THE ROLE OF COBALT-60 SOURCE IN INTENSITY MODULATED RADIATION THERAPY: FROM MODELING FINITE SOURCES TO TREATMENT PLANNING AND CONFORMAL DOSE DELIVERY

By

Sandeep Kaur Dhanesar

A thesis submitted to the Department of Physics, Engineering Physics & Astronomy

In conformity with the requirements for

the degree of Doctor of Philosophy

Queen’s University

Kingston, Ontario, Canada

(August, 2013)

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Abstract

Cobalt-60 (Co-60) units played an integral role in radiation therapy from the mid-1950s to the 1970s. Although they continue to be used to treat cancer in some parts of the world, their role has been significantly reduced due to the invention of medical linear accelerators. A number of groups have indicated a strong potential for Co-60 units in modern radiation therapy. The Medical Physics group at the Cancer Center of the Southeastern Ontario and Queen’s University has shown the feasibility of Intensity Modulated Radiation Therapy (IMRT) via simple conformal treatment planning and dose delivery using a Co-60 unit.

In this thesis, initial Co-60 tomotherapy planning investigations on simple uniform phantoms are extended to actual clinical cases based on patient CT data. The planning is based on radiation dose data from a clinical Co-60 unit fitted with a multileaf collimator (MLC) and modeled in the EGSnrc Monte Carlo system. An in-house treatment planning program is used to calculate IMRT dose distributions. Conformal delivery in a single slice on a uniform phantom based on sequentially delivered pencil beams is verified by Gafchromic film. Volumetric dose distributions for Co-60 serial tomotherapy are then generated for typical clinical sites that had been treated at our clinic by conventional 6MV IMRT using Varian Eclipse treatment plans. The Co-60 treatment plans are compared with the clinical IMRT plans using conventional matrices such as dose volume histograms (DVH). Dose delivery based on simultaneously opened MLC leaves is also explored and a novel MLC segmentation method is proposed. In order to
increase efficiency of dose calculations, a novel convolution based fluence model for
treatment planning is also proposed.

The ion chamber measurements showed that the Monte Carlo modeling of the
beam data under the MIMiC MLC is accurate. The film measurements from the uniform
phantom irradiations confirm that IMRT plans from our in-house treatment planning
system are deliverable. Comparing the Co-60 dose distributions and DVHs to the IMRT
plans from the clinic indicates that Co-60 is able to provide similar dose conformality to
targets and dose sparing to critical organs. The results of the novel MLC segmentation
algorithm and the photon fluence model proposed in this work compared well with the
Monte Carlo calculations.

In summary, the investigations presented in this thesis confirm that Co-60
tomotherapy is indeed capable of providing state-of-the-art conformal dose delivery. We
have shown that the perceived beam limitations often identified with Co-60 (e.g., lower
penetration, source size artifacts under small field collimation, and larger penumbra) are
negligible when using intensity modulated techniques.
Co-Authorship

Chapter 3 contains a version of an article submitted to Medical Physics as: *Dhanesar S, Darko J, Joshi C J, Kerr A, and Schreiner L J “Cobalt-60 tomotherapy: clinical treatment planning and phantom dose delivery studies.”* Sandeep Dhanesar performed the Monte Carlo simulations, experiments, and treatment planning and data analysis, interpreted the results and wrote the manuscript. Johnson Darko and Chandra Joshi provided guidance with the Monte Carlo simulations and edited the manuscript. Andrew Kerr provided guidance on the treatment planning software and edited the manuscript. The first version of the treatment planning software was developed by Nick Chang who was supervised by Andrew Kerr. Sandeep Dhanesar further developed the program for the purposes of this project. John Schreiner supervised the project, advised on data interpretation, and edited the manuscript.

Chapter 4 contains a version of an article submitted to Medical Physics as: *Sandeep Dhanesar, Johnson Darko, Chandra P. Joshi, and L. John Schreiner “Three dimensional intensity modulated dose distributions for clinical cases using Cobalt-60 based serial tomotherapy approach.”* Sandeep Dhanesar developed the third version of the treatment planning system for three dimensional planning, performed Monte Carlo simulations, experiments, treatment planning, and data analysis, interpreted the results and wrote the manuscript. Johnson Darko provided clinical insight, advised during treatment planning software development, and edited the manuscript. Chandra Joshi provided clinical insight on the project. John Schreiner supervised the project, advised on data interpretation, and edited the manuscript.
Chapter 5 contains a version of an article that is published in Medical Physics as: 

*Dhanesar S, Darko J, and Schreiner L J “Aperture superposition dose model versus pencil beam superposition dose model for a finite size Cobalt-60 source for tomotherapy deliveries” Med. Phys. 39, 206-215.* The Medical Physics journal has granted permission to reproduce this article as part of this thesis. The permission letter is available in Appendix A. Sandeep Dhanesar developed the model, performed experiments, Monte Carlo simulations and data analysis, interpreted the results and wrote the manuscript. Johnson Darko and John Schreiner supervised the project, advised on data interpretation, and edited the manuscript.

Chapter 6 contains a version of an article that will be submitted to Medical Physics as: *Sandeep Dhanesar, Johnson Darko, and L. John Schreiner “Multi-source Cobalt-60 model for photon fluence calculations used in convolution/superposition dose calculation method.”* Sandeep Dhanesar developed the model, performed Monte Carlo simulations and experiments, completed the data analysis, interpreted the results and wrote the manuscript. John Schreiner and Johnson Darko supervised the project, advised on data interpretation, and edited the manuscript.
Acknowledgements

"Sometimes our light goes out but is blown into flame by another human being. Each of us owes deepest thanks to those who have rekindled this light", Albert Schweitzer.

There are certain individuals in the world whose trust, belief and guidance helps us overcome most unattainable tasks. I would like to take a moment to appreciate countless hours of mentorship, supervision and guidance of my supervisors, Dr. John Schreiner and Dr. Johnson Darko. Without their guidance, support, and encouragement this thesis and my research would not have been possible. Dr. Schreiner took a young Physics student and ensiled her with confidence and knowledge. Under his supervision, I learned how to expand on my basic knowledge of Physics; I learned in-depth knowledge of how to examine issues, research and present them. I sincerely thank Dr. Chandra Joshi and Dr. Andrew Kerr for teaching me the clinical aspects of medical physics. I would also like to acknowledge the support of my colleagues and friends Greg Salomons, Chris Peters, Lynda Mowers, Nick Rawluck, Matthew Marsh, Amy MacDonald, Daxa Patel, Tim Olding, Dr. Xiangyang Mei, Nick Chng, and Lourdes Garcia. I could not have wished for a supportive or friendlier working environment.

Also, I wish to thank my family and friends. It is due to their constant support and encouragement, I have remained focused and motivated to finish my PhD. Thanks also to many other people who have helped me in other ways during my study. I would like to give my deepest gratitude to my husband Harmeet who has been my constant pillar of strength and motivation. Lastly, and most importantly, God, thank you for your countless blessings.
Statement of Originality

I hereby certify that all of the work described within this thesis is the original work of the author. Any published (or unpublished) ideas and/or techniques from the work of others are fully acknowledged in accordance with the standard referencing practices.

(Sandeep Kaur Dhanesar)

(June, 2013)
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Glossary

**Absorbed Dose** The energy absorbed per unit mass in tissue when the incident radiation undergoes interactions with atoms in the patient. The SI unit for absorbed dose is the Gray (Gy), where 1 Gy = 1 J/kg.

**Cancer** A generic term for more than 100 different diseases that are characterized by uncontrolled abnormal growth of cells.

**Collimator** A mechanical device, installed on a radiotherapy unit along the path of a beam to define the shape and size of the radiation beam. A typical collimator is constructed from a heavily attenuating material like tungsten.

**Conformal Radiation Therapy** A radiation therapy technique that uses multiple radiation beams that are each custom-shaped to conform the high dose to the tumour while sparing the normal tissue. It is also known as 3-D or conformational radiation therapy.

**Conventional Radiation Therapy** Radiation therapy based on relatively broad beams with simple, standard beam shapes and a limited number of gantry angles.

**Convolution-Superposition Dose Calculation Method** A model based dose calculation method that takes into account the contribution from primary and secondary radiation. In this method the photon dose is calculated as the convolution of the energy released at each interaction site and a spatially invariant photon dose kernel.

**Cumulative Dose-Volume Histogram** A graph that shows the volume of a structure receiving greater than the specified dose. It is commonly referred to simply as “DVH”.

**Dose Calculation** The process that provides an approximate dose distribution that arises
in a patient or in a phantom due to a radiotherapy treatment. The dose distribution is determined based on the geometry of the radiotherapy unit, beam energy, and the patient anatomy.

**Dose Distribution** Isodose curves, which are drawn at regular intervals of absorbed dose and are expressed either as a percentage of the dose at a reference point or the absolute dose value.

**External Beam Radiation Therapy** A form of radiation therapy in which radiation is directed at the patient from a source external to the patient.

**EGSnrc system** (Electron-Gamma Shower) is a Monte Carlo code used for the simulation of electron and photon transport through an arbitrary geometry.

**Fan Beam** A narrow, slit-shaped divergent beam.

**Field** The area of the radiation beam that projects at the isocenter on the plane perpendicular to its axis.

**Forward Treatment Planning** A technique of determining the dose distributions based on manually selected field orientations and beams.

**Gamma Comparison Method** A dose comparison method, based on the dose and distance criteria, assigns a value of 1 or less to the regions where there is a good agreement between two distributions and a value of 1 or higher where the agreement fails.

**Gamma Rays** A form of radiation emitted via the decay of nuclei of radioactive materials such as Cobalt-60.

**Gantry** The head of the radiation therapy that houses a radiation source, collimation system, beam monitoring system. It can rotate completely around a patient, to direct a
radiation beam from any direction.

**Heterogeneity** Refers to the condition when the volume of the patient or a phantom consists of different composition and density.

**Intensity Modulated Radiation Therapy (IMRT)** An advance method of conformal radiation therapy in which the intensity within a radiation beam is modulated during treatment to spare more adjoining normal tissue than is spared during conventional or 3-D conformal radiation therapy. Because of this, an increased dose of radiation can often be delivered to the tumor.

**Inverse Treatment Planning** An automated process, which begins with a desired dose distribution and comes up with a set of optimized beam parameters that make it possible to deliver a dose distribution similar to the desired one.

**Isocentre** The centre of rotation of a radiation beam.

**Kernel** A function that describes the spread of energy due to local scattering in the medium.

**Medical Linear Accelerator (LINAC)** A type of high-energy x-ray machine most commonly used for external beam radiation treatments. A medical linac uses high frequency electromagnetic waves to accelerate electrons to high energies through a linear waveguide. The high energy electron beam may be used directly for treating superficial tumors, or it can strike a high atomic number material to produce x-rays for treating deep-seated tumors.

**Milan-Bentley Dose Calculation Method** A dose calculation method based on empirical dose data measured in water for calculating the dose in the patient. It is based on the concept of diverging fan lines that radiate from a source and which intersect depth lines located at
selected distances below the phantom. The dose distributions are obtained by rapidly sampling measured data sets consisting of a central axis component and an off-axis component for different source-to-surface distances.

**MIMiC** A binary multi-leaf collimator manufactured by Best Nomos. It consists of 20 leaves, projecting a smallest size of 1x1 cm\(^2\) at the isocenter of the machine (80 cm).

**Monte Carlo Dose Calculation Method** A pure model-based technique that models the dose distribution in a medium by simulating photon and electron transport. It simulates the stochastic nature of photon interactions by the means of sampling randomly from known cross sections of photon interactions. The trajectory of the photon is simulated until its energy falls below an energy threshold or when it leaves the volume of interest.

**Multi-leaf Collimator (MLC)** A special type of collimator which can define irregularly shaped radiation fields. An MLC has narrow, interlaced metal blocks (“leaves”) that can be independently driven in or out of the beam to regulate the amount of radiation passing through.

**Objective Function** A function that evaluates the planned dose distribution with respect to the desired dose distribution. The goal is to achieve a minimum difference between the planned and the desired dose distribution.

**Pencil Beam** A ray-like radiation beam of small field size.

**Penumbra** The width of the edge of the field defined by the distance between the 80% and 20% of the dose.

**Percentage Depth Dose (PDD)** The percent depth dose is equal to the ratio of dose at a particular depth over dose at a reference depth normalized to 100. The reference depth is usually the depth where maximum dose occurs. These measurements are taken along the central axis of the beam.
**Radiation Therapy (Radiotherapy)** Medical treatment of cancer and other diseases using ionizing radiation from x-ray machines or radioactive materials, with the primary aim to kill undesired cells.

**Target Volume** A 3-D region in a patient’s anatomy (e.g. tumor) for which a high radiation dose is prescribed.

**Tomotherapy** A rotational form of IMRT that uses intensity modulated fan beam of radiation. The fan beam revolves around the patient while the patient is being translated through the gantry.

**Tumour** An abnormal mass of cancerous tissue.
# Nomenclature

**Symbols**

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Co-60</td>
<td>Cobalt-60</td>
</tr>
<tr>
<td>C</td>
<td>central axis dose</td>
</tr>
<tr>
<td>d</td>
<td>depth</td>
</tr>
<tr>
<td>dmax</td>
<td>depth at the maximum dose</td>
</tr>
<tr>
<td>D</td>
<td>dose</td>
</tr>
<tr>
<td>F</td>
<td>objective function</td>
</tr>
<tr>
<td>FS</td>
<td>field size</td>
</tr>
<tr>
<td>E</td>
<td>energy (in eV)</td>
</tr>
<tr>
<td>K</td>
<td>energy deposition kernel</td>
</tr>
<tr>
<td>M</td>
<td>the total number of voxels in cm³</td>
</tr>
<tr>
<td>µ</td>
<td>linear attenuation coefficient</td>
</tr>
<tr>
<td>µ/ρ</td>
<td>mass attenuation coefficient</td>
</tr>
<tr>
<td>N</td>
<td>the total number of pencil beams</td>
</tr>
<tr>
<td>ρ</td>
<td>density in g/cm³</td>
</tr>
<tr>
<td>R</td>
<td>off-axis-ratio</td>
</tr>
<tr>
<td>T</td>
<td>terma</td>
</tr>
<tr>
<td>W</td>
<td>jaw/aperture function</td>
</tr>
<tr>
<td>x</td>
<td>beam weights</td>
</tr>
<tr>
<td>xₚₑ</td>
<td>distance from the central axis to the lateral edge of the primary collimator.</td>
</tr>
</tbody>
</table>
Φ photon fluence
Ψ energy fluence
Ω source distribution function for Co-60
γ gamma index

**Acronyms**

AAS Aggressive active set
AS Aperture Superposition
CCSEO Cancer Center of the Southeastern Ontario
CG Conjugate gradient
CT Computer tomography
CTV Clinical target volume
Co-60 Cobalt-60
DVH Dose volume histogram
ECUT electron cut-off energy
FSPB Finite Size Pencil Beam
GTV Gross tumour volume
KERMA Kinetic energy released in the medium
IMRT Intensity modulated radiation therapy
MLC Multi-leaf Collimator
Ni-60 Nickel-60
NTCP Normal tissue complication probability
OAR Organ at risk
<table>
<thead>
<tr>
<th>Acronym</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>OD</td>
<td>Optical density</td>
</tr>
<tr>
<td>PCUT</td>
<td>Photon cut-off energy</td>
</tr>
<tr>
<td>PDD</td>
<td>Percentage depth dose</td>
</tr>
<tr>
<td>PTV</td>
<td>Planning target volume</td>
</tr>
<tr>
<td>RDF</td>
<td>Relative dose factor</td>
</tr>
<tr>
<td>SAD</td>
<td>Source-to-axis distance</td>
</tr>
<tr>
<td>SSD</td>
<td>Source-to-surface distance</td>
</tr>
<tr>
<td>TERMA</td>
<td>Total energy released in the medium</td>
</tr>
<tr>
<td>TCP</td>
<td>Tumour control probability</td>
</tr>
<tr>
<td>TPS</td>
<td>Treatment planning system</td>
</tr>
</tbody>
</table>
Chapter 1: General Introduction

Cancer is a class of diseases characterized by the growth and malignant spread of tumors. A cancerous growth is initiated by damage to the deoxyribonucleic acid (DNA) resulting in mutations to genes that initiate inappropriate activation of proteins that normally suppress cell division. These mutations are triggered by agents called carcinogens, which could be certain chemicals, viruses, radiations, and/or environmental agents (e.g. tobacco smoke and alcohol).

Cancer is one of the leading causes of death in Canada. On the basis of current incidence rates, the Canadian Cancer Society estimated 186,400 new cases of cancer (excluding 81,300 non-melanoma skin cancers) and 75,700 deaths from cancer in Canada in 2012 (Canadian Cancer Society 2012). The main modalities of cancer treatment include surgery, chemotherapy, radiation therapy, and hormonal therapy. The choice of therapy depends on the location and stage of the cancer as well as the performance status of the patient. Often different types of treatments are used in combination, either simultaneously or sequentially.

1.1 Review of Radiation Therapy

Radiation therapy uses radiation for the treatment of cancer, particularly solid tumors such as cancers of skin, brain, breast, prostate, or cervix. It may be used with curative intent, or as palliative treatment, where cure is not possible and the aim is for symptom relief. Of the people who develop cancer, about one half will receive radiation therapy sometime during treatment of their disease (Schreiner, 2006; Foroudi et al., 2003).
Radiation therapy works by damaging the DNA of tumour cells (Hall, 2011). Although damage to DNA is the primary mechanism of cell killing by an ionizing radiation through direct or indirect action, other mechanisms such as damage to the membrane have also been identified (Hall 2011). Direct ionization occurs when atoms constituting the biological material are ionized by the dense radiations such as neutrons and alpha particles. Indirect ionization occurs via ionization of the water molecules since approximately 80% of a cell is composed of water. Indirectly ionizing radiations such as x-rays, γ-rays, and electrons ionize water molecules, leading to creation of free radicals, atoms or molecules with an unpaired electron in the outer orbit. The free radicals such as the hydroxyl radical (OH•) are highly reactive and have the ability to diffuse far enough from the initial interaction site to reach and damage the other biological sites.

Cellular response to radiation depends on many factors such as repair of radiation injury, redistribution of cells in the cell cycle, repopulation by stem cells and the re-oxygenation status. Slow dividing cells respond slowly to radiation (late response) as opposed to rapidly dividing cells (acute response). Cancerous tumors are considered rapidly responding tissues and thus are more sensitive to radiation damage.

Although radiation is directed at the tumor with the intent of destroying the tumor and curing the disease, it is not possible to completely spare the normal tissues surrounding the tumor. During the process of treatment, they are also potentially damaged. Normal tissues generally are late responding tissues; therefore, they have greater capacity to repair the radiation injury than the rapidly proliferating tumor. Because of this, radiotherapy treatment is delivered in a course of many small fractions delivered over a period of several weeks (Hall, 2011). When a tumor is surrounded by
acutely responding normal and critical tissues, the amount of radiation dose delivered to the tumour is often compromised to certain extent.

Radiation therapy is available in two modes - external beam radiation therapy (EBRT) and brachytherapy (Podgorsak, 2005; Van Dyk, 1999). EBRT involves directing a beam of high energy radiation from outside the body, towards cancer site to be treated. In brachytherapy the radiation source (e.g. Iridium-192) is placed directly within or in close vicinity of cancerous volume in the patient’s body. The discussion in this thesis is limited to EBRT only.

1.2 Technology in External Beam Radiation Therapy

The therapeutic use of ionizing radiation began soon after the discoveries of x-rays (1895) and radioactivity (1896). Initial x-ray tubes delivered x-rays using accelerating potential of 40-70 kV and by 1922 the Coolidge x-ray machines operating at accelerating potentials of 200 kV and 750 kV (cascade tubes) became available. The developments in high energy x-ray technology, artificial radioactivity and particle accelerators continued since then, yielding mature modern technology catering to the use of ionization radiation in cancer treatments. Today various types of ionizing radiations such as photons (x-rays and γ-rays), electrons, protons and neutrons are used to deliver radiation therapy to cancer patients. Two of the photon based EBRT modalities - Cobalt-60 (Co-60) teletherapy and medical linear accelerators (linacs) are discussed in this chapter.

1.2.1 Cobalt-60 Teletherapy

Between 1936 and 1941 extensive research on identification and production of Cobalt radioisotopes resulted in discovery of radioactive cobalt isomers with half-lives
of 10.7 minutes and 5.3 years (Livingood et al 1938, 1941). As described in Eq. (1.1),
the radioactive cobalt isotope, $^{60}_{27}Co$, is produced by neutron bombardment of $^{59}_{27}Co$ in a
nuclear reactor through interaction.

$$^{59}_{27}Co + \frac{1}{0}n \rightarrow ^{60}_{27}Co + \gamma \quad (1.1)$$

The radioisotope undergoes negative beta decay under which the nucleus emits a
beta particle (Fig. 1.1). This leads to an excited state of Nickel-60 (Ni-60), which gives
up its energy by the emission of two gamma rays with energies of 1.17 and 1.33 MeV.
Since these two energies are almost the same, the Co-60 source is usually considered to
be a mono-energetic gamma ray source with an average energy of 1.25 MeV (Johns and
Cunningham, 1983; Khan, 1984).

![Decay scheme of Co-60 radionuclide. (Adapted from Johns and Cunningham, 1983).](image)

Figure 1.1: Decay scheme of Co-60 radionuclide. (Adapted from Johns and Cunningham, 1983).
In 1946, the use of Co-60 $\gamma$-rays as a radium substitute for the EBRT was suggested (Mitchell 1946). Between 1949 and 1951, three research groups from the USA and Canada independently worked on the development of Co-60 teletherapy units. The first Co-60 teletherapy unit based on H. E. John’s design was installed in August, 1951 at Saskatchewan Cancer Institute, Saskatoon, Canada (Greenstok 1995, Robison 1995). The other two units were installed at London cancer clinic of Ontario Cancer Foundation London, Canada (October, 1951) and M. D. Anderson Cancer Center, Houston, USA (September, 1953).

Within a short period, Co-60 units became highly popular worldwide and became the workhorse of every radiation therapy clinic. Many countries, including Canada, France, Russia, India, the Czech Republic and Argentina began manufacturing their own Co-60 EBRT units. The Co-60 units produced by Atomic Energy of Canada (now Best Theratronics) are the most utilized worldwide. These units use 185 TBq (5000 Ci) to 555 TBq (15000 Ci) radio-activity in a cylindrical Co-60 source capsules with a packing density of approximately 5.88 g/cm$^3$ (Glasgow 1999). They are available with gantry mounted source heads with source to isocentre distances of 80 cm and 100 cm. The main components of Co-60 EBRT units are gantry, source head, collimator system, and treatment couch. Figure 1.2 shows a picture of a clinical T780c Co-60 unit (MDS Nordion, Canada). This unit uses a 1.5 cm or 2.0 cm diameter cylindrical source capsule placed in a source drawer housed in the source head shielded with lead (Pb) and depleted Uranium. Radiation fields of up to 35x35 cm$^2$ at the treatment distance of 80 cm are defined by multi secondary collimators using adjustable tungsten leaves (Glasgow 1999).
Figure 1.2: A clinical Theratronics 780c Co-60 EBRT unit (Best Theratronics, Kanata, Ontario, Canada).

For decades, the Co-60 unit was the most popular radiation therapy unit used worldwide, with the ability to provide energy in the MeV range (1.25 MeV) (Schreiner et al., 2003). However, due to recent advances in radiotherapy such as the linac, IMRT and tomotherapy, the Co-60 units have almost disappeared from North America. Compared to the new linac machines, they have been perceived to be inferior due to some of their beam characteristics (Van Dyk, 1999). Also unlike new linac units with effectively point-like sources of radiation, the Co-60 unit has a cylindrical radiation source with a diameter 2 cm and height 2.8 cm. This gives rise to a broader penumbra, the width of the radiation field edge, which may weaken the conformation of the
delivered dose distribution to the target volume. Another perceived problem is that the Co-60 beam is of low energy and less penetrating when compared to the beams typically obtained with linacs. This means that when treating deep-seated tumors using a single or very few beams, one may deliver an excess dose to the superficial normal tissue in order to deliver the required amount of radiation to the tumour or less than the required amount of dose to the tumor in order to minimize the dose to the superficial normal tissue.

1.2.2 Medical Linear Accelerators

Medical linacs are the mainstay of radiation therapy today. Linacs use microwave RF fields to accelerate electrons to kinetic energies from 4 to 25 MeV in special evacuated structures called accelerating waveguides. The RF fields are produced by magnetrons or klystrons with typical frequency of 2.856 GHz. Since the introduction of first medical linacs, there have been tremendous technological developments that have led to highly sophisticated linacs in comparison with those used in the 1960s. A typical state-of-the-art medical linac is equipped with features such as: dual or triple x-ray energies (from 4 MV to 25 MV), multiple electron energies (e.g. 6, 9, 12, 16, 20 and 25 MeV), independent collimator jaws, dynamic wedge, dynamic multileaf collimator, electronic portal imaging device and computer controlled operation etc. (Karzmark 1993, Podgorsak 1999, Podgorsak 2005).

1.3 Dose Delivery Techniques in External Beam Radiation Therapy

In EBRT, a conformal radiation dose is delivered to the target volume using several different techniques. The conventional approach of delivery dose via EBRT involves a combination of several fields at different gantry angles, delivered with
relatively broad beams with simple, standard custom shapes. This approach provided limited sparing to the surrounding normal tissue when delivering radiation to the deep seated tumors. However, with recent advances in radiation therapy it is now possible to accurately target the tumor with higher doses of radiation, while minimizing damage to the healthy tissue. Many of these have been fueled by improvements in computers, imaging, dose computations and dose delivery hardware. In the subsequent sections, three commonly used techniques referred as three dimensional conformal radiation therapy (3DCRT), IMRT, and tomotherapy are discussed.

1.3.1 3DCRT

Three-dimensional (3-D) conformal radiation therapy (Purdy and Starkschall, 1999; Ezzell et al., 2003; Van Dyk, 1999) uses multiple radiation beams that are each custom shaped to conform the dose to the tumour (see Fig. 1.3). The fields are shaped to the target using a multi-leaf collimator (MLC), which is made up of individual “leaves” of a high atomic number material, usually tungsten that can move in and out to produce complex field shapes or beam apertures. While MLC leaves are allowed to shape the field, they are not allowed to move in and out within that particular field. Therefore, the shape of the beam for each field stays the same during the treatment (Siochi and Celi, 2000) and the intensity of a particularly field is not modulated.

1.3.2 Broad Beam IMRT

The 3DCRT was the standard radiation delivery mode up until recently. However, Intensity-Modulated Radiation Therapy (IMRT) is now the new generation of conformal therapy (Webb, 2003; Ezzell et al., 2003; Bortfeld et al., 1994). IMRT is more complex than the 3-D conformal radiation therapy in the sense that it allows
increased degree of freedom to modulate in-field intensity. In early 1980s, Brahme and Cormack (Brahme, 1988; Cormack, 1987) independently demonstrated that the use of non-uniform intensities can provide a better dose conformity to the tumor and at the same time spare the healthy organs from unwanted radiation. Unlike 3-D conformal radiation therapy that is delivered using a single radiation beam, IMRT is delivered as a sequence of many small beams that enter the body from many angles. With these multiple thin pencil beams the intensity of radiation within each field can be modulated to achieve a better conformity to the tumour (Fig. 1.3). In IMRT, intensity of radiation is modulated using the leaves of the MLC that can move in and out of the field to produce a sequence of complex field shapes or beam apertures. The cumulative effect is a complex intensity-modulated radiation beam.

Figure 1.3: Comparisons between two forms of radiation therapy: IMRT vs. 3DCRT (Siochi and Celi, 2000).
1.3.3 Tomotherapy

Tomotherapy is a rotational form of IMRT, which uses a fan beam (slit-shaped) of radiation instead of broad beam (Mackie et al., 1993). The term tomotherapy is derived from tomography, which means slice therapy. In tomotherapy, a fan beam of radiation revolves around the patient as the patient is translated through the ring gantry. Modulation of the fan beam intensity is achieved by a binary MLC that consists of a series of tungsten leaves. Another modality similar to tomotherapy is known as volumetric-modulated arc therapy (VMAT). VMAT delivers radiation to a large volume by rotating the gantry in one or more arcs instead of slice by slice approach used in tomotherapy. As it rotates, a number of parameters such as the MLC aperture shape, the fluence-output rate (dose rate), the gantry rotation speed, and the MLC orientation are varied. However, since the work in this thesis employs tomotherapy approach, the VMAT approach is not discussed further.

Two different approaches to tomotherapy have been investigated. These are referred to as serial and helical deliveries. Serial-tomotherapy was the first to be used in the clinics (Carol, 1995; Mackie, 2006). It required a retrofit to an existing linear accelerator and used a MIMiC (Best Nomos, Pittsburgh, PA, USA) binary MLC mounted to the head of a commercial linac. The MIMiC had a set of 20 tungsten leaves, projecting a maximum field size of 20x2cm$^2$. Serial tomotherapy can irradiate two slices at once with a narrow rotating beam of radiation, modulated by two sets of the MIMiC MLC leaves. After one complete rotation of the gantry, the couch is translated in discrete steps to deliver the next two slices.
IMRT provided by serial-tomotherapy is very sensitive to patient movement and indexing of the patient for the treatment of next slice. Due to rapid dose changes in IMRT treatments, slight patient shifts can have a significant impact on the overall dose delivery accuracy. The patient indexing method used by serial-tomotherapy has particularly been an issue since several studies have shown that it can introduce dose uncertainties at the junctions of the slices (Carol et al., 1996; Low, 1997). With this problem in mind, a research group at the University of Wisconsin developed a different approach to tomotherapy, known as helical-tomotherapy (Mackie et al., 1993; Mackie, 2006). Dedicated commercial helical-tomotherapy machines became available in 2003 (TomoTherapy Corp., Madison WI).

Unlike serial-tomotherapy, the couch in helical-tomotherapy is translated as the fan beam of radiation rotates around the patient. This reduces the potential of under- or over-dosing the region in the volume as junctions between slices are removed. The geometric properties of helical-tomotherapy are similar to those of a helical CT scanner. The unit uses a binary mini-MLC (TomoTherapy Corp., Madison, WI) to modulate the intensity of the radiation beam. The mini- MLC consists of 32 leaf pairs, projecting a fan-beam of radiation up to 40 cm in length (in-plane) and 5 cm in width (cross-plane). With this larger fan beam, a greater patient area can be treated.

1.4 Treatment Planning

Treatment planning is the process during which a team consisting of radiation oncologists, medical physicists and medical dosimetrists plan the appropriate external beam radiotherapy or internal brachytherapy treatment technique for a patient with cancer. In particular for external beam radiotherapy treatment planning, this process
consists of prescribing the radiation dose, determining the tumour and critical organ contours, setting the beam orientation and beam modifier specifications, and performing dose calculations and dose optimization. Traditional treatment planning methods involved the use of forward treatment planning approach. In forward planning treatment technique, the dose distributions are determined by manually selected field orientations and beams. However, clinical cases can be very complex and cannot be planned using manually-selected beam geometries. This is particularly true when using the tomotherapy technique since due to its requirement for thousands of beams, it cannot be performed manually. Intensity-modulated plans involving complex geometry are generated using sophisticated computing methods to perform automated optimization of the beam intensity patterns to achieve a desired conformal dose pattern. This type of treatment planning is known as inverse treatment planning (Mageras and Mohan, 1993; Ezzell, 1996; Gallant, 2006; Webb, 1989; Hristov and Fallone, 1997; Spirou and Chui, 1998; Chng, 2005; Holmes et al., 1995). In this technique, rather than calculating the dose distribution based on selected beam geometries, the treatment planning begins with a desired dose distribution and then a computer program optimizes the beams to achieve that dose pattern. In this section an overview of the dose calculation and optimization techniques is provided.

### 1.4.1 Dose Calculation Methods

Dose calculations methods are a vital part of any treatment planning system. The dose calculation methods can be classified as either correction-based or model-based methods (Mackie et al., 1996; Van Dyk, 1999). The correction-based methods determine dose from reference dose measured under the standard conditions in a water
phantom with some adjustments to account for specific treatment conditions such as patient contour and inhomogeneities, e.g., regions with different tissue composition and density. The model-based methods determine dose from radiation transport through the patient from first principles.

1.4.1.1 Milan Bentley method

Traditionally, the patient dose calculations of photon beams were mainly based on correction-based methods due to their inherent simplicity (Mackie et al., 1996). A simple correction-based method was proposed for computerized planning by Milan and Bentley in the early 1970s (Milan and Bentley, 1974). The Milan and Bentley method requires empirical dose data measured in water for calculating the dose in the patient. It is based on the concept of diverging fan lines that radiate from a source and which intersect depth lines located at selected distances below the phantom surface. The dose distributions are obtained by rapidly sampling measured data sets consisting of a central axis component and an off-axis component for different source-to-surface distances (SSD). The dose at a point \((x, y, d)\) can be calculated as follows:

\[
(x, y, d) = C(0, 0, d)R(x, y, d)(\frac{SSD}{SSD+h+d})^2
\]  

(1.2)

where \(C(d)\) is the central axis dose at depth \(d\) and \(R(x; d)\) is the off axis ratio at depth \(d\) and off axis distance \(x\) and \(y\). The last term accounts for changes in the SSD from those in the reference conditions. These changes are taken into account using the inverse square law, which states that the photon fluence is inversely proportional to the square of the distance from the source. The parameter \(h\) represents the thickness the tissue. From this one can determine how much the actual SSD differs from an initial SSD. The
central axis dose is commonly referred to as a percentage depth dose curve (PDD). A PDD curve is obtained by normalizing the dose at a depth with respect to the dose at a reference depth.

The Milan and Bentley method was a widely used in early computational planning in the 1970s and 1980s. Although the Milan and Bentley method provides a quick way for calculating dose since the calculations are based on tabulated beam data, it offers limited accuracy when the dose calculation conditions (e. g., patient curvature and heterogeneities) are different from the conditions that were used to gather beam data (e. g., flat surface, homogeneous water phantom). The original Milan-Bentley model in Eq. (1.2) was modified over the years to account for changes in surface curvature and basic inhomogeneity corrections more effectively (Booth, 2002). Dhanesar (2008) additionally modified the method to account for scatter at each SSD. In the original Milan-Bentley algorithm, the measurements are taken at only one SSD and then using those data, doses at other SSDs are calculated using the inverse square law. Hence, the algorithm assumes that the contribution from scattering remains the same at all SSDs and the inverse square law alone is sufficient for calculating doses at different SSDs. This assumption was found to be insufficient for the Co-60 measurement conditions. Through several pencil beam measurements made by Dhanesar (2008), it was found that scatter made a higher contribution to the measurements taken at smaller SSDs. This is because in the experimental set-up used in the Co-60 dose deliveries, the pencil beam collimator and the phantom were placed very close to each other. Therefore, the scattered electrons that originated within the collimator made a significantly higher contribution to the measurements made at smaller SSDs than the measurements made at
larger SSDs. In order to take into consideration the effect of scattering at all SSDs, Eq. (1.2) was modified to include an SSD-dependent scattering function as in Eq. (1.3).

\[ (x, y, d) = C(0, 0, d)R(x, y, d)(\frac{SSD}{SSD + h + d})^2 K_s(SSD, d) \]  

\( K_s(SSD, d) \) is the scatter component which was well modelled using an exponential function. The scatter component is generally difficult to model using the Milan-Bentley algorithm. However, since the focus of this model was restricted to one pencil beam and a limited range of SSDs, a reasonable approximation could be made. The details of the fitting parameters are presented in the work by Dhanesar (2008).

1.4.1.2 Convolution-Superposition method

Model-based methods provide a better dose accuracy since they are based on the fundamentals of radiation transport. Advances in computer technology enabled these model based algorithms, such as the convolution-superposition approach to become the standard in the clinical treatment planning (Mackie et al., 1996; Verhaegen and Seuntjens, 2003; Andreo, 1992).

The convolution-superposition method has been extensively used to calculate photon dose distributions (Mackie et al., 1985; Mohan et al., 1986; Ahnesjo et al., 1987; Boyer and Mok, 1985; Nilsson and Knoos, 1992; Mackie et al., 1996). The method takes into account dose contributions from primary and secondary radiation. The primary radiation consists of photons that are incident upon the surface of the patient and interact with the tissue for the first time. The incident photons could be the photons originated from the radiation source directly or produced due to interactions within the treatment head (Task Group 65, 2004). The scattered radiation, on the other hand,
consists of photons that have interacted at least once in the medium and reach the destination point by indirect multiple pathways within the patient.

The convolution-superposition method calculates the photon dose as the convolution of the energy released at each primary interaction site, also referred to as TERMA, the total energy released in the medium due to primary photon interactions and a spatially invariant photon dose kernel. This can be seen in Eq. (1.4), where the dose, \( D(x,y,z) \), at a point \((x,y,z)\) in a medium, is calculated by convolving the TERMA, \( T(x,y,z) \), with the photon dose kernel at that point \( K(x,y,z) \):

\[
D(x, y, z) = \iiint T(x', y', z')K(x - x', y - y', z - z')dx' dy' dz' \tag{1.4}
\]

The dose kernel describes the spread of energy due to local scattering in the medium. The kernel is spatially invariant when irradiating a homogeneous medium with a mono-energetic photon beam. However, for poly-energetic photon beams, a collection of weighted kernels are used to account for the energy spectrum. This general approach is known as the superposition method. The photon dose kernel is generally computed using a Monte Carlo method and the TERMA (Eq. 1.5) is calculated based on the photon fluence, \( \Phi(E, x, y, z) \), at a point \((x, y, z)\) and the attenuation coefficient, \( \frac{\mu}{\rho} \).

\[
T(x, y, z) = \sum_E \frac{\mu(E)}{\rho} E\Phi(E, x, y, z) \tag{1.5}
\]

The photon fluence within the phantom at point \((x,y,z)\) is calculated from the in-air photon fluence at the phantom surface, as well as by accounting for the diverging field geometry and attenuation within the medium. This is a standard convolution-superposition approach used for point sources. Currently there is no published convolution-superposition method for finite size sources method. During this project,
studies were performed to develop a similar model for Co-60 source, which has a finite source size. The studies indicate that this method can be extended to model doses from finite size sources such as the Co-60 source if the photon fluence model is adjusted to account for the source size. This is achieved by convolving the source fluence distribution function with the window function representing the field size:

$$\phi(x, y, z) = \int \Omega(x', y', zp)[\phi(x, y, z)_{full}W(x - x', y - y', zp)]dx'dy'$$  \hspace{1cm} (1.6)

where, $\Omega(x', y', zp)$ is the source fluence distribution at the point of calculation, $\phi(x, y, z)_{full}$ is the photon fluence from the source at the isocenter for an open (or full) field size calculated via a Monte Carlo simulation, and $W(x - x', y - y', zp)$ is a rectangular window function corresponding to the field size. Primary and scatter photon fluences are modeled separately. Chapter 6 provides in-depth description of the development of this model.

Most of the treatment planning systems used in clinics today use a convolution-superposition method for dose calculations. The main reason is that this method takes into account a large amount of the physics that actually occurs as a radiation beam passes from the head of treatment machine through the patient. It explicitly incorporates the effects of beam shaping and beam modification in the modeling process. However, the superposition-convolution method does have some disadvantages. The main one is the time required for calculations. Another disadvantage stems from the beam commissioning process. The convolution-superposition method is not a pure model-based algorithm; it requires that some empirical parameters be determined from the dose measurements in a water phantom (Purdy and Starkschall, 1999). However, since the
medium in which the dose is calculated is generally different from water, there may be slight variations in dose calculations in certain regions of the phantom.

1.4.1.3 Monte Carlo method

The Monte Carlo technique is a pure model-based technique that determines the dose distribution in a medium by simulating fundamental photon and electron transport (Rogers et al., 1995; Verhaegen and Seuntjens, 2003; Andreo, 1992; Purdy and Starkschall, 1999). It simulates the stochastic nature of photon interactions by the means of sampling randomly from known cross sections of photon interactions (Andreo, 1991; Mackie et al., 1996). The trajectory of the photon is simulated until its energy falls below an energy threshold or when it leaves the volume of interest. A dose distribution can be achieved by following the histories of millions of such photons and electrons, keeping track of the energy deposition within tissue elements of the body. Hence, the use of Monte Carlo method for treatment planning in radiotherapy is a two-step process; first, the beam output of the radiotherapy is modeled and second, the dose distribution in the patient or phantom is calculated by using the beam model created in the first step. Apart from the precision of the cross section data, the accuracy of the Monte Carlo calculations depends mainly on the correctness of the information about the starting condition of the radiation transport, the geometry of the setup and the materials used.

In the past, the Monte Carlo technique had a limited application in radiotherapy due to its high demands for computing power. However, since computing power is not a big concern these days, the Monte Carlo method is slowly making its way into the radiation treatment planning systems. One of the main reasons is that this is the only known method that takes into account all relevant physical interactions. Clinically it is
considered to be the most accurate method for calculating dose since it allows one to calculate dose even in the regions not well accommodated by other dose algorithms, for example in the interface between tissues of very different electron densities (Sauer, 1995; Yu et al., 1995; Arnfield et al., 2000; Neuenschwander et al., 1995). Details of the Monte Carlo details specific to this thesis are fully described in the next chapter.

1.4.2 Optimization Methods

Due to a large number of beams, treatment planning for tomotherapy is not possible using manual forward planning techniques. Tomotherapy requires an automated technique such as those used in the inverse treatment planning systems, where doses are optimized based on the user specified desired dose distribution. Specifically, this is achieved by numerically optimizing an objective function.

An objective function evaluates the planned dose distribution with respect to the desired dose distribution. The goal is to achieve a minimum difference between the planned and the desired dose distribution. The objective function can be based entirely on dose (i.e. absorbed dose arising from interaction of radiation within tissue) or it can use a radiobiological model (Van Dyk, 1999). The biological objective functions assume that optimization of a plan should be based on the biological effects produced by the underlying dose distribution. The objective of the treatment is to maximize the tumour control probability (TCP) while maintaining the normal tissue complication probability (NTCP) to within acceptance levels. A TCP is related to a dose distribution by a dose response function, which relates the effects of the radiation dose to tissue, tumour, and organ. However, this relationship is not sufficiently understood (Bortfeld et al., 1996; Wu and Mohan, 2000; Van Dyk, 1999). Therefore, the majority of the optimization algorithms are based on dose-based objective functions. A well-established and the
most commonly used dose-based objective function is the quadratic model (Van Dyk, 1999). The quadratic model minimizes the sum of the squared differences between the delivered and the desired dose distribution (Webb, 1989; Burkelbach et al., 1990; Xing and Chen, 1996; Mageras and Mohan, 1993). Since the quadratic objective function is convex (Choi and Deasy, 2002; Van Dyk, 1999), its minimum can be computed easily. This aspect is of great interest for gradient based optimization algorithms. The major drawback is that it is very sensitive to initialization and can get trapped in local minima. Despite this, it has a wide application in clinical treatment planning systems. Some of the major treatment planning systems that are based on this objection function are Pinnacle3 (Philips Medical Systems, Bothell, WA), Eclipse (Varian Medical Systems, Palo Alto, CA) and Corvus (Nomos, Sewickely, PA).

There are many proposed physical optimization algorithms (based on physical objective functions) for tomotherapy treatment planning; however, not all of them are practical since typical tomotherapy treatment involves thousands of pencil beams. The best optimization technique for tomotherapy is one that is robust, flexible, fast, and requires less memory. Iterative techniques are believed to satisfy most of these requirements. In iterative techniques, the plan is updated such that the value of the objective function decreases at each iteration. In the literature, the iterative techniques are classified as either stochastic or deterministic. The stochastic algorithms are based on randomly modifying a variable and then testing to see whether the objective function has been reduced. In deterministic algorithms, the variables are not modified randomly; the sequence of iterations is based on the starting point of the optimization, which is determined from the objective function. The discussion of this thesis is limited to deterministic algorithms only.
The examples of deterministic algorithms are linear programming and gradient-based algorithms. Linear programming was first used in radiation therapy treatment planning in 1968 by Bahr et al. as a mathematical tool for solving a set of linear equations to achieve positive solutions and minimization of a linear objective function (Bahr et al., 1968). In order to achieve desired uniformity of the dose in the target volume, linear programming technique uses minimum and maximum dose limits. The advantage of this technique is that it produces only positive solutions and therefore, one does not need to worry about beam-weights being negative. However, the difficulty is that by restricting to non-negative beam weights, it constrains the solution too conservatively. Because of these constraints, it either finds a solution or it fails.

The gradient based algorithms are preferred since they use a quadratic objective function. These are performed in two stages. Firstly, the direction of the optimization is chosen and secondly, along the chosen direction, a line of minimization is performed. The algorithm itself does not limit the solution to be non-negative. Most of the IMRT gradient based dose optimization algorithms known to date (Hristov and Fallone, 1997; Spirou and Chui, 1998; Chng, 2005) either impose positivity constraints to the solution (in order to ensure their physical meaning), or truncate the optimization when negative beam-weights are encountered.

1.5 Motivation of Research and Potential of Co-60 IMRT

Cancer is one of the leading causes of deaths in the world with more than two thirds of cancer deaths occurring in the low or middle income countries (WHO 2006). The increase in number of deaths has been indicated to be related to lack of cancer treatment technology in those countries. The International Atomic Energy Agency
(IAEA) reported that the developing countries only have 33% of the total radiation therapy facilities even though they make up of about 85% of the entire worldwide population. Their report indicated that out of about 5000 radiotherapy units required in the entire developing world, only 2200 such units are currently available (IAEA, 2003). The IAEA predicted that this shortage could grow to at least 10,000 by year 2015.

Co-60 units have the potential to fill the shortage of radiotherapy machines in the developing countries due to their simple and robust design, requirement for minimum power, maintenance, and technical expertise to operate. Although the basic Co-60 units are commercially available, some clinics in the developing world are reluctant to acquire such units as they perceive this technology as having limited potential for clinical advancement. Several research groups have been motivated by this background and have begun investigating ways to modernize Co-60 units (Schreiner et al., 2003; Warrington and Adams, 2002; Warrington and Adams, 2002, 2008; Cadman et al., 2007, 2011; Fox et al., 2008; Joshi et al. 2001, 2008). These groups indicated that issues such as low energy, low dose rate, and broader penumbra may not be problematic if the Co-60 units are modernized with state of the art devices. Studies have shown that the beam penetration and penumbra problems faced with Co-60 can be minimized if a large number of radiation fields from multiple directions are used (Warrington and Adams, 2002, Cadman et al., 2007, Fox et al., 2008). Furthermore, Joshi et al. (2001, 2008) noted that the problem of low dose rate, arising from the relatively lower energy of Co-60 and the activity of the source, can be solved if the unit is redesigned to include multiple sources and/or a different shaped source with an increased source activity.

The Medical Physics research group at the Cancer Centre of Southeastern Ontario (CCSEO) and Queen's University is investigating IMRT with Co-60 source,
based on multiple beam angles. The focus has been on Co-60-based tomotherapy as the form of delivery. The “ring" shape of the gantry of the tomotherapy machine is ideal for housing multiple Co-60 sources. Therefore, with this geometry and the potential for multiple sources, many of the perceived disadvantages of Co-60 can be mitigated (Schreiner et al., 2003). The initial Co-60 tomotherapy research was performed using a simple translation-rotation bench top apparatus, in conjunction with a conventional Cobalt-60 radiotherapy machine (Theratron 780C, MDS Nordion, Kanata, ON). The intensity modulation was achieved by scanning a phantom across a stationary pencil beam, mimicking a behavior similar to that of an MLC (Kerr et al., 2000), which modulates the beam by moving its leaves in and out.

Preliminary results showed that Co-60 based tomotherapy has the potential to provide highly conformal dose distributions (Salomons et al., 2003). These studies were based on a forward treatment planning technique and hence generating tomotherapy plans was not practical. There was no commercial inverse treatment planning system for Co-60 tomotherapy. In 2005, a graduate student at Queen’s University in Kingston, Ontario, Canada developed a 2-D inverse treatment planning program specifically for Co-60 tomotherapy in order to automatically generate complex conformal plans that had a clinical relevance (Chng, 2005). This program was successfully able to generate treatment plans for simple 2-D target volume geometries. Dhanesar (2008) further improved the in-house program to plan complex non-clinical patterns and validated them experimentally on film. The dose model was based on a crude dose calculation method, Milan-Bentley (Milan and Bentley, 1989) that provided reasonably accurate dose calculation for a small pencil beam of size 1x1 cm² in water equivalent homogeneous phantoms. The inverse treatment planning system was then used to
optimize the doses to conform them to the tumour and spare the critical structures. The agreement between the plan and the delivery was within the desired tolerance, which suggested that the in-house inverse treatment planning system can accurately generate conformal plan. The success of the non-clinical deliveries with Co-60 showed that Co-60 tomotherapy has the potential to play a significant role in radiation therapy.

The work in this thesis is undertaken as a continuation of previous Co-60 IMRT work at CCSEO and explores areas that need further developments. The results of the investigations presented in this thesis is expected to lead to improved and redesigned Co-60 unit, which has the potential to provide a reliable, low cost, low maintenance and state-of-the art radiation therapy. This may be of particular interest to the developing countries where radiotherapy resources are scarce but cancer patient population is greater.

1.6 Research Goals

The simple pencil beam collimator based experimental set-up described in Section 1.5 is very useful when investigating the feasibility of Co-60 based tomotherapy. However, the pencil beam collimator is not a practical choice for the development of a clinical Co-60 tomotherapy unit. This is because the delivery using a scanned pencil beam takes a very long time to complete. Hence, a practical choice would be to use a unit incorporating an MLC. The Medical Physics group at the CCSEO acquired the MIMiC collimator (Best NOMOS Corporation, Sewickly, PA) for conformal Co-60 dose deliveries. As discussed earlier, the MIMiC has 20 tungsten leaves that can modulate the intensity of 20 beamlets simultaneously. The use of a MIMiC collimator at the CCSEO for Co-60 tomotherapy treatment planning and
delivery has been limited due to the lack of a dose calculation method. The Milan-Bentley approach discussed earlier could not be used for this type of planning since it was limited to a single pencil beam only. Thus, a Monte Carlo based dose calculation method was used for preliminary MIMiC based studies. Dhanesar (2007) simulated a radiation treatment system, consisting of a MIMiC collimator and Theratronic 780C Co-60 radiotherapy unit, using the EGSnrc Monte Carlo program (Walters et al., 2007) and calculated the dose directly in a water-equivalent, circular phantom. The doses were then optimized for several non-clinical patterns. Although the delivery results obtained were promising, they could not be generalized to “true” clinical cases where the target volumes are of different shapes, contain heterogeneities and are 3-dimensional.

Also, in the preliminary work, the dose behavior of the MIMiC was characterized to achieve dose delivery via individually opened beamlets (one leaf open at a time). Therefore, the treatment plans were prepared for the individually opened beamlets only. Although these plans are sufficient to display the potential of Co-60 tomotherapy for clinical scenarios, the method of using individual beamlets defeats the purpose of using an MLC for dose deliveries. For a time efficient delivery, the MLC is intended to be operated with as many leaves open as possible at one instance. For point source based IMRT calculations, this is done by simply adding the doses for the pencil beams. This method is known as the finite size pencil beam superposition method (Bourland JD and Chaney, 1992). However, this method does not work well for finite size sources.

All these aspects are well worth studying in order to initiate the development of a commercial Co60-based tomotherapy unit. Thus the goals of this thesis work are to:
1. extend previous 2D Co-60 tomotherapy planning for simple phantoms to actual clinical tomotherapy using patient CT data,

2. extend 2D studies to three dimensions and evaluate true clinical cases planned with Co-60 tomotherapy to modalities being used in clinics today. This requires further developments to the current in-house inverse treatment planning software.

3. develop a dose summation method that will work for large sized sources such as the Co-60 source, and

4. develop a convolution based Co-60 dose calculation method so that the Monte Carlo dose calculation method can be avoided for treatment planning due to its requirement for long computation time.

1.7 Thesis Outline

Chapter 2 presents a review of the EGSnrc Monte Carlo system which was used for patient dose calculations. The focus is limited to the components and treatment units used in this work. A review of the in-house inverse planning system is also provided. A step by step procedure is described so that the readers of this thesis can follow the work presented here.

Chapter 3 presents the results of our investigations on 2D dose delivery and clinical treatment planning. The dose deliveries are performed on cylindrical homogeneous phantoms to verify the in-house inverse planning system and to show that conformal deliveries are possible with Co-60 source. The 2D Co-60 tomotherapy plans for targets in head and neck and pelvic region are generated using the EGSnrc Monte
Carlo system and the verified in-house planning system. Treatment plan comparisons are made to the actual patient plans used for treatments at the CCSEO.

Chapter 4 extends the treatment planning work presented in Chapter 3. It first reviews what dose delivery factors are to be considered in order to minimize the effect of the Co-60 beam penumbra, particularly the junctioning problems associated with serial tomotherapy approach. Then 3D treatment plans are presented for various clinical sites such as in the brain, eye, head and neck, abdomen and pelvic. The Co-60 tomotherapy treatment plans are compared to IMRT plans used for actual patient treatments at the CCSEO.

Chapter 5 demonstrates the issues with the tomotherapy intensity modulated fan beam segments when using conventional pencil beam superposition method. It demonstrates that doses for IMRT or tomotherapy segments for point sources can be calculated based on the finite size pencil beam superposition method. However, for finite sources such as the Co-60 source, the pencil beam superposition method has limitations; therefore the Co-60 tomotherapy segments need to be defined based on a new aperture superposition purposed in this chapter.

The apertures discussed in Chapter 5 have been calculated in Monte Carlo. However, there can be numerous aperture possibilities depending on the complexities of the cases used for treatment planning. Chapter 6 proposes a fluence calculation dose model for Co-60 source based on convolution approach. This approach can calculate doses for any Co-60 beam quickly. Intensity modulated segments can then be generated using this method in conjunction with the aperture superposition model presented in Chapter 5.
Chapter 7 summarizes the progress made on Co-60 tomotherapy during this research period and outlines some steps that need to be taken in order to extend this work further.
1.8 References


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Chapter 2: Review of the EGSnrc Monte Carlo Code and the CCSEO Co-60 Tomotherapy Treatment Planning System

Chapters 3 through 6 present dose calculations and treatment planning results obtained using the EGSnrc Monte Carlo system and the in-house inverse treatment planning system. Since these four chapters present manuscripts published or submitted to refereed journals, the details regarding the methodologies used are presented concisely. This chapter presents more in-depth details regarding the tools and methods used during this research project.

2.1 The EGSnrc Monte Carlo Code

The EGSnrc system (Electron-Gamma Shower) is a Monte Carlo code used for the simulation of electron and photon transport through an arbitrary geometry (Kwarkow, 2000; Kwarkow and Rogers, 2000). It is an upgraded version of the EGS4 system, incorporating enhancements such as an improved multiple-scattering theory which includes relativistic spin effects in the cross section, a more accurate boundary crossing algorithm and improved sampling algorithms for a variety of energy and angular distributions. The detail description of the enhancements can be found in the EGSnrc online manual (Kwarkow, 2000).

The EGSnrc Monte Carlo system consists of random number generators, user codes, input simulation files, cross-section data for different particle interactions in
different materials, geometric model of the system, the particle transport algorithms, variance reduction methods, and the data analysis tools for the review of simulation results. A description of some of these components is provided in the following sections.

2.1.1 Random Number Generator

A random number is defined as a number that would be chosen by chance, without any correlation with other numbers. In Monte Carlo methods, random number generators are an integral component since random sampling determines how well the probabilistic nature of a physical process is simulated. Pseudo random numbers, generated by a computer code, are widely used in MC simulations. They use an initial seed of the sequence to produce a sequence of uncorrelated numbers uniformly distributed over a predefined range (Bevington and Robinson 1992). The EGSnrc Monte Carlo code used in this thesis employs two pseudo random number generators - RANMAR and RANLUX. These random number generators can produce the same random number sequence on different machines and can be initialized to guarantee independent random number sequences when doing parallel runs. More details of these generators can be found in literature (James, 1994; Luscher, 1994; Marsaglia and Zarn, 1991; Marsaglia et al. 1990).

2.1.2 The User Codes

The EGSnrc code system offers various user codes that have been extensively used in radiation physics (Kawrakow and Rogers 2003, Rogers et al 1995, Rogers et al 2005, Walters et al 2005). These codes provide an interface for the users to create accelerator geometries, input files with user defined simulation variables, dose calculations phantoms and perform data analysis. Three of the user codes, namely
BEAMnrc, BEAMDP, and DOSXYZnrc, have been used for the work presented in this thesis. Figure 2.1 presents a flow diagram of how these user codes are utilized in the EGSnrc Monte Carlo System.

![Flowchart of BEAMnrc and DOSXYZnrc processes](image)

Figure 2.1: Flowchart of BEAMnrc and DOSXYZnrc processes (Adapted from EGSnrc Online manual).

2.1.2.1 BEAMnrc Code

The EGSnrc simulation process starts with the BEAMnrc code. The BEAMnrc code is used to model the radiation therapy units, characterize and simulate radiation beams, and finally calculate the phase-space data in a user defined plane. The phase-space contains information such as particle position, energy, direction cosine, and charge and history tag of each particle in the specified plane. The following is a brief step by step process of a BEAMnrc simulation outlined in Figure 2.1:
Step 1: Specify and build treatment units: The BEAMnrc code contains various geometric structures known as component modules. One must choose from these components to build a radiotherapy unit model. Some of the commonly used components are FLATFILT, JAWS, MLCS, PYRAMIDS, and SLABS. They can be used for modeling treatment unit components such as the flattening filter, secondary jaws, multi-leaf collimator, beam blockers, and slabs of desired material, respectively. Once all the desired components modules have been added in the order in which they appear in the treatment unit, the user can build the treatment unit from the compile command in the BEAMnrc program.

Two clinical radiation therapy units modeled in BEAMnrc for the research work in this thesis were the T780c Co-60 unit (Best Theratronics, Ottawa, Canada) and CL2100EX linac (Varian Medical Systems, Palo Alto, USA). As shown in the 3D illustration in Fig. 2.2, T780c Co-60 unit consists of a 2 cm diameter cylindrical source encapsulated into a lead and tungsten housing. The beam is collimated by the primary collimator and a series of lead adjustable collimators. The trimmer bars were removed from the physical Co-60 unit to accommodate the fan beam block collimator or the MIMiC collimator.

The opening for each of the adjustable collimator of the Co-60 is calculated based on the primary collimator opening and the field size (Fig. 2.3). For example for an adjustable collimator $i$, where $i$ is the number of the adjustable collimator varying from 1 to 4, half of the top ($t$) or bottom ($b$) opening can be calculated as:
where, $Z_{xcit,b}$ is the distance between the bottom of the source to the top or bottom of the adjustable collimator $i$. The subscript $x$ corresponds to the $x$-collimator. The $y$ collimator openings are calculated similarly, but with a different $z$-axis location. $Z_{pc}$ is the distance between the source and the top of the primary collimator. $X_{pc}$ is the distance from the central axis to the lateral edge of the primary collimator. $X_{c4,b}$ is half of the bottom opening of the 4th (lowest) adjustable collimator and $Z_{xc4,b}$ corresponds to the $z$-axis location from the bottom of the source to the bottom of adjustable collimator 4. $X_{c4,b}$ is calculated as following:

$$X_{c4,b} = \frac{FS}{2SSD} Z_{xc4,b}$$

where, SSD is source to surface distance and FS is field size.

The CL2100EX linac shown in Fig.2.4 consists of a thin target where photons are produced by bremsstrahlung emission and a forward peaked x-ray beam is produced. This x-ray beam is collimated by a primary collimator and filtered through the flattening filter to achieve a flat and symmetric dose distribution at a depth in the patient or phantom. The size of the photon beam is defined by secondary collimators or the MIMiC collimator (not shown in the representation). The model of the MIMiC collimator includes the tongue and groove design to minimize inter-leaf radiation leakage. Different views of the model of the MIMiC collimator are shown in Fig. 2.5. The validation of these modeled treatment units is presented in Chapter 3 and also by Joshi (2008).
Figure 2.2: 3D representation of the T780c Co-60 unit (Best Theratronics, Ottawa, Canada) in the EGSnrc-Beamnrc Monte Carlo system. The diagram has been adapted from Joshi (2008).

Figure 2.3: Schematic diagram of the configuration used for modeling Theratronic T780C Cobalt 60 in BEAMnrc Monte Carlo program. The setup shows a cylinder source with 2 cm diameter, primary collimator and four adjustable collimators.
Figure 2.4: 3D representation of the CL2100EX linac (Varian Medical Systems, Palo Alto, USA).

Figure 2.5: The model of the MIMiC collimator (Best Nomos, Pittsburgh, PA) in the EGSnrc-BEAMnrc Monte Carlo system.
Step2: User defined input file:  An input file is what contains the simulation parameters. In the previous step the steps to build a treatment unit were described; however, the geometry (e.g. size, shape, material composition, and location) must be defined by the user via input file. Besides geometry of the component modules, the input file also contains the source type (energy, location, shape, spectra etc.), the MC calculation parameters (number of histories, random number seeds, cut-off energies, variance reduction techniques, etc.) and finally the desired output quantities and their location (Kawrakow and Rogers 2003).

The source types used for this work were ISOURC1 (Isotropic Point Source), ISOURC3 (Interior Isotropic Cylindrical Source), and ISOURC21 (Phase Space Source). The ISOURC1 was used to model an isotropic point source positioned along the Z-axis (the same as the axis of the radiation beam). This source is particularly useful for modeling point sources used in linac machines. The ISOURC3 was used to model isotropic sources in a cylindrical geometry such as a Co-60 source. Finally, the ISOUCRC21 was utilized when a phase space file generated at some scoring plane was used as a source itself. The ISOURC21 is particularly useful for doing repeated simulations of lower portions of a treatment unit model, where upper part is not changed from simulation to simulation. By not having to transport the particles in the upper portion of the treatment head each time, the simulation time can be reduced significantly. The dose and fluence resulting from a phase space source are normalized by the number of initial particles in the original, non-phase space
source (i.e. the source that generated the phase space source) (Rogers et al., 2005).

**Step 3: Cross section data**: In the EGSnrc system, radiation transport is controlled by the information generated by the PEGS4 code. PEGS4 is a stand-alone program that is used to construct and store the precise material cross-section data and hence the probabilities of a particular interactions, which are then extracted to the media data files to be used by the EGSnrc system for MC simulations. The PEGS4 cross-section data for many commonly-used media are included in the EGSnrc installation package. These data are based on the density effect corrections reported in ICRU Report 37 (ICRU, 1984).

**Step 4: Simulation, Output, and analysis**: Once all input parameters have been defined, a simulation can be started from the BEAMnrc interface or from the operating system’s command prompt. As described previously, the output of a simulation is a phasespace file, which contains the interaction history of the particles, their fluence and their direction cosines. The fluences can be extracted using analysis software known as the BEAMDP.

2.1.2.2 DOSXYZnrc Code

The DOSXYZnrc code is another standalone code in the EGSnrc system, which simulates the photon and electron transport in volume elements and scores the dose deposition in specified 3D phantoms. As shown in Fig. 2.1, the DOSXYZnrc code utilizes the phasespace file from the BEAMnrc program as its radiation source. The user, however, can opt to use the other standard source codes available within the DOSXYZnrc code as the radiation source. The phantom geometries can be simple
homogenous water-equivalent material slabs or complex geometries with heterogeneous composition representing a patient (Walters et al., 2005).

2.1.3 Radiation Transport in the EGSnrc System

With any Monte Carlo code, the EGSnrc system requires detailed information regarding the interaction properties of particles and the media through which it travels. A particle history is started as soon as the particle is created with position and energy coordinates according to a specified source distribution. The particle travels a certain distance before undergoing an interaction. The choice of interaction type and the resulting energy transfer is governed by probability distributions that depend on the total interaction cross-sections. Depending on the type of interaction, secondary particles may also be created. These particles are also transported in a similar manner and this process continues until all particles have deposited their energy within the medium or left the geometry.

An electron can undergo hundreds of thousands of interactions as it slows down in a medium. If the electron transport is considered to be event-by-event, the computation time would make it impractical to do such a simulation. The EGSnrc code utilizes the condensed history technique to deal with this limitation. As described by Berger (1963), the condensed history method involves condensing a large number of individual interactions into a single step. Since in most cases a single interaction causes little change in the energy and direction of an electron, this method provides a reasonable approximation. The EGSnrc code has been shown to perform particle transport to within 0.1% accuracy with respect to analytical solutions to transport situations (Kawrakow, 2000; Seuntijens and Kawrakow, 2002). However, as with all Monte Carlo codes, the overall accuracy is limited by the accuracy of the underlying
cross-sections. The reader is referred to the work by Kawrakow and Bielajew (1998) for more details regarding the condensed history technique.

In the implementation of the condensed history technique, the EGSnrc uses energy cut-off values ECUT and PCUT, and production threshold energies AE and AP for secondary electrons and photons, respectively. A cutoff energy is necessary to ensure the Monte Carlo simulation does not continue for a long time. Transport of a particle is terminated when the particle’s energy falls below the cutoff. The parameters ECUT and PCUT are required to be greater than or equal to AE and AP since AE and AP serve as lower limits on ECUT and PCUT. AE and AP are defined when material data set is created using PEGS4 data. AE and AP represent the lowest energy for which the material data are generated. In other words, AE and AP determine the cutoff energy at which secondary particles are created and anything below these numbers would stop transport irrespective of the values of ECUT and PCUT. Typical values for these parameters are provided in the input file example presented in Appendix B.

2.1.4 Efficiency Enhancement and Variance Reduction Techniques

A full Monte Carlo simulation can be prohibitively time-consuming, especially when tens of millions of particle histories must be considered in order to achieve the desired precision. For this reason, most Monte Carlo codes employ efficiency enhancement and variance reduction techniques. Efficiency enhancing techniques (EET) introduce acceptable bias (approximations) to reduce simulation times (Sheikh-Bagheri et al. 2006). The condensed history technique discussed previously and the range rejection technique are some of the main EETs used in the EGSnrc code (Kawrakow and Rogers 2003, Sheikh-Bagheri et al. 2006).
Variance reduction techniques (VRT) reduce simulation times without introducing a systematic error. A true variance reduction technique improves the efficiency of a calculation while providing an unbiased estimate of the quantity of interest. The examples of the VRT are bremsstrahlung splitting, photon splitting and Russian roulette. In bremsstrahlung splitting, an electron is forced to produce many bremsstrahlung photons which are set in motion with appropriate adjusted (lower) weights and their energy is reduced by an amount equal to the energy of one of the emitted photons. The technique can save a lot of simulation time. The Russian roulette and particle splitting are used when the region of interest of a given application comprises only a small fraction of the geometry of the simulation. One photon (or an electron) can be “split” into many photons as it approaches a region of interest, these photons will later play “Russian roulette” as they leave that region to determine which one will survive, again with appropriate particle weight carried along to preserve overall physics of the interaction process. The readers are referred to Kawrakow and Rogers (2003) and Sheikh-Bagheri et al. (2006) for more details regarding the variance reduction and efficiency enhancing techniques.

2.1.5 Parallel Computing

As noted in the previous sections, the Monte Carlo simulations can be extremely time consuming. The EGSnrc system allows parallel computing so that a single simulation can be executed simultaneously on multiple computers in order to reduce the elapsed time from start to finish of the simulation. During parallel-processing in the EGSnrc, each computer is initially assigned with a small fraction of the total number of histories. The parallel-processing subroutine keeps track of how many particles have been simulated and how many are remaining. As soon as a CPU finishes a group of
histories, it is assigned with further number histories to execute until the required total number has been reached. When all histories have been completed, the individual output files from the different CPUs are combined to give a final output file. The simulations during this project were carried on 56 to 92 CPUs running on the PBS queuing software (PBS works, Altair Engineering, Detroit, MI, USA).

2.2 Co-60 Tomotherapy Treatment Planning

2.2.1 Current Process

The Co-60 tomotherapy treatment planning process is based on various in-house MATLAB based programs and open sources codes available for radiotherapy research and dose calculations. Figure 2.6 outlines the steps that are followed to generate conformal Co-60 plans. The current treatment planning is based on Monte Carlo method.

Figure 2.6 Flowchart of the Co-60 tomotherapy treatment planning process at the Cancer Center of Southeastern Ontario (CCSEO).
**Step 1: Export patient data from Eclipse TPS:** To investigate the clinical potential of Co-60 tomotherapy, actual patient data from the Eclipse treatment planning system (TPS) was used. The treatment planning data were exported from the TPS in Digital Imaging and Communications in Medicine (DICOM) format, which is the standard for communications between different diagnostic and therapeutic modalities. The exported files included the CT images containing patient geometry and anatomy information, the radiotherapy plan (DICOM RP) file, radiotherapy dose (DICOM RD) file, and the radiotherapy structures (DICOM RS) files.

**Step 2: Read and re-format patient data in CERR:** The dose calculations for patient treatment planning in this project are performed in the DOSXYZnrc component of the EGSnrc Monte Carlo program in order to take into account the patient heterogeneities. However, the DOSXYZnrc cannot read the DICOM format data exported from the Eclipse TPS. In order to read and re-format the DICOM data, an open source MATLAB based software called Computation Environment for Radiotherapy Research (CERR) program was used. The CERR used the dicom patient data and consolidated them into a MATLAB data file commonly referred as *planC* in CERR terminology. The planC file contains information such as the CT images, structure outlines, planning parameters, optimized dose distribution, and beam delivery parameters for the modality for which the plan was originally created for.

The *dicomrt_ctcreate* subroutine of the CERR was used to create a file called *CT phantom*, which is a phantom geometry and composition file required by
the DOSXYZnrc Monte Carlo program for dose calculations. The resolution of the CT phantom was $0.25 \times 0.25 \times 0.25 \text{cm}^3$.

**Step 3: Prepare dose calculation input files using a CCSEO program:** Considering a large number of pencil beams (between 1000-11000) used for dose calculations, the DOSXYZnrc input files were not created one by one using the DOSXYZnrc gui. Instead an in-house MATLAB program was created that can generate the input files automatically. The input files for the DOSXYZnrc program contained information such as the radiation source, number of particles, the CT phantom file location, isocenter, gantry angle, and the dose calculation efficiency parameters. The readers are referred to the online manual of the DOSXYZnrc program for a detailed list of parameters and their description.

**Step 4: Calculate pencil beam doses in the EGSnrc system:** As described previously, the DOSXYZnrc program can take a phasespace file as a source or the particles can be simulated from the original radiation source. To reduce overall dose calculations times in this project, the phasespace file for each of the MIMiC pencil beams were scored below the MIMiC using the BEAMnrc program, and these phasespace files were used as the primary radiation source file for the DOSXYZnrc program. Doses were calculated for each of the MIMiC defined pencil beams in the CT phantom. The jobs were parallelized on a large number of CPUs to reduce the calculation time. Currently a cluster of 96 CPUs is available at the CCSEO for Monte Carlo dose calculations; however, some of the initial work presented in this thesis was based on 56 CPUs. The PBS queuing software was used to manage the jobs.
Step 5: Compute total un-modulated dose distribution using a CCSEO program:

Once each of the pencil beam dose calculations were complete in the DOSXYZnrc program, total doses from all pencil beams from different directions were calculated using the in-house MATLAB software. For example, let $N$ be the total number of pencil beams with $n$ corresponding to the indices of the pencil beams ($1 \leq n \leq N$) and $M$ be the total number of voxels with $m$ representing the indices of each voxel ($1 \leq m \leq M$). If the dose calculated at a point from pencil beam $n$ to voxel $m$ per unit of energy fluence is represented as $d_{nm}$, then the total dose, $d_m^d$, delivered to that voxel by all pencil beams is calculated as:

$$
  d_m^d = \sum_n d_{nm} x_n
$$

(2.3)

where $x_n$ represents the beam-weights that describe the quantity of fluence delivered to that beam (Van Dyk, 1999). The beam-weights in the context of this project correspond to the irradiation time. For initial dose distribution, the beam weights for all these pencil beams were the same, therefore, a total unmodulated dose distribution is determined. Figure 2.7 shows an example of a single slice of the unmodulated dose distribution for a prostate case when equal weight is set to all beams used. This dose distribution is then optimized to conform doses to target, as will be discussed later in this section.

Step 6: Extract clinical structure outlines using a CCSEO program: The Co-60 tomotherapy treatment planning in this project was based on the clinical structures outlines used for actual patient treatment planning at the CCSEO. The structure contours were drawn by the radiation oncologists for targets and critical structures. The contours are defined according to the standards recommended by the Report 62 of the International Commission on Radiation Units and Measurements (ICRU).
Figure 2.7 Total dose for a prostate case prior to optimization. It is obtained by adding dose from each of the equally weighted beamlets calculated in DOSXYZnrc Monte Carlo program.

The tumour volume that can be identified visibly by looking at the CT image is defined as the gross tumor volume (GTV). The GTV does not fully take into account the whole tumour volume. Clinical practice and experience indicates that there is generally a microscopic spread of the disease, which is not seen on the CT image (Van Dyk, 1999). Therefore, the oncologists will often draw another contour, known as the clinical target volume (CTV), slightly outside of the GTV. These two contours are sufficient if the patient remains stationary during the treatment and there are no set-up errors. However, this is not possible as patient motion and set-up errors are unavoidable. Hence another contour, defined as the planning target volume (PTV), is included to account for errors that tend to shift the CTV. Along with targets, organs at risk (OAR) are also drawn. The purpose of defining such contours is to limit the dose to healthy organs, which may otherwise be damaged.

The clinical contours were exported from the Eclipse and were saved into a matlab data file using the CERR. The contours to be used for Co-60 tomotherapy planning were extracted using an in-house MATLAB program.
Step 7: Define dose objectives for optimization: Each contoured structure is prescribed with a desired dose. However, since it is impossible to achieve a dose distribution with exact dose prescriptions, a tolerance of 5% is specified to allow the dose to vary slightly. After defining structures and prescribing doses, importance factor for each of the structures considered during optimization are provided. Optimization parameters such as the step size and number of iterations are also provided.

Step 8: Optimize doses using the CCSEO Planning System: Using the initial selection of beam weights, the treatment planning system computes a dose distribution. The calculated dose distribution is then compared to the desired dose distribution using an objective function. The in-house inverse planning system uses a quadratic objective function to evaluate the dose distribution achieved at each iteration. The quadratic function computes the sum of the squares of the difference between the prescribed and the current dose (Van Dyk, 1999) and is defined as:

\[
F_{obj} = \sum_i (d^p_m - d^d_m)^2
\]  

where, \(d^p_m\) is the prescribed dose and \(d^d_m\) is the dose delivered to each voxel, which can be computed using Eq. 2.3. The goal is to minimize the objective function. The minimization of the objective function is achieved using the gradient based optimization method. In general, the gradient optimization algorithms are performed in two steps. First, a search direction along which the objective function decreases is established and second, the length of the step
along that direction is determined. At each iteration, comparison is made the desired dose distribution; if the calculated dose distribution does not satisfy the specified criteria, new beam weights are used to recompute the dose distribution. The optimization is terminated if the new dose distribution satisfies the given criteria. Otherwise, another set of beam weights are computed, and the optimization continues to iterate until a desired dose distribution is achieved. The readers are referred to the work by Chng (2005) for detailed description of the inverse planning system used in this thesis.

**Step 9: Create MLC leaf sequence files for dose delivery:** The beam weights for each of the pencil beams determined during optimization are then used to calculate dwell times. This is done using the dose rate for the Co-60 source. As will be described in Chapter 3, the deliveries for this work were carried out with the pencil beam block. Therefore when irradiating, the gantry of the Co-60 unit remained stationary at 90 degrees and the phantom rotated and translated in front of the beam, mimicking an intensity modulated fan beam similar to tomotherapy. The effective intensity is modulated by varying the dwell time at each translational step. The phantom is allowed to dwell for a longer period when a higher dose is required and a shorter period when a lower dose is required.

### 2.2.2 Future process

Although MIMiC collimator is available for dose delivery studies, current dose delivery studies have been based on a pencil beam block collimator as mentioned in Step 9 of the previous section and also Chapter 3. One of the reasons is that the finite size pencil superposition method used for tomotherapy dose calculations is not accurate for simultaneously opened MLC leaves. An aperture superposition model was proposed
in this thesis to calculate dose for tomotherapy deliveries using the MIMiC MLC. Although the details of these investigations are presented in Chapter 5, a summary of modeling details is provided here to keep this chapter self-contained.

According to the pencil beam superposition model, the total absorbed dose to a point \((x, y, z)\) is the sum of the doses delivered by multiple beamlets such as \(P_1, P_2... P_n\), where \(P_n\) is the dose delivery by pencil \(n\). The generalized Eq. (2.5) is same as Eq. (2.3).

\[
D = \sum_{n=1}^{N} w_n P_n (x, y, z) \tag{2.5}
\]

In this equation, \(N\) is the number of beamlets and \(w_n\) is a relative weight of the \(n\)th beamlet. For simplicity, the beamlet weights in this example are considered to be the same.

The model proposed in Chapter 5 does not sum individual pencil beams; rather it adds up the ‘apertures’. An aperture is defined as series of consecutive opened leaves with no closed leaves in between. A full modulated fan beam is considered as the sum of apertures \((A_1, A_2... A_n)\). The total dose absorbed to point \((x, y, z)\) is specified by Eq 2.6.

\[
D = \sum_{n=1}^{N} w_n A_n (x, y, z) \tag{2.6}
\]

The apertures and pencil beams in Chapter 3-5 are calculated in Monte Carlo. While this method is accurate, it is time consuming. Hence, Chapter 6 proposes a dedicated convolution method for Co-60 source.

The aperture superposition model will be incorporated into the leaf segmentation model, which currently is based on the pencil beam superposition model.
2.3 References


Chapter 3: Delivery Validation and 2D Co-60 Tomotherapy Treatment Planning

A version of this chapter with content formatted for publication has been accepted for publication as: Dhanesar S, Darko J, Joshi C J, Kerr A, and Schreiner L J Cobalt-60 tomotherapy: clinical treatment planning and phantom dose delivery studies Med. Phys in press.

3.1 Abstract

Investigations have shown that a Cobalt-60 (Co-60) radioactive source has the potential to play a role in intensity modulated radiation therapy (IMRT). In this paper, conformal dose delivery potential is evaluated by delivering conformal dose plans on a cylindrical homogeneous phantom containing clinical structures similar to those found in a typical head and neck cancer. Also, the clinical potential of Co-60 tomotherapy is investigated by generating 2D clinical treatment plans for head and neck (H&N) and prostate anatomical regions. These plans are compared with the 6 MV based treatment plans for modalities such as linac-based tomotherapy and broad beam IMRT, and 15 MV based conformal radiation therapy (3DCRT).

For experimental validation studies, clinical and non-clinical conformal dose patterns were delivered on circular, homogeneous phantoms containing GafChromatic film. For clinical planning study, dose calculations were performed with the EGSnrc Monte Carlo program, where a Theratronics 780C Co-60 unit and a 6 MV linear accelerator were modeled with a MIMiC binary multi-leaf collimator. An in-house inverse treatment planning system was used to optimize tomotherapy plans using the
same optimization parameters for both Co-60 and 6 MV beams. The IMRT and 3DCRT plans for the clinical cases were generated entirely in the Eclipse treatment planning system based on in-house IMRT and 3DCRT site specific protocols.

The doses delivered to the homogeneous phantoms agreed with the calculations, indicating that it is possible to deliver highly conformal doses with the Co-60 unit. The dose distributions for Co-60 tomotherapy clinical plans for both clinical cases were similar to those obtained with 6 MV based tomotherapy and IMRT, and much more conformal compared to 3DCRT plans. The dose area histograms showed that the Co-60 plans achieve the dose objectives for the targets and organs at risk (OAR).

These results confirm that Co-60 tomotherapy is capable of providing state-of-the-art conformal dose delivery and could be used for the treatment of targets in both small and larger separation anatomical regions.

3.2 Introduction

Cobalt-60 teletherapy (Johns et al, 1952a; Johns et al, 1952b) was the most popular radiation therapy unit from mid-1950s to 1960s and introduced high energy photon radiation treatment. In the past five decades, tremendous research and innovation has gone into improving radiation therapy, particularly, through the development of intensity modulated radiation therapy (IMRT) techniques. Most of these advances have been implemented on medical linear accelerators (linacs). The modernization of Co-60 units has been minimal. The lack of significant development in Co-60 radiation therapy technology may be attributed to some perceived disadvantages such as relatively lower photon energy (average 1.25 MeV), lower radiation output, lower beam penetration, and larger beam penumbra (Glassgow, 1999; Schreiner et al, 2003). This lack of development resulted in the nearly complete fall off of Co-60 units.
in clinics in the developed world. It has also introduced reluctance in some clinics in the
developing world to acquire basic Co-60 units, since they perceive this technology as
having limited potential for clinical advancement.

A number of authors have indicated the importance of Co-60 units due to their
simple and robust design, and requirement for minimal maintenance and technical
expertise to operate as compared to the technologically complex linacs (Schreiner et al,
2003; Warrington and Adams, 2002; Van Dyk and Battista, 1996). These authors have
pointed out that if the Co-60 unit was modernized with state of the art devices, it may be
able to provide similar quality of radiation therapy as provided by linacs. Further
literature suggests that the beam penetration and penumbra problems faced with Co-60
become negligible when one goes to rotational radiation therapy, where a large number
of radiation beams from multiple directions are used (Johns and Cunningham, 1983;
Laughlin et al, 1986; Sternick et al, 1997). Laughlin (1986) and Sternick (1997) have
shown that the advantages of high-energy beams decrease and become negligible with
advanced multi-beam treatment modalities.

In recent years, there has been increased interest in Co-60 based 3D conformal
radiation therapy (3DCRT) and IMRT (Fox et al, 2008). Adams and Warrington (2008)
compared Co-60 plans based on customized blocks to 6 MV based IMRT plans. Fox et
al (2008) performed computer based treatment planning analysis on Co-60 broad beam
IMRT, comparing plans with those obtained with IMRT based linac beams. They
specifically focused on how increasing the number of beams can improve the
conformality of the Co-60 plans.

These observations have encouraged our group to investigate Co-60 based
tomotherapy. Tomotherapy is a rotational implementation of IMRT using a fan beam
and can generate highly conformal plans (Mackie and Holmes, 1993; Mackie, 2006; Oliver, 1999). Clinical implementation of tomotherapy is available in two modes: serial and helical tomotherapy. In serial-tomotherapy, also referred to as sequential-tomotherapy, a volume is irradiated with a narrow rotating beam of radiation (Carol, 1995; Beavis, 2004). Modulation of the fan beam intensity is achieved by a fan beam multi-leaf collimator (MLC) that consists of a series of tungsten leaves. These slices are then translated sequentially over the volume. In helical tomotherapy (Mackie et al., 1993), the radiation source revolves continuously around the patient while the patient is translated smoothly through the modulated fan beam. Cadman et al (2006) has done a study on Co-60 based tomotherapy and proposed a collimator design for multi-slice helical tomotherapy, which provides a way to reduce overall treatment time by allowing multiple planes to be treated concurrently. Cadman and group have also investigated optimal beam parameters, plan quality, and treatment times based on computer modeling (Cadman, 2007; Cadman and Bzdusek, 2011).

The focus of our research has been based on serial tomotherapy since it is easy to implement in a laboratory-setting. Previous studies by our group confirmed a strong potential for Co-60-based tomotherapy, using multi-field techniques for simple non-clinical dose patterns (Schreiner et al, 2009), as well as for homogeneous phantoms containing clinically relevant structures (Joshi et al, 2009). In addition, imaging studies have shown a potential for adaptive radiation therapy via image guided techniques such as Co-60 megavoltage fan beam CT, Co-60 tomosynthesis, and Co-60 cone beam CT (Schreiner et al 2003; Schreiner et al 2009; MacDonald et al, 2009; Rawluk et al, 2010). Studies have also been done on methods of improving the dose output of Co-60 machines (Joshi et al, 2008). The dose output findings indicate that the problem of low
dose rate, arising from the activity of the source, can be solved if the unit is redesigned to include multiple sources and/or a different shaped source with an increased source activity.

In this work, we report further developments of Co-60 based tomotherapy investigations. Two key elements are discussed:

- The first element concentrates on conformal dose irradiations obtained on a phantom translated with computer-control through a collimated pencil beam. Previous computer simulation work has shown conformal Co-60 delivery is possible (Fox et al., 2008; Cadman and Bzdusek, 2011); however, the previous work was not validated by measurements. We extend this computational work to conformal measurements. Previous reports (Schreiner et al., 2009) by our group on Co-60 tomotherapy conformal dose delivery have been for non-clinical structures. In this paper we present dose delivery results obtained for a cylindrical homogeneous phantom containing structures similar to those found in a typical head and neck cancer.

- The second element of this paper concentrates on treatment plans of heterogeneous cases obtained with pencil beams from the multi-leaf collimator instead of the scanning pencil beam. Although previous planning results on phantoms containing structures similar to prostate and head and neck looked promising, they were incomplete (Joshi et al., 2009). The H&N plan presented by Joshi et al. (2009) was not based on real anatomy, via CT data, rather the dose was calculated on a homogeneous water phantom. Also the previous modeling for the prostate case did not simulate a full
tomotherapy implementation since only 21 equally spaced gantry angles could be simulated due to the limitation of our Monte Carlo dose engine at that time. This number is significantly smaller than 50 plus angles typically used for tomotherapy plans. In this paper new results are presented for investigations on two clinical cases (prostate and H&N cancer) considering all anatomical and tissue heterogeneities using CT datasets and a Monte Carlo dose calculation method. These sites provide anatomical regions with a larger separation (pelvic region) and complex target geometry that is proximal to critical structures (H&N). The Co-60 treatment plans are compared with the three linac based techniques: 6 MV linac-based tomotherapy, broad beam IMRT, and 15 MV based 3DCRT.

We believe that a continual flow of reports establishing the potential of Co-60 based IMRT will motivate developers to modernize Co-60 clinical units and make them attractive throughout the world. This may make modern conformal techniques more widely accessible. Co-60 tomotherapy has the potential to reinvigorate Co-60 in the developed world. This may also remove the reluctance of clinics in the developing world to purchase basic Co-60 units because vendors will be seen to be committed to continuous support and development of a sustained Co-60 technology.

3.3 Material and Methods

The experimental methods discussed in this section enabled us to extend the computational work previously reported in the literature (Fox et al, 2008; Cadman and Bzdusek, 2011; Schreiner et al, 2009) to experimental validation via homogeneous
phantoms irradiations. The homogenous phantom work is then extended to patient treatment planning based on Monte Carlo dose calculation methods to investigate the potential of Co-60 tomotherapy for clinical cases, such as the treatment of prostate and head and neck cancer.

3.3.1 Benchtop Tomotherapy Apparatus with Pencil Beam Block Collimator and MIMiC

All irradiations were performed on a Co-60 T780C unit (Best-Theratronics, Kanata, Canada) modified by the addition of a special purpose computer-controlled rotate translate benchtop apparatus, as shown in Fig 3.1. This first-generation tomotherapy test system includes a rotation-translation stage, which can move a phantom to be irradiated through a 1×1 cm\(^2\) Co-60 pencil beam. The pencil beam is generated using the block collimator as shown in the diagram. IMRT is performed in a single slice by varying the velocity of the phantom as it moves through the pencil beam. By this simple approach, we are well able to imitate the beam delivery from a multi-leaf collimator (MLC).

The use of the custom pencil beam block collimator was necessary at this stage because the available MIMiC MLC (Best Nomos, Pittsburgh, PA) could not be auto controlled from outside the treatment room. Although the MIMiC was fully implemented in the in-house treatment planning system, as will be discussed later in this paper, it was not ready for full 2D dose deliveries as this would have required the operator to enter the treatment room each time a different MLC sequence was delivered, which is not practical since the number of MLC sequences in tomotherapy deliveries can be very large. Therefore, a pencil beam collimator was used for the delivery validation work.
Figure 3.1 The first generation in-house tomotherapy bench-top apparatus incorporated onto a Theratronics T780C Co-60 unit with a cylindrical film phantom made from Lucite glass slabs. The pencil beam is generated by a block collimator. In this setup, the pencil beam stays stationary at all times and the phantom is positioned on a system enabling computer controlled rotation and translation across the pencil beam, mimicking serial 3D tomotherapy delivery.

3.3.2 Treatment Planning and dose calculation method

The treatment planning of the tomotherapy delivery in this study was carried out using an in-house inverse treatment planning program (Chng, 2005). This program, written in MATLAB (MathWorks, Natick, MA, USA), is based on the conjugate gradient active set algorithm (Hristov and Fallone, 1997). In this implementation, all beamlets that do not go through the targets are set to a zero beam weight during the first step and only those remaining are then optimized via an iterative approach. This reduces the overall optimization time for the tomotherapy treatment planning. In the current implementation of the software, the first step is to calculate the dose in the phantom from equally weighted pencil beams and then to optimize the pencil beam weights until desired dose distribution is achieved. The dose for individual pencil
beams is calculated via two different methods depending on the type of phantom: Milan-Bentley (Milan and Bentley, 1974) for the plans assuming homogenous phantoms and Monte Carlo (Rogers and Bielajew, 1990; Han et al, 1987; Rogers et al, 1998) for plans based on heterogeneous phantoms, i.e. representing real human anatomy via CT data sets. The Milan-Bentley model is based on diverging fan lines that radiate from a source and which intersect depth lines located at selected distances below the phantom. The dose distributions are obtained by rapidly sampling measured data sets consisting of a central axis component and an off-axis. This simple approach was used for calculating dose for homogeneous, circular phantoms used for conformal dose delivery on film in order to validate the optimization aspects of the in-house inverse treatment planning system. However, because this method has limitations for calculating dose in heterogeneous medium of non-circular shapes, a Monte Carlo method has been adapted for calculating dose in heterogeneous medium, particularly for phantoms defined by the patient CT data sets. The MIMiC collimator was used for patient treatment plans.

The T780C Co-60 unit (Best Theratronics, Kanata, Canada) and CL2100EX 6 MV linear accelerator (Varian Medical systems, Palo Alto, USA) were modeled using the BEAMnrc Monte Carlo code.\textsuperscript{31-33} This code is part of the EGSnrc Monte Carlo code system that has been widely used for modelling and calculating doses from the radiation therapy units (Rogers and Bielajew, 1990; Han et al, 1987; Rogers et al, 1995; Rogers et al, 1998; Mora et al, 1999; Ding, 2007; Walters et al, 2005). The modeling of the cobalt and linac units was based on the manufacturer’s specifications and included a mounted MIMiC MLC. The Monte Carlo design of the MIMiC system included the details of the tongue and groove geometry. The Co-60 unit consisted of a 2 cm diameter cylindrical source, primary collimator, jaws, and the MIMiC collimator. A Co-60 source spectrum
(labelled as ‘bareco60’in BEAMnrc) with 1.17 and 1.33 MeV mono-energetic $\gamma$–photons was used in the simulations (Han et al, 1987; Rogers et al, 1998; Mora et al, 1999). The 6 MV linear accelerator consisted of an electron target, primary collimator, flattening filter, monitor chamber, jaws, and the MIMiC collimator. The global electron cut-off energy (ECUT) of 0.521 MeV, global photon cut-off energy (PCUT) of 0.010 MeV, and the maximum fractional energy loss/step (ESTEPE) of 0.25 (i.e. 25%) were used in all simulations. The boundary crossing and electron step algorithms used were PRESTA I and PRESTA II, respectively. All the BEAMnrc simulations were performed with uniform bremsstrahlung photon splitting (using a splitting factor of 10) and with the Russian Roulette variance reduction technique enabled. For all other parameters, the default BEAMnrc and EGSnrc settings were used (Rogers et al, 1995).

A phase-space plane was defined below the MIMiC collimator, located at 40 cm from the source for the Co-60 unit and 51 cm from the target for the 6 MV unit. The phase space file contains an output from the BEAMnrc simulation that records for each particle reaching the scoring planes its charge, energy, position, direction, and interaction history. Separate phase space planes were scored for each of the individual 20 beamlets (opened to form a 1x2 cm$^2$). The central field sizes from 2x2 cm$^2$ to 20x2 cm$^2$ with an increment of 2 cm were also simulated for both Co-60 and 6 MV units. Note that in all of our simulations we assume we open 2 sets of MIMiC leaves to get 2x1 cm$^2$ pencil beams. Although the smallest field size achievable with the MIMiC collimator is 1x1 cm$^2$, 2 sets of 1x1 cm$^2$ leaves were used so that the center of the 2x1 cm$^2$ field corresponds to the central axis.

In order to keep statistical error to within 1%, up to 8 billion histories were simulated depending on the beam size. Since the Monte Carlo method is
computationally intensive, a parallel computing cluster consisting of 56 CPUs running PBS queuing software (PBS works, Altair Engineering, Detroit, MI, USA) was implemented and used to reduce the overall calculation time.

The DOSXYZnrc (Watlers et al., 2005) code, also part of the EGSnrc Monte Carlo code, was used to generate the 3D dose data. The DOSXYZnrc code required two inputs for the dose calculations: the phase space data for each beamlet of the MIMiC and a data file containing the material information, dimensions and position of the CT phantom in which dose was to be calculated. The DICOM CT datasets for clinical cases were converted to the input format required by DOSXYZnrc using the DICOM-RT library available in the radiation therapy treatment planning interface program called Computational Environment for Radiotherapy Research (CERR) (Deasy et al., 2003). Note, for this work the dose optimization was limited to a single slice of thickness 2.5 mm because of computing power limitations. Therefore, the phantom was positioned so that the center of the desired CT slice (z coordinate) was aligned with the isocentre of the machine (80 cm for the Co-60 unit and 100 cm for the 6 MV unit). The positioning of the CT phantom in x and y directions corresponded to the x and y isocentre coordinates used in the Eclipse Treatment planning systems (Varian Medical Systems, Palo Alto, CA), for the cases also planned using broad beam IMRT techniques.

For each beamlet, 750 million particles were simulated in DOSXYZnrc with a voxel resolution/grid spacing of the calculated dose distribution of 2.5 x 2.5 x 2.5 mm³. The 2D dose distributions from each of the 20 beamlets (or less for smaller targets) of the MIMiC were calculated individually from 50 equidistant orientations (angles). The doses from the individual beamlets were then summed.
3.3.3 Pencil Beam Block and MIMiC MLC Measurements

Various measurements were performed to commission the pencil beam block based on the Milan-Bentley method and to validate the accuracy of the Monte Carlo modeling for predicting the actual Co-60 delivery.

3.3.3.1 Pencil Beam Block Commissioning

The first set of measurements were used to commission the pencil beam block for the homogeneous dose deliveries. As required by the Milan-Bentley method, several measurements were performed for the 1x1 cm$^2$ pencil beam of radiation in tissue-equivalent materials at varying depths. These measurements were performed in the Blue Phantom water tank (IBA Dosimetry, Germany) using an ion chamber and a diode detector. The ion chamber used for these measurements was a small cylindrical, type CC01 (IBA Dosimetry) chamber with an active volume 0.01 cm$^3$. The diode detector used was an electron field diode detector (IBA Dosimetry) with an active area measuring 2 mm in diameter. The dose measurements were taken at several different source-to-surface distances (SSDs). At each SSD, the percent depth dose (PDD) was measured at 1 mm interval. To determine the off axis dose, in-plane and cross-plane profiles were obtained at each SSD at depths 0.5, 1, 2, 5, 10, 15, and 20 cm. The data was stored in a look-up table as per the Milan-Bentley method.

3.3.3.2 MIMic Collimator Model Validation

A validation study for the Monte Carlo modeling of the MIMiC MLC was also performed. Doses under the MLC set to define 2x2 cm$^2$ to 20x2 cm$^2$ fields were calculated along the central axis using the Monte Carlo code and then measured. From the phase space data, the 3D dose was calculated in a 30×15×35 cm$^3$ water phantom.
using the DOSXYZnrc Monte Carlo code. A voxel resolution of 2.5x2.5x2.5 mm³ provided adequate dose resolution for both percentage depth dose (PDD) and profiles (cross-plane and in-plane) for the various field sizes and depths. The corresponding dose measurements were obtained in a water phantom (IBA Dosimetry) using a small ion chamber (IBA Dosimetry) chamber with an active volume of 0.01 cm³.

3.3.4 Conformal Delivery on Homogeneous Phantoms

3.3.4.1 Treatment Planning of Test Phantom

Figure 3.2 shows an example of a cylindrical test phantom that contains treatment planning structures resembling a simple head and neck case. The clinical target volume (CTV) represents the target to be treated, and the left and right nodes represent the posterior nodes where the disease has spread. The CTV was prescribed to a dose of 300 cGy [to accommodate the response of the GafChromic films (International Specialty Products, NJ, USA)] and the left and right posterior nodes were prescribed to 95% and 65% of the CTV, respectively. The spinal cord represents the organ at risk (OAR) to which the maximum dose was limited to 57% of the CTV dose. Both the CTV and OAR were assigned with high relative importance factors to achieve the treatment planning objectives. A 2D tomotherapy distribution was generated using 31 pencil beams from 51 equidistant orientations similar to that typically used in commercial applications of tomotherapy. Note that 31 pencil beams here are actually 31 translational steps that the film phantom would take while moving across the stationary beam defined by the pencil beam block collimator. A variation of ±5% between the desired dose and optimized dose was considered acceptable.
Figure 3.2 Treatment planning structures for the test head and neck case. The CTV is the primary target and the right and left nodes are secondary targets. The cord is an organ at risk.

3.3.4.2 Conformal Dose Measurements

The conformal dose distributions were experimentally verified on a cylindrical, homogenous polystyrene phantom by irradiating GafChromic EBT film. Since the dose response of the film is non-linear (Cheung et al, 2005; Devic et al, 2004) film was first calibrated using 12 rectangular pieces of film with dimensions 11 cm x 11 cm. These pieces were placed between solid-water slabs in a direction perpendicular to the radiation beam and irradiated with known doses uniformly with a radiation beam of field size 10 x 10 cm$^2$. Each film contained a fiducial mark to identify and maintain the same orientation. The optical densities were obtained with an EPSON Expression 10000XL (Long Beach, CA) flatbed scanner. A paper mask was used so that all films could be scanned in the same orientation and at the same position on the scanner bed. Scanning was done 24 hours after irradiation to minimize post-coloration effect of the GafChromic film (Cheung et al, 2005; Devic et al, 2004; Devic et al, 2005). The calibration procedure was performed four times to ensure consistency across trials. The validation film was cut into a circular shape with a diameter 19.2 cm similar to the film phantom shown in figure 1 and was placed at the centre of the phantom. Scanning of
this film was done in a similar manner as the calibration film procedure described above.

### 3.3.5 Treatment Planning Incorporating Tissue Heterogeneities

In order to further investigate the clinical potential of Co-60 tomotherapy, planning studies were extended that incorporated tissue heterogeneities from patient CT data sets. Two cases considered for this study were head and neck (H&N) and prostate. As mentioned previously, although full 3D dose calculations were performed using the Monte Carlo modelling, dose optimization was performed on a single CT slice only due to computing power limitations. The Monte Carlo dose calculations were performed for equally weighted pencil beams for a desired number of orientations. The doses were then optimized using the in-house inverse planning software based on a set criterion and clinical contours. Figures 3.3a-b show the CT images for the H&N and prostate anatomy, respectively, containing the appropriate treatment planning structures. The structure contours are the original contours used in the clinical IMRT plans. The structure information was extracted using an in-house MATLAB based program and were then incorporated into the inverse planning program for tomotherapy treatment planning with Co-60 and 6 MV beam data.

The target structures in the H&N case (Fig 3.3a) were PTV70, PTV63, and PTV56. The planning target volume (PTV) is obtained by adding a margin of 5 mm around the CTV to account for uncertainties arising from the setup error or motion of the patient. Note that the CTV contours are not shown in Fig 3.3a) to maintain the clarity of the figure. The OAR structures were the spinal cord, mandible and left and right parotid glands. The remaining area was considered as normal tissue. A dose of 70 Gy was prescribed to treat the primary cancer (PTV70) while doses of 63 Gy and 56 Gy
Figure 3.3 a) Transverse CT slices displaying a) H&N anatomy and b) prostate anatomy. The contours represent targets (e.g. PTV70, PTV63, PTV56 for H&N case and PTV for the prostate case), critical structures (e.g. spinal cord, mandible, and parotids for H&N case and rectum, bladder, and femurs for prostate case), and healthy tissue (body).

...to the posterior neck nodes (PTV63 and PTV56) were considered adequate for treatment of the nodal disease. The dose to the spinal cord was limited to a point dose maximum of 40 Gy. It can be seen from Fig 3.3a) that PTV70 is overlapping the PTV63, PTV56, mandible, and parotids. During optimization high priority was assigned to the targets in order to ensure sufficient tumour coverage. Due to this, mandible and parotids glands were allowed to receive higher doses than those recommended by the QUANTEC (2010) as long as the doses to these structures did not exceed those achieved with the clinical IMRT plan approved by the attending oncologist.

In the prostate case, the prostate was outlined as the CTV while the rectum, bladder, and right and left femoral heads were outlined as the OARs (see Fig 3.3b). The PTV is obtained by expanding the CTV by 5 mm to account for setup uncertainty and patient motion. The rest of the volume was considered as normal tissue. A dose of 78 Gy was prescribed to the prostate. Only a portion (20%) of the rectum was allowed to
receive a maximum dose of 70 Gy or more. The upper limit of the volume receiving a maximum dose of 70 Gy or more for other OAR structures was 20% to the bladder and 60% to the right and left femoral heads. The main reason for allowing some portions of the rectum and bladder to receive high doses was because there is some overlap between the PTV and these structures.

The Co-60 and 6 MV distributions were generated using the same in-house treatment planning objectives to perform a true comparison between the two energies. The IMRT and 3DCRT plans were generated in the Eclipse treatment planning based on in-house IMRT and 3DCRT clinical protocols for H&N and prostate cancer. These were the actual plans used for patient treatments. The dose prescriptions for IMRT cases and other dose objectives were similar to the tomotherapy dose objectives stated previously. Unlike tomotherapy plans, these Eclipse plans were prepared by optimizing the full 3D volume. For direct comparison to the tomotherapy plans, the dose distribution for the slice of interest was extracted from the 3D dose distribution. Both the H&N and prostate clinical IMRT plans were generated using 7 fields, 6 MV sliding window techniques in Eclipse. The H&N 3DCRT plan was generated using a multi-phase lateral parallel opposed pair (POP) technique with a cord shield at 40 Gy and a posterior neck boost from two lateral electron beams. The 3DCRT plan for the prostate was a standard 15 MV 4 field box.

3.4 Results and Discussions

The work presented in this section is intended to show the potential for Cobalt conformal delivery with tomotherapy approach through irradiations on homogeneous phantoms, as well as a treatment plan comparison between Co-60 tomotherapy and commonly used radiation therapy modalities in the cancer clinics. In order to make the
report flow better, the discussion of each experiment is provided as the results are reported.

### 3.4.1 Pencil Beam Block Commissioning

Dose measurements for the 1x1 cm\(^2\) pencil beam were performed in order to commission the in-house inverse planning system for the dose delivery studies on homogeneous cylindrical phantoms. Two measurement techniques were used to measure central axis and off axis doses at several SSDs. The data were stored in a look up table, which was accessed by the inverse planning system during optimization. Figure 3.4 shows a comparison of the PDD curves (a) and profiles (b) obtained with an ion chamber and a diode detector at an SSD of 80 cm. Both of the measurements are within 1.9% agreement of each other. Since the error bars were very small, they are not displayed on the PDDs and profiles.

### 3.4.2 Validation of the Monte Carlo Model of the MIMic Collimator

The data for the validation of the Monte Carlo results under the MIMiC MLC are shown in Fig 3.5. Figures 3.5a and 3.5b show the simulated and measured dose profiles and PDD curves for a 2x2cm\(^2\) field size obtained by opening two pairs of central leaves. The outer profiles are in the crossplane direction and the inner profiles are in the inplane direction; both profiles are taken at 0.5 cm depth of water. Figures 5c and 5d show similar results for 20x2cm\(^2\) field size, which is obtained by opening all leaves of the multi-leaf collimator. The profiles for both fields were individually normalized to 100% at central axis. The dose profile results matched within 1 mm in the high dose–high gradient region and within 2% in high dose–low gradient and low dose–low gradient regions. A comparison of the simulated and measured PDD curves shows agreement.
Figure 3.4 1x1 cm$^2$ pencil beam measurements with various measurement techniques (a) a comparison of the PDD at an 80 cm SSD and (b) a comparison of the profile at 80 cm SSD and 5 cm depth.

Figure 3.5 Relative dose profiles and percentage depth dose comparisons between Monte Carlo calculations (black dashed lines) and ion chamber measurements (solid red lines) for 2x2 cm$^2$ (a-b) and 20x2 cm$^2$ (c-d) field sizes of Co-60 fields defined by the MIMiC MLC. The outer profiles for both fields correspond to the crossplane direction of the field and the inner profiles correspond to the inplane direction of the field. Data for both fields were individually normalized to 100%.
within 1% beyond the build-up region. Similar agreement was obtained for all other field sizes falling between 2x2cm2 and 20x2cm2. The 6 MV MIMiC Monte Carlo data could not be validated with the measurements as the MIMiC was not installed for use on the CL2100EX unit. However, the depth value for the maximum dose and penumbra width were consistent with those typically obtained for a 2x2cm2 6 MV beam.

3.4.3 Homogeneous Phantom Treatment Plan and Dose Delivery Test

As noted previously, doses to a large number of differently shaped targets were optimized with the inverse treatment planning system to determine whether the program is able to generate accurate dose distributions. Figure 3.6a shows the conformal dose distribution generated for the head and neck structures in a cylindrical homogeneous water-equivalent phantom. Figure 3.6b shows the corresponding film measurement, which visually shows a high degree of dose conformity. The quantitative results are presented in Figures 3.6c and 3.6d. Figure 6c compares the dose area histograms (DAHs) of the calculated and delivered dose. A majority of the structure volumes show a good agreement with the delivery; slight differences are seen in the nodal regions. These two distributions are further studied using a gamma map (Low et al, 1998; Low et al, 2003) of the simulated and delivered dose distributions. A gamma value less than or equal to 1 suggests that the agreement between delivery and simulation is within the specified criteria. While a common clinical criterion for gamma analysis uses a difference of up to 3% in dose and 3 mm in distance for 95% of the volume, we have used a slightly less stringent gamma criteria of 4%, 4 mm at this preliminary stage. These relaxed criteria have been adopted in part because the planning and delivery in this particular experiment are based on an initial crude Milan-Bentley beam model and a simple proof of principle delivery system. As shown in Fig 3.6d, over 95% of the pixels
were below gamma of 1, indicating a successful dose delivery. We also performed analysis using the 3%, 3mm gamma criteria to see how far our experimental deliveries are from clinically acceptable deliveries. We were able to get 87% of the pixels to be below gamma of 1, which we believe is acceptable at this stage considering the limitations of the dose calculation method and the dose delivery system.

Figure 3.6 (a) Co-60 tomotherapy dose distribution simulated using in-house inverse treatment planning system for head and neck test phantom using a tomotherapy approach. (b) GafChromic film measurement of the delivered plan, (c) dose area histogram comparison between the delivered and the simulated doses, and (d) gamma comparison between the delivered and simulated dose distribution using 4%, 4mm criteria.
Measurements were also performed with the MIMiC collimator to show that the pencil beam block with computer-controlled phantom can truly mimic an MLC delivery. Figure 3.7 shows two examples of the MIMiC deliveries in which selected MIMiC leaves were opened one at a time to deliver a modulated fan beam. The schematic diagram of the MIMiC collimator on each profile shows which leaves were opened. The dose was delivered to radiochromic film and comparisons are made with the fan beam calculated using the FSPB method (Bourland and Chaney, 1992), where individual leaf doses are calculated using a Monte Carlo technique. Other than the penumbral regions, the agreement between the calculated and delivered dose is 2%. Penumbral dose differences are up to 3.5%.

Figure 3.7 Examples of deliveries in which select MIMiC leaves were opened one at a time to deliver modulated fan beams. The schematic diagram of the MIMiC collimator on top of each profile shows which leaves were opened. In a) all open leaves have the same beam weights. In b) open leaves have differing beam weights. In the schematic diagram of the MIMiC collimator, white squares represent closed leaves; light and dark gray squares represent opened leaves (dark gray means MLC leaves are opened for the longest period).
These results show that the test conformal dose plans generated with our in-house treatment planning system agree well with the measurements performed with our bench top apparatus. The treatment planning system now can be used to plan more complex cases of clinical relevance, as discussed in the next section.

3.4.4 Treatment Planning Incorporating Tissue Heterogeneities

Our previous planning work (Joshi et al, 2009) could not mimic true clinical scenarios since it was not possible to account for tissue heterogeneities or optimize plans based on a large number of beams due to computing power limitations. The computational results presented now extend the studies to clinically relevant planning that include tissue heterogeneities by incorporating CT data sets into the Monte Carlo dose calculations.

3.4.4.1 Treatment Planning of Head and Neck Cancer

The H&N example chosen for this planning study was a challenging case from our clinical practice particularly chosen because of complexity of the target and its proximity to critical structures. The goal was to see if conformity of dose could be achieved with Co-60 tomotherapy for this case and to provide a comparison with the linac based modalities. Fig. 3.8 shows that Co-60 tomotherapy is able to deliver highly conformal treatment doses to the target structures and low doses to the OAR and the normal tissue regions. Furthermore, the Co-60 tomotherapy is considerably superior to the 3DCRT delivery when it comes to dose conformality.

A comparison of the DAHs in Fig 3.9 further quantifies and emphasizes the similarity of the Co-60 plan to the 6 MV linac tomotherapy plan for various volumes of interest. Table 3.1 provides statistics such as the mean, maximum and minimum doses
Figure 3.8 Dose distributions for the H&N case generated with (a) Cobalt-60 based tomotherapy, (b) 6 MV Linac based tomotherapy, (c) 6 MV 7 Field broad beam IMRT, and (d) 6 MV POP 3DCRT with electrons.

to each of the structures considered. The minimum dose to PTV70 for both Co-60 and 6 MV tomotherapy plans is less than 70 Gy due to a small overlap with the critical structures, however, the area receiving 70 Gy or more is greater than 95%, therefore this plan is acceptable. The average dose to PTV70 with 6 MV IMRT and 3DCRT is also in the neighbourhood of 70 Gy. The maximum dose to PTV63 and PTV56 is higher than 105% of the prescribed dose for both of the tomotherapy plans. This is because these structures overlapped with the PTV70s as seen in Fig 3.3a. As mentioned above, the priority was set such that PTV70 had sufficient coverage. To achieve this, a portion of
the nodal volume was allowed to receive a maximum dose of 70 Gy. In all plans other than the 3DCRT, the maximum doses to the cord (see Fig 3.9e) were significantly lower than the upper 40 Gy dose objective set for the spinal cord; the lowest dose to the cord was achieved in the Co-60 plan. The 3DCRT plan has a low dose to the body since it used only two parallel opposed photon beams. In fact, even 3DCRT had a low dose to body since it only used two photons beams. However, this came at the cost of high doses to both parotids as shown in Figures 3.9g-h and not enough coverage of PTV63.

The Co-60 plan had very low doses to mandible, similar to those obtained with 6 MV tomotherapy. We recognize that although the parotid gland doses for Co-60 plan were comparable to the 6 MV plans (tomotherapy and broad beam IMRT) used for comparison, the mean doses were higher than those recommended by QUANTEC (2010). This is due to the fact that there was quite a bit of overlap between the targets and the parotid glands. In order to achieve sufficient dose coverage to the target, the parotid glands were allowed to receive more than usual doses. The dose objectives and structures were based on those used for the clinical IMRT plan which had been approved by the attending oncologist.

Figure 3.10 compares 2D lateral and anterior-posterior central profiles for the respective H&N plans. The Co-60 tomotherapy is able to produce high dose gradients despite the larger penumbra. This is due to the large number of pencil beams, the weight of the beams can optimized such that doses can be conformed around the target. These are similar to those obtained with 6 MV tomotherapy and IMRT. Thus we conclude that Co-60 tomotherapy is able to provide doses to both target and OARs comparable to those obtained with other sophisticated techniques.
Figure 3.9 Dose area histograms for H&N case: targets (a-c), normal tissue (d) and critical structures (e-h). These are generated with Co-60 and 6 MV based tomotherapy, IMRT, and 3DCRT. The histograms are based on calculations to a single slice only thus, the term Dose Area Histogram (DAH) is used instead of Dose Volume Histogram.
Table 3-1 Maximum, minimum and average doses to each structure of H&N case with Co-60 and 6 MV tomotherapy, 6 MV IMRT, and 6 MV 3DCRT.

<table>
<thead>
<tr>
<th>Structure</th>
<th>Co-60 Tomo Doses (Gy)</th>
<th>6 MV Tomo Doses (Gy)</th>
<th>6 MV IMRT Doses (Gy)</th>
<th>6 MV 3DCRT Doses (Gy)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dmax</td>
<td>Dmin</td>
<td>Davg</td>
<td>Dmax</td>
<td>Dmin</td>
</tr>
<tr>
<td>PTV70</td>
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Figure 3.10 2D (a) lateral and (b) anterior-posterior dose profiles for H&N case taken from the four dose distributions generated with Co-60 Tomotherapy, 6 MV tomotherapy, 6 MV IMRT, and 3DCRT.
3.4.4.2 Treatment Planning for Prostate Cancer

A Co-60 tomotherapy treatment plan of a prostate clinical case was generated to evaluate its potential for treatments in sites with deep seated targets. Figure 3.11 shows the conformal dose distributions for a prostate case generated using Co-60 and 6 MV linac based tomotherapy, as well as 7 field broad beam IMRT and 15 MV based 4 field 3DCRT. In the Co-60 tomotherapy plan, high doses are concentrated within the prostate region as desired and only a small portion of the rectum and bladder receives a high dose. The dose distribution is similar to that obtained with the linac based tomotherapy and IMRT, and is much more conformal compared to 3DCRT treatment for this type of cancer.

These prostate plans are quantitatively compared in Fig 3.12 using dose area histograms for the planned slice and numerically in Table 3.2. It can be seen that all four plans give similar dose to the prostate. The total dose to the body in the Co-60 plan is slightly higher than in the 6 MV linac based tomotherapy and IMRT plan. The difference is 4 Gy between the Co-60 average dose to the body and the 6 MV tomotherapy and IMRT doses. The effect of this extra dose is felt to be insignificant to the tissue. Only 12% of the rectum received a dose of 70 Gy or more in the Co-60 plan. The Co-60 plan shows equivalent sparing of other critical structures compared to the 6 MV tomotherapy and produces better sparing compared to the 6 MV IMRT plan. Compared to the 3DCRT plan, Co-60 tomotherapy is much better in all aspects at sparing critical structures and gives much less dose to the normal tissue.

Figure 3.13 compares 2D lateral and anterior-posterior profiles for all four treatment modalities. It clearly shows the extent of conformality Co-60 tomotherapy provides in the high dose regions. The results from the dose distributions show that it is
possible to provide a highly conformal radiation therapy via Co-60 based tomotherapy when treating deep-seated tumors situated in anatomical regions with large separations.

Figure 3.11 Dose distributions for the prostate case generated with (a) Cobalt-60 based tomotherapy, (b) 6 MV linac based Tomotherapy, (c) 6 MV 7 Field IMRT, and (d) 15 MV 4 field box with 3DCRT.
Figure 3.12 Dose area histograms for prostate case: target (a), normal tissue (b) and critical structures (c-f).
Table 3-2 Dose statistics stating maximum, minimum and average dose to each structure of the prostate plans for Co-60 tomotherapy, 6 MV tomotherapy, 6 MV IMRT, and 15 MV 3DCRT.

<table>
<thead>
<tr>
<th>Structure</th>
<th>Co-60 Tomo</th>
<th>6 MV Tomo</th>
<th>6 MV IMRT</th>
<th>6 MV 3DCRT</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(D_{\text{max}})</td>
<td>(D_{\text{min}})</td>
<td>(D_{\text{avg}})</td>
<td>(D_{\text{max}})</td>
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</table>

Figure 3.13 2D (a) lateral and (b) anterior/posterior dose profiles for prostate case.

Computational modeling of Co-60 beams has shown a strong potential for conformal dose delivery (Fox et al., 2008; Cadman and Bzdusek, 2011). In this work we have extended the previous computational foundation by commissioning an in-house treatment planning system based on measured beam data from an actual Co-60 test bed that mimics a Co-60 tomotherapy delivery. We have validated the treatment planning
system via conformal dose measurements in a homogeneous phantom containing structures similar to those found in a typical head and neck cancer.

We have also shown clinical relevance of Co-60 tomotherapy by showing clinical treatment plans for pelvic and head and neck sites. These plans were compared to the plans generated with linac based treatment modalities such 6 MV-tomotherapy, 6 MV broad beam IMRT, and 6 MV conventional treatment plus electrons. The detailed qualitative and quantitative analysis of the treatment plans showed that Co-60 tomotherapy can provide comparable plans to 6 MV linac based tomotherapy and standard 6 MV IMRT treatments and superior plans compared to the conventional 3DCRT plans.

3.5 Conclusions

Contrary to the perceived belief that Co-60 is limited by its relatively lower energy and larger penumbra we have shown in this paper that beam properties have negligible effects when dose is delivered from multiple directions and hence, clinically acceptable tomotherapy plans are possible for the treatment of cancers located in smaller (e.g. H&N) as well as larger (e.g. prostate) anatomical regions using Co-60 tomotherapy. Further research is underway to extend these studies to three dimensions.
3.6 References


Chapter 4: Volumetric Co-60 Tomotherapy

Treatment Planning

A version of this chapter is being prepared for submission to Medical Physics as: Sandeep Dhanesar, Johnson Darko, Chandra P. Joshi, and L. John Schreiner “Three dimensional intensity modulated dose distributions for clinical cases using Cobalt-60 based serial tomotherapy approach.”

4.1 Abstract

Cobalt-60 (Co-60) radiation units played a significant role in the history of radiation therapy. However, due to lack of development, Cobalt’s usage today is limited. In order to revive Co-60 therapy, our group has investigated the potential for Co-60 intensity modulated radiation therapy (IMRT). The purpose of this paper is to report results of investigations on the clinical potential of Co-60 based tomotherapy.

One of the key elements to volumetric Co-60 tomotherapy planning was determining how the couch will need to be incremented for dose delivery via serial tomotherapy approach. Therefore GafChromic film measurements were performed to determine couch increment. An in-house inverse treatment planning system was used in conjunction with the EGSnrc Monte Carlo program for creating Co-60 serial tomotherapy treatment plans for anatomical sites such as brain, eye, head and neck (H&N), hard palate, stomach, and prostate. The Monte Carlo modeling was based on experimental measurements on Theratronics 780C Co-60 teletherapy unit retrofitted with a MIIMiC binary multi-leaf collimator. Comparisons were made to the 6 MV based broad beam IMRT plans used for patient treatments at our clinic. The IMRT plans were generated in the Eclipse treatment planning system based on in-house IMRT site
specific clinical protocols. The quality of the Co-60 plans was assessed based on 3D dose distributions, dose volume histograms (DVHs), and mean, maximum and minimum dose statistics for each of the plans. The treatment delivery times for Co-60 tomotherapy were also estimated.

The results of the experiments showed that an increment of 1.71 cm minimized both under-dosing and over-dosing at the beam junctions for serial tomotherapy. The 3D dose distributions for Co-60 tomotherapy plans look comparable to broad IMRT plans. The DVHs show that it is possible to conform high doses to the targets and spare critical structures within the acceptable dose tolerances specified by in-house site-specific clinical protocols. Treatment delivery times for Co-60 tomotherapy using the currently available sources are higher than those with the broad beam IMRT; however these can be reduced with the modified Co-60 source or dual dose delivery as reported in literature.

Treatment planning study confirms that Co-60 tomotherapy is capable of providing state-of-the-art conformal dose delivery similar to that provided by the standard clinical IMRT delivery.

4.2 Introduction

Cobalt-60 (Co-60) teletherapy (Johns et al., 1952) revolutionized radiation therapy in the 1950s. It was the first ever unit to provide high energy beams in the megavoltage range which could be used for the treatment of deep seated tumours, minimizing the damage to superficial healthy tissue and surrounding organs. Further research led to the development of medical linear accelerators (linacs), which could provide much higher energy beams than Co-60 units. The linacs provided high radiation output, increased beam penetration, and sharp beam penumbra. In part because
of these characteristics, Co-60 units eventually lost their popularity and slowly disappeared from clinics in Europe and North America (Glasgow, 1999; Schreiner et al., 2003). Subsequently, continuous development of linacs resulted in more innovative radiotherapy solutions; for example, intensity modulated radiation therapy (IMRT). IMRT provided highly conformal dose distributions by the use of multi-leaf collimators (MLC) in conjunction with linacs and the application of computerized optimization techniques in treatment planning. Comparable developments did not occur on Co-60 units.

Recent literature suggests that if the Co-60 unit is revised and upgraded, it has the potential to once again play an important role in radiation therapy, especially due to its simple and robust design, and requirement for minimal maintenance and technical expertise to operate as compared to the technologically complex linacs (Schreiner et al., 2003; Warrington et al., 2002; Van et al., 1996). The beam penetration and penumbra problems faced with Co-60 become negligible in multiple angles, rotational radiation therapy (Laughlin et al., 1986; Johns et al., 1983; Sternick et al., 1997). The work by Joshi et al. (2008) showed that dose output problem can be solved by increasing the packing density of the source or by re-designing its shape to maximize the dose output for fan beam deliveries. These reports have motivated a number of groups to evaluate Co-60 based three dimensional (3D) conformal therapy (3DCRT) and IMRT. Adams and Warrington (Adams et al., 2008) showed that conformal Co-60 plans based on customized blocks are not too far behind 6 MV based IMRT plans. Fox et al. (2008) investigated Co-60 based IMRT and determined optimal number of fields needed for Co-60 broad beam IMRT to achieve plans of similar quality to 6 MV based IMRT plans. This study showed promising results for a number of different treatment sites. Cadman
and Bzdusek (2011) performed a treatment planning study focused on plan quality and treatment times for clinical cases using Co-60 helical tomotherapy (Mackie et al., 1993) (rotational form of IMRT) approach. Their study was limited to 24 angles instead of 51 angles used in the commercial TomoTherapy® (Tomotherapy, Inc., Madison, WI) Hi-Art® unit. The dose volume histograms (DVHs) presented in their paper showed a close comparison between the Co-60 and 6 MV IMRT plans. Although this work has been predominately computer model based, the studies above have shown a strong potential for Co-60 based IMRT. In recent work, we have extended computational Co-60 research to conformal dose deliveries on test phantoms (Schreiner et al., 2009; Dhanesar et al., 2013). The purpose of the work was twofold: to validate that our in-house treatment planning system for Co-60 tomotherapy, and to show that Co-60 conformal deliveries were possible. With fully validated Co-60 based in-house treatment planning system, we were able to evaluate the clinical potential of the Co-60 tomotherapy using clinical cases treated at our clinic (Dhanesar et al., 2013; Joshi et al., 2009). However, one limitation of our previously reported work on clinical treatment planning was that it was limited to single slices only, due to computing power constraints. In this report we now extend our previous treatment planning work to three dimensions (3D) and present a planning comparison study between Co-60 tomotherapy and standard broad beam IMRT for targets in the central nervous system (CNS), head & neck (H&N), abdomen, and pelvic. These cases were actual clinical cases treated at our clinic. The results of our investigations on dosimetric considerations for 3D dose delivery via serial tomotherapy approach and treatment delivery times are also reported.
4.3 Material and Methods

4.3.1 Dosimetric Considerations for Serial Tomotherapy Approach

Co-60 tomotherapy investigations reported in this work are based on serial tomotherapy delivery approach since it is easy to implement in a laboratory-setting. It was also the initial approach used in the early days of linac IMRT. In serial tomotherapy, a narrow rotating beam of radiation is used to irradiate one slice of the patient at a time (Machie et al., 1993; Schreiner et al., 2009). The couch is then translated sequentially to treat the next slice until the entire volume of interest is treated. This technique was used prior to helical tomotherapy (Adams et al., 2008) (method used in commercial tomotherapy units) in which the radiation source revolves continuously around the patient while the patient is translated smoothly through the modulated fan beam. In both approaches, modulation of the fan beam intensity is achieved by a narrow slit MLC.

The drawback of serial tomotherapy approach is that it can introduce over- or under-dosing at the junctions of adjacent slices (Dhanesar et al., 2012). This is particularly a problem for Co-60 since the broader beam penumbra can cause dose discrepancies at the beam junctions. In order to minimize this effect, an experiment was performed on Theratronics 780 Co-60 unit (Best Theratronics, Kanata, Ontario) with a binary MLC, MIMiC (Best Nomos, Pittsburgh, PA), to determine an optimal couch index for Co-60 tomotherapy deliveries. An unmodulated fan beam of size 20x2cm\(^2\) defined by all opened leaves of MIMIC was oriented at the GafChromic film (International Specialty Products, NJ, USA) placed at the isocenter of the machine (80 cm). The film data was then analysed in MATLAB (MathWorks, Natick, MA, USA), and different increments were applied to determine the acceptable increment.
4.3.2 Treatment Planning Study

Six patients previously irradiated at the Cancer Center of Southeastern Ontario using standard broad IMRT technique on Clinac 21EX with 120 leaf Millennium MLC (Varian Medical Systems, Palo Alto, CA) were selected for treatment planning study using Co-60 tomotherapy. The disease sites of interest extended over a range of anatomies including brain, eye, head H&N, hard palate, stomach and prostate.

4.3.2.1 Linac based IMRT plans

The IMRT plans were generated in the Eclipse treatment planning system (Varian, Palo Alto, CA, USA) using our clinical site-specific protocols. The brain patient had a right frontal lobe tumor and was treated with 40 Gy in 15 fractions using five 6 MV IMRT beams oriented from 300°, 270°, 240°, 192°, and 350°. The PTV had a significant overlap with the optic chiasm, brainstem, and right optic nerve. The eye patient had a right orbital tumour and was treated to 35 Gy in 20 fractions using four 6 MV IMRT beams from gantry angles 3°, 330°, 5°, and 40°. The main organs at risk (OARs) in this case were the brain and the left eye. Although both brain and orbital tumors were treated to relatively low doses, these cases were challenging due to critical structures situated in close proximity to targets. The H&N case was also challenging as it involved three PTVs nested in a number of OARs. It was treated using seven 6 MV IMRT beams from gantry angles 150°, 90°, 45°, 0°, 310°, 250°, and 200°. The primary PTV was treated to 66 Gy in 33 fractions while the secondary targets were treated to 60 and 54 Gy. The OARs were the spinal cord, mandible, larynx, and left and right parotid glands. Another H&N case, involving the treatment of hard palate, was also chosen for Co-60 tomotherapy planning. The patient received adjuvant radiotherapy consisting of
35 Gy in 20 fractions from six 6 MV broad IMRT beams from gantry angles 145°, 90°, 45°, 0°, 310°, 250°. The last two cases chosen for this Co-60 tomotherapy treatment planning study involved deep seated tumours in the abdomen and pelvic region since these tumors are often perceived to be poor candidates for low energy beam treatments with Co-60. The patient with stomach cancer was treated to 30 Gy in 20 fractions using 4 broad IMRT beams from gantry angles 150°, 90°, 45°, and 0°. The OARs in this case were liver, left kidney, left lung, and cord. The prostate cancer patient was treated to 78 Gy in 38 fractions using 7 broad IMRT beams from gantry angles 110°, 80°, 40°, 355°, 310°, 280°, and 250°. The main OARs considered during planning of this case were rectum, bladder, and femoral heads.

4.3.2.2 Co-60 tomotherapy planning

For Co-60 tomotherapy treatment planning, the EGSnrc Monte Carlo program (Rogers et al., 1990; Rogers et al., 1995; Walters et al., 2005) was used for dose calculations. Computed tomography (CT) images used for broad IMRT planning were transferred to open source treatment plan analysis software known as Computational Environment for Radiotherapy Research (CERR) (Deasy et al., 2003). The CERR was used for creating a patient anatomy based phantom data file, commonly known as CT phantom, and dose calculation input files for the EGSnrc Monte Carlo program. The BEAMnrc (Rogers et al., 1995) component of the EGSnrc Monte Carlo program was used for modeling Theratronics 780 Co-60 teletherapy unit with the MIMiC collimator. The details regarding the Monte Carlo model of the treatment unit and its validation have been presented in our previous work (Joshi et al., 2008; Dhanesar et al., 2013; Dhanesar et al., 2012). Radiation transport was simulated for each of the MIMiC leaves
and the phasespace plane was scored immediately below the MLC and before the CT phantom. A phasespace file contains information such as the photon fluence, energy, location, direction, and interaction history. The DOSxyznrc (Walters et al., 2005) component of the EGSnrc program then used the phasespace files and calculated doses in the CT phantom from 51 beam directions (projections) of each of the equally weighted pencil beamlets defined by the openings of the MLC. The voxel resolution/grid spacing of the CT phantom was 2.5x2.5x3mm³ for patient dose calculations. The CT phantom was then translated to the next slice using serial tomotherapy approach based on the optimal increment determined by the procedure in Section 4.3.1. Table 4.1 provides details regarding how many translations (full gantry rotations) were required to cover the PTVs considered in this work. The total dose deposited in the CT phantom was calculated in MATLAB by adding doses from each of the individual pencil beamlets from all beam directions and translations (couch increments). Total time for dose calculations varied from 48 hours to 1.5 weeks depending on the size of the PTVs. The least and most amount of dose calculation time was taken by the eye and H&N case, respectively. This time was based on in-house parallel computing cluster consisting of 96 CPUs running PBS queuing software (PBS works, Altair Engineering, Detroit, MI, USA).

Once the dose calculations were completed, in-house inverse planning software (Chng, 2005) was used to optimize the pencil beam weights until the desired dose distribution was achieved. The in-house inverse treatment planning software, written in MATLAB, is based on the conjugate gradient active set algorithm (Hristov et al., 1997). In this implementation, all beamlets that do not go through the targets are set to a zero
beam weight during the initial step and only those remaining are then optimized via an iterative approach. This reduces the overall optimization time for the tomotherapy treatment planning. The in-house planning system used least-square minimization as a cost function. All PTV and OAR contours were taken directly from the clinical IMRT plans. The dose prescriptions and constraints were also the same as those ones used in the Eclipse IMRT plans. Doses were optimized to ensure at least 95% of the PTV received the minimum prescribed dose and the OARs and normal tissue were spared as much as possible based on the in-house clinical site-specific protocol. The optimization times varied from 4-18 hours depending on the complexity of the case.

Table 4-1 Target volume and beam parameters for each of the anatomical sites planned using Co-60 tomotherapy. H&N had three targets - PTV66, PTV60, and PTV54 (in the order specified in the table), all others had one PTV. The number of translations corresponds to the full gantry rotations, where each rotation had 51 projections and 20 pencil beamlets per projection. A beamlet is a pencil beam defined by individual MLC leaf openings.

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<th># of projections</th>
<th># of MLC leaves</th>
<th># of translations</th>
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<td>H&amp;N</td>
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<td>20</td>
<td>11</td>
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4.3.3 Co-60 Tomotherapy Treatment Times

In order to determine treatment delivery times for Co-60 tomotherapy plans, we calculated the maximum time required for each of the 51 segments in a full gantry rotation and repeated the process for the number of gantry rotations (slice widths). The total treatment time was calculated by adding the maximum times for each of the segments. As Cadman and Bzdusek reported, the dose rate for the commercially available 2 cm diameter cylindrical Co-60 source (Best Theratronics, Ltd., Ottawa, Ontario, Canada) is approximately 250 cGy/min at 100 cm source to axis distance (SAD) for a beam of 10x10 cm$^2$. Therefore, the dose rate for the 80 cm SAD T780 Co-60 unit discussed in this work is approximately 390 cGy/min for a beam of 10x10 cm$^2$. It should be noted that the treatment times estimated in this work were based on beam-on time only and the time that would be required for gantry to rotate between successive fields, data transfer of the MLC delivery files and error in the estimated rotation time was not considered. The time for these factors is not known since Co-60 tomotherapy unit has not yet been manufactured. A similar method was used to calculate treatment time for broad IMRT. The total beam-on time for IMRT deliveries was estimated by multiplying the dose rate to the number of Monitor Units (MU) per field. The dose rate for broad IMRT delivery was 400 MU/min.

4.4 Results

4.4.1 Serial Tomotherapy Dose Delivery Consideration

Figure 4.1 shows two examples of the couch indexing study. In the first case (Fig 4.1a), an increment of 2 cm was used since the maximum projection of the fully opened MIMiC collimator in the longitudinal direction is 2 cm at the isocenter. The
dose distribution and vertical profile (Fig 4.1a and 4.1b) clearly show that the 2 cm increment is not optimal and gives rise to severe underdosing at the junctions of the beams. After considering several increments, 1.71 cm couch index was found to be optimal as shown in Figs. 4.1c-d. The couch index below this number causes overdosing at the junctions.

**Figure 4.1 Illustration showing optimal increment for Co-60 serial tomotherapy delivery.** In (a) 2 cm couch index is used and the corresponding vertical profile is shown in b). c) shows a dose distribution obtained for an “optimal” indexing of 1.71 cm and d) show corresponding profile to 1.71 cm index.

### 4.4.2 Treatment Planning Comparison

The quality of the Co-60 tomotherapy plans for various clinical sites such as the brain, eye, H&N, hard palate, stomach, and prostate was assessed by comparing the calculated dose distributions, DVHs, and mean, minimum, and maximum dose values to the standard clinical IMRT plans used for patient treatments at our clinic. Figures 4.2-
4.4 show that conformity of doses is possible using Co-60 tomotherapy without giving excess doses to the surrounding tissues and OARs. All Co-60 tomotherapy dose distributions are comparable to those for clinical IMRT plans based qualitative analysis.

The DVHs in Fig 4.5 and dose statistics in Table 4.2 provide more in-depth quantitative comparison of the Co-60 treatment plans to the 6 MV based broad IMRT plans. As shown in Fig 4.5a, 95% of the PTV received the prescription dose of 40 Gy in both Co-60 and IMRT brain case. The mean doses to the PTV for Co-60 and 6 MV IMRT plans were 40.7 Gy and 40.6 Gy, respectively. A challenge in this case was the overlap between the target and OARs such as optic nerves, optic chiasm and brainstem. Since the right optic nerve and optic chiasm were prone to receive high doses due to their proximity to the target, high priority was set on these structures compared to the other critical structures. As a result, the right optic nerve and optic chiasm were better spared in Co-60 plan compared to the brainstem, brain, and right orbit, which received higher doses than 6 MV IMRT plan. However, these differences were mainly in the low dose areas and were within acceptable clinical tolerances. Fig 4.5b shows DVHs for the eye plan with Co-60 and 6 MV based IMRT. The Co-60 plan was normalized according to the 6 MV IMRT plan so that 90% of the PTV would receive the prescription dose of 35 Gy. A slightly colder plan was accepted in this case in order to limit doses to the optic chiasm and optic nerves. The optic chiasm and right optic nerve in Co-60 plan did receive higher doses than the IMRT plans, however, these doses were within the acceptable dose levels based on the in-house clinical protocol for this site. The H&N case was perhaps the most difficult case since not only did it have many critical structures in close vicinity; it also had 3 different targets that were prescribed to different dose levels. The Co-60 plan was normalized such that the volumes of PTVs
receiving the prescription doses were same as in the IMRT case. It can be seen in Fig. 4.5c and Table 4.2 that this was achieved for two of the targets. However, a slightly colder coverage to PTV66 was accepted in order to provide better sparing to critical structures as long as over 95% of the clinical target volume (CTV) received the prescribed dose. The Co-60 plan in general provided better sparing to the critical structures than the IMRT plan. It should be noted that parotid glands in both plans received higher doses since there was a significant overlap between the parotids and the targets. Fig 4.5d shows another H&N case, particularly for the treatment of hard palate, where the parotid gland doses were lower since there was no overlap with the PTV. In this case, the PTV coverage in the Co-60 plan was similar to the 6MV plan. All critical structures satisfied the clinical dose tolerances. Figs. 4.5e-f show DVHs for stomach and prostate case, respectively. For both sites, the mean PTV doses obtained with Co-60 tomotherapy and 6 MV based IMRT are similar. Although Co-60 tomotherapy plans are slightly colder and less homogeneous than the IMRT plans, these were acceptable since nearly 95% of the PTV volumes received the prescribed doses. The Co-60 plan, particularly for prostate cancer, provided better sparing to the critical structures compared to the clinical IMRT plans.
Figure 4.2 Co-60 tomotherapy and 6 MV based broad beam IMRT dose distributions for brain and orbit cancer. Full 3D comparison between Co-60 and 6 MV IMRT is presented by transverse, sagittal, and coronal views.
Figure 4.3 Co-60 tomotherapy and 6 MV based broad beam IMRT dose distributions for two H&N cancer sites. Full 3D comparison between Co-60 and 6 MV IMRT is presented by transverse, sagittal, and coronal views.
Figure 4.4 Co-60 tomotherapy and 6 MV based broad beam IMRT dose distributions for stomach and prostate cancer. Full 3D comparison between Co-60 and 6 MV IMRT is presented by transverse, sagittal, and coronal views.
Figure 4.5 Dose volume histograms (DVHs) for brain (a), eye (b), H&N (c), hard palate (d), stomach (e), and prostate (f). Co-60 tomotherapy DVHs are shown by solid lines and 6 MV based broad beam IMRT DVHs are shown by dashed lines.
Table 4-2 Dose statistics for targets and main critical structures for various anatomical regions for Co-60 tomotherapy plans and 6 MV broad IMRT plans. Mean, minimum, and maximum absolute and relative (shown in brackets) dose values are shown. Relative dose values are normalized to the primary dose prescriptions e.g. 45, 35, 66, 35, 30, and 78 Gy for brain, orbit, H&N, hard palate, stomach, and prostate, respectively.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Co-60 plan doses (Gy)</th>
<th>IMRT plan doses (Gy)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>D&lt;sub&gt;mean&lt;/sub&gt;</td>
<td>D&lt;sub&gt;min&lt;/sub&gt;</td>
</tr>
<tr>
<td>(a) Brain Cancer PTV</td>
<td>102%</td>
<td>101%</td>
</tr>
<tr>
<td>Optic Chiasm</td>
<td>101%</td>
<td>87%</td>
</tr>
<tr>
<td>R. Optic Nerve</td>
<td>96%</td>
<td>4%</td>
</tr>
<tr>
<td>Brainstem</td>
<td>43%</td>
<td>37%</td>
</tr>
<tr>
<td>(b) Orbit Cancer PTV</td>
<td>8%</td>
<td>28%</td>
</tr>
<tr>
<td>Brain</td>
<td>11%</td>
<td>1%</td>
</tr>
<tr>
<td>L. Optic Nerve</td>
<td>19%</td>
<td>58%</td>
</tr>
<tr>
<td>R. Optic Nerve</td>
<td>89%</td>
<td>58%</td>
</tr>
<tr>
<td>Optic Chiasm</td>
<td>8%</td>
<td>28%</td>
</tr>
<tr>
<td>L. Orbit</td>
<td>1%</td>
<td>3%</td>
</tr>
<tr>
<td>(c) H&amp;N Cancer PTV66</td>
<td>102%</td>
<td>83%</td>
</tr>
<tr>
<td>PT60</td>
<td>96%</td>
<td>75%</td>
</tr>
<tr>
<td>PT54</td>
<td>86%</td>
<td>77%</td>
</tr>
<tr>
<td>Cord</td>
<td>21%</td>
<td>0%</td>
</tr>
<tr>
<td>Mandible</td>
<td>59%</td>
<td>0%</td>
</tr>
<tr>
<td>L. Parotid</td>
<td>57%</td>
<td>0%</td>
</tr>
<tr>
<td>R. Parotid</td>
<td>33%</td>
<td>0%</td>
</tr>
<tr>
<td>(d) Hard Palate Cancer</td>
<td>105%</td>
<td>74%</td>
</tr>
<tr>
<td>PTV</td>
<td>13%</td>
<td>3%</td>
</tr>
<tr>
<td>Brainstem</td>
<td>11%</td>
<td>0%</td>
</tr>
<tr>
<td>Cord</td>
<td>15%</td>
<td>0%</td>
</tr>
<tr>
<td>L. Parotid</td>
<td>46%</td>
<td>14%</td>
</tr>
<tr>
<td>R. Parotid</td>
<td>54%</td>
<td>23%</td>
</tr>
<tr>
<td>(e) Stomach Cancer PTV</td>
<td>103%</td>
<td>62%</td>
</tr>
<tr>
<td>L. Lung</td>
<td>11%</td>
<td>0%</td>
</tr>
<tr>
<td>L. Kidney</td>
<td>12%</td>
<td>0%</td>
</tr>
<tr>
<td>Liver</td>
<td>38%</td>
<td>0%</td>
</tr>
<tr>
<td>Cord</td>
<td>6%</td>
<td>0%</td>
</tr>
<tr>
<td>(f) Prostate Cancer PTV</td>
<td>102%</td>
<td>72%</td>
</tr>
<tr>
<td>Rectum</td>
<td>47%</td>
<td>1%</td>
</tr>
<tr>
<td>Bladder</td>
<td>43%</td>
<td>4%</td>
</tr>
<tr>
<td>L. Femoral Head</td>
<td>23%</td>
<td>3%</td>
</tr>
<tr>
<td>R. Femoral Head</td>
<td>23%</td>
<td>4%</td>
</tr>
</tbody>
</table>
4.4.3 Treatment Delivery Times

The treatment delivery times for Co-60 tomotherapy plans were estimated and compared to those obtained for 6 MV based clinical IMRT plans. These times are reported in Table 4.3. As mentioned previously, the treatment times reported here are based entirely on beam-on time and do not include contribution coming from gantry rotation, table translation, and MLC data transfer. As expected, Co-60 times are longer than the standard clinical IMRT delivery times. One reason is that the Co-60 plan is based on 102 to 561 fan beam IMRT segments depending on the volume being treated as oppose to 5-7 broad beam IMRT segments. This increased degree of modulation for Co-60 tomotherapy plans results in longer treatment time. Another reason is that the dose rate for 6 MV broad IMRT broad deliveries is relatively higher than the Co-60. Cadman and Bzdusek (2011) reported Co-60 tomotherapy plans that required relatively lower treatment delivery times than reported here. Although, some of the difference in treatment times comes from the type and size of target being treated, the main difference stems from the fact that the plans presented in this study were generated based on a narrow fan beam of slice width 1.71 cm and 51 projections as compared to the slice width of 4.8 cm for single slice, 9.6 cm for dual slice delivery, and 24 projections used in Cadman and Bzdusek study.

<table>
<thead>
<tr>
<th>Case</th>
<th>Brain</th>
<th>Eye</th>
<th>H&amp;N</th>
<th>Hard Palate</th>
<th>Stomach</th>
<th>Prostate</th>
</tr>
</thead>
<tbody>
<tr>
<td>Co-60</td>
<td>6.2</td>
<td>2.6</td>
<td>9.8</td>
<td>5.6</td>
<td>6.5</td>
<td>8.6</td>
</tr>
<tr>
<td>6MV IMRT</td>
<td>2.5</td>
<td>0.8</td>
<td>2.8</td>
<td>1.1</td>
<td>1.2</td>
<td>2.5</td>
</tr>
</tbody>
</table>
4.5 Discussion

It has been reported in literature that Co-60 units can play an important role in modern radiation therapy if modern dose delivery techniques are investigated (Schreiner et al., 2003; Warrington et al., 2002; et al., Van Dyk et al., 1996). Computational modeling (Adams et al., 2008; Fox et al., 2008; Cadman et al., 2011; Dhanesar et al., 2013; Joshi et al., 2009) and dose delivery studies (Schreiner et al., 2009; Dhanesar et al., 2013) have shown strong potential for Co-60 IMRT. Despite these reports, there is still reluctance among the medical physics community regarding the role of Co-60 in modern radiation therapy. A continual flow of reports establishing the potential of Co-60 based IMRT are needed in order to convince the clinicians and motivate developers to modernize Co-60 clinical units and make them attractive throughout the world. In our effort to reinvigorate Co-60, we investigated the clinical potential of Co-60 tomotherapy using experimentally validated in-house inverse treatment planning system dedicated to Co-60 planning. The Co-60 tomotherapy plans were generated using beam parameters similar to those used in commercial 6 MV based tomotherapy for various small and large typical clinical sites and were compared to 6 MV based broad beam IMRT plans used for patient treatments at our clinic. In-depth comparison using 3D dose distributions, DVHs and dose statistics showed that Co-60 tomotherapy can provide comparable plans to standard 6 MV based broad beam IMRT treatments.

The Co-60 tomotherapy treatment times were found to be longer than the IMRT treatment times due to the use of increased number of beams in the tomotherapy delivery approach and lower Co-60 source dose output. A number of studies indicate that Co-60 treatment times can be further improved. Cadman and Bzdusek (2011) reported Co-60 tomotherapy plans that required relatively lower treatment delivery
times by using an MLC allowed the treatment of larger slices. Although it is realised that dose conformality will be compromised with increased slice width, Cadman and Bzdusek (2011) showed that these plans were not too far off the IMRT plans. Joshi et al (2008) investigated alternative methods which could yield lower treatment times. In particular it was suggested that increasing the source diameter or packing density can significantly increase the output, which would allow the treatment time to be lower. They also suggested a significant gain in radiation output maybe obtained if efficient source shape designs for Co-60 tomotherapy deliveries are considered instead of using standard cylindrical sources.

4.6 Conclusion

Co-60 tomotherapy has the potential to provide state-of-the art conformal dose delivery for complex small and large clinical treatment sites as that provided by the modern treatment modalities currently available in clinics. The low energy and larger beam penumbra of the Co-60 beams has negligible effect on plan quality when dose is delivered from multiple directions.
4.7 References


Chapter 5: Dose superposition considerations for Co-60 tomotherapy Deliveries

A version of this chapter with content formatted for publication has been published as:

5.1 Abstract

The finite size pencil beam (FSPB) superposition method is a commonly used dose calculation method in intensity modulated radiation therapy (IMRT). The FSPB model assumes that dose for a broad intensity modulated beam can be calculated by superposition of dose from small, pencil-like beams. However, this model is limited to point-like radiation sources and is not valid for finite size sources, such as a Cobalt-60 (Co-60) source of 2 cm diameter. In this paper we present results that show the limitation of this model and propose an alternative model, namely the aperture superposition (AS) model, to calculate photon dose for intensity modulated beams arising from finite size radiation sources.

The AS model is based on adding beam apertures rather than pencil beams. Each aperture is defined as a series of adjacently opened leaves of a multi-leaf collimator with no closed leaves in between them. The apertures are calculated using the EGSnrc Monte Carlo program. The accuracy of the AS model was tested for dose
calculations of fan beams, as encountered in tomotherapy treatment plans. The results were compared to the FSPB model and GafChromic film measurements. The measurements and simulations were performed for a clinical Theratronics 780C Co-60 unit with MIMiC binary multi-leaf collimator mounted on it.

The comparisons between the AS model and film measurements show agreement better than 1.5% in the high dose regions and 3.7% in the low dose regions. On the contrary, film measurement comparisons to the FSPB model show that the FSPB model underestimates the dose by up to 7% for small field sizes such as 2x2 cm$^2$ and 20% for larger field sizes such as 20x2 cm$^2$.

The results presented in this paper indicate that the AS model provides better accuracy than the FSPB model when calculating dose for fan beams from large radiation sources. The implementation of this model to the current treatment planning systems has the scope of advancing Co-60 based IMRT and tomotherapy.

5.2 Introduction

Intensity modulated radiation therapy (IMRT) continues to attract clinical interest throughout the world and has become the standard of care for a number of treatment sites. Cobalt-60 (Co-60) based IMRT, particularly the tomotherapy approach, has been under study at our center. We believe this modification to the current Co-60 unit can make conformal dose delivery more accessible worldwide as it is relatively low in cost and is easy to maintain. The development of conformal Co-60 units may also have benefit for smaller cancer clinics in developing countries which have limited access to resources such as water and power, since it will indicate that Co-60 radiation therapy is under renewed development and is supported by vendors. Using a simple approach based on a pencil beam superposition model to imitate the fan beam delivery
with a binary multi-leaf collimator (MLC), we have demonstrated a strong viability of Co-60 based tomotherapy (Schreiner et al., 2009; Joshi et al., 2009; Dhanesar et al., 2010). There have been a number of other studies that have also indicated a strong potential of Co-60 in modern radiation therapy (Cadman et al., 2011; Cadman et al., 2007; Fox et al., 2008; Adams et al., 2008).

However, during our studies, the limitation of using the pencil beam superposition model to obtain an intensity modulated fan beam has become apparent. The pencil beam superposition model, formally known as the finite-size pencil beam (FSPB) model, was proposed by Bourland and Chaney in 1992. This algorithm assumes that the radiation beam can be geometrically divided into identical, small, finite size, pencil like beams that allow the dose calculation of the broad beam by superposition. Each pencil beam has a weighted total photon fluence which may vary with position in the full broad beam. A key assumption made in this model is that the origin of radiation is a point source. Because of this assumption, and the lack of accounting for partial radiation transmission from the MLC leaf sides and ends, Bourland and Chaney’s method produces steep dose gradients, in other words dose profiles with essentially zero penumbra. Ostapiak et al. (1997) modified the FSPB model to account for the fact that the source has a finite size and that there will be some transmission due to the components of the treatment head and the MLC leafs, both of which will give rise to the penumbra region. They also introduced a radiation beam with a spectrum of photon energies instead of the mono-energetic beams in the initial model and adopted the Fast Fourier Transform convolution technique to speed the calculations of the FSPBs. Over the years there have been numerous other studies that have proposed further changes to improve the accuracy of the dose calculations (Knoos et al., 1995; Sharpe et al., 1995;
Bergman et al., 2004; Jelen et al., 2007). In particular, Liu et al. (1997) and Ahnesjo et al. (2005) improved the accuracy of pencil beam dose calculations by introducing a multi-source model in which the primary radiation is modeled separately from the scatter radiation.

In all these studies the focus for development has always been on linear accelerator (linac) based beams. Studies that were based on Co-60 beams have been on hypothetical units that used a point source (Bourland et al., 1992; Ostapiak et al., 1997), not based on the actual clinical Co-60 units that have a large source (diameter 2cm). Although the modified FSPB takes into account the source size, it has a limitation when modeling Co-60 intensity modulated beams. This is because the modified FSPB model correction is limited to a linac with a typical source diameter of about 0.2 cm. As shown in Fig. 5.1a, the diameter of the linac source is smaller than the width of the single leaf opening (0.52cm) of the MLC. Hence from the MLC’s eye-view, the full source can be seen from just the single leaf opening. However, as shown in Fig. 5.1b, the Co-60 source diameter is comparatively larger than the single leaf width and only a portion of the source (approximately 1.1 cm diameter) is seen from a single leaf opening. Hence, the source diameter limitation of FSPB will be shown to be the cause of the failure of the model in calculating Co-60 MLC beam. Cadman and Bzdusek (2011) has also pointed out the source size problem and used correction factors to account for dose differences (Cadman et al., 2011).
Figure 5.1 Schematic diagrams showing the difference between the source sizes for linac (a) and for Co-60 units (b). The linac source is small enough to be fully seen from a single leaf opening from the MLC’s eye-view, whereas for the Co-60 unit, only a portion of the source is seen. The MLC is placed such that a field size of 1x1 cm$^2$ is projected at the isocentre, which is 100 cm from source for a linac and 80 cm from source for a Co-60 unit. Note, the ‘partial source’ problem is resolved at a field size of approximately 3.5 x 3.5 cm$^2$, at which point the full source is unobstructed.

The work presented in this paper has two purposes. The first purpose is to investigate whether the dose for Co-60 intensity modulated beams can be calculated based on the FSPB model. To achieve this purpose, the FSPB model based dose calculations for Co-60 beams are compared with the film measurements and Monte Carlo simulations to quantify the dose differences. The Co-60 FSPB profiles are compared with those calculated for linac beams, which originate from a relatively smaller source, to assess the magnitude of the dose difference.

The second purpose is to propose a dose calculation algorithm that provides accurate dose calculations of intensity modulated beams from finite size sources that are
larger than the width of a single MLC leaf and present validation work done with comparison to film measurements. The work presented is limited to tomotherapy approach, which uses intensity modulated fan beams instead of broad beams.

5.3 Material and Methods

The individual finite size pencil beams can be obtained by two methods: 1) via a look-up table method where the pencil beam parameters are pre-computed and stored for later use, 2) via the convolution dose calculation method, in which the calculations are done on the fly based on the pre-calculated pencil beam kernels. We currently do not have a convolution dose model for the Co-60 source, hence for this study, a Monte Carlo (MC) method was used to calculate pencil beams or larger beams that were then stored in look-up tables. Although the MC method provides better dose calculation accuracy than the convolution method, it is computationally extensive; therefore, it cannot currently be used to calculate every possible beam combination. As part of future work, further studies are underway to develop a faster convolution based dose calculation method.

5.3.1 Monte Carlo Simulations

The Monte Carlo simulations were done for a clinical T780C Co-60 unit (Best Theratronics, Kanata, Canada) with a 2 cm diameter cylindrical source, and for a 6 MV photon beam from a CL2100EX (Varian Medical systems, Palo Alto, USA) linear accelerator. Both of these units were modeled using the EGSnrc/BEAMnrc code (Rogers et al., 1995), with source to axis distances corresponding to the actual clinical distances for these units (80 cm SAD for Co-60 unit and 100 cm for CL2100EX). For intensity modulation, both of these units were simulated with the MIMiC collimator
Once the phase space data were calculated for these units, the EGSnrc/DOSXYZnrc code (Walters et al., 2006) was used to calculate the 3D dose distribution in water. A dose grid of 2.5 mm$^3$ was used.

### 5.3.2 Dose Model

Figure 5.2 shows an example of an intensity modulated pattern from the MIMiC collimator. The MIMiC collimator has a set of twenty (20) 8 cm thick tungsten leaves, which project the smallest field size of 1x2 cm$^2$ at the isocentre for a 2 cm delivery mode (treating 2 slices at a time) and a maximum field size of 20x2 cm$^2$. According to the FSPB model, the total absorbed dose to a point $(x, y, z)$ is the sum of the doses delivered by multiple beamlets such as $P_1$, $P_2$...$P_8$ shown in Fig. 5.2a. This can be generalized as (Eq 5.1).

$$D = \sum_{n=1}^{N} w_n P_n(x, y, z) \tag{5.1}$$

$N$ is the number of beamlets and $w_n$ is a relative weight of the $n$th beamlet. For simplicity, the beamlet weights in this example are considered to be the same.

The model that we propose does not sum individual pencil beams; rather it adds up the ‘apertures’. As shown in Fig. 5.2b, an aperture is defined as series of consecutive opened leaves with no closed leaves in between. The full modulated fan beam in this example is considered as the sum of 5 apertures ($A_1$, $A_2$...$A_5$). The total dose absorbed to point $(x, y, z)$ for a generalized case is specified by Eq 5.2.

$$D = \sum_{n=1}^{N} w_n A_n(x, y, z) \tag{5.2}$$

As mentioned previously, the dose for each individual aperture is calculated using the Monte Carlo method. Since our investigations are focused on Co-60
tomotherapy, the superposition of apertures is only applicable in the crossplane
direction. The inplane direction is not modulated since our tomotherapy deliveries are
based on irradiating 2 slices (2 cm long) at a time.

Figure 5.2 An example of the intensity modulated beam from a binary multi-leaf
collimator (MIMiC). Opened leaves correspond to gray boxes and closed leaves
correspond to white boxes. The numbers above the leaf pairs represent the identification
of the MLC leaf pairs. A leaf pair is made up of top and bottom leaves. The modulated
fan beam in (a) is represented by the sum of individual beamlets and in (b) as the sum of
apertures.

5.3.3 Measurements

The Monte Carlo modeling of the treatment units was validated by comparing
the simulated percentage depth dose (PDD) curves and dose profiles with the ion
chamber measurements. An IC-10 (0.13 cm$^3$) cylindrical ionization chamber (Wellhofer
Dosimetrie, Germany) was used for broad beam measurements and an A-16 (0.007 cm$^3$)
EXRADIN micro point chamber (Standard Imaging, USA) was used for fan beam
measurements. The Co-60 unit modeling was verified with and without the MIMiC
mounted on the unit. However, for the 6 MV units, it was not possible to mount the
MIMiC collimator due to mechanical limitations; therefore the verification was only
done for the treatment head without the MLC. Comparison of the Monte Carlo
simulations and ion chamber measurements showed agreement better than 1% and 2% for the high dose and low dose regions, respectively. In this work, we mainly discuss profiles in crossplane direction since no modulation is done in inplane direction. Detailed verification results, as well as a comparison of crossplane and inplane profiles have been published in our previous work (Joshi et al., 2008).

The film measurements were performed with Gafchromic EBT (International Specialty Products, NJ, USA) radiochromic film. The film was calibrated carefully based on the protocols established by Cheung et al. (2005) and Dhanesar (2008). The measurements were then done for several different MIMiC leaf combinations. The film was placed at a depth of 5 cm within a 15 cm solid water phantom. Fiducial marks were used on 4 corners of the film to ensure accurate alignment.

5.4 Results and Discussion

5.4.1 FSPB model for Co-60 versus 6 MV Beams

Figure 5.3 shows an example of three leaf sequences. In the first sequence (a), only the leaf pair labeled as “10” is open, the rest of the leaves are closed. In the second sequence (b), only leaf pair “11” is open. The third sequence shows both leaf pairs 10 and 11 open and all others closed. In most treatment planning systems, particularly for those designed for linac machines, the dose arising from sequences (a) and (b) can be added to get the dose for sequence (c) with some minor corrections to account for penumbra. As mentioned previously, this is known as the FSPB method.

Ideally, there is no radiation transmission through the closed leaves. In reality, this is not the case; and dose models generally use a transmission factor to account for any leakage radiation. In order to study the effect of leakage radiation in this work, film
measurements and Monte Carlo simulations were performed on the fully closed MIMiC collimator (all 20 leaves closed) and results were compared to those obtained for the fully open field (all 20 leaves opened). The film measurement showed a maximum dose of 2.6cGy for the fully closed field versus 325cGy for the fully opened field for an equal amount of irradiation time. Thus, from the film measurement, the maximum dose from leakage is about 0.8%. The Monte Carlo calculation gives a corresponding leakage dose of less than 0.2%. These doses agree within the uncertainty of the film measurements. Currently, we consider this transmission perturbation small, and do not include a correction in our dose model. In any final system, a transmission correction factor would have to be measured more carefully and suitable correction accommodated.

![Figure 5.3 MLC leaf sequences of opened (gray boxes) and closed (white boxes) leaf pairs. A leaf pair is made up of top and bottom leaves. In the first leaf sequence (a), only leaf pair 10 is open; in the second sequence (b), only leaf pair 11 is open; in the second sequence (b), only leaf pair 11 is open; in the third sequence (c), both leaf pairs 10 and 11 are open.](image)

Figures 5.4 and 5.5 show the FSPB model results for fan beams from the Co-60 and 6 MV units, where 1x2 cm² individual beam profiles (gray-dashed lines/markers) are added up as per the FSPB model to yield 2x2 cm², 4x2 cm², 6x2 cm², and 20x2 cm².
fan beam profiles (solid black lines/markers). The individual profiles are calculated by Monte Carlo (gray-dashed lines) and also measured with film (gray squares) for Co-60 beams. In the case of Monte Carlo, FSPB superposition profiles are obtained by simply adding the individual profiles. The film FSPB profiles are obtained by irradiating the film by opening one MLC leaf at a time until the desired fan beam is achieved. The results are compared with the fan beam profiles obtained from the “open fields” (solid red lines/markers). An open field is defined as a field that consists of multiple opened beamlets, or in other words a broad fan beam. The individual profiles are normalized relative to the open field profiles.

For the Co-60 beams, the FSPB model underestimates dose by 7% for the 2x2 cm$^2$ field size in the high dose regions around the central axis, 11% for the 4x2 cm$^2$ field size, and as much as 16% for the 6x2 cm$^2$ field size when compared to the open field profiles for these field sizes. These differences increase even more for the larger fields such as the 20x2 cm$^2$, where the differences are above 20% (Fig. 5.5). Note that Monte Carlo calculations and film measurements agree to better than 1.5% in the high dose regions and 3.7% in the low dose regions for all individual profiles, FSPB method profiles, and open field profiles. For the 6 MV beams, the peak dose for the open field profiles agree well with the peaks seen in the profiles obtained with the FSPB model. The penumbral regions of the beamlets do give rise to ‘dips’ in the 6 MV FSPB profiles (added from individual profiles); however these can be easily corrected by renormalizing them based on the maximum dose.

The differences seen in the Co-60 beams cannot be a result solely of the large penumbra of the Co-60 beamlets since if this was the case similar ‘dips’ in the profiles as those seen for 6 MV beams would have been observed for Co-60 beams and there
would be no dose differences in the peaks of the profiles for the open field and the FSPB method. However, Figs. 5.4-5.5 clearly show dose discrepancies in the peaks of the Co-60 profiles. These differences arise from the size of the Co-60 source. As noted previously, the individual beamlets of the collimator on the Co-60 unit only “see” a portion of the source when the source size is larger than the size of the beamlet. Hence, when the beamlets are added up sequentially according to the FSPB model, the total source size considered is smaller than that seen from a broad beam and the dose output is lower.

Figure 5.6 shows similar results for the off-axis leaves. The far end leaves show further reduction in dose as the amount of source seen lessens even more due to the narrow gap of the MLC “seen” from an off-axis position (Cadman et al., 2011). Results thus far clearly demonstrate the limitation of the FSPB model for Co-60 beams. Since accuracy in dose calculations is vital in radiation oncology, the differences in dose due to this model cannot be ignored. To achieve an acceptable accuracy in dose, Co-60 source geometry or dose calculation method need to be re-thought. While the FSPB calculation problem could be minimized by redesigning Co-60 units with smaller sources, this is likely not clinically feasible, since the dose rate from Co-60 units is already at a low limit of about 250cGy/min (Joshi et al., 2008; Cadman et al., 2010). Reducing the source size further would reduce the dose rate even more. Thus, the redesign of the source geometry may not be the ideal solution at this time, although it may be useful in the future design of newer cobalt units (Joshi et al., 2008).
Figure 5.4 A comparison of relative dose profiles obtained with Co-60 beams and 6 MV beams for 2x2 cm$^2$ (a and b), 4x2 cm$^2$ (c and d) and 6x2 cm$^2$ (e and f) field sizes. Co-60 beams are calculated in Monte Carlo (dashed/solid lines) and are compared with film measurements (square markers). 6 MV profiles are obtained with Monte Carlo only. Above each set of profiles, a schematic diagram of the MLC shows the leaf pairs that were opened for that particular comparison. Note red lines and markers correspond to the ‘open field’ profiles, black lines and markers correspond to the FSPB superposition profiles, and gray lines and markers correspond to individual 1x2 cm$^2$ profiles. To avoid cluttering figures, these colour legends are not provided explicitly. The colour scheme applies to subsequent Figs. 5.5 through 5.7.
Figure 5.5: A comparison of relative dose profiles obtained with Co-60 beam (a) and 6 MV beam (b) for 20x2 cm² field size (all 20 leaves are opened as shown by the schematic diagram above (a)). Co-60 beams are calculated in Monte Carlo (dashed/solid lines) and are measured with film.
Another possibility that we considered was to increase the leaf width to ensure it was relatively larger than the size of the source. In this study, a hypothetical MLC design was considered, in which the leaf width was 2 to 3 times more than the actual leaf width of the MIMiC collimator. Figure 5.7 shows similar dose profile comparisons as in Figs.5.4-5.6 but each beamlet now projects a minimum field size of 2x2 cm² for case (a) and 3x2 cm² for case (b). As can be seen, the FSPB dose calculation accuracy improves as the beamlet size increases. However, beamlets of this size are inadequate for clinical IMRT dose delivery and would result in reduced conformity and plan quality.

Another solution to the ‘partial source’ problem would be to increase the source to collimator distance (SCD). The source to collimator distance for the Theratronics 780 Co-60 with binary MLC retrofit is approximately 42 cm (the bottom of the MLC is at 50 cm after adding 8 cm of MLC thickness). In order to see full source, the SCD can be increased to approximately 54 cm to see a 2 cm diameter source. Although this would minimize the underdose problem, it leaves a very small distance between the bottom of the MLC and the SSD of the unit (80cm), making the unit unusable for phantom studies or patient treatments. For these reasons, rather than increasing leaf width or SCD, we propose an aperture based model for which the results are presented in the next section.
Figure 5.6 Relative off-axis dose profiles for Co-60 beams obtained using Monte Carlo calculations and the FSPB model. The profiles in (a) are obtained by opening leaf pairs 1-10 and profiles in (b) are obtained by opening leaves that are away from the central axis (leaf pairs 1-4).

Figure 5.7 This diagram shows the results of two hypothetical MLC designs for which the leaf width is increased 2 to 3 times compared to the MIMiC collimator. Open field profiles for (a) 4x2 cm² and (b) 6x2 cm² field sizes are compared with the FSPB model calculation. Each pencil beam is 2x2 cm² (a) and 3x2 cm² (b) instead of 1x2 cm² in the previous case. The schematic diagrams above the profiles show which leaves were opened.
5.4.2 Aperture Superposition Model

In the aperture superposition (AS) dose model, the consecutive opened beamlets are not considered as the sum of the individual beamlets, but instead as broad beam apertures. Figures 5.8a-d show examples of the intensity modulated Co-60 beams calculated using the AS model and the FSPB model. The calculations are compared against the film measurements. The results show an excellent agreement between the film measurements and the AS model. Most regions are well within 1.5% dose difference. The comparisons to the FSPB model again show that the FSPB model underestimates dose considerably, particularly for larger fields, where a dose difference up to 11% can be seen. It should be noted that the FSPB model is the same as the AS model when an aperture is defined by a single leaf opening. This would be the case when the MLC leaves adjacent to the opened leaf are closed.

The results can be extended to apertures with varying beam weights as shown in Fig. 5.9a. Figs. 5.9b-d show the 2D dose distributions and Fig. 5.9e quantifies the dose differences based on the 1D dose profiles. The high dose regions in the dose distribution generated using the FSPB model are smaller in size compared to those obtained from the film results and AS model. The two tall peaks in Fig. 5.9e further show this difference in dose.

Figure 5.10 shows the gamma maps (Low et al., 2003) between the (a) film measurement and AS model and (b) film measurement and FSPB model. To be consistent with the clinical standards, we used gamma criteria of 3%, 3mm. Regions that are greater than 1 (shown by red on the distributions) indicate disagreement between the calculation and measurement and regions that are less than 1 (shown as blue) indicate
Figure 5.8 A comparison of the intensity modulated fan beam profiles for several MIMiC combinations (a)-(d) obtained with the aperture superposition (AS) and the FSPB model and compared with GafChromic film measurements. The MIMiC schematic diagram for each profile shows which leaf combination was considered for this study. It should be noted that the apertures used for AS model and pencil beams used for FSPB model are calculated by Monte Carlo.
Figure 5-9 (a) An example of the intensity modulated fan beam with varying beam weights. 2-D dose distributions from (b) the film measurement, (c) AS model, and (d) FSPB model. (e) shows 1-D cross-plane profile comparison between the 3 methods.

good agreement. In Fig. 5.10a all gamma values are below 0.8, therefore, there is a good agreement between the AS and the film measurement. Figure 5.10b contains regions that have gamma values as high as 2, indicating a failure in those regions. Overall, the 2D comparisons with film re-establish the dose agreement with the AS model and indicate the under-dosing nature of the FSPB model for Co-60 source. Since this intensity modulated beam consisted of a small number of beams, the differences seen in dose are still relatively small. However, for a full tomotherapy plan, which inherently uses thousands of beams, these differences could further grow, making the FSPB model unacceptable in clinics for Co-60 dose calculations.
Figure 5.10 The percent dose difference maps between (a) the AS model and film and (b) the FSPB model and film. A gamma criterion of 3% and 3mm was used. Regions greater than 1 indicate disagreement between the calculation and measurement. The same color scale is used for both maps. Blue regions in above gamma distributions correspond to gamma values less than 1 and red correspond to gamma values more than 1. Note that in a) gamma values are below 0.8, which means there is no failure and in b) gamma values are as high as 2, indicating discrepancies in dose.

5.5 Conclusions

In this work, a commonly used dose model for photon dose calculations of IMRT beams is investigated. In particular, film measurements and Monte Carlo simulations were performed to determine if the pencil beam superposition model will work for finite size radiation sources such a Co-60 source. The results show that the FSPB dose model underestimates dose notably for non-point like sources. A new model called the aperture superposition model is proposed to calculate dose from a Co-60 source. The results show that the aperture superposition dose model provides accurate dose calculations of intensity modulated beams from the finite size Co-60 source. The work in this paper was limited to fan beams used in tomotherapy deliveries. The application of the aperture superposition model can likely be extended to commercial treatment planning systems for broad beam IMRT. This investigation is currently underway.
5.6 References


Chapter 6: Multi-Source Photon Fluence Model for Co-60 Beams

A version of this chapter is being prepared for submission to Medical Physics as: Sandeep Dhanesar, Johnson Darko, and L. John Schreiner “Multi-source Cobalt-60 model for photon fluence calculations used in convolution/superposition dose calculation method.”

6.1 Abstract

The convolution-superposition method is a commonly used dose calculation method in commercial treatment planning systems for calculating photon dose from point-like radiation sources. In order to implement this method for finite size sources, such as encountered in a Cobalt-60 (Co-60) unit, the photon fluence model needs to be modified. In this paper, we present a multi-source photon fluence model that can be used for finite size sources.

The photon fluence model proposed in this paper is based on the simulation of contributions from one primary radiation source and two scatter radiation sources. Fluence is calculated by the convolution of two functions – the source distribution function and the aperture function. The source distribution functions for different radiation components were determined using a Monte Carlo (MC) simulation for the maximum field size defined by the Theratronic T780 Co-60 unit (Best Theratronics, Kanata, ON). The validity of the MC model was verified with the ion chamber measurements made in a water phantom. The aperture function is a rectangular window function representing the field size. The fluence model was evaluated by comparing the calculated photon fluence to the MC simulated fluence.
The comparisons between the photon fluence calculated using the model proposed in this paper and the MC simulation results show agreement, better than 2% in the in-field region of all large and small central fields. In the tail and sharp dose gradient regions, this agreement is better than 5% for the large field sizes and 1.5% for the small field sizes. The primary and scatter fluence (from source region) output factor are within 1% for fields greater than 4x4 cm\(^2\) and 2% for small fields such as 1x1 cm\(^2\).

The results presented in this paper indicate that the photon fluence models presented here can accurately determine fluence for large and small radiation beams from finite size sources.

6.2 Introduction

Cobalt-60 (Co-60) units played a significant role in radiation therapy worldwide from the mid 1950’s, when they inaugurated high energy external beam therapy, until the late 1970s. However, their current usage has decreased considerably due to introduction of modern technology such as the linear accelerator (linac) based intensity modulated radiation therapy (IMRT) machines (Cadman et al., 2011). Studies suggest that Co-60 units have the potential to play an important role if they are upgraded to deliver dose via methods such as 3D conformal radiation therapy (3DCRT) or IMRT (Van Dyk et al., 1996; Warrington et al., 2002). The new developments would motivate the development of simpler conventional units that could be of interest to small clinics that do not have the manpower to operate complex modern machines.

A number of groups have indicated a strong potential of Co-60 based IMRT. The focus of these studies has been on treatment plan evaluation (Cadman et al., 2011; Adams et al., 2008; Fox et al., 2008; Joshi et al., 2009; Cadman, 2007), treatment delivery (Schreiner et al., 2009; Dhanesar et al., 2007), and on-board imaging (Dhanesar et al.,
Treatment planning has mostly been based on dose modeling methods such as the simple look up table method proposed by Milan and Bentley (Milan et al., 1974), modified linac based methods that use correction factors (Cadman et al., 2011), and Monte Carlo methods (Joshi et al., 2009). A recent study by Dhanesar et al. (2012) proposed an aperture superposition model over the traditional pencil beam superposition model for Co-60 based tomotherapy dose calculations in order to take into account the finite size of the Co-60 source. The doses under apertures in this treatment planning method were pre-calculated by Monte Carlo modeling. While Monte Carlo methods are generally considered the most accurate for treatment planning, the calculations are time consuming; hence they are not clinically practical for calculating dose delivery with different apertures. This mandates the development of a practical dose calculation method for Co-60 beams.

Convolution-Superposition methods are widely used in commercial radiation therapy treatment planning systems for modeling the dose for linac beams that originate from point like sources. These methods are relatively fast compared to the Monte Carlo methods. One of the fundamental components of the convolution method is the source kernel, also known as the source distribution function. The source kernel defines the distribution of the photon fluence at the point where the radiation originates. Often the source kernel is separated into multiple kernels to distinguish between the radiation coming from the target, known as primary radiation, and from the other components of the treatment head, designated the secondary radiation (Liu et al., 1997; Ahnesjo et al., 2005). In a linac the primary source of radiation is effectively a point source. However, the secondary source of radiation, particularly from the flattening filter, is a three dimensional (3D) distributed source. Since the secondary source contributes a significant
amount of fluence to the total fluence, extensive studies have been done on the modeling of the flattening filter source (Liu et al., 1997; Chaney et al., 1994; Ahnesjo, 1992; Ahnesjo, 1994; Kase et al., 1986).

A Co-60 source, which also is a 3D distributed source, can be modeled using somewhat similar principles. [Note that from here onwards, Co-60 source will be referred to as a “Co-60 disk” to avoid confusion with the word “source” used for different components of radiation such as the “primary source” or “secondary or scatter source”]. In this paper we propose a multi-source fluence model that could be used in the convolution-superposition dose calculation method. The fluence model is based on one primary radiation source and two scatter radiation sources. The source distribution functions are calculated using the results of a onetime simulation of the maximum field size in Monte Carlo. The particles reaching the isocenter plane are projected back to their source positions and the source distributions are derived.

6.3 Materials and Methods

6.3.1 Monte Carlo Simulation of Co-60 Beam

A Theratronic T780C Co-60 unit (Best Theratronics, Kanata, Canada) was modeled in BEAMnrc, which is part of the EGSnrc Monte Carlo program (Roger et al., 1995). The Co-60 unit consists of a 2 cm diameter cylindrical source, a fixed primary collimator, and four adjustable collimators. The primary collimator is made of tungsten with thickness of 6.2 cm. It is located at 1.5 cm from the bottom of the source and projects a maximum field size of 35x35 cm$^2$ at the isocenter, which is 80 cm away from the source. The adjustable collimator consists of a series of lead leaves in the x and y directions located below the primary collimator (see Fig 2.3 in Chapter 2).
A beam of field size 35x35 cm$^2$ was simulated at the isocenter using a jaw opening formalism similar to that used by Mora et al (1999) and Joshi et al (2008). The purpose of simulating this open field beam was to model the source distribution for the maximum field size. Further details regarding its usage will be provided in Section 6.3.2.

Ten billion particles were simulated in order to keep the uncertainty of the fluence less than 1.5%. Each simulated particle that reached the scoring plane was recorded in a phasespace file. The phasespace file contained detailed history of the particles such as the particles’ charge, energy, position ($x$, $y$), direction cosine ($u$, $v$), and $LATCH$ (a variable that contains the interaction history of the particle). $LATCH$ is set during the simulation based on the region where the particle is created and the regions where it has interacted. Based on $LATCH$ it was determined if a particle reaching the scoring plan was a primary photon or created from an interaction within the source, primary collimator, or the adjustable collimators.

6.3.2 Convolution Dose Calculation Method

In the convolution-superposition method conventionally used for point sources, the three dimensional dose at a point $(x,y,z)$ in a medium (such as a phantom), $D(x,y,z)$, is calculated by convolving the $TERMA$ (total energy released in medium), $T(x,y,z)$, with the photon dose kernel kernel at that point $K(x,y,z)$ (Liu et al., 1997; Ahnesjo et al., 2005).

$$D(x,y,z) = \iiint T(x',y',z')K(x-x',y-y',z-z')dx' dy' dz' \quad (6.1)$$

The photon dose kernel is generally computed using a Monte Carlo method and the $TERMA$ (Eq. 6.2) is calculated based on the photon fluence, $\Phi(E,x,y,z)$, at a point $(x, y, z)$ and the attenuation coefficient, $\frac{\mu}{\rho}$,
\[ T(x, y, z) = \sum E \frac{\mu(E)}{\rho} E \Phi(E, x, y, z) \]  

(6.2)

The photon fluence within phantom at point \((x, y, z)\) is calculated from the in-air photon fluence at the phantom surface, accounting for the diverging field geometry and attenuation within the medium.

This is a standard convolution-superposition approach used for point sources. This method can be extended to model doses from finite size sources such as the Co-60 source if the photon fluence model is adjusted to account for the source size.

6.3.2.1 Multi-Source Photon Fluence Model

In the convolution superposition method, fluence from particles originating from different components of the treatment head is modeled through a multi-source fluence model. Fig. 6.1 shows a representation of the multi-sources. These source models can be obtained by a onetime simulation described in Section 6.3.1. Using the \textit{LATCH} variable of the phasespace file, the origin of the particles reaching the scoring plane can be determined. Based on the interaction history, the particle fluences are separated into 5 categories:

1. Primary photon fluence \(\Phi_{ps}\): fluence of photons that have reached the scoring plane without interacting with the components of the treatment head;
2. source scatter photon fluence \(\Phi_{ss}\): fluence of photons that have undergone interaction within the source and the shielding around it;
3. primary collimator photon fluence \(\Phi_{pc}\): fluence of photon that are generated due to interactions in the primary collimator;
4. secondary collimator photon fluence $\Phi_{jaws}$: fluence of photons that are generated due to interactions with the adjustable collimators (jaws); and

5. contamination particles fluence $\Phi_{elec}$: fluence of electrons or positrons created anywhere in the treatment head.

Based on the categories above, the total fluence can be obtained as in Eq. (6.3),

$$\Phi = \Phi_{ps}(x, y, z) + \Phi_{ss}(x, y, z) + \Phi_{pc}(x, y, z) + \Phi_{jaws} + \Phi_{elec}(x, y, z).$$  \hspace{1cm} (6.3)

For the fluence model proposed here, only the first four components will be used (we will show in Results section that the contribution from the contamination electrons is negligible).

Figure 6.1: Multi-source representation for Co-60 fluence model. The primary source, located at $Z_{pc}$, is defined as $S_{ps}$. Source scatter source, located at $Z_{ss}$, is defined as $S_{ss}$, the primary collimator source is defined as $S_{pc}$.  

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Liu et al. (1997) used imaging theory to calculate extra focal fluence at the isocenter, where the source kernel for the flattening filter was convolved with a rectangular window function or jaw aperture function. However, for a Co-60 beam one more term was necessary to account for in-field lateral variations in the beam’s energy spectrum, particularly, off-axis softening. In the modified method used, the jaw aperture function is first multiplied by the open field fluence defined by the maximum possible field size and then convolved with the source kernel:

\[
\Phi_{sp}(x, y, z) = \int \Omega_{sp}(x', y', zp) [\Phi(x, y, z)_{sp_{full}} W(x - x', y - y', zp)] dx'dy'
\] (6.4)

In this equation, \(\Omega_{sp}(x', y', zp)\) is the source fluence distribution at the point of calculation, \(\Phi(x, y, z)_{sp_{full}}\) is the photon fluence from the primary source at the isocenter for 35x35 cm\(^2\) field size calculated via a Monte Carlo simulation, and \(W(x - x', y - y', zp)\) is a rectangular window function corresponding to the field size. In Eq. (6.4) \(\Phi_{ss}(x, y, z)\) and \(\Phi_{pc}(x, y, z)\) are calculated similarly, except the photon fluence at the isocenter for the maximum field size and the source distribution functions, \(\Omega_{ss}(x', y', zp)\) and \(\Omega_{pc}(x', y', zp)\) are different. \(\Phi_{jaw}(x, y, z)\) is not calculated based on a source distribution function. It is assumed that the contribution from the jaws increases only the weight of the total fluence. Therefore, to obtain the distribution of the fluence from the jaw region, the first three components’ weight is increased according to the output factors for the jaws.

6.3.2.1.1 Source fluence distribution function

The source fluence distribution is the fundamental component of the fluence model. In order to calculate the source fluence distribution at the calculation plane, the
source fluence distribution at the location of the source needs to be determined. This is particularly difficult when the source is distributed in 3D because it is complex to implement distributed sources in the convolution model. Therefore, we make a simplifying assumption that since the distance between the isocenter and the source is much greater than the dimension of the source itself, the source can be approximated as a 2D disk. The same assumption will be made for both the scattering source and the primary collimator source.

In order to use this disk approximation it is important to set the location of the disk relative to the actual components in the head of the unit. This can be done by determining where the maximum number of photons originate. The distribution of photon fluence as a function of $z$-axis position can be determined from the EGSnrc phasespace variable $ZLAST$. As discussed above, the $ZLAST$ specifies the $z$-axis position of the photons’ last interaction before it reaches the phasespace plane. This $ZLAST$ distribution for the Co-60 unit was analyzed for all particles reaching the scoring plane based on the radiation categories specified earlier in this section. The position of the source was chosen based on where the maximum number of particles originate from.

Once the positions for all three sources are determined, the source distributions were determined by back projecting the particles reaching the scoring plane to the effective source location. This was calculated using an in-house MatLab script, which took the 2D phasespace file as an input and projected the particles from each radiation component to the user defined planes. Figure 6.2 shows the geometry used to project the particles to their source locations. $x_1$, $y_1$, and $z_1$ are positions of a particle at the scoring plane. $x_2$, $y_2$, and $z_2$ are new locations of the particle when projected back. $z_2$ is determined using $ZLAST$ as mentioned in the previous paragraph. $x_2$ and $y_2$ are
determined as a function of the direction cosines of \( x \), \( y \) and \( z \) coordinates. The direction cosines for \( x \), \( y \), and \( z \) are:

\[
\cos \theta_x = \frac{\Delta x}{p}, \quad \cos \theta_y = \frac{\Delta y}{p}, \quad \cos \theta_z = \frac{\Delta z}{p} \tag{6.5}
\]

where, \( \Delta x \), \( \Delta y \), and \( \Delta z \) are the lengths of vectors \( x \), \( y \), and \( z \) and \( p \) is the distance between the points \( A \) and \( B \). Using relations, \( \Delta x = x_2 - x_1 \), \( \Delta y = y_2 - y_1 \), \( \Delta z = z_2 - z_1 \), and \( p = \frac{(x_2 - x_1)}{\cos \theta_z} \), \( x_2 \) and \( y_2 \) can be determined by Eqs. (6.6).

\[
x_2 = x_1 - (z_1 - z_2) \frac{\cos \theta_x}{\cos \theta_z}, \quad y_2 = y_1 - (z_1 - z_2) \frac{\cos \theta_y}{\cos \theta_z} \tag{6.6}
\]

The direction cosines of \( x \) and \( y \) are obtained from the phasespace data, where they are defined by variables \( u \) and \( v \) as discussed previously. The direction cosine of \( z \) can be found using,

\[
\cos \theta_z = \sqrt{1 - (\cos^2 \theta_x + \cos^2 \theta_y)}, \tag{6.7}
\]

which is derived from the relationship \( \cos^2 \theta_x + \cos^2 \theta_y + \cos^2 \theta_z = 1 \).

![Figure 6.2: The projection geometry used to obtain a particle’s position (x and y coordinates) at the new z-axis position.](image)
Now that $x_1$ and $y_2$ are determined at the source plane, the source photon fluence can be determined. Since this source location is the initial location, we will refer to the $x$ and $y$ positions as $x_0$ and $y_0$. The source fluence at the location of the source, $\Omega(x_0, y_0, z_3)$, is determined as:

$$\Omega(x_0, y_0, z_3) = \sum_{\cos \theta_s} \frac{G}{c(x_0, y_0)}$$

(6.8)

where, $G$ is the particle’s weight and $c(x_0, y_0)$ is the area of a ring on which the particles are falling into. The particle weight is also is read from the phasespace data. This source fluence was achieved by reading each particle history one by one using an in-house Matlab program.

To determine how the source distribution is viewed from the phantom surface, the pinhole method (Mohan et al., 1987) from imaging is used to project the source image at the calculation point. As shown in Fig. 6.3, if the aperture is reduced to an ideal pinhole then the source image at the isocenter can be determined as following.

$$\Omega_{ps}(x, y, z_p) = \Omega_{sp} \left( x_o \frac{z_c - z_{ps}}{z_p - z_c}, y_0 \frac{z_c - z_{ps}}{z_p - z_c}, z_p \right)$$

(6.9)

The term $\frac{z_c - z_{ps}}{z_p - z_c}$ takes into account the source divergence though the aperture. $z_{ps}$ is the effective source position for the primary source (ps), $z_c$ is the position from the source location to the top of the lowest aperture, and $z_p$ is the distance from the source to the phantom surface. $\Omega_{ss}(x, y, z_p)$ and $\Omega_{pc}(x, y, z_p)$ are determined similarly but with different source locations and source photon fluence.
6.3.2.1.2 Photon fluence output factor

Due to the finite size of the Co-60 source, only a portion of the source is seen from the phantom surface when the field size is small. The MLC apertures block number of photons reaching the center of the field. Since small fields are commonly encountered in IMRT for dose conformality, it is important to account for the apertures blocking the number of photons reaching the center of the field in order to predict the photon fluence correctly.

There are two ways to calculate output factors. One method is to determine the area of the source seen from the isocenter using the similar triangle relationship (Jaffray et al., 1993; Sharpe et al., 1995; Zhu et al., 1994). The other option is to project the source fluence on the isocenter via a pinhole method (Mohan et al., 1987). Using the later method, the ratio between the primary fluence for a particular field at the center of the field and the total fluence for the maximum open field can be determined.
6.10a-c) provide relative output factors for primary source, scatter source, and primary collimator:

\[ r_{f_{ps}} = \frac{\phi_{ps,f}(x=0,y=0,z_p)}{\phi_{total,open}(x=0,y=0,z_p=80)} \]  
\[ r_{f_{ss}} = \frac{\phi_{ss,f}(x=0,y=0,z_p)}{\phi_{total,open}(x=0,y=0,z_p=80)} \]  
\[ r_{f_{pc}} = \frac{\phi_{pc,f}(x=0,y=0,z_p)}{\phi_{sp,open}(x=0,y=0,z_p=80)} \]  

\[ r_{f_{jaw}} \] is determined by fitting output factors for several open beams simulated using the Monte Carlo into a two degree polynomial function.

6.4 Validation of the Photon Fluence Model

6.4.1 Monte Carlo Simulations

Using jaws definitions described in Section II.A, various symmetric fields such as 30x30 cm\(^2\), 25x25 cm\(^2\), 20x20 cm\(^2\), 15x15 cm\(^2\), 10x10 cm\(^2\), and 7x7 cm\(^2\), as well as asymmetric fields such as 7x30 cm\(^2\), 7x20 cm\(^2\), and 15x7 cm\(^2\) were Monte Carlo simulated to compare with fluence from modeling results. The number of particles simulated was between 10 and 30 billion depending on the field size in order to keep uncertainty less than 1%. As mentioned previously, the scoring plane for these fields was set at the isocenter. As in the simulation of the 35x35 cm\(^2\) field, each component module for these beams was tagged with a LATCH number to identify each phasespace particles with the different source categories.

Since the adjustable collimators of the Co-60 unit define a minimum field size of 7x7 cm\(^2\), several small field simulations were performed using a block collimator. A tungsten block collimator of thickness 8 cm was placed at a distance 42 cm away from
the source. The design of this collimator, for example the divergence, is similar to the binary MIMiC multi-leaf collimator (Best Nomos, Pittsburg, PA) we implemented for our previous investigations (Dhanesar et al., 2012). The binary multi-leaf collimator was used for tomotherapy deliveries where fields varied from 1x1 cm$^2$ to 20x2 cm$^2$, defining a fan beam of radiation. The block was not limited to defining fan beams but could also provide small square fields such as 5x5 cm$^2$, 4x4 cm$^2$, and 3x3 cm$^2$. The purpose of simulating small fields was to determine if the photon model is able to predict the fluence accurately when the full Co-60 source cannot be seen from the measurement plane. Several off axis fields, particularly 1x1 cm$^2$ and 2x2 cm$^2$ were also simulated to test the applicability of the model for such beams. The simulation parameters were similar to those used for open fields except that a large number of particles were used to obtain reasonable fluence at the isocenter.

6.4.2 Monte Carlo Simulation Validation

The Monte Carlo modeling of the unit was tested rigorously against measurements. The phasespace data from BEAMnrc was used by the DOSxynrc program (Walters et al., 2006) to calculate dose in a water equivalent phantom. DOSxynrc is also part of the EGSnrc Monte Carlo system that provides dose information in different phantom materials. The phantom was defined to be of water equivalent density with dimensions 30cm x 30cm x 30cm. The voxel size was 2 mm$^3$. Some select open field phasespace data, as well as certain fan beam data defined by the tungsten block were calculated in water for comparison to measurements made in a water phantom.

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6.4.3 Measurements

The Monte Carlo simulation results from the DOSxyznrc program were validated by comparing the simulated percentage depth dose (PDD) curves and dose profiles for various field sizes with the ion chamber measurements. An IC-10 (0.13 cm$^3$) cylindrical ionization chamber (Wellhofer Dosimetrie, Germany) was used for the broad beam measurements and an A-16 (0.007 cm$^3$) EXRADIN micro point chamber (Standard Imaging, USA) was used for fan beam measurements. The broad beam data was also verified without any add-on collimator. The block collimator was verified by measuring certain fan beams defined by the multi-leaf collimator. Both in-plane and cross-plane profiles were measured, as well depth dose curves. In this paper only broad beam verification results are presented. The fan beam verification results are presented in our previous work (Dhanesar et al., 2012).

6.5 Results and Discussion

6.5.1 Monte Carlo Model Validation

Figure 6.4a compares the Monte Carlo simulated profiles to the ion chamber measurements. For 10x10 cm$^2$, in-field agreement is better than 0.8% and the out of field agreement, particularly at low dose gradients, is 1.5%. For the beam of field size 20x20 cm$^2$, in-field agreement is still less than 1%, however, low dose gradient discrepancy is slightly higher, 2.2%. The percentage depth dose comparison at the central axis is shown in Fig. 6.4b. Other than the buildup region, the agreement for both field sizes is within 1.5%.
Figure 6.4 Comparison of Monte Carlo simulation results to the ion chamber measurement performed in a water tank. Crossplane profiles for 10x10 cm\(^2\) and 20x20 cm\(^2\) are compared in (b) and percentage depth dose (PDD) curves are compared in (b).

### 6.5.2 Open Field Simulation Characteristics

Figure 6.5 shows the results of the open field simulation for the maximum field size of 35x35 cm\(^2\). The contribution from each of the radiation components are normalized to the total fluence. Out of the total number of photons reaching the isocenter, 64.5% are primary photons, 24.6% are scattered photons from the source region, 7.3% are scattered photons from the primary collimator, 3.2% are scattered photons from adjustable collimators (jaws), and 0.4% are contamination electrons. Since the number of electrons contributing to the total fluence is very small, they are not considered while calculating the fluence using the proposed multi-source model discussed in this paper. It is assumed that the contribution from the jaws will only increase the relative intensity of the fluence so their exact extent via source distribution is not determined.
Figure 6.5 Photon fluence profiles through the center of the field size 35x35 cm$^2$ separated into different radiation components.

6.5.3 Source Locations

Figure 6.6 shows the $z_{last}$ distribution for all particles reaching the isocenter for 35x35 cm$^2$ Co-60 beam. It represents the fluence contribution from different locations of the treatment head along the central axis. The region was limited between -10 and 40 cm as there is no component in the treatment head before or after these locations. This allowed a finer bin resolution as BEAMDP program of EGSnrc system limits the number of bins to 200. As specified in the figure, the tallest curve represents the photons coming from the source region. Both primary and scattered photons are included together. The small curve shown is for primary collimator photons. A close-up view is presented to show the contribution from the jaw regions. Figures 6.7a) and 6.7b) separate the primary fluence and scatter fluence from the source region to determine where the maximum fluence contribution is coming from. Based on these results it was determined that both the primary source and the scatter source (from source regions) are located at the same position, which is at -0.17 cm. The bottom of the source is a good estimate for the
source position. For the primary collimator, three positions (top, middle and bottom) were chosen to evaluate which one will give a better primary source model.

Figure 6.6 The zlast distribution of all particles reaching the isocenter for 35x35 cm² Co-60 beam. A close-up view is shown to display the contribution from the jaw region.

Figure 6.7 The zlast distribution for primary (a) and scatter source from source region (c).

6.5.4 Source Fluence Distributions

Figure 6.8 shows the source distributions on the source planes estimated in the previous section. The fluence is normalized to the total of fluence from source region. Fig. 6.8b) shows the fluence distribution of the primary source and Fig. 6.8c) shows the source distribution of the scattered photons from the source region. Fig 6.8a) is the total
of the fluence in b) and c). From the source profiles shown in Fig. 6.9, it can be seen that the primary and scatter source diameter defined by the 50% of the respective fluence intensity is roughly 1.8 cm and 1.7 cm, respectively. It should be noted that although the source in our unit has a physical diameter of 2 cm, the effective diameter in the model will be less since the primary collimator for Theratronic 780C Co-60 unit was originally designed for a source diameter of 1.5 cm. Therefore, “full source” seen from the isocenter of the largest possible fields is still going to be less than 2 cm.

Figure 6.8 Total source distribution (a) obtained with the sum of the primary source distribution (b) and distribution of the scattered photons from the source region (c).

Figure 6.9 Crossplane profiles of the source fluence distributions presented in Fig. 6.8a-b).
Figure 6.10 shows the source distributions of the primary collimator estimated at three different positions. The left distribution corresponds to the top of the primary collimator. As expected, there is no contribution from the center of the primary collimator as there is no material in the center that could generate fluence. The middle distribution shows a relatively smaller area with no photons. This is also expected as some of the photons created before this plane would have diverged into this region. The distribution on the right is achieved by projecting the primary collimator photons to the bottom of the primary collimator. All these three distributions are evaluated to determine which one would give the best primary collimator model.

Figure 6.10 Primary collimator distribution at the top (a), middle (b), and bottom (c) of the primary collimator.

### 6.5.5 Fluence Output Factors

Figure 6.11 shows the output factors for field sizes 35x35 cm$^2$ to 1x1 cm$^2$. The normalization is based on the total output of the maximum field size. The primary radiation output factors from field sizes 7x7 cm$^2$ to 35x35 cm$^2$ are all the same, indicating the full source can be viewed from the calculation point. As the field size decreases, particularly to field sizes on the order of 2x2 cm$^2$ or smaller, the primary radiation output
decreases drastically. This is expected as the block used to define the small fields is blocking some of the primary radiation. Similar observation can be made for scatter radiation from the source region. The uncertainty in the convolution method and the Monte Carlo method is less than 1% when the full source can be viewed from the calculation point and it increases as the field size decreases. For example, the primary output factor for a beam of field size 1x1 cm$^2$ from the Monte Carlo method is 0.42 compared to 0.39 from the convolution method. The scatter output from the source for the same field size is 0.14 and 0.16 for the MC and convolution method, respectively. The primary collimator and jaw scatter output factor are within 2% for fields above 7x7 cm$^2$ but discrepancies increase up to 6% for smaller fields. It should be noted that these percentage differences are relative to the primary collimator and jaw scatter factors, therefore, the uncertainty in the total fluence due to these two components is very small. As discussed before, the scatter output factors from the jaw region is calculated from a polynomial function which was fitted using the Monte Carlo simulated data.

![Output factors diagram](image)

Figure 6.11 Output factors calculated from convolution (Conv.) method and Monte Carlo (MC) method. The total fluence output factor is represented as “all”, and the output factors for primary and scatter radiation are also shown.
6.6 Photon Fluence Distributions

6.6.1 Open Field Distributions

Figure 6.12 shows the total fluence at isocenter for 30x30 cm$^2$ field size. As shown in Fig. 6.12a), the total fluence is normalized to 100%. Fig 6.12b) shows the dose difference map obtained by subtracting Monte Carlo calculated fluence from the fluence calculated using the convolution model. The differences in fluence are within 1% in the in-field region. There are some hot and cold spots of as much as 4% in the high and low fluence gradient regions. The extent of the regions with discrepancies can be seen from the histogram presented in Fig 6.12c. The majority of the difference are within 3%. The lines through figure 6.12a represent the profile locations (crossplane, inplane, and diagonal) presented in Fig. 6.12d-e). These profiles show comparison between the convolution model and the Monte Carlo simulations for the different radiation components, as well as the total fluence. To avoid making profile plots look too crowded, the primary collimator and jaw fluence profiles are plotted as a sum. Profiles from all three locations for the primary and scattered fluence (from source) show that the dose differences are within 2%. As mentioned before, these are in the high and low dose gradients. Small distal difference in the sharp gradient regions will make the dose difference look more pronounced as seen on the dose difference map. The primary collimator fluence was calculated using the source distribution obtained at the bottom of the primary collimator plane. However, the total contribution from the jaws and primary collimator showed a discrepancy in the sharp fluence gradient region more than the two components discussed above. This is because the assumption that jaw fluence only increases the weight of the intensity rather than the overall fluence distribution may not
be completely true. However, because these contributions are relatively small compared to the other two components, the differences in the overall fluence are not prominent.

Open field fluence calculations were also done for various other field sizes. The agreement between the convolution and Monte Carlo calculations improved as the field size got smaller. An example of a 10x10 cm$^2$ field size comparison is shown in Fig. 6.13. The maximum difference is approximately 2%, which is in the high gradient region. The fluence in the low gradient region and in the in-field region is within 0.5 to 1%.

6.6.2 Photon Distributions for Small Fields

As mentioned previously, another MC model of the treatment unit was used to calculate fields less than 7x7 cm$^2$ since the clinical units cannot provide smaller field. Figure 6.14 shows the fluence calculations for a 4x4 cm$^2$ beam. The majority of the regions are within 1.5%. The primary and source scatter fluence has improved, particularly in the high dose gradients, however, the total fluence from the primary collimator and jaws shows worse agreement based on the inplane, crossplane, and diagonal profiles. This shows that these models have limitations for small fields. However because the primary collimator fluence decreases quite a bit with field size, these discrepancies do not contribute much to the overall fluence. Figure 6.15 shows another example of a small field (2x2 cm$^2$). The differences in fluence are again within 1.5%. These are in the high and low dose gradients.
Figure 6.12 Photon fluence calculations for a 30x30 cm² beam: a) 2D fluence distribution calculated using convolution (Conv) method, b) dose difference map between Conv and Monte Carlo (MC) method, c) histogram of the absolute dose difference, (d-f) crossplane, inplane and diagonal profiles for various radiation components. To avoid cluttering figures, only profiles in (d) contain the radiation component labels. The same labels are applicable to the other figures.
Figure 6.13 Photon fluence calculations for a 10x10 cm$^2$ beam: a) 2D fluence distribution calculated using convolution (Conv) method, b) dose difference map between Conv and Monte Carlo (MC) method, c) histogram of the absolute dose difference, (d-f) crossplane, inplane and diagonal profiles comparing conv and MC results for various radiation components.
Figure 6.14 Photon fluence calculations for a 4x4 cm\(^2\) beam: a) 2D fluence distribution calculated using convolution (Conv) method, b) dose difference map between Conv and Monte Carlo (MC) method, c) histogram of the absolute dose difference, (d-f) crossplane, inplane and diagonal profiles comparing conv and MC results for various radiation components.
Figure 6.15 Photon fluence calculations for a 2x2 cm$^2$ beam: a) 2D fluence distribution calculated using convolution (Conv) method, b) dose difference map between Conv and Monte Carlo (MC) method, c) histogram of the absolute dose difference, (d-f) crossplane, inplane and diagonal profiles comparing conv and MC results for various radiation components.
6.6.3 Off-axis photon fluence distributions

The results discussed to this point have been based on symmetric large and small fields. We have also calculated various asymmetric fields to evaluate the applicability of the convolution model in these situations. This is of particular interest as the source distribution model in this paper does not depend on the off-axis position but only on the jaw aperture size. The asymmetric field sizes defined by large beams did not have much difference in fluence. For example the in-field agreement for fields 20x7 cm$^2$ and 15x7 cm$^2$ was within 1.5% and the dose high gradient regions were within 2.2%. To test the extreme cases, the 2x2 cm$^2$ and 1x1 cm$^2$ beams were shifted away from the central axis and their results were compared to the MC simulated data. As shown in Fig. 6.16a, as the 2x2 cm$^2$ beam is shifted to the left from the central axis, the discrepancy in the regions of sharp dose gradient increases. In Fig. 6.16b, it can be seen that this discrepancy is greater when the 1x1 cm$^2$ beam is shifted away from the central axis. This indicates that the amount of source that is obscured in this simulation increases as the field is shifted further away from the central axis. Therefore, while the model presented here provides a fast method for calculating fluence for large and small symmetric fields, and large asymmetric fields (fields greater than 4x4 cm$^2$), further work is needed to account for this increasing obscuring of the source with off-axis position. Work is already underway by our group to modify the source distribution functions for small off-axis fields. This will extend the applicability of this model for IMRT dose calculation, which inherently uses small on-axis and off-axis beamlets to achieve conformal dose. These results will be presented in a follow-up report.
Figure 6.16 In (a) 2x2 cm$^2$ beam is shifted off-axis by 5 cm and 9 cm. In (b) 1x1 cm$^2$ is shifted off-axis by 5.5 cm and 9.5 cm. Shifts refer to opening off-axis leaves of the MLC.

6.7 Conclusions

In this work, a multi-source photon fluence model is investigated for a finite size sources such as the Co-60 source. This model is derived from the principles of convolution-superposition method commonly used in the commercial treatment planning systems. It is not only valid for large beams but also works well for small on-axis beams which are generally difficult to calculate due to some part of source being obscured by the jaws. The results show that this model provides accurate dose calculations for large and small symmetric fields. Although it works well for large asymmetric fields, a discrepancy is found in the sharp gradient regions for small off-axis fields. This is due to a simplifying assumption made in the source distribution function. Work is underway to extend the model to IMRT dose calculations.
6.8 References


Chapter 7: Conclusions

7.1 Potential of Co-60 Tomotherapy

Co-60 teletherapy units were the mainstay of external beam radiation therapy in mid-1950s due to their ability to provide megavoltage treatment. However, since the development of newer linear accelerators x-ray units, the role of Co-60 units in the developed world has been substantially reduced. Although it has been perceived that the fall-out of Co-60 units is due to their beam characteristics such as low energy, low dose rate and broader penumbra compared to linac machines, it has been pointed out in literature that these issues are not problematic if Co-60 units are re-developed to include modern delivery techniques such as asymmetric jaws, multi-leaf collimators and image guidance features (Schreiner et al., 2003; Warrington and Adams, 2002; Suit, 1986).

Several groups in North America have been investigating the potential of Co-60 IMRT (Warrington and Adams, 2002, Cadman and Bzdusek, 2011, Fox et al., 2008). The Medical Physics research group at the Cancer Center of the Southeastern Ontario (CCSEO) is one of them. The focus of the CCSEO group has been on serial tomotherapy since it is easy to implement in a laboratory-setting. Previous studies by the CCSEO group confirmed a strong potential for Co-60-based tomotherapy, using multi-field techniques for simple non-clinical dose patterns, as well as for homogeneous phantoms containing clinically relevant structures. However the clinical practicality was not explored due to lack of accurate dose calculation methods and limitations of the in-house inverse planning system.

The goals of this thesis were to: a) extend the treatment planning capabilities to include clinical cases using computed tomography simulation images, b) further develop
the in-house inverse planning system so that 3D clinical plans could be generated, c) explore dose delivery issues, especially those pertaining to fan beam deliveries, and e) explore an efficient dose calculation model dedicated to Co-60.

The goal of extending Co-60 treatment planning to clinically relevant cases was successfully achieved. Chapter 3 summarized the treatment planning results for two clinical sites: prostate and head and neck. Comparisons of these plans to the plans generated using the clinical modalities showed that Co-60 tomotherapy can provide conformal radiation therapy. The work presented in Chapter 3 was limited to one slice only. The in-house planning system was further developed during this research period to extend single slice planning to multi-slice planning. Chapter 4 presented 3D volumetric treatment plans. In this chapter, plans for various clinical sites such as the brain, eye, head and neck, stomach, and prostate were presented. The Co-60 beams are often criticized for having broader penumbra, which can worsen dose conformality to targets. This is particularly viewed to be a problem for small structures wrapped around the critical structures since a high degree of conformality of the high treatment dose to targets is required in order to spare the surrounding healthy tissues and organs. The plans for brain, eye and head and neck sites, sites which all present targets in close proximity to the critical structures, showed that the penumbra of the Co-60 beams is not a major concern when using tomotherapy delivery approach. Comparisons to clinical 6 MV beam IMRT plans showed that Co-60 plans of similar quality can be obtained for these sites. Also, as mentioned previously, another concern with the Co-60 beams is low energy, which is known to deposit high surface doses when using conventional delivery approaches. Because of this, the deep seated tumours such as in pelvic region are perceived to be poor candidates for Co-60 based radiation therapy. The treatment plans for stomach and
prostate showed that this is not true for Co-60 based tomotherapy plans since doses to normal tissue and critical structures in the Co-60 plans are similar to those achieved with the clinical IMRT plans.

Chapter 5 demonstrated the issues with the Co-60 tomotherapy fan beam deliveries using the conventional pencil beam superposition dose method. It was shown that although doses for IMRT or tomotherapy segments for point sources can be calculated based on the finite size pencil beam superposition method, the same cannot be used for the Co-60 source because of its large finite size. A new aperture superposition dose model was developed, which provided better agreement between calculations and measurements. The apertures used this model were based on look tables created using Monte Carlo simulations. The use of Monte Carlo based data for such modeling is not efficient since there can be numerous MLC aperture possibilities depending on complexity of the beam modulations required for good dose coverage. Chapter 6 proposed a fluence calculation dose model for Co-60 source based on a convolution approach. This method provided fast and accurate aperture calculations.

Overall, the goals of this thesis work have been met. The investigations have continued to show a strong potential for Co-60 tomotherapy. The results of the investigations made during this project have attracted attention from the medical physics community. The CCSEO is now collaborating with a radiotherapy industry leader to push the development of the Co-60 units so that modern radiation can be made more accessible in the world.
7.2 Future Work

The work presented in this thesis has made significant contributions to Co-60 tomotherapy treatment planning and dose calculation methods. This work can be further extended as following:

- Co-60 conformal dose delivery has been achieved using simple pencil beam blocks. In order to move towards more efficient delivery methods, a multi-leaf collimator MIMiC has been incorporated into a treatment unit. Conformal dose deliveries for plans based on MIMiC need to be delivered.

- Chapter 5 showed that doses can no longer be added using pencil beam superposition method for finite size sources. An aperture superposition method was proposed. This aperture superposition method needs to be incorporated into the MLC segmentation algorithm. This step would be necessary before conformal dose deliveries can be achieved with the MIMiC collimator.

- In chapter 6, a convolution based fluence model was proposed for fast and accurate dose calculations. This work, in conjunction with work in Chapter 5, was for dose calculations using tomotherapy approach. Preliminary work has been performed to extend dose calculation model to Co-60 based broad beam IMRT using a more flexible two dimensional MLC. This model needs to be evaluated with Co-60 IMRT broad beams calculated in Monte Carlo. This work is now planned at the CCSEO through a project with Best Theratronics to upgrade the T780 cobalt unit with features making broad beam conformal delivery including IMRT possible.
7.3 References


Appendix A
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Appendix B

BEAMnrc Input File Example

An example of the BEAMnrc input file for Theratonics T780C Co-60 units with MIMiC MLC

gTomo780c_sdMIMiC Leaf 19 ab (1x2 finger) MIMiC Tongue and Groove
#!GUI1.0
AIR521ICRU
0, 0, 0, 0, 1, 3, 0, IWATCH ETC.
50000000000.0, 29, 65, 400, 1, 10, 2, 0, NCASE ETC.
0, 3, 0, 1, -2.91, -0.11, 0.0, 0.0, 0.0, 0.0, IQIN, ISOURCE + OPTIONS
1, SPECTRUM
X:/HEN_HOUSE/spectras/bareco60.spectrum
1
0, 0, 0.7, 0.01, 0, 2, 0.7, 0, ECUT, PCUT, IREJCT, ESAVE
0, 0, 0, 0, 0, PHOTON FORCING
1, 9, SCORING INPUT
0, 1
0, DOSE COMPONENTS
-8.67, Z TO FRONT FACE

************** start of CM FLATFILT with identifier srchead **************
10, RMAX
Cobalt 60 source and housing with lead shileding
-8.67, ZMIN
5, NUMBER OF LAYERS
1, 5, # CONES, ZTHICK OF LAYER 1
1.165,
1.165,
2, 0.12, # CONES, ZTHICK OF LAYER 2
1, 1.165,
1, 1.165,
2, 0.64, # CONES, ZTHICK OF LAYER 3
1, 1.165,
1, 1.165,
2, 2.8, # CONES, ZTHICK OF LAYER 4
1, 1.165,
1, 1.165,
2, 0.11, # CONES, ZTHICK OF LAYER 5
1, 1.165,
1, 1.165,
0.7, 0.01, 0, 2,
P521ICRU
0.7, 0.01, 0, 2,
PB521ICRU  
0.7, 0.01, 0, 2,  
W521ICRU  
0.7, 0.01, 0, 2,  
STEEL521ICRU  
0.7, 0.01, 0, 2,  
W521ICRU  
0.7, 0.01, 0, 2,  
STEEL521ICRU  
0.7, 0.01, 0, 2,  
W521ICRU  
0.7, 0.01, 0, 2,  
STEEL521ICRU  
0.7, 0.01, 0, 2,  
W521ICRU  
0.7, 0.01, 0, 2,  
STEEL521ICRU  
0.7, 0.01, 0, 2,  
W521ICRU  
0.7, 0.01, 0, 1,  
CO521_588  
0.7, 0.01, 0, 2,  
STEEL521ICRU  
0.7, 0.01, 0, 2,  
W521ICRU  
0.7, 0.01, 0, 2,  
STEEL521ICRU  
0.7, 0.01, 0, 2,  
W521ICRU  
0.7, 0.01, 0, 2,  
*********** start of CM PYRAMIDS with identifier pricoll ***********
10, RMAX
Primary Collimator
1, 0, # LAYERS, AIR OUTSIDE
1.524, 7.724, 1.072, 2.253, -1.072, -2.253, 1.072, 2.253, -1.072, -2.253, 10, 10,
0.7, 0.01, 0, 8, ECUT ETC. FOR AIR
0.7, 0.01, 0, 3,
W521ICRU

*********** start of CM JAWS with identifier leaf1 ***********
10, RMAX
Leaf 1
2, # PAIRED BARS OR JAWS
Y
8.034, 10.653, 1.085, 1.091, -1.085, -1.091,
X
10.97, 13.749, 2.121, 2.43, -2.121, -2.43,
0.7, 0.01, 0, 8,
0.7, 0.01, 0, 4,
PB521ICRU
0.7, 0.01, 0, 4,
PB521ICRU
10, RMAX
Leaf 2
2, # PAIRED BARS OR JAWS
Y
14.066, 16.128, 1.098, 1.102, -1.098, -1.102,
X
16.447, 18.512, 2.73, 2.959, -2.73, -2.959,
0.7, 0.01, 0, 8,
0.7, 0.01, 0, 4,
PB521ICRU
0.7, 0.01, 0, 4,
PB521ICRU

10, RMAX
Leaf 3
2, # PAIRED BARS OR JAWS
Y
18.828, 20.89, 1.108, 1.112, -1.108, -1.112,
X
0.7, 0.01, 0, 8,
0.7, 0.01, 0, 4,
PB521ICRU
0.7, 0.01, 0, 4,
PB521ICRU

10, RMAX
Leaf 4
2, # PAIRED BARS OR JAWS
Y
23.59, 25.652, 1.118, 1.122, -1.118, -1.122,
X
25.972, 27.638, 3.788, 3.973, -3.788, -3.973,
0.7, 0.01, 0, 8,
0.7, 0.01, 0, 4,
PB521ICRU
0.7, 0.01, 0, 4,
PB521ICRU

15, RMAX
atch
3, 0, #LAYERS, AIR OUTSIDE
37.1, 38.6, 5.75, 5.75, -5.75, -5.75, 0.9, 0.9, -0.9, -0.9, 9, 5.25,
************ start of CM VARMLC with identifier vmlc ************

10, RMAX
MIMIC - Binary Multileaf Collimator
0, 3, ORIENT, NGROUP
40.7, ZMIN
8, ZTHICK
1, 3
20, 0.514858
1, 3
-8.230114, START
0.0, 0.0, WSCREW, HSCREW
0.028194, 3.19024, 43.08538, WTONGUE, HTONGUE, ZTONGUE
0.028194, 3.2004, 43.0803, WGROOVE, HGROOVE, ZGROOVE
0.00508, LEAFGAP
1, ENDTYPE
-11.48719761, ZFOCUS or RADIUS of leaf ends
-11.48719761, ZFOCUS of leaf sides
0, 0, 1
0, 0, 4
-0.432, 0.432, 5
0, 0, 2
-0.432, 0.432, 5
0, 0, 4
0, 0, 1
0, 0, 0, 8.
AIR521ICRU
20, 0, 0, 6, 0,
W521ICRU

************ start of CM SLABS with identifier slab2 ************

10, RMAX
optional slab1
1, NSLABS
79.5, ZMIN
0.5, 0.7, 0.01, 0, 7, 0
AIR521ICRU

******************************** end of all CMs ********************************
Global ECUT= 0.7
Global PCUT= 0.01
Global SMAX= 5
ESTEPE= 0.25
XIMAX= 0.5
Boundary crossing algorithm= PRESTA-I
Skin depth for BCA= 0
Electron-step algorithm= PRESTA-II
Spin effects= On
Brems angular sampling= Simple
Brems cross sections= BH
Bound Compton scattering= Off
Pair angular sampling= Simple
Photoelectron angular sampling= Off
Rayleigh scattering= Off
Atomic relaxations= Off
Electron impact ionization= Off

:Stop MC Transport Parameter: