigid rod used to assess the gantry and collimator axis of rotation, (b) the method used to test the rotation axis of the gantry, (c) film star shots for radiation isocentre determination and (d) film irradiation for light/radiation coincidence assessment.

provided by Best Theratronics to indicate the isocentre distance (when magnetically mounted to the head) was pointing at the cross-hairs center, then, the couch was moved back so the rigid rod pointer would not hit the couch when the gantry was rotated 360 degrees. A marker was placed at the end of the couch and its pointer touched the end of the rigid rod pointer at the isocentre. Next, the gantry was rotated and the separation between the two points was observed. For axes of rotation of the couch and collimator, the field size template was placed on the couch at 100 cm SSD. Both the field size template and the rigid rod pointer were placed at the center of the light field cross-hairs. The couch or collimator was rotated through a range of angles and observed if the rigid rod pointer leaves the circle with a radius greater than 1 mm on the field size template.
Radiation rotation isocentre: The gantry, collimator and couch rotation axes were determined by EBT3 Gafchromic film (Ashland Specialty Ingredients, Bridgewater, NJ, USA) irradiation (star shots) placed between plastic water phantoms (CIRS, USA). The irradiation was performed with the film set at the mechanical isocentre and marked using either the pin on the magnet mounted rod (gantry) or a marker dot at the cross-hairs. The jaws were set to give a narrow, symmetric fan beam coinciding with the respective axis (gantry/collimator/couch). The irradiation of the film was performed at different angles of the respective axis. The film was analyzed to find the intersection of the radiation beams and compare to marks on film indicating the mechanical isocentre of the gantry, collimator and couch.

Light/radiation coincidence: The EBT3 Gafchromic film was used to evaluate the light and radiation field coincidence. Films were irradiated at 100 cm SSD beneath the clear plastic phantom with an embedded grid of BBs. The field size was adjusted so that the light field edges lie on the grid edges of the BB phantom. The radiation field edge positions were compared to the BB shadow positions in the film to determine the relationship between light and radiation fields.

Other assessments: The accuracy of the angle read out at different angles of the gantry, collimator and couch was verified using a spirit level. To test the accuracy of the field size indicated by the upgraded Equinox-100, various field sizes were projected at the field size template and measured for verification. The accuracy of the optical indicator was tested over the range of its readout by projecting the indicator on the top of plastic phantom with different heights. Couch deflection was measured with a distributed load of 75 kg. Figure 4.2 depicts different methods and materials used for the mechanical assessment of the upgraded unit. These measurement methods were according to the AAPM Task Group 45 report [Nath, 1994] for clinical linacs.
4.2 Dose measurements

Dose measurements were performed to serve two purposes. Firstly, the pre-commissioning data were acquired to get initial data to commission the TPS. As mentioned in Chapter 3, beam data acquired in water phantom are used to establish physical parameters of the beam in the TPS model used to calculate the dose from that specific treatment unit. The type and amount of measurements for beam commissioning are reviewed in AAPM Report 106 [Das, 2008].

Secondly, the post-commissioning data were acquired under various clinical conditions to validate the TPS. The validation was performed in various clinical conditions and included point dose measurements, the one-dimension dose distribution at specific depth acquired through profiles and the two-dimensional (2D) dose distribution on a film. This section describes the process of dose measurement for both commissioning and validation.

4.2.1 Plastic Phantom Measurements

Prior to validation point dose measurements, a few absolute dose measurements were acquired to determine shutter time (transit time), which accounts for the finite time required for the Co-60 source to move from the fully shielded position to the fully exposed position. Shutter time correction was determined using the graphical method as proposed by Orton and Seibert [Orton, 1972]. Absolute dose validation measurements were performed to test the accuracy of the TPS model to estimate the absorbed dose at a point (0D measurements) in a patient for irradiations delivered by different beam configurations. For instance, treatments in radiation therapy can be performed while the gantry is stationary (static mode) or moving around a patient (arc mode). While in static mode, dose is delivered at a single gantry angle, whereas in arc mode the gantry angle changes while dose is being delivered. These beam configurations affect the dose
deposition of photons at a point relative to the gantry angle that the beam enters the patient. Although the model may easily predict dose accurately delivered in static mode with beams perpendicular to the surface of the patient, it gets complicated for oblique beams delivered at different gantry angles. This is due to the difference in the depth that photons must traverse to reach the point of measurement under oblique conditions. Also, under oblique geometry the estimation of the absorbed dose due to scattering lateral to the point of interaction is affected by the possibility of the scattering photons.

Point dose measurements were performed in a $30 \times 30 \times 12 \text{ cm}^3$ plastic water phantom with isocentre at 6 cm depth using a $0.6 \text{ cm}^3$ Farmer type chamber (PTW-Freiburg, Germany) connected to an electrometer Max 4000 (PTW-Freiburg, Germany) operating at -300V. The point dose validation of the 0D Eclipse calculations was performed using various square and rectangular field settings in static mode with single beams, two beam parallel opposed pairs (POP), and four-field boxes. For the arc mode, the dose was delivered with the gantry angle set to

Figure 4.2: Experimental setup for point dose measurements.
Figure 4.2 shows the experimental setup for these absolute dose measurements. Similar plans were created in Eclipse and the two results were compared.
Figure 4.3 presents 3D dose distributions generated in the Eclipse TPS during the point dose measurements. The dose distributions describe an advantage of using multiple beams in which the high dose is delivered to the target volume (Fig. 4.3 (a)). On the other hand, the arc beam delivers the high dose to small volume of the target (Fig. 4.3 (b)). From these dose distributions, it is evident that the Co-60 beam can be used to produce conformal dose deliveries if multiple or arc beams are used.

### 4.2.2 Blue Water Phantom Measurements

The beam commissioning data including percentage depth dose (PDD), profiles (in-line, cross-line and one diagonal) and relative dose factors (RDFs) were acquired with a scanned ion chamber in a water tank. The PDD measurements provide information to the TPS regarding beam penetration by defining the percentage dose at any depth relative to a fixed reference depth along the central axis of the beam (depth at which the maximum dose is given). The beam profiles describe the dose fall-off laterally from the central axis at set depths while the RDFs provide information about the beam output relative to a reference field size of 10 × 10 cm² at 100 cm SAD. The Eclipse requires that the commissioning beam data be acquired with field sizes ranging from 3 × 3 cm² to a maximum of 40 × 40 cm². The measurements from field sizes smaller than 3 × 3 cm² are not used by the configuration program even when included because they do not significantly affect the beam parameters calculated by Eclipse model [Torsti, 2013].

All the dose measurements were collected using a 48 × 48 × 41 cm³ Blue water phantom (IBA Dosimetry, Germany), which is commonly used for data acquisition required for a radiation therapy unit commissioning [Das, 2008]. Two CC13 (0.13 cm³) ionization chambers (IBA
Figure 4.4: The Blue Phantom water tank setup for various beam commissioning measurements.

Dosimetry, Germany) were used in the process of data acquisition; one as a reference and another as a field detector. The field ionization chamber moved through the water tank in a scan pattern programmed to acquire the data. It moved in either Y- or X-directions of the gantry in the water phantom according to the vendor specifications. The reference ionization chamber is usually placed in the radiation beam to remove any instantaneous fluctuations that occur in linac beam output during data acquisition. Although this is not needed for the Co-60 beam, the software does not allow the use of one ionization chamber to acquire measurements. The positioning of the reference ionization chamber was above the water tank at a beam edge to avoid obscuring the downstream beam measurement. The profiles were also acquired using a smaller CC01 (0.01 cm³) ionization chamber (IBA Dosimetry, Germany) in field sizes of 3 x 3 cm² and 10 × 10 cm² for comparison with the CC13 measurements. The field and reference ionization chambers were connected to a computer with OmniPro-Accept 7.4C software (IBA Dosimetry, Germany) to facilitate data acquisition. The OmniPro-Accept software enables fast step-by-step and continuous dose measurements. All PDD and profile measurements were acquired with a non-stop moving field ionization chamber (continuous acquisition mode).
Figure 4.5: (a) the experimental setup for oblique incidence measurements in the Blue phantom and (b) Different depths traversed by photons due to oblique geometry ($d_\parallel$ long to $d_\perp$ short depths).

The PDD measurements were measured along the central axis at SSD of 100 cm and normalized at depth of maximum dose (0.5 cm). The PDD curves were smoothed in Matlab (MathWorks, USA) and compared to PDD curves reported in BJR supplement 25 (BJR 25) [McKenzie, 1996] to ensure accuracy of the measurements before used for commissioning. The beam profiles of all the square field sizes were measured at different depths at SSD of 100 cm and normalized to the maximum dose at the central axis. Because of the size of the water tank with respect to field sizes $35 \times 35$ cm$^2$ and $40 \times 40$ cm$^2$, only half of the profiles were acquired in order to obtain the best characteristics of the dose fall-off at the edge of the field. The diagonal profiles of the largest field size ($40 \times 40$ cm$^2$) were also measured at different depths at SSD of 100 cm. The beam profiles in OmnPro-Accept software were used to quantify the beam penumbra of the upgraded Equinox-100. The RDF measurements of both square and rectangular field sizes were performed at a depth of 5 cm and normalized by the dose from a $10 \times 10$ cm$^2$ field size. Figure 4.4 show the measurement setup for the PDDs, beam profiles and RDFs.
To validate the Eclipse model dose calculations once the model had been commissioned, simple treatment plans were delivered to the phantom. Profiles that express the 1D dose distribution from the Co-60 beam were acquired using various clinical conditions. The accuracy of the Eclipse model to predict the dose distribution across the beam in homogenous water was investigated through measurements in square and rectangular open fields. In addition to the standard fields, the validation of the open fields also performed with oblique beams. The oblique beams incorporate different distances traveled by the photons to reach the same depth in a medium. As depicted in Fig. 4.5 (b), the variation of distances traveled by photons affects the dose deposition at that depth in the medium as opposed to perpendicular beams where photons travel approximately same distances. To test the accuracy of the Eclipse model to predict the dose deposition under oblique incidence, cross-line profiles at gantry angles 20° and 30° were acquired in a water phantom at a depth of 5 cm.

It is common in radiation therapy to modify the radiation beams using shielding materials to protect the healthy tissues that are in the vicinity of the tumour during treatments. These
modifiers change the photon fluence in the beam which affects the dose distribution in the
medium. To validate the accuracy of the Eclipse model to predict the delivered dose in these
modified fields, various beam profiles were acquired at 5 cm depth in a water tank using two lead
blocks of physical dimensions of $3 \times 2 \times 7$ cm$^3$ (Block1) and $8.5 \times 3 \times 7$ cm$^3$ (Block2) placed in
the beam. Figure 4.6 shows one of the blocks mounted on the block tray and placed in the beam
as used for the measurements. The blocks were mounted at the center of a block tray and then
placed in the block tray slot on the Co-60 unit head making the distance from the source to the
top of the block to be 54.6 cm. The transmission factors for the block tray and lead blocks were
0.93 and 0.01, respectively.

Finally, 1D dose distributions were obtained to validate against the Eclipse model for
inhomogeneity correction. Inhomogeneity validation tests the accuracy of the Eclipse model to
predict the dose in different tissues and between interfaces between tissues. Beam profiles were
acquired at depth of 6 cm in the water tank with two different cylindrical materials mimicking
patient inhomogeneities for bones and lungs. The 2.8 cm diameter inserts were placed in the
radiation beam along the central axis at 3 cm depth. Figure 4.7 shows the experimental set up for
inhomogeneity correction measurements.

**Figure 4.7:** The inhomogeneity correction measurement setup

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4.2.3 Film Measurements

The previous model validation methods in Secs 4.2.1 and 4.2.2 do not provide complete details to confirm Eclipse ability to calculate dose distributions in a patient. To better evaluate the clinical accuracy of the Eclipse model to predict dose distribution from the upgraded Equinox-100, simple treatment plans were delivered to a radiochromic film in the plastic phantom. Radiochromic film is one of the common 2D dosimetric methods used to validate treatment plans in radiation therapy. Its applications have been extensive even in high dose gradients plans such as intensity modulated radiotherapy (IMRT) [Kairn, 2011; Chung, 2013].

Radiochromic film is made of polyester materials that have an electron density equivalent to tissue. In this study Gafchromic EBT3 film (Ashland Specialty Ingredients, Bridgewater, NJ, USA) were used. When irradiated, the radiochromic film becomes darker. The amount of darkening is related to optical density which is defined as the \( \log_{10}(I_0/I) \) with \( I_0 \) as initial photon intensity and \( I \) as transmitted photon intensity. Since the relationship between optical density and dose is not linear, the film was calibrated to establish a relationship between the dose and optical density. For the purpose of the calibration, up to six film pieces were irradiated at known doses in plastic water phantom perpendicular to the beam in a 10 × 10 cm\(^2\) field size under reference conditions. The films were scanned with an EPSON 10000XL scanner (Epson Canada, Markham, Ontario) after 48 hours post irradiation, which allows film stabilization before being developed.

The Eclipse validation was performed for single treatment plan of a four-field box in 10 × 10 cm\(^2\) at 0° and 180° and 3 × 3 cm\(^2\) at 90° and 270° delivered on a single Gafchromic EBT3 film in the plastic water phantom described in Sec 4.2.1. The film pieces were then scanned using an EPSON
Table 4.1: Summary of all the measurements performed in this thesis, source to surface distance was set at 100 cm.

<table>
<thead>
<tr>
<th>Purpose</th>
<th>Measurement</th>
<th>Field Size (cm²)</th>
<th>Depth (cm²)</th>
<th>Gantry Angle (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Beam Data</td>
<td>PDD</td>
<td>3×3 to 40×40</td>
<td>0 to 30</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>Profiles</td>
<td>3×3 to 40×40</td>
<td>0.5, 5, 10, 15, 20, 25, 30</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>RDF</td>
<td>3×3 to 40×40</td>
<td>5</td>
<td>0</td>
</tr>
<tr>
<td>Validation</td>
<td>Absolute Dose:</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Static</td>
<td>10×10</td>
<td>6</td>
<td>0, 90, 180, 270</td>
</tr>
<tr>
<td></td>
<td>Arc</td>
<td>10×10</td>
<td>5</td>
<td>270-90, 90-270</td>
</tr>
<tr>
<td></td>
<td>Square Fields</td>
<td>3×3, 5×5, 10×10, 20×20</td>
<td>5</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>Rectangular Fields</td>
<td>20×3, 30×5</td>
<td>5</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>Oblique Beams</td>
<td>10×10</td>
<td>5</td>
<td>20, 30</td>
</tr>
<tr>
<td></td>
<td>Blocked Beams</td>
<td>8×12, 10×10, 15×15, 15×25</td>
<td>5</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>Inhomogeneities</td>
<td>10×10</td>
<td>6</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>Film</td>
<td>3×3 and 10×10</td>
<td>2</td>
<td>0, 90, 180, 270</td>
</tr>
</tbody>
</table>

10000XL scanner as previously described. Film calibration process to establish dose function used for dose determination is given later in this chapter. Similar plans were performed in Eclipse and profiles were generated and compared with the film measurements.

4.3 Beam Configuration

Beam configuration in Eclipse TPS model requires the adjustment and settings of particular physical parameters of the treatment unit to enable subsequent dose calculations for treatment plans. It involves entering measured beam data and calculating reproduced data in order to generate parameters specific for the treatment unit. All the water tank commissioning data including PDDs, profiles and RDFs were each imported from OmniPro-Accept software to their respective workspaces in Eclipse TPS and the beam were calculated to establish the modeling
parameters. Beam calculation was preceded by a primary calibration performed to establish the reference dose rate of the Co-60 source.

The beam configuration generates physical parameters of the primary and secondary photons of the beam. The primary and scattering photons in the Eclipse TPS are modeled as virtual primary and second sources respectively. The physical effect of the finite size in both X- and Y-directions of the primary source is modeled through an effective spot size (ESS). For linacs the ESS value ranges from 0 to 2.5 mm while for Co-60 source it ranges from 1 mm to 10 mm. The appropriate values of the ESS in collimator X- and Y-directions produces a good match of the measured and the model calculated profiles [Torsti, 2013]. To investigate the appropriate ESS parameters, profiles at ESS values of 4 mm to 8 mm were generated in Eclipse in a $10 \times 10$ cm$^2$ field sizes at the depths of 5 cm. These profiles were compared to measurements of the same field size and depth to obtain the ESS value that produces a profile in good agreement with measurement.

Modeling of the scattered photons was performed using open beam parameters which include the second source. While for linacs the second source models scattered photons primarily from the flattening filter and primary jaws, for a Co-60 unit the scattered contribution is essentially from the primary and secondary jaws in the head of the unit. The open beam parameters must be adjusted to achieve an acceptable match of the measured and Eclipse model calculated beam profiles especially in the penumbra and outside the field. One of the parameters that mostly affects the beam is the second source intensity (SSI), which is given as a percentage of the primary photon fluence. Table 4.2 shows the list of all open beam parameters in the Eclipse TPS. To achieve a suitable SSI, PDDs and profiles were generate with different SSI values to account for the amount of scattered photons in various field sizes. The generated profiles were compared to measurements and the appropriate values were selected.
Table 4.2: The default parameters calculated by the AAA configuration algorithm for the Co-60 beam.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Dimensions</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distance of second source from target (mm)</td>
<td>120</td>
</tr>
<tr>
<td>Distance from Y-jaw top surface from target (mm)</td>
<td>200</td>
</tr>
<tr>
<td>Distance from Y-jaw bottom surface from target (mm)</td>
<td>300</td>
</tr>
<tr>
<td>Distance from X-jaw top surface from target (mm)</td>
<td>300</td>
</tr>
<tr>
<td>Distance from Y-jaw bottom surface from target (mm)</td>
<td>400</td>
</tr>
<tr>
<td>Distance of MLC from target (mm)</td>
<td>510</td>
</tr>
<tr>
<td>Size of second source (mm)</td>
<td>23.3</td>
</tr>
<tr>
<td>Relative intensity of second source</td>
<td>0.035</td>
</tr>
<tr>
<td>Mean energy of second source (MeV)</td>
<td>0.83</td>
</tr>
</tbody>
</table>
Results and Discussion

This chapter presents the results and discussion of the measurements outlined in Chapter 4. The first section shows the results for the different measurements performed to commission the upgraded Equinox-100 unit. This section includes the mechanical assessment, determination of the shutter correction time and beam commissioning data. The second section describes the modeling of both primary and second source parameters for the Co-60 beam in the Eclipse treatment planning system (TPS). Finally, the chapter presents various validation results showing how well the Eclipse TPS predicts the patient delivered dose from the upgraded Equinox-100 unit through validation of delivery of various treatment plans.

Figure 5.1 shows a typical beam profile from water tank measurements provided as a guide of the different dose regions of the profile that are referred in this discussion. At the top of the profiles,
there is a high dose region which represents the dose from 100% to 90% of the dose at the central axis. The top corners in the high dose region will be identified as the shoulders. The penumbra region lies between 90% and 10% of the central axis dose. Below the penumbra is the foot of the profile which represents the region between the 10% and 2% dose of the central axis. The region outside the field stands for the dose region below 2% of the central axis dose. The analysis in this thesis is based on the local difference between the measured and calculated data, given by:

\[
\text{Difference(\%)} = \frac{Dose_{Eclipse} - Dose_{meas}}{Dose_{meas}} \times 100\%.
\]  

[5.1]

5.1 Commissioning of the Upgraded Co-60 Unit

Prior to the commissioning of the radiation beam in the Eclipse TPS a number of measurements were performed to determine the mechanical characteristics and integrity of the upgraded Equinox-100. This section presents the results for determining of the shutter correction time and the mechanical accuracy assessments of the upgraded Equinox-100 unit.

5.1.1 Shutter Time Correction

The shutter time corrects for the systematic error in effective treatment time required to deliver the prescribed dose from the Co-60 beam. Figure 5.2 shows the plot of measured output versus time settings used to determine the shutter time correction. Due to the stability of the Co-60 beam, the ion chamber standard deviations were very small and hence the error bars cannot be seen on the plot. From a linear regression fit of the data, the shutter time was determined to be -0.58 ± 0.08 seconds.
5.1.2 Mechanical Assessment

The mechanical commissioning of the Equinox-100 unit involved the tests described in Chapter 4 to ensure that the treatment unit meets the manufacturer’s specifications for accurate mechanical motions and machine performance. Such tests are performed for newly acquired or substantially
upgraded radiation treatment equipment before being put into clinical service. All the mechanical tests for consistency and accuracy were completed according to AAPM Task Group 142 report [Klein, 2009].

Table 5.1 shows the results of various mechanical tests performed for the upgraded Equinox-100 and their respective tolerance from the manufacturer as well as those conventionally adopted in commissioning of clinical linacs. Note, that the clinical linac criteria shown here are based on recommendations of the AAPM Task Group 142 report [Klein, 2009] for intensity modulated radiation therapy (IMRT) and stereotactic radiosurgery (SRS) or stereotactic body radiation therapy (SBRT). Since SRS or SBRT delivers high doses to small volumes, some of the tests for clinical linacs are shown with extra small tolerances. The Co-60 results showed that the gantry performances passed the entire manufacturer and clinical linac tolerance criteria. The observations that the Co-60 unit characteristics were with tolerances similar to those achieved with clinical linacs suggest that the gantry of the upgraded unit could accurately handle delivery techniques such as IMRT used on the linac. Similarly, the collimator parameters performed well regarding the manufacturer and clinical linac tolerances except for the light/radiation coincidence for SRS or SBRT (bolded in the table). This is expected given the large physical size of the Co-60 source that produces a penumbra beyond the field light even when the source and the light are equal distance from the isocentre. In this regard, the manufacturer’s criterion for light/radiation coincidence is 3 mm, which aims to accommodate the effect of the penumbra and hence the failure for the 1 mm linac criterion is expected. For the remainder of the mechanical tests the upgraded Co-60 collimators performed within the clinical linac criteria.
**Table 5.1:** Various mechanical tests performed for the upgrade Co 60 unit showing an excellent performance against the manufacturer and clinical linac tolerances.

<table>
<thead>
<tr>
<th>Test</th>
<th>Upgraded Equinox-100 Performance</th>
<th>Manufacturer Tolerance</th>
<th>Clinical Linac Tolerance [Klein, 2009]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gantry</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Angle Readout</td>
<td>0 ± 0.043°</td>
<td>0.5°</td>
<td>1°</td>
</tr>
<tr>
<td>Rotation Mechanical Isocentre</td>
<td>1 ± 0.5 mm</td>
<td>2 mm</td>
<td>1 mm</td>
</tr>
<tr>
<td>Rotation Radiation Isocentre Size</td>
<td>0.23 ± 0.5 mm</td>
<td>-</td>
<td>1 mm</td>
</tr>
<tr>
<td>Rotation Mechanical to Radiation Isocentre Distance</td>
<td>0.56 ± 0.5 mm</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Collimator</td>
<td>Rotation Mechanical Isocentre</td>
<td>&lt;1 ± 0.5 mm</td>
<td>1 mm</td>
</tr>
<tr>
<td>Angle</td>
<td>0.5 ± 0.043°</td>
<td>0.5°</td>
<td>1°</td>
</tr>
<tr>
<td>Field Size Readout</td>
<td>&lt;1 ± 0.5 mm</td>
<td>2 mm/Jaw 4 mm/Field</td>
<td>1 mm</td>
</tr>
<tr>
<td>Optical Distance Indicator</td>
<td>0 ± 0.5 mm at 100 cm 0 ± 0.5 mm at 80 cm 4 ± 0.5 mm at 120 cm</td>
<td>1 mm at 100 cm 2 mm at 80 cm 4 mm at 120 cm</td>
<td>1 mm at 100 cm 2 mm at end of range</td>
</tr>
<tr>
<td>Light/Radiation Coincidence</td>
<td>&lt;2 ± 0.5 mm (5, 10, 20 cm² fields)</td>
<td>3 mm</td>
<td>2 mm/1 mm</td>
</tr>
<tr>
<td>Rotation Radiation Isocentre Size</td>
<td>0.44 ± 0.5 mm</td>
<td>-</td>
<td>1 mm</td>
</tr>
<tr>
<td>Rotation Mechanical to Radiation Isocentre Distance</td>
<td>0.70 ± 0.5 mm</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Couch</td>
<td>Isocentric Angle</td>
<td>0.5 ± 0.043°</td>
<td>0.5°</td>
</tr>
<tr>
<td>Rotation Mechanical Isocentre</td>
<td>&lt;1 ± 0.5 mm</td>
<td>1 mm</td>
<td>1 mm</td>
</tr>
<tr>
<td>Positioning</td>
<td>&lt;1 ± 0.5 mm</td>
<td>1 mm</td>
<td>2 mm/1 mm</td>
</tr>
<tr>
<td>Rotation Radiation Isocentre Size</td>
<td>0.05 ± 0.5 mm</td>
<td>-</td>
<td>1 mm</td>
</tr>
<tr>
<td>Rotation Mechanical to Radiation Isocentre Distance</td>
<td>0.19 ± 0.5 mm</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
The couch of the upgraded Equinox-100 also performed within criteria recommended by manufacturer and expected for clinical linacs. This suggests that the 4 degrees of freedom (longitudinal, lateral, vertical and couch rotation angle) couch can be used to position a patient for treatment accurately. Currently, the upgraded Co-60 couch can be used for 3D conformal radiation therapy with a mobile portal imaging as it has been shown in previous work [Marsh, 2017]. Although, Co-60 megavoltage computed tomography (MVCT) has been shown to produce images with good visualization comparable to a 6 MV linac reliable for Co-60 tomotherapy [Schreiner, 2003], it’s applications require the incorporation of detectors on the Co-60 unit useful for IMRT delivery techniques. Additionally, the couch is limited to motion initiation from the control panels mounted on the couch or from the hand control. Thus, it is currently not possible to use the dose delivery techniques that require simultaneous couch and gantry translations during dose delivery.

5.1.3 Beam Commissioning Data

Before the measured data were used for beam commissioning, an assessment to evaluate their accuracy was performed by comparing the measurements and the available standard data. The PDDs were evaluated by comparing the depth dose measurements with the data from the BJR Supplement 25 (BJR 25) [McKenzie, 1996]. For the dose profiles, extra measurements acquired by a small volume ion chamber (CC01) with active volume 0.01 cm³ were compared with the measurements acquired by the CC13 ion chambers with active volume 0.13 cm³. The differences between the measurements and the BJR 25 or CC01 data are based on their local differences, in which the calculated dose represents the BJR and the CC01 ionization chamber measurements.
5.1.3.1 Measured Percentage Depth Dose

Figure 5.3 shows a comparison of the measured and BJR 25 PDD curves in different field sizes. The difference between the measured and BJR 25 PDD curves was less than 2% for depths up to 20 cm, with the ion chamber measuring lower doses than those reported in the BJR 25. At depths deeper than 20 cm the variation increases up to about 5% local difference. This behavior is common for all field sizes measured at the large depths where there are low doses. The variation is not expected considering the decrease of dose in these depths with the decreased signal-to-noise ratio.

Figure 5.3: The measured and BJR PDD curves acquired in different field sizes, the plots at the bottom shows increasing differences with depths.
5.1.3.2 Measured Dose Profiles

The size of the detector may affect the accuracy of the measurements. To check for the accuracy of the CC13 data, measurements using CC01 were acquired and compared. The results depicted in Fig. 5.4 show good agreements between the measurements acquired by the two ion chambers. The variation between them is less than 2% in the high dose region for both 3 × 3 cm² and 10 × 10 cm² field sizes. Good agreement is also observed in all regions of the penumbra except in the foot region of the dose profiles (i.e. dose between 10% and 2% of the central axis) in which the differences increase to about ±30% in both field sizes. This increase begins slowly at about 50% dose of the central axis, with the CC13 measuring slightly higher doses than the CC01. Notice however, the noise in the CC01 measurements that could be contributing to the higher differences observed in low dose regions such as the foot and outside the field. Nevertheless, these differences in the shaded regions suggest that the CC13 data are affected by the partial volume effect as opposed to CC01. The partial volume effect is caused by the volume averaging between the sensitive volume of the detector and the measured dose gradient [Wuerfel, 2013]. The CC13 with an internal diameter of 6 mm is most likely to result in volume averaging to measurements compared to the CC01 with an internal diameter of 2 mm. The large volume of the detector causes the low dose regions such as the penumbra to be overestimated and as a result, the profiles in this region broaden. The detector volume effect on measurements in the penumbra (80% - 20%) has shown that the width measured with a 6 mm diameter ion chamber such as the CC13 would be larger by 2 mm than the width measured by a 3 mm volume detector for the Co-60 beam [Dawson, 1985]. Since all the commissioning and some of the validation measurements used in this study were acquired by the CC13, it may contribute to disagreement in the low dose regions during validation.
Figure 5.4: The comparison of the measurements of the CC13 against the CC01, where the high variation in the foot of the profile is highlighted.
5.2 Establishing Co-60 model in Eclipse

As discussed previously, the modeling of the primary and second sources in the Eclipse model requires appropriate setup of an effective spot size (ESS) and a second source intensity (SSI) respectively. This section presents the results regarding the selection of the appropriate ESS and SSI required for modeling the Co-60 beam in the Eclipse TPS. Note that, all the Eclipse generated data in this thesis were calculated with a grid size (i.e. resolution) of 1.5 mm.

5.2.1 Modeling of the primary Co-60 source

The primary radiation source in the Eclipse Analytical Anisotropic Algorithm (AAA) model is not a point, but an ellipse with finite transverse (X) and longitudinal (Y) dimensions, which is referred to as an effective spot size (ESS) in the Eclipse model. As a reminder, the ESS values are not physical dimensions of the source rather the parameters that represent the physical dimensions of the radiation source of the treatment unit in the Eclipse TPS. Because of the small size of the linac targets, the size of the source model in the conventional clinical linac AAA settings in the Eclipse TPS can be set to values between 0 and 2.5 mm. But, owing to the physical size of the Co-60 source, the Co-60 AAA configuration of Eclipse allows the source dimensions to vary between 1 mm to 10 mm. This range is still small compared to the actual size of the Co-60 source and therefore, the Eclipse model may have limitations to predict the delivered dose in some regions of beams, especially for smaller field sizes.

Figure 5.5 shows the measured and various dose profiles generated with ESS values of 4 mm, 5 mm, 6 mm, 7 mm and 8 mm in both X and Y-directions. Note that when a single ESS value
Figure 5.5: The dose profiles of the measurements and the Eclipse model generated with different ESS values at 5 cm depth, where the variation of each ESS value with the measurements are shown at the bottom of the plot.

(i.e 2 mm) is used for X and Y, the source model is assumed circular with equal dimensions in the two directions. During the ESS selection, the SSI value to be used in the AAA Co-60 model was set to 3.5% for reasons explained in the next section. It can be observed that dose profiles generated with the ESS values of 4 mm and 5 mm overestimate the dose in the shoulder edges of the profiles, while underestimating the dose in the foot region. This suggests that the 4 mm and 5 mm are not appropriate ESS values to model the primary Co-60 source. The Eclipse model using a 6 mm ESS value gives data with good agreement to the measurements in the high dose region including the shoulders of the field. However, the AAA calculations using the 6 mm ESS gives lower doses than the measurements in the foot. While, the AAA calculations with the ESS of 7 mm and 8 mm give good agreement in the high dose region, they underestimate the dose at the shoulders and the penumbra of the field. Because the 6 mm ESS value the Eclipse calculations
agree better with the measurements in the high dose region in which a small variation in a clinical situation can cause a potential dose deviation, the 6 ESS is therefore considered to provide an appropriate model of the Co-60 source. At this stage the SSI value was set at 3.5%.

5.2.2 Modeling of the Second Source Intensity

The second source intensity (SSI) of the AAA algorithm models the fluence generated from a virtual source intended to account for photons scattered in the head of the unit. Figure 5.6 depicts the position of the second source in the AAA model. The dashed lines indicate the ideal beam edges defined by the field size if the source were a point. Since the radiation source is not a point, the sharpness of the field edges is lost due to divergence of the photons from the primary source (red). As defined previously, the distance from the dashed to the red line at any point in the medium is known as geometrical penumbra. As it can be observed, its size is naturally dependent on the size of the radiation source. For the Eclipse model, the penumbra is further extended because of the divergence of the photons (blue) coming from the second source which is positioned below the primary source. As mentioned before, the second source influence is largely depending on its intensity which in the Eclipse TPS ranges from 0 to 20% of the primary photon intensity. For a linac model, the Eclipse TPS is typically between 2% and 3.5%.

Figure 5.7 shows different percentage depth dose (PDD) profiles generated using a combination of the 6 mm ESS and various SSI values in different field sizes. As shown in a local difference curve at the bottom of the plot, there are small differences between the PDD curves generated with various SSI values. Note that the curves are similar to each other regardless of the field size and it can be seen that the differences (see the zoomed portion in the insert) of using the 3.5% and 15% in the 20 × 20 cm² (see Fig. 5.7 (c)) is only about 0.5%. This is not surprising based on the
Figure 5.6: The positions of both the primary and second sources in the Eclipse model are shown with the extended penumbra due to the position of the second source photons (blue).

The influence of the SSI values is more pronounced when looking at the dose profiles at the edges and outside of the beam (see Fig. 5.8). It can be observed that the measurements and Eclipse model calculations vary depending on SSI value as well as the field size. In the $3 \times 3$ cm$^2$ field size (see Fig. 5.8 (a)), all dose profiles are almost undistinguishable regardless of the SSI value used. This observation was expected because the estimated amount of scattering photons in small fields is small. As the field size increases, the influence of the SSI is observed especially in
Figure 5.7: The percentage depth dose profiles generated in various field sizes with different SSI values, the small variations to measurements are shown at the bottom of the plots.

In the $10 \times 10 \text{ cm}^2$ field size, the AAA algorithm is largely underestimating the dose in the foot and outside of the beam. However, the underestimation is significantly reduced when the Eclipse model uses large SSI values. Similar patterns are observed in the $20 \times 20 \text{ cm}^2$ (see Fig. 5.8 (c)), in which the Eclipse model predicted even more doses in the foot that affects the local difference relative to the SSI used. It is clear that the SSI value has a significant influence of the Eclipse TPS modeled dose in the foot of the profiles, and is dependent on the field size. Since the SSI values have negligible influence in the small field sizes, then the appropriate value can be chosen based on the agreement of the measured and Eclipse model calculated data in the foot of the large field size. While in open beams the large values of SSI do
Figure 5.8: The dose profiles generated in different field sizes at 5 cm depths with different SSI values, their respective local differences to measurements are shown at the bottom of the plots.

not affect the PDD and produce a good match of the profiles, it is however important to consider the presence of beam modifiers inside the beam during the Co-60 beam modeling. Usually beam modifiers are made of high density materials such as lead that naturally increase the amount of scattering photons in the beam due to photon interactions.

Figure 5.9 presents the measured and the Eclipse calculated profiles acquired with a lead block at the center of a 20 × 20 cm² field size. Notice that the effect of the SSI value on the Eclipse model is up to about 30%. Likewise in the penumbra and outside the field, the overestimation is clearly seen. This suggests that in the presence of beam modifiers, the use of large values of SSI
overestimates the dose under the block and in the foot of the beam. The percentage of the overestimation depends on the SSI value used. At higher SSI values, the Eclipse model predictions under the block and in the penumbra are suboptimal.

In summary of Sec 5.2, the Eclipse model parameters required to model the upgraded Co-60 beams have been established. The primary source with a 6 mm ESS provides a better model of the Co-60 beam and gives good agreements of the measured and the Eclipse calculated doses. It was also shown that the variation of the SSI values has no influence for the central axis dose. Although the large SSI values used in the model improve the agreement of the dose profiles in the foot, these values result in suboptimal Eclipse model estimations in the modified beams. Thus, for the rest of this thesis the combination of the 6 mm ESS and 3.5% SSI is used for the Eclipse model calculations.

Figure 5.9: The dose profiles generated at 5 cm depth in a blocked $20 \times 20 \text{ cm}^2$ field size with different SSI values, their differences to measurements are shown at the bottom of the plot.
5.3 Eclipse dose calculation

This section provides the results of the validation plans as described in Chapter 4. The first part of the section presents the results for relative 1D dose measurements including the percentage depth doses (PDDs), dose profiles, oblique beams, inhomogeneity corrections and blocked beams. The second part describes the point dose measurements in which absolute doses were measured and compared to Eclipse calculations in multiple beams and in various beam configurations reproducing situations used in clinics. The last part of this section focuses on a typical clinical delivered dose, displayed using a relative 2D dose distribution.

5.3.1 1D Measurements

The 1D-dose measurements include the percentage depth dose (PDD) and various dose profiles acquired in different beam configurations. These data are compared with the corresponding Eclipse TPS calculations performed in the same configurations.

5.3.1.1 Percentage Depth Dose

Figure 5.10 shows examples of the PDD curves of different square field sizes. The figures also show the local differences between the measured and the Eclipse TPS PDDs below the curves. In this study, the analysis for the buildup region (beam entrance region) is not performed because of the size of the ion chamber used to acquire the measurements.
Figure 5.10: The PDD curves in various square field sizes, the good agreements up to 20 cm depths are shown by the local differences at the bottom of each curve.

Figure 5.11: The PDD curves in rectangular field sizes, the variations are shown by the local differences at the bottom of each curve.
The region beyond the maximum dose in both the square and rectangular field sizes show an excellent agreement between the measured and Eclipse TPS calculated doses. The variation between the curves is less than 2% for depths of up to 20 cm, where the Eclipse predicts higher doses than the measurements. At depths beyond 20 cm, variations in both square and rectangular fields increase to about 4%. Notice that this is the same pattern shown when the measurements were compared to BJR 25 data and as previously mentioned the increased variation in the deeper depths is due to a decreased dose, leading to an increased noise-to-signal ratio. The recommended clinical criteria for the dose along central axis is less than 2% which was achieved [Fraass, 1998; Venselaar, 2001; IAEA, 2004]. Although the agreement of ±2% for the depths beyond the maximum dose is not as good as ±1% reported for the linac beams [Fogliata, 2006; Van Esch, 2006], the agreement observed suggests that the Eclipse model can accurately predict the delivered dose of the Co-60 beam along the central axis.

5.3.1.2 Dose Profiles

Figure 5.12 shows dose profiles of the measurements compared to the Eclipse TPS calculations in different open square fields at 5 cm depths. The dose profiles show an excellent agreement in the high dose region of the beams with the average variation of less than ±2% in all cases. Similar agreements are observed in the high dose region of the elongated rectangular fields (see Fig. 5.13). These results correspond to the recommended clinical criteria for dose agreements in the high dose region of 2%.

However, the agreement is suboptimal in the penumbra of the small fields (see Figs. 5.12 (a) and (b)). The Eclipse model underestimates the dose up to -40% with the highest distance to agreement being 2.2 mm in the penumbra of the 3 × 3 cm² field size. But, the agreement does
improve as the field size increases. As shown Fig. 5.12 (b), a maximum difference in the $5 \times 5$ cm$^2$ field size is -35% with a highest distance to agreement of 2 mm. In the large field sizes (see Figs. 5.12 (c) and (d)), the agreement improves to less than -20% or distance to agreement of 1.2 mm in the $20 \times 20$ cm$^2$. The agreement is good in the rectangular fields shown in Fig. 5.13. The maximum variation for the short sides is observed to be about -20% with a distance to agreement of 1.6 mm in the short side of the $20 \times 3$ cm$^2$ (see Figs. 5.13 (a)). In the long sides, the maximum difference is observed to be about -20% with a distance to agreement of less than 1 mm. The clinical recommended criteria in the penumbra region is 10% or 2 mm [Fraass, 1998; Venselaar, 2001].

The maximum variation between the measurements and the AAA model calculations is observed in the foot (region between 10% and 2%) of the profiles. The local difference in this region is up to -60% for the $3 \times 3$ cm$^2$ and -40% for the $5 \times 5$ cm$^2$ field size. The local variation improves for the region outside the beams. Although, the local differences provide the variations of the data happening at each point in the beam, the clinical interest however, is the variation in relation to the dose at the central axis (global variation). The global variations are recommended for comparison of doses in low dose regions where the accuracy of the measurements is inadequate [Venselaar, 2001], the calculation is given by:

$$\text{Difference(\%)} = \frac{Dose_{\text{Eclipse}} - Dose_{\text{meas}}}{Dose_{\text{meas,CA}} \times 100\%}. \quad [5.2]$$

The global variations are shown in the lower plots in each quadrant of Fig. 5.14. The maximum variation is about -4% observed in the foot region of the small field sizes. The agreement improves to less than 2% further outside the field. Although, the variations in the foot and outside the beam for the rectangular fields are not calculated globally, they are assumed to follow the same distribution as the square fields. The suggested clinical criteria for the region outside the field are 3%.
Figure 5.12: Dose profiles in various square field sizes at 5 cm depths, where the shaded regions indicate the area with high differences between the curves shown at the bottom.

Figure 5.13: Dose profiles in rectangular field sizes acquired at 5 cm depths, where the differences between the curves is shown at the bottom.
As mentioned earlier, the problem of the agreement in the edges of the profiles is primarily due to geometrical size of the Co-60 source. However, the influence of the Eclipse modeled scattered dose based on the field size is observed. For example, in the small field size in which the amount of modeled scattering photons is small, the variation is very pronounced. As the field size increases, the amount of the Eclipse modeled scattered dose also increases, which leads to improved agreement in the region. Although the agreement of the Eclipse Co-60 calculated to the measured data is not as observed with linacs, these observations suggest that the Eclipse TPS can be used to predict the dose of the Co-60 beam for large field sizes greater or equal to 5 × 5 cm². More quantification using small volume detectors may be required to make a conclusive judgement on the application of the Co-60 beam in the small field sizes.

Figure 5.14: Dose profile in various field sizes at 5 cm depths, where the global variations are shown at the bottom of each plot.
5.3.1.3 Oblique incidence validation

The 1D measurements were performed to test the accuracy of the Eclipse TPS to estimate the dose delivered by Co-60 oblique beams. The results are presented in Fig. 5.15. There is an excellent agreement of the measured and the Eclipse model estimated profiles in the high dose region. The local difference in the high dose region of both beams at both $20^\circ$ and $30^\circ$ gantry angles is less than 2%. This agreement corresponds to recommended clinical criteria of 2%, which indicates that the AAA algorithm can accurately predict the delivered dose in the high dose region of the oblique beams from the Co-60 beam.

Similar variations observed in the standard normal beams are also seen in the penumbra region for the oblique delivery. The only difference is that the Eclipse model underestimation in oblique beams starts at about 90% dose of the central axis dose in both beams at $20^\circ$ and $30^\circ$ gantry angles. The variation is up to a maximum of -35% observed in the far side of the penumbra (outer penumbra) in the oblique beam at the $20^\circ$ gantry angle. A maximum distance to agreement is about 2 mm up to 20% dose of the central axis, but increases to about 8 mm at 10% dose of the central axis. This variation is because the outer side of the beam in the phantom is moved further away from the normalization point, causing lower doses compared to the same region in beams at $0^\circ$. This is evident when the variations are calculated globally, as the maximum differences between the measurements and the Eclipse model estimations is less than 10%. Also, note that the measured distances are normal to the obliquity which results in large distances observed.

On the other hand, the AAA algorithm overestimates the dose delivered in the inner side of the beam (inner penumbra). The maximum variation is about 2.6% and 5.8% in the inner penumbra of the beam at the $20^\circ$ and $30^\circ$ gantry angles respectively. The overestimation in the inner penumbra is caused by the limitation of the AAA algorithm to account for the reduced dose due
Figure 5.15: Oblique profiles acquired in a $10 \times 10$ cm\(^2\) field sizes at 5 cm depths, both local and global variations are shown at the bottom the plots.

to distances traverse in the phantom. The agreement outside the beam is as good as it is less than 2\% (shown by the green line at the bottom of each plot). Generally, the AAA algorithm can accurately predict the delivered dose using Co-60 oblique beams. The clinical criteria recommended for the penumbra in the oblique beams is 15\% or a distance to agreement of 3 mm [IAEA, 2004].
5.3.1.4 Inhomogeneity validation

A further test of the integrity of the Eclipse dose calculations was performed for inhomogeneity corrections. The Eclipse model was validated through calculations in lung density equivalent materials to represent low density tissues and in bone equivalent materials to represent high density tissues. Figure 5.16 shows the measured profiles acquired with lung and bone inhomogeneities in the beam against the Eclipse calculated profiles. There is a good agreement between the measured and Eclipse TPS calculated doses in the lung inhomogeneity as shown in Fig. 5.16 (a). A maximum variation is observed to be less than 1%. However, the Eclipse tends to overestimate the dose near the lung inhomogeneity up to 4.7%. The clinical criteria near the lung is 3% [Venselaar, 2001; IAEA, 2004]. The approximation in the inhomogeneity is related to the ability of the AAA algorithm to accurately estimate the depth-dependent (primary photons) and lateral scatter (secondary electrons) components of the photon beam. It seems that the AAA photon algorithm has a poor estimation of the lateral scattered photons after interaction of the primary photons in the lung inhomogeneity. This incorrect estimation of the dose near the lung inhomogeneity has shown to depend on the beam energy. In a study to test the AAA algorithm photon dose calculation for the 6 and 18 MV, the AAA overestimated the dose near the lung inhomogeneity up to 7% for the 6 MV while no overestimation was observed for the 18 MV [Van Esch, 2006]. It is stated in that reference that the AAA algorithm produces large amount of scattered photons with large divergence angles for low energies as opposed to high energies. This could explain the poor agreement for the Co-60 beam near the lung inhomogeneity. Therefore, a careful consideration should be taken when the Co-60 beam is used to treat tumours near the lung to avoid a dose deficiency.
Figure 5.16: Inhomogeneity profiled acquired in a $10 \times 10$ cm$^2$ field sizes at 6 cm depth, both local and global variations are shown at the bottom of the plots.

On the other hand, there is good agreement between the measured and calculated dose in the bone shown in Fig. 5.16 (b). A maximum variation in the bone inhomogeneity is about 3.6% while near the bone the difference is less than 1%. The overestimation of the dose in the bone inhomogeneity is related to failure of the AAA algorithm to predict the attenuation of the primary photon for materials with density greater than 1 g/cm$^3$. The variations in the bone were also observed in the study to evaluate bone inhomogeneity correction by the AAA algorithm using the
It was found that the algorithm was overestimating the dose inside the inhomogeneity up to 5%. The agreement in the penumbra and outside the field size in these simple inhomogeneity corrections is similar to Sec 5.3.1.2. In general, the AAA algorithm provides an excellent estimation of the Co-60 beam inhomogeneity correction. The poor agreements observed in these results demonstrate the common limitations of the AAA algorithm for the low energies that have been reported even for the linac beams [Van Esch, 2006; Robinson, 2008].

5.3.1.5 Modified Beam validation

In addition to the open beams, the validation was also performed to test the TPS for modified beams. A number of blocked beam measurements were acquired in both square and rectangular field sizes and compared to the Eclipse model calculations. The results are shown in Fig. 5.17. It can be observed that the Eclipse TPS is generally able to estimate the Co-60 dose under the block. The difference between the measured and the Eclipse calculated is in this region is less than 3%. The variation is caused by the limitation of the Eclipse TPS to accurately estimate the transmitted primary photons in the block. The clinical criteria recommended for agreement under the block is 4% [IAEA, 2004]. The agreement within the clinical recommended criteria in the region under the blocks suggests that the Eclipse model can accurately estimate the amount of primary photons from the Co-60 beam transmitted through the blocks.

The measured and Eclipse model estimations also agreed well in the high dose gradients inside the beam. A maximum difference between the measured and Eclipse calculated profiles is less than 15%, except in Fig. 5.17 (b) in which the difference is about 20%. Notice that in all cases the maximum variations appear at the corners of the blocks, which could be caused by limitations
of the Eclipse TPS to properly model doses at edges of the non-divergent blocks used in this study. The variations in the penumbra and outside the beam are similar to open beams. These results are in the range of the clinical criteria of the agreement in the high gradient regions inside the beam and in the penumbra of 15% or distance to agreement of 2 mm.

Figure 5.17: Blocked dose profiles acquired at 5 cm depths in various field sizes. The agreements are shown using both local and global variations between the measured and Eclipse calculated.
Generally, the results demonstrate an excellent agreement throughout the beam between the measurements and the AAA algorithm predictions of the dose delivered by the Co-60 modified beams. Although, these were simple plans, they have provided an overview of the dose estimation under modified beams. The results indicate that the Eclipse model could provide accurate estimations of the dose even under the MLC used for IMRT delivery techniques using the Co-60 beam. However, more treatment plans under clinical situation plans may be required to further understand the feasibility of the modified beams using the Co-60 beam.

5.3.2 Point Dose Measurements

The point dose measurements represent an absolute dose of interest delivered at a point. The measurements were performed to mimic various beam configurations used to deliver radiation treatment in clinical situations. Given the prescribed time and dose rate, the Eclipse model was tested to reproduce the measured dose in the same beams configurations. The doses were delivered at a 5 cm depth in plastic phantom using either static or arc beams in various configurations as described previously. Table 5.2 shows an excellent agreement of the results for the four-field box measured in different field sizes for both static and arc beams. As mentioned previously, the ion chamber deviations were too small to include in the measurements because of the stability of the Co-60 beam. All static beams for the four-field box for both square and rectangular fields show an excellent agreement. The maximum difference in both square and rectangular fields in different configurations is -1.7%. It is interesting to see that the good agreement is observed in both the small and large field sizes despite the difference of the amount of scattering photons. A maximum average variation of -1.2% is observed for all square and rectangular field sizes at different angles in a $5 \times 5$ cm$^2$. This suggests that the AAA algorithm can accurately predict point doses of the Co-60 beam inside the field regardless the field size.
Table 5.2: The excellent agreements of the point measurements shown by local differences (%) acquired in various beam configurations at 5 cm depths.

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For the arc beams, measurements were acquired at gantry angles 90°-270° and 270°-90° at a depth of 5 cm. Similar to the static beams, the arc beams in square and rectangular fields also show good agreements between the measured and the Eclipse calculated doses. However, the variations in the arc beams observed in the rectangular field size are relatively larger with a maximum variation of ~2.1%. The discrepancies are mainly due to arc beam configurations in which photons at different angles interact at different layers of a medium. Nevertheless, the maximum average variation in all plans is -2%.

5.3.3 2D Measurements

Finally, the test for accuracy of the Eclipse model to predict the delivered dose in a 2D dose distribution was performed. The 2D dose distribution was performed through irradiation of the EBT GaChromic films and the results for a four-field box treatment plan are depicted in Fig. 5.18. It is clearly seen that the high dose is delivered in the target. However, the intended dose in the targeted volume has an influence to dose delivered at the entrance of the beams. As shown in Fig. 5.19, in this plan the beam entrance dose is up to 80% dose of the target volume. However, in conformal radiation therapy, this is avoided by the application of multiple beams to deliver the prescribed dose to the target.

Figure 5.19 also shows good agreement between the measured and the AAA algorithm calculated dose profiles. As described earlier, the major discrepancies are observed at the edges of the beams. However, the differences have significantly decreased due to the application of more than a single beam. As can be observed from the differences at the bottom of plot, the maximum local difference in these regions for both the short and long sides is about -4%. The distance between the two curves in the penumbra is less than 1 mm. It is a surprising that only four beams can
Figure 5.18: A four field box dose distribution showing high dose at the target volume but also at the beam entrances.

Figure 5.19: Four-field box of two opposite $10 \times 10$ cm$^2$ and $3 \times 3$ cm$^2$ field size at 5 cm depth, where the local differences are shown at the bottom.
potentially improve the agreement even for the $3 \times 3 \text{ cm}^2$ between the measurements and the Eclipse model calculations. However, the results also indicate the effect of volume averaging in the ion chamber measurements especially in the low dose regions as opposed to film. Although this was a simple treatment plan, the results are consistent to what has be reported that the Co-60 beam penumbra is significantly reduced when the conformal delivery techniques such as IMRT are used [Fox, 2008; Wooten, 2015].
6

Conclusions

6.1 Summary and conclusions

Despite the dominant use of Cobalt-60 (Co-60) radiation units in the early development of megavoltage radiation therapy, Co-60 based radiation therapy has significantly declined due to lack of improvements to handle the modern radiation therapy techniques. However, due to operational simplicity and lower related costs, Co-60 units account for the majority of the external radiation therapy in the lower and middle income countries. Alternatively, clinical linear accelerators (linacs) systems which have benefited from technological advances in radiation therapy have become the standard device for precise radiation dose deliveries in developed countries. Yet, studies suggest that the Co-60 unit can be modernized to produce precise dose deliveries similar to linacs [Kerr, 2000; Joshi, 2008, 2014; Schreiner, 2009; Rawluk, 2010; Dhanesar, 2013]. Recently, the Cancer Center of Southeastern Ontario (CCSEO) modified its T780C Co-60 unit to an upgraded Equinox-100, which provided a less expensive treatment unit that can support the current dose delivery techniques. This current development was one of the continuous efforts conducted by the medical physics research group of CCSEO to improve Co-60 unit radiation therapy. This thesis has focused on the commissioning of the upgraded Equinox-100 unit in the Varian Eclipse TPS following the mechanical acceptance tests of the unit.

The Co-60 beam dose calculations in the Eclipse TPS was performed by the Anisotropic Analytical Algorithm (AAA), which is a model-based algorithm that uses pencil beam convolution to provide 3D photon dose calculations in a patient. The AAA model divides the
photon beam into three components: the primary photons which come from the primary radiation source and reach the patient without prior interaction, the scattered photons from the flattening filter and the head of the unit and the electron contamination created as a result of photon interactions in the head of the unit. The AAA algorithm was established primarily for linac dose calculations before it was upgraded to a version that can perform dose calculations for a Co-60 beam. For the AAA to perform dose calculations of the treatment unit, the unit must be characterized and modeled under reference conditions. The beam modeling of the treatment unit requires dose measurements acquired in water to establish physical parameters such as energy spectrum and photon fluence. Details of various beam measurements and modeling processes are found in Chapter 4 and 5. To validate the Eclipse model, measurements in simple treatment plans such as the PDDs, profiles, oblique incidence, inhomogeneity, blocked field, point dose and film measurements were acquired and compared to Eclipse calculations.

The results analysis shows an excellent agreement between the measurements and the Eclipse model calculations. Most of the local percentage differences between measurements and the Eclipse calculations as shown in Chapter 5 were within the recommended criteria from different authors [Fraass, 1998; Venselaar, 2001; IAEA, 2004]. Table 6.1 summarizes all the results in this thesis against various clinical recommended criteria. The main challenge, however, is noticed in the region between 10% - 2% of the dose at the central axis (foot) of the profiles. A local variation in this region is more than 30% in small fields. The disagreement in this region is not surprising because of the finite size of the Co-60 source.

In conclusion, the mechanical acceptance tests suggest that a modification of the old Co-60 units to a modern treatment unit is possible with an excellent mechanical accuracy. Furthermore, the results in this thesis suggest that the AAA in Eclipse TPS can accurately predict the delivered
Table 6.1: The summary of all the results from this thesis as compared to different recommended criteria, the bolded values indicate the failed criteria of all the authors

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</tr>
<tr>
<td>5</td>
<td>Inhomogeneity</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Within inhomogeneity</td>
<td></td>
<td>3.6</td>
<td>5</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Inner field</td>
<td></td>
<td>4.2</td>
<td>7</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Penumbra (90%–10%) (Distance to agreement)</td>
<td>20</td>
<td>2 mm</td>
<td>7 mm</td>
<td>3 mm</td>
</tr>
<tr>
<td></td>
<td>Outer region of the field</td>
<td></td>
<td>2</td>
<td>7</td>
<td>5</td>
</tr>
<tr>
<td>6</td>
<td>Blocked field</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Under the Block</td>
<td></td>
<td>3</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Inner field</td>
<td></td>
<td>2</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Penumbra (90%–10%) (Distance to agreement)</td>
<td>15</td>
<td>2 mm</td>
<td>2 mm</td>
<td>15</td>
</tr>
<tr>
<td></td>
<td>Outer region of the field</td>
<td></td>
<td>2</td>
<td>5</td>
<td>5</td>
</tr>
</tbody>
</table>

Comment: For the results in the foot region, the subscripts represent local (l) and global (g) differences.

* The results are for dose between depth of maximum dose (0.5 cm) and 20 cm depth.
dose in various clinical situations from the Co-60 beam for field sizes greater than $5 \times 5 \, \text{cm}^2$. This indicates that the Co-60 units can play a potential role in modern conformal radiotherapy if accommodated with equipment that supports such delivery techniques. To our knowledge, this thesis is the first to demonstrate the capability of the upgraded Equinox-100 unit and the accuracy of the Eclipse TPS AAA algorithm in modeling the Co-60 beam.

6.2 Future work

Currently, the Eclipse model prediction of the Co-60 dose delivered in small fields is suboptimal. To extend this work it would be worthwhile to characterize the small fields using measurements acquired by a small volume detector to provide clearer conclusions. The characterization may also involve Monte Carlo simulations of the upgraded Co-60 beam to provide a standard comparison.

Furthermore, the availability of 3D Co-60 treatment planning opens up future possibilities to investigate the role of Co-60 units in intensity modulated radiation therapy (IMRT) techniques. Previous studies have suggested the feasibility of the Co-60 beam in IMRT if the unit is developed [Fox, 2008; Schreiner, 2009; Dhanesar, 2013]. With the upgraded unit, it would be worthwhile to equip multi-leaf collimators (MLC) to support the IMRT techniques on the Co-60 unit.

Finally, to ensure the accuracy of the TPS, it would be valuable to develop the treatment planning system based on Co-60 configurations. For example, the establishment of the TPS based on conjugal gradient active set algorithm [Hristov, 1997] such as the CCSEO in-house inverse treatment planning capable of tomotherapy planning [Chng, 2006; Dhanesar, 2013]. Moreover, it
would be valuable to investigate the possibility of commissioning the Co-60 unit into other commercially available TPSs such as NOMOS. This would provide multiple TPS options for Co-60 unit users and ensure the availability of Co-60 unit 3D treatment planning.
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