Intensity Modulation of Therapeutic Photon Beams Using a Rotating Multileaf Collimator

by

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Abstract

This thesis describes the development and implementation of a novel method of delivering intensity modulated radiation therapy (IMRT) that provides greater accuracy and spatial resolution than currently available methods. Through improvements in multileaf collimator (MLC) based fluence generation, a dose distribution may be generated that conforms more closely to the tumour target volume. Healthy tissue surrounding the target volume will therefore receive less dose, reducing the probability of side effects and allowing the physician to increase the prescribed tumour dose (dose escalation).

MLC based IMRT techniques are well established but suffer several physical limitations. Dosimetric spatial resolution is limited by the MLC leaf width, interleaf leakage and tongue-and-groove effects degrade dosimetric accuracy and the range of leaf motion limits the maximum deliverable field size. Based on observations from a linear systems model of dosimetric spatial resolution degradation it is hypothesized that, by rotating the entire MLC between each sub-field, improvements will be obtained in spatial resolution, dosimetric accuracy and maximum deliverable field size. To generate arbitrary fluence maps in this way, a series of unique algorithms were developed that are capable of determining the necessary rotated MLC segments. These IMRT fields may be delivered statically (with the collimator rotating to a new position in between sub-fields) or dynamically (with the collimator rotating and leaves moving simultaneously during irradiation). A full description of the rotational leaf motion algorithms is provided.
An analysis of the rotating leaf motion calculation algorithms with focus on radiation efficiency, the range of collimator rotation and number of segments is provided. The mechanical and radiation producing characteristics of standard linear accelerators under collimator rotation conditions are also investigated. The technique is evaluated by characterizing the ability of the algorithms to generate rotating leaf sequences for desired fluence maps. Comparisons are also made between our method and conventional sliding window and step-and-shoot techniques. Results show improvements in spatial resolution, reduced interleaf effects and maximum deliverable field size over conventional techniques. Clinical application of these enhancements can be realized immediately with static rotational delivery although improved control of the MLC will be required for dynamic delivery.
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Chapter 1

INTRODUCTION

1.1 Radiation Therapy

Cancer is a disease defined by the uncontrolled growth and spread of mutated host cells. Although it is not the main cause of fatalities it is currently the main source of potential years of life lost in developed countries. Currently, 130,000 and 1,200,000 people are diagnosed with cancer in Canada and the United States respectively each year [1].

Radiation Therapy is the method of treating a disease through the delivery of radiation. Radiation Oncology is the sub-specialty of radiation therapy specific to cancer treatment. Cancerous tissue is destroyed primarily through damage caused by ionizing radiation. Cell death results from molecular changes in DNA and other critical cell components caused by direct or indirect ionization [2]. The amount of energy per unit mass deposited through ionization at a point is defined as the radiation dose (Gray). The goal of radiation therapy is to generate an accurate distribution of dose in a specified target volume while minimizing the amount of radiation to the surrounding healthy tissue. For approximately half of cancer patients the aim of radiation therapy treatment is to completely eradicate the disease (curative). For the other half the aim is to minimize pain and suffering (palliative). Other forms of treatment include surgery or chemotherapy and are often given in combination with radiation to increase tumour control. In this chapter an introduction to various aspects of radiation therapy relevant to the thesis are presented. For a more detailed
introduction to radiation oncology the reader is referred to Jacob Van Dyk's compendium 'The Modern Technology of Radiation Oncology' (Medical Physics Publishing, 1999) [3].

1.1.1 Historical Background

X-rays were discovered by Wilhelm Roentgen in 1895, signaling the birth of an entirely new branch of physics research. Shortly after his discovery it was realized that high energy photons would be useful in medical applications. Firstly, the penetrating characteristics of x-rays could be used in a non-invasive way to acquire internal images of the body. Secondly, in combination with investigations performed by Henri Becquerel and Marie Curie into radioactivity, a new method of treating disease was invented. The non-invasive and penetrating nature of radiation was seen as an alternative to surgery and, in 1899, the first patient to be cured by radiation was reported [4].

Initial uses of radiation for cancer treatment consisted mainly of surface malignancies where dose was delivered by radioactive sources applied directly to the lesion [5]. In 1922 x-ray tubes capable of generating a photon spectrum with maximum energy of 200 keV were introduced, allowing for the treatment of more deeply located disease. Higher energy particle accelerators were developed in the 1940s which provided x-ray beams with peak energies in the MeV range. Cobalt-60 sources, produced in a nuclear reactor, were introduced in 1951. Having a high specific activity, average photon energy of 1.25 MeV and a half life of 5.26 years Cobalt-60 quickly became the most popular method of radiation treatment and is still used in developing countries as well as some larger radiation therapy centres in the western world.

The most important development in radiation production as applied to radiation therapy was the introduction of linear accelerators in the 1960's. Today, the
linear accelerator (linac) is the most common device used to generate megavoltage energy photons and electrons for cancer treatment.

1.1.2 Goal of Radiation Therapy

The goal of radiation therapy is to destroy cancerous cells without harming the surrounding normal tissue. Unfortunately, it is inevitable that some normal tissue cells are irradiated. The amount of cancerous cell destruction is therefore balanced by the limits imposed by healthy tissue.

1.1.3 Tumour/Healthy Tissue Response to Dose

The response of different cell types to dose is complex. Radiation Biology is the study of cellular response to radiation. The mechanism of cell death resulting from dose deposition has been studied in detail but is still an area requiring investigation. Tumour or tissue response may be defined as the percentage probability that the tumour or tissue is permanently destroyed through the process of irradiation. A classical model of tumour response to radiation derived from in vitro experimental data is characterized by the sigmoidal dose-response curve \[^2\] in Figure 1.1(a). At low doses, the response of the tumour is negligible. Only at some minimum threshold, in this case 30 Gy, is there enough dose to kill a significant amount of cancerous tissue. Increasing the dose further results in a dramatic increase in the number of cells that are destroyed. Finally, as the percentage of cell death approaches 100\%, the slope tapers off and asymptotically approaches complete tumour control. A normal tissue dose-response curve is also displayed in Figure 1.1(a). It follows the same sigmoidal form of the tumour curve but is offset to higher dose, showing that normal tissue is less sensitive to radiation. The dose selected for treatment will be at a level between the two curves so that tumour and normal tissue response are maximized and minimized respectively (see example Figure 1.1(a)).
Figure 1.1: A classical representation of healthy tissue and tumour response to radiation dose is shown in (a). Both curves have a sigmoidal form. At low doses the amount of cell kill is negligible but increases dramatically at a given threshold. In the classical representation the tumour response is considered to be greater than normal tissue. By choosing a dose midway between the two curves adequate tumour control can be achieved resulting in only a small amount of healthy tissue damage. A more realistic clinical representation is shown in (b). The normal tissue curve is now to the left of the tumour curve, indicating that it is more sensitive to dose. Also, the tumour curve is less steep and plateaus before 100%, due most likely to heterogeneity of the tumour cells and prior spread of metastatic disease.

Dose response curves derived from the limited clinical data available suggest that this simplified model is misleading in some respects. A more plausible clinical representation of dose response is shown in Figure 1.1(b) [3]. The normal tissue and tumour response curves have the same form but show some important relative differences. The normal tissue curve and tumour curves have similar low dose thresholds but the normal tissue curve has a steeper slope. Normal tissue is therefore more sensitive to lower doses than the tumour. The reduced slope of the tumour curve results most likely from the heterogeneity of cancerous cells throughout the
tumour volume. Finally, it has been suggested that the tumour curve achieves its upper threshold at less than 100% control due to metastatic disease not located at the tumour site [3]. Although the cells in the original tumour can be completely destroyed the tumour may re-grow from the metastatic cells not located at the tumour site.

The model presented in Figure 1.1(b) emphasizes the need to reduce dose to healthy tissue in radiation delivery. Due to the steep slope of the healthy tissue curve only a small increase in dose will result in a significant increase in healthy tissue damage. There is a clear necessity for improved accuracy and spatial resolution in dose delivery and it has therefore been a focus of research throughout the history of radiation therapy. This thesis is devoted to the development of an improved method of dose delivery that has increased spatial resolution and accuracy over currently available techniques. It should be noted that the quantification of improvement and benefit in radiation therapy is challenging. The merits of any ‘improvement’ have to be measured in terms of an increase in patient life span and quality of life. The relationship between an improved dose distribution that conforms more closely to the tumour volume and the overall benefit to the patient is not well defined. Only after years of post irradiation follow-up with statistically large enough groups of patients can any technical improvement be related to a clear benefit.

1.1.4 Treatment Planning

With the development of penetrating megavoltage photon beams it became necessary to estimate, at least two-dimensionally, the distribution of dose that would result from multiple beams entering the patient from different directions. The arrangement of beams used to treat a given patient must be chosen so that the dose distribution will conform to the radiation oncologist’s prescription. Before the advent of computers, atlases of 2-dimensional dose distributions were used to predict the contribution from
each radiation beam [6]. The final distribution was the weighted sum of each beam obtained from the atlas. With computers it became possible to eliminate tedious manual calculation of multiple beam geometries. Dose calculation methods were developed to predict dose deposition for arbitrary geometries (see section 1.2.4), allowing more flexibility and accuracy when planning a treatment with different radiation fields.

1.1.4.1 CT/MR Imaging

One of the benefits of radiation therapy is that it is relatively non-invasive. In order to preserve this benefit it is necessary that the location of the target volume also be determined through non-invasive means.

Initially, physicians were guided by simple techniques such as conventional radiography and surface palpation. In the 1970s a new 3-dimensional imaging modality known as computed tomography (CT) scanning was invented [7]. By using radiographic projections obtained at varying angles around the patient the 3-dimensional distribution of physical density inside the patient is calculated. With this improved anatomical information physicians were able to define the location of the target volume in 3-dimensional space with significantly greater accuracy. Magnetic Resonance Imaging (MRI) was introduced in the 1980s providing additional 3-dimensional information that was complementary to CT image data [7].

Three-dimensional anatomical information combined with improvements in computer technology provided an environment for modeling different beam combinations on patient anatomy prior to treatment. By modeling different potential treatments using the patient’s image data it is possible to determine the beam configuration that more closely meets the physician’s prescription.
1.2 Dose Deposition

The mechanism of photon dose deposition involves the transfer of energy from the photon to tissue through particle interactions in the tissue. In the 0 to 25 MeV energy range produced by clinical linear accelerators there are six basic types of interactions that can occur. They are: Rayleigh scattering, photoelectric effect, Compton effect, pair production, triplet production and nuclear photodisintegration. Rayleigh scattering is elastic and therefore does not contribute to dose. The photoelectric effect, Compton effect and pair production occur with the greatest probability in this energy range.

1.2.1 Photon Interactions

The photoelectric effect occurs when there is a collision between a photon and an atom resulting in a bound electron being ejected from that atom. A diagram of the photoelectric process is shown in Figure 1.2(a). The kinetic energy transferred to the electron, $E_{\text{trans}}$, is given by the difference between the incident photon energy, $h\nu$, and the binding energy of the electron, $BE$,

$$E_{\text{trans}} = h\nu - BE. \quad (1.1)$$

The probability of a photoelectric interaction occurring is greatest at photon energies that are slightly above the binding energy of the electron.

Compton interactions occur when a photon interacts with a loosely bound or free electron. In this interaction the electron absorbs some of the photon energy while the rest remains with the scattered photon as shown in Figure 1.2(b). The energy transferred to the electron is given by

$$E_{\text{trans}} = h\nu - h\nu' \quad (1.2)$$
where $h \nu'$ is the energy of the scattered photon.

\begin{itemize}
  \item[(a)] Photoelectric
  \begin{itemize}
    \item Incident Photon
    \item Ejected Electron
  \end{itemize}
  \item[(b)] Compton
  \begin{itemize}
    \item Incident Photon
    \item Scattered Photon
    \item Compton Electron
  \end{itemize}
  \item[(c)] Pair Production
  \begin{itemize}
    \item Incident Photon
    \item Positron
    \item Electron
  \end{itemize}
\end{itemize}

\textit{Figure 1.2: For photons having an incident energy in the clinical range (0 to 25 MeV) the photoelectric effect (a), Compton effect (b) and pair production (c) are the most common types of interactions resulting in a transfer of energy to electrons in the medium.}

Pair Production occurs when an incident photon is replaced by an electron-positron pair as shown in Figure 1.2(c). The interaction occurs when a photon is stimulated by the electromagnetic field of an atom nucleus. Because the rest mass
energy of the electron and positron, $m_e$, must be created in the interaction, the resulting energy transferred to the positron-electron pair is:

$$E_{\text{trans}} = h \nu - 2m_e$$  \hspace{1cm} (1.3)

### 1.2.2 Electron Energy Transfer – Stopping Power

The mechanism of energy transfer to electrons was introduced in the previous section. Energy is transferred in the form of kinetic energy, thereby imparting a velocity to the electron that brings it to a different location in the medium [6]. Electric forces act on the electron as it passes through the medium, causing it to lose energy and slow down. The stopping power of the medium is defined as the rate of particle energy loss per unit thickness. Energy losses can be divided into two categories, ionizational and radiative. The former results in further ionization of electrons along the initial electron trajectory. Each one of these ejected electrons will also undergo their own ionizational and radiative losses until they come to rest. Radiative losses occur due to Bremsstrahlung processes that result in photon production and do not contribute directly to energy deposition in the medium.

### 1.2.3 Fluence-Dose relationship

Before discussing the relationship between photon fluence and absorbed dose it is necessary to present the quantity kerma, $K$, defined by the kinetic energy transferred from photon to electrons per unit mass.

$$K = \Phi \left( \frac{\mu}{\rho} \right) E_{\nu}$$  \hspace{1cm} (1.4)
where $\Phi$ is the input photon fluence (see equation 1.8), $\left( \frac{\mu}{\rho} \right)$ is the mass attenuation coefficient (or photon interaction probability cross section) and $E_{tr}$ is the average energy transferred to the medium from each interaction.

Kerma differs from dose in that it is a measure of the energy transferred to the medium at a point and not the final energy absorbed at that point [6]. In the case where there is an equilibrium between kerma and absorbed dose (i.e. the number of electrons entering the point of interest is equal to the number that are set in motion) the relationship between absorbed dose and fluence may be written:

$$D = \Phi \left( \frac{\mu}{\rho} \right) \bar{E}_{ab} \quad (1.5)$$

where $\bar{E}_{ab}$ is equal to the average energy absorbed in the medium from each interaction. In general, the equilibrium condition is not completely satisfied and there is no simple method of expressing dose in terms of fluence. Monte Carlo calculation techniques which model individual interactions for large numbers of photons and electrons are the most accurate method of calculating dose. Unfortunately, calculation times on currently available computers are not yet adequate to make this method clinically useful, although commercial systems\(^1\) are currently being developed.

\(^{1}\) Peregrine, Livermore, CA
MDS Nordion, Kanata, ON
1.2.4 Pencil Beam Dose Deposition

A simplified model of dose deposition has been suggested by many investigators that considers the photon beam as a series of infinitely thin pencil beams as shown in Figure 1.3(a). Due to photon scatter and electron transport, when the pencil beam interacts with tissue it will deposit dose in and around the area of interaction. The distribution of radiation resulting from one pencil beam is a symmetric function referred to as the Dose Spread Kernel (DSK) [8, 9]. A radiation field is made up of multiple pencil beams. Each pencil contributes to the total dose in a cumulative fashion. The final dose distribution is therefore the sum of each DSK for each pencil beam as described in Figure 1.3(b). Each pencil beam is infinitely small making the number of DSKs that need to be summed infinite as well. The limit of the summing operation as the number of pencil beams approaches infinity is a convolution.

![Figure 1.3: The pencil beam model of dose deposition. A single infinitely thin pencil beam of photons will generate a distribution of dose in and around the point of interaction as shown in (a). The spread of dose resulting from the pencil beam is the Dose Spread Kernel (DSK). (b) With multiple pencil beams the total dose is the sum of all DSKs. In the limit as the number of pencil beams approaches infinity the calculation becomes a convolution of the incident fluence by the DSK (c).](image-url)
Therefore, by convolving the incident photon fluence by the DSK it is possible to approximate the final dose distribution [10-12] as shown in Figure 1.3(c). This simplified model assumes various characteristics of the photon beam and medium that are not completely realistic. It is assumed that the photon energy spectrum is spatially invariant and that the medium is composed of a homogenous material. Variations on the basic model have been introduced in commercial treatment planning systems to account for these limitations [13].

1.3 Basic Dose Delivery Techniques

There are several methods of delivering radiation that try to meet the goal of providing a uniform dose to the target volume and minimal dose to surrounding tissue. The focus of this section is to describe the relevant methods of dose delivery used with high-energy photon beams.

A plot of dose as a function of depth for a 10cm x 10cm 6 MV (peak energy of 6 MeV) photon beam is shown in Figure 1.4. Near the surface there is a region where electronic equilibrium has not yet been attained and dose is being deposited primarily downstream (buildup region). The location of dose maximum occurs at 1.5 cm depth after which the dose slowly decreases due to attenuation and increasing distance from the source. Due to the decrease in dose with depth it is not possible to arrive at a uniform dose in the tumour volume. Multiple fields must be used to achieve the treatment goals.
Figure 1.4: A plot of absorbed dose versus depth for a square 10cm x 10cm 6 MV photon beam. Dose is given as a percentage of the maximum located at a depth of 1.5 cm.

1.3.1 Multiple Fields

The linac is mounted on a gantry that rotates about the patient as shown in Figure 1.5(a). Dose is delivered to the patient from multiple directions with the center of rotation located inside the tumour volume [5]. The dose inside the tumour is the cumulative sum of the contribution from all beams. The dose in the surrounding tissue is the contribution from a subset of those fields that depends on the orientation and position of each field. A typical prostate treatment dose distribution is shown in Figure 1.5(b). In this case the prostate receives the total contribution of dose from two sets of opposing fields while the surrounding tissue receives a maximum dose given by the sum of only one of the opposing beam pairs. Choice of beam orientation is a manual procedure and often requires multiple iterations before an adequate dose distribution is obtained. Full computer optimization of beam orientations has been investigated by others [14, 15] but has not yet achieved clinical implementation.
Figure 1.5: The linac rotates about the isocenter as shown in (a), allowing delivery of radiation to the target volume from different angular directions. An axial CT slice showing four fields of a conformal field prostate treatment plan is presented in (b). The maximum dose is located in the target volume where the four fields intersect. The majority of surrounding tissue receives dose from only two of the four fields.
1.3.2 Arcs

In addition to delivering multiple fields at fixed gantry angles, radiation may also be delivered while the linac gantry is rotating [5]. In some cases the location of critical healthy tissue structures is such that by spreading the dose to the surrounding tissue throughout an arc, the dose to those structures will be minimized. The work presented in this thesis involves a treatment method using fixed gantry angles, therefore arc treatment techniques will not be discussed further.

1.4 Field Shaping

When the linac electron beam interacts with the target (photon source), high energy photons radiate from the target preferentially in the forward direction [6]. A ‘flattening filter’ placed downstream from the target is used to modify the initial fluence so that the output fluence is relatively constant over its spatial extent. Beneath the flattening filter is a fixed ‘primary’ collimator that defines the maximum possible field size. Below that is a secondary collimator consisting of two pairs of attenuating metal blocks that define the field used to treat the target volume (Figure 1.6(b)). These ‘jaws’ may be moved in and out of the beam thereby defining an arbitrary rectangular field shape up to a maximum of 40 cm x 40cm. Because tumours grow into a variety of shapes, a rectangular field will be limited in its ability to conform to the treatment volume and reduce dose to the surrounding healthy tissue.

One technique used to improve the field conformity is to create custom metal alloy blocks that match the shape of the tumour [5] as shown in Figure 1.6(c). By using low temperature alloy materials (e.g. Cerrobend) high attenuation blocks may be built that conform to the target volume. The blocks are mounted below the secondary jaws. Because the projected shape of the tumour is different at each gantry angle, a different set of blocks must be built for each field. This process of field shaping is cumbersome, requiring the apparatus and manpower to generate custom
blocks for each patient. Also, because there is a separate set of blocks for each field it is necessary to enter the room and change them after each field.

**Figure 1.6:** A collimation system located below the source shown in (a) is used to shape the radiation field to the target volume. Two sets of translatable jaws are used to define a rectangular field shape (b). A more tightly conforming field shape may be obtained by adding alloy blocks below the secondary collimator as shown in (c)

1.4.1 **Multileaf Collimator**

In recent years a new device for two-dimensional field shaping has been introduced. The Multileaf Collimator (MLC) consists of a series of tungsten alloy leaves that move parallel to each other in and out of the radiation field as shown in Figure 1.7(a). The MLC is typically located below the secondary jaws. A picture of a partially disassembled MLC is shown in Figure 1.7(b). Each leaf is connected to a separate computer controlled motor allowing it to be positioned independently of the other leaves [16]. A field that conforms to the desired treatment shape is generated by moving the leaves to a position where the projection of the end of each leaf abuts the edge of the treatment volume.
Figure 1.7: (a) Each leaf of the multileaf collimator (MLC) is translated individually in and out of the radiation field using a separate motor. By abutting the leaf edges with the edge of the treatment volume, field shaping conforming to the tumour is created. A photograph of an MLC assembly is shown in (b).

Due to the point source geometry of photon production, the physical leaf dimensions are smaller than their projection at the target. Typical leaf width projections are on the order of 5 and 10 mm at the isocenter (100 cm from the source). The actual physical leaf widths are roughly half that width (2.5 mm and 5 mm). The Varian MLC leaves used in this thesis are composed of a Tungsten Alloy and are 6 cm deep, attenuating greater than 95% of the incident photon beam. For a typical maximum field size of 40x40 cm² (defined at the isocenter) a minimum of 80 leaves and motors (40 on each side) are required. Smaller leaf width MLCs are available but have a more limited field size. A more detailed discussion of MLC characteristics is presented in Chapter 3.

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2 Varian Oncology Systems, Palo Alto, CA
1.5 Intensity Modulated Radiation Therapy

The previous section presented methods of conforming a relatively uniform photon beam to a target volume through the use of collimators. A uniform dose may be obtained in the target volume with the appropriate choice of beam orientation. In some cases, due to the particular combination of beam orientation or irregularities in the patient surface, it is also necessary to modify the intensity of the beam across its two-dimensional extent [5, 6]. For example, a wedge shaped attenuator may be placed in the field to create a one-dimensional gradient at some orientation of the collimator. The appropriate choice of gradient will improve the dose distribution and bring the plan closer to the treatment goals.

Historically, two-dimensional modification of photon beam intensity has been limited to simple gradients and irregular surface compensators. In the last decade there has been a large body of research and development devoted to methods for generating complex arbitrary 2-dimensional intensity maps. The rationale behind these developments is that with 2-dimensional intensity modulation there will be increased flexibility in generating dose distributions thereby improving the likelihood of reaching the treatment goals.

1.5.1 Complex Fluence Generation

Intensity $I$ is defined as the energy passing through unit area, $a$, per unit time, $t$.

$$ I = \frac{dE_{_{kx}}}{dadt} $$

In the case where the photon energy spectrum remains constant throughout intensity modification the process can more simply be described as fluence modulation. The Fluence $\Phi$ is the number of photons, $N$, passing through unit area.
Henceforth, the terminology intensity modulation and fluence modulation refers to the same basic process.

Generation of arbitrary two-dimensional fluence distributions is performed using the attenuation properties of photons as they pass through matter. The one-dimensional wedge gradient example described at the beginning of this section is created using an attenuator with varying thickness in one dimension. Extending to two dimensions, a custom-machined metal block may be constructed to modify the fluence for each field. Unfortunately, the machining process is typically too cumbersome for routine clinical use on large numbers of patients and the fluence and spatial resolution are inadequate in many cases.

The method most commonly used to generate complex fluence maps at fixed gantry angles involves the cumulative sum of multiple uniquely shaped sub-fields as shown in Figure 1.8. Although the fluence resulting from an individual sub-field is relatively constant over its spatial extent, the size and shape of each sub-field is different. At locations where two sub-fields overlap, the fluence will be the sum of each sub-field contribution. Conversely, at locations where only one field is open to the source, the contribution to the total fluence will be from that field only. In this way, by using an adequate number of sub-fields, it is possible to generate arbitrary fluence maps.

\[
\Phi = \frac{dN}{da}
\] (1.8)
1.5.1.1 Multiple MLC Fields

Due to the ease with which field apertures may be modified with the MLC it is the most common method for defining the sub-fields used in IMRT [18]. Each aperture is defined by the MLC leaf positions. Because the leaves are computer controlled, the entire sequence of sub-fields may be loaded into the MLC controller at once. After
the delivery of each sub-field the leaves are immediately moved into position for the next field, making the total delivery time short enough for clinical use. Radiation may also be delivered with the leaves moving continuously throughout delivery [19, 20]. In this case the fluence is divided into a large number of continuously varying sub-fields (>100) which provide superior overall fluence resolution. The fluence map is delivered by controlling the speed of each leaf and/or the dose rate of the linac. A detailed description of MLC based IMRT delivery including the limitations of current leaf positioning methods is presented in Chapter 3. In this thesis a new method of controlling the multileaf collimator is described that improves on conventional MLC based IMRT methods.

1.5.2 Plan Optimization

Although IMRT increases the flexibility of generating dose distributions it complicates the treatment planning process. The number of combinations and permutations involved when each fluence ‘pixel’ of each field can be modified is too large to be evaluated by a human observer. For this reason with the development of fluence modulation came the introduction of treatment planning optimization methods [21-23]. The purpose of the optimization is to calculate the optimal fluence map for each field such that the physician’s prescription is satisfied. In general, the fluence map for each field is derived with the intention of providing a uniform dose to the target volume while limiting dose to surrounding structures. Although plan optimization is an inherently three-dimensional problem, the fluence map required for each field is two-dimensional. Once a satisfactory plan has been obtained, the problem becomes the generation of that two-dimensional fluence map. Derivation of the optimal fluence maps and calculation of the apertures needed to generate those fluence maps are therefore two separate processes. During plan optimization it is assumed that the linac will be capable of delivering the desired fluence. In this way, the problem of deriving the optimal plan, and therefore the optimal fluences, is de-
coupled from the problem of generating the appropriate apertures needed to generate those fluences. This thesis is focused entirely on the problem of aperture generation and the concept of plan optimization is introduced in this section for completeness only.

1.5.3 Leaf Sequencing

Once the desired fluence maps have been calculated the MLC apertures used to generate those fluence maps must be derived. Numerous leaf-sequencing techniques have been proposed in the literature [24-26]. In general, the fluence profile between each leaf and its opposing pair is considered separately. By modifying the position and size of the gap between the leaves for each sub-field it is possible to produce virtually any fluence profile. A more detailed description of conventional leaf sequencing methods is described in section 4.1.

1.5.4 Delivery

The leaf positions for each sub-field are transferred from the planning system to the MLC control computer prior to treatment. Once the patient is in position, the gantry angle, collimator jaw positions and MLC leaf positions are set for the first sub-field segment. When the beam is activated at the control console the MLC controller computer moves the leaves to each required position. Once the sequence has finished the gantry is moved to the next field. The process is repeated until all fields have been delivered.

1.6 Factors Affecting Spatial Resolution in Dose Delivery

An ideal dose distribution is one where the cancerous tissue receives 100% of the prescribed dose and the surrounding tissue receives nothing. Currently, the most important factor affecting our ability to achieve this goal is the limited information
available to the oncologist when determining the location of cancerous tissue [27-29]. Advances in positron emission tomography (PET) and functional MRI show promise in providing metabolic information not currently available with standard imaging techniques [30, 31]. The radiation oncologist estimates the planning target volume (PTV) from the available diagnostic information. It is then the job of the physicist and/or dosimetrist to plan the patient treatment. In practice there are several specific factors that limit our ability to deliver an ideal dose distribution to the PTV.

1.6.1 Imaging Resolution

Although not the most important source of spatial resolution degradation, the resolution of diagnostic images places an upper bound on how precisely the PTV can be defined. Pixel resolution in the transverse imaging plane (cut through the body horizontally) is typically on the order of 0.5 mm. In the superior-inferior dimension the spacing ranges from roughly 1mm to 1cm.

1.6.2 Patient Immobilization

Although there is significant emphasis placed on reproducing the patient set-up for treatment, there is no guarantee that the patient will be in precisely the same position as when diagnostic images were obtained. Because many organs in the body are mobile (e.g. lungs and heart) and radiation is not delivered instantaneously, there will always be some additional error during each treatment due to patient/organ motion. Finally, the target volume itself will change over time. Although the degree of variation is different in each case there will inevitably be some error due to a mismatch between the treatment fields and the true PTV at that instant.

The magnitude of these effects have been studied by others [32] and is highly site dependent [32, 33]. For example, in the head and neck region an accuracy of 1 to
2 mm is achievable [34] while in the lung region errors as high as 3 cm have been noted [35].

1.6.3 Delivery Technique

In addition to the spatial resolution degrading factors described in the previous sub-sections the specific technique or mechanism used in delivering dose to the PTV is important. The tools available for generating an ideal dose distribution suffer from certain limitations. The photon beam has certain physical characteristics, photon scatter and electron transport in particular, that make it impossible to conform perfectly to the PTV [36]. Also, mechanical properties of the field-shaping device (e.g. MLC) provide additional limits on spatial resolution and dosimetric accuracy [37].

As part of this thesis a rigorous evaluation of the spatial resolution limiting characteristics of conventional delivery techniques is presented. By using linear systems analysis it is shown how different delivery parameters affect our ability to deliver an ideal dose distribution. From the results of this investigation a new way of delivering IMRT is proposed that has potentially higher spatial resolution than existing techniques. Finally, the bulk of the thesis is devoted to the development and testing of this novel dose delivery method.

1.7 Thesis Objectives and Summary

Previous investigators have shown that by improving the spatial resolution of fluence delivery dose distributions may be generated that conform more closely to the tumour volume and reduce dose to the surrounding healthy tissue [38, 39]. The primary objective of this thesis is the development of a novel method for the delivery of IMRT that has several advantages over current techniques including higher spatial resolution fluence modulation.
1.7.1 Spatial Resolution Degradation

It was recognized in the preliminary stages of this thesis that there was a need for a more rigorous method for investigating the spatial resolution capabilities of dose delivery techniques. For this reason a new analysis method was developed by expanding on a study by Bortfeld et al. [37] applying linear systems analysis for quantifying dosimetric spatial frequency degradation (Chapter 2). Modeling dose delivery as a linear system has two main advantages. Firstly, the spatial resolution capabilities of various components in the dose delivery chain can be assessed separately. This provides insight into which component provides the best or worse spatial resolution. Secondly, with the aid of spatial frequency analysis, it is now possible to evaluate spatial resolution changes independent of the volume that is being treated. Typically, new delivery technology is assessed using relatively arbitrary treatment scenarios (e.g. a specific treatment site and beam configuration). By quantifying spatial resolution properties of the delivery technique independent of the treatment volume an unbiased evaluation may be achieved.

1.7.2 IMRT Delivery With MLC Rotation

Through observations obtained from the linear systems analysis described in Chapter 2 a new method of IMRT delivery is proposed. The method consists of rotating the entire MLC between each IMRT sub-field. These IMRT fields may be delivered statically (with the collimator rotating to a new position in between sub-fields) or dynamically (with the collimator rotating and leaves moving simultaneously during irradiation). With this fundamental increase in flexibility advantages in spatial resolution are theoretically obtainable. In addition, there are potential advantages in reduced systematic interleaf errors and generating larger maximum field sizes. The majority of Chapter 3 is devoted to investigating the mechanical and radiation producing characteristics of standard linacs under collimator rotation conditions.
Specific experiments are used to evaluate each component in the delivery chain that may affect dosimetric accuracy. Through this investigation any dosimetric error in rotational IMRT delivery due to basic limitations of current linac hardware are determined.

1.7.3 Leaf Motion Derivation

Although there are several theoretical advantages to delivering IMRT with MLC rotation, the realization of those advantages is not trivial. Deriving the appropriate rotated MLC apertures for arbitrary fluence maps is significantly more complex than with conventional fluence generation methods. The most challenging aspect of this thesis was the development of a series of algorithms capable of deriving the leaf positions for each rotated MLC aperture. Chapter 4 is devoted to a full description of the rotational leaf motion algorithms. Included is a description of conventional leaf positioning methods as well as an analytic model of the rotational leaf motion derivation problem.

1.7.4 Rotating MLC Evaluation

In Chapter 5 a series of experiments that evaluate the rotational technique in both its static and dynamic modes is presented. First, the rotational leaf motion algorithms are characterized in terms of their ability to derive leaf motions for a variety of clinical fluence maps. In particular, the dependence of the algorithms on various fixed delivery parameters is investigated. Next, the dosimetric accuracy of the rotational technique is evaluated using a series of varying complexity fluence maps. Finally, a series of experiments are presented that focus on the advantages of using MLC rotation in IMRT delivery. The experiments described in each section are repeated for two separate linacs having different MLC models. For comparison, fluence maps are generated using both rotational and conventional delivery techniques.
Chapter 2

SPATIAL RESOLUTION DEGRADATION

Flexibility and complexity in patient treatment due to advances in radiotherapy techniques necessitate a simple method for evaluating spatial resolution capabilities of the dose delivery device. The purpose of the following investigation is to evaluate a model that describes the ability of a linear accelerator to deliver a desired dose distribution. The model, developed as part of this thesis and published in the journal Medical Physics [40], is based on linear systems theory and is analogous to methods used to describe resolution degradation in imaging systems. A qualitative analysis of spatial resolution degradation using the model is presented in the spatial and spatial frequency domains. The ability of the model to predict the effects of geometric dose conformity to treatment volumes is evaluated by varying multileaf collimator leaf width and magnitude of dose dispersion. Dose distributions for three clinical treatment shapes, circular shapes of varying diameter and one intensity modulated fluence map are used in the evaluation. It is shown that the model accurately predicts the dependence of dose conformity on these parameters. The spatial resolution capabilities of different radiation therapy devices can be quantified, providing a simple method for comparing different treatment machine characteristics. Also, because different treatment sites have different resolution requirements, the model may be used to tailor machine characteristics to each specific site [40-43].

Validation of the model consists of two parts. The first part is a qualitative investigation into each component focusing on effects to spatial resolution in dose delivery. Included as examples are a range of typical clinical shapes varying in size
and complexity. The second part consists of a quantitative analysis of the dose conformity to the same clinical treatment shapes indicated above. Results are discussed in the context of the qualitative investigation and are used to evaluate the relevance of the model.

Through observation of various aspects of the model, specifically the benefits of collimator rotation, a method of improving the spatial resolution of IMRT distributions is proposed. This chapter provides a rigorous theoretical investigation of the limits of spatial resolution in dose delivery. It lays a foundation for the development of a new IMRT delivery technique described throughout the remainder of the thesis.

2.1 Linear Systems Theory

When characterizing the response of a given system to some form of stimulus it is advantageous if the entire system may be modeled as linear. For a system to be linear it is required that each component that makes up the system respond in a linear fashion [44]. Expressed mathematically,

$$H[k_1f_1(x, y) + k_2f_2(x, y)] = k_1H[f_1(x, y)] + k_2H[f_2(x, y)].$$

(2.1)

Where $H$ is an operator describing any component of the system, $k_1$ and $k_2$ are constants and, $f_1(x, y)$ and $f_2(x, y)$ are any two-dimensional functions describing a stimulus to the system. A system that adheres to equation 2.1 satisfies both the property of additivity

$$H[f_1(x, y) + f_2(x, y)] = H[f_1(x, y)] + H[f_2(x, y)]$$

(2.2)

as well as homogeneity

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2.1.1 Spatial Invariance

For two-dimensional systems it is advantageous if each component adheres to an additional constraint. If the location of the stimulus affects the system output then evaluating the system for arbitrary stimulus will be significantly more complex. The system must therefore also have the property of spatial invariance. Expressed mathematically, for a system output $g(x, y)$, it is required that

$$H[k_1f_1(x, y)] = k_1H[f_1(x, y)].$$

(2.3)

2.1.2 Application To Imaging Systems

Many imaging systems can be considered linear and may be approximated as spatially invariant for the purposes of evaluation. Some examples include: photography, diagnostic x-ray imaging and computed tomography [7, 45-48]. In imaging applications, the ability of the imaging device to detect varying resolution objects defines its utility. By modeling the device as a linear system it can be compartmentalized into a chain of separate resolution degrading components. Each component has a linear effect on the previous link in the chain. The advantage of this formulation is two fold.

Firstly, it allows for the characterization of the imaging system by a spatial resolution degradation function. This function is not dependent on the object being imaged. Removing this dependence allows for an evaluation of the imaging system without relying on arbitrary test images. Different imaging systems may then be...
compared in a robust fashion, with the linear degradation function of each system defining its performance.

Secondly, because each component of the system can be separated, it is possible to evaluate each component individually. Different components will have different effects on spatial resolution. By evaluating the mechanisms of spatial resolution reduction from each component, it may be possible to determine methods of improving the system as a whole. The "weakest link" in the chain may also be determined in this fashion, allowing investigators to focus on components that will result in the greatest improvement.

2.1.3 Fourier Model Of Spatial Resolution Degradation (Spatial Frequency)

A large body of work has been devoted to the study of imaging systems in both the spatial and spatial frequency domains. The Fourier transform is used to convert spatial information to the frequency domain. For a 2-dimensional function \( f(x,y) \) the corresponding function in the Fourier (or frequency) domain, \( F(\mu, \nu) \), is given by

\[
F(\mu, \nu) = FT[f(x,y)] = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} f(x,y) e^{-i2\pi(\mu x + \nu y)} dx dy
\]  

(2.5)

and the inverse Fourier transform is

\[
f(x,y) = FT^{-1}[F(\mu, \nu)] = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} F(\mu, \nu) e^{i2\pi(\mu x + \nu y)} d\mu d\nu.
\]  

(2.6)

Analysis of a function in the frequency domain provides direct information about the magnitude of lower and higher frequency components. Higher frequency
components are formed by high resolution aspects of the spatial function [44, 49]. The ability of an imaging system to preserve the high spatial frequency characteristics of an object relates directly to the spatial resolution capabilities of the system. The Fourier transform of a function will contain real and imaginary parts. When converting a spatial function to its Fourier representation, absolute position information is contained in the phase component of the transformed function. The Fourier function may be written:

\[ F(\mu, \nu) = |F(\mu, \nu)|e^{i\phi(\mu, \nu)}. \]  

(2.7)

The phase angle, \( \phi(\mu, \nu) \), is equal to

\[ \phi(\mu, \nu) = \tan^{-1}\left( \frac{I(\mu, \nu)}{R(\mu, \nu)} \right) \]  

(2.8)

where \( R \) and \( I \) are the real and imaginary parts respectively. When describing spatial resolution characteristics of a spatially invariant system it is only necessary to evaluate the frequency spectrum, \( |F(\mu, \nu)| \).

\[ |F(\mu, \nu)| = \left[ R^2(\mu, \nu) + I^2(\mu, \nu) \right]^{1/2} \]  

(2.9)

The spatial frequency spectrum \( |F(\mu, \nu)| \) contains only real components and is used for all spatial frequency domain plots displayed in this chapter.

2.1.3.1 Modulation Transfer Function Concept

A method of quantifying spatial resolution degradation is to supply an input to the system that contains all spatial frequencies and measure the resulting output spatial frequency spectrum. If all spatial frequencies are present in the input, the amplitude
of spatial frequencies in the output will represent the ability of the system to transfer any spatial frequency.

A function that contains all spatial frequencies is the Dirac delta function, as shown in Figure 2.1(a). The Fourier transform of the delta function is a function equal to unity throughout the frequency domain (i.e. it contains an equal contribution of all spatial frequencies) as shown in Figure 2.1(b). When the delta function is passed through the system it becomes a function often referred to as the point spread function, which has different spatial and spatial frequency characteristics (see Figure 2.1(c)). By obtaining the Fourier transform of the point spread function, it is then possible to evaluate how well the spatial frequencies have been preserved by the system. The modulation transfer function (MTF) is defined as the absolute value of the output spatial frequency spectrum shown in Figure 2.1(d) scaled to its value at the zero frequency position \([7, 49]\). It is therefore written as:

\[
MTF(\mu, \nu) = \frac{|T(\mu, \nu)|}{T(0,0)}.
\]  

The MTF describes the spatial frequency transfer capabilities of a system independently of the object being imaged. Also, by applying the MTF to a given input to the system it will be known \textit{a priori} what to expect in the output. In this thesis, the same theory will be applied in the context of delivering dose distributions. The input will be the ideal prescribed dose distribution and the output will be what can be attained with the given dose delivery device and technique. An analogue to the MTF used in imaging systems is developed. It describes mathematically the ability of the dose delivery system to generate the desired dose distribution, just as the MTF describes the spatial resolution abilities of an imaging system.

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Figure 2.1: Illustration of modulation transfer function concept. The Dirac delta function in (a) is input to the imaging system. Its Fourier transform (denoted by FT) is a constant throughout the frequency domain and is shown in (b). The system output $t(x,y)$ is shown in (c). In the frequency domain, the Fourier transform of the $t(x,y)$ is displayed (d). All spatial frequencies are input to the system but some higher frequency information has been reduced in the output. The amount of spatial frequency reduction represents the spatial resolution capabilities of the system.

2.1.4 Sampling Theory

In imaging there are many systems that produce an output of finitely spaced pixels. The image formed is not a continuous function but is sampled at discrete intervals. The actual function and its sampled representation have different properties. Spatial resolution of the system will be affected by the sampling frequency at the image [7, 49]. Mathematically, the sampling process can be modeled by the continuous function of the input multiplied by a series of shifted delta functions, also known as a
comb function (III), as shown in Figure 2.2(b) and (c). A multiplication operation in the spatial domain corresponds to a convolution operation in the frequency domain. The Fourier transform of a comb function is also a comb function in the frequency domain. Therefore, the frequency representation of the continuous function is convolved by a sequence of delta functions. The result is a series of adjacent repeated copies of the original frequency spectrum as shown in Figure 2.2 (c).

2.1.4.1 Aliasing

When the sampling interval is increased, the spacing of delta functions in the frequency domain decreases. At some point, adjacent frequency spectra will overlap. When they overlap the frequency components are summed, which degrades the true frequency spectrum of the continuous function as shown in Figure 2.2. This effect is known as aliasing [49]. It becomes more severe for higher resolution functions and as the sampling interval increases.

2.1.4.2 Nyquist Criterion

If a function has a finite spatial frequency extent (band-limited) with maximum spatial frequency given by $\mu_{\text{max}}$, then it is possible to fully describe that function through sampling if the sampling frequency is equal to or greater than $2\mu_{\text{max}}$. Commonly referred to as the Nyquist criterion [44], it is derived from the duplication effect of frequency spectra due to sampling shown in Figure 2.2. At the Nyquist frequency, adjacent spectra will be as close as possible without overlapping, preserving all spatial frequencies throughout the sampling process. By fully reproducing the spatial frequency spectrum, the spatial function will also be preserved in its entirety, causing no degradation of spatial resolution as a result of sampling.
Figure 2.2: The function shown in (a) is sampled at a frequency $1/\Delta x$ through multiplication by the comb function (b). The result is a discrete representation of the original function shown in (c). A multiplication in the spatial domain corresponds to a convolution in the spatial frequency domain. The frequency spectrum is therefore convolved by the Fourier transformed comb function, producing multiple shifted copies of the original spectrum. When the sampling frequency is too low, adjacent frequency spectra overlap, causing a degradation of the original frequency spectrum known as aliasing.

In this chapter sampling effects are investigated in the context of generating dose distributions. The theory introduced here is applied to radiation delivery, which allows a better understanding of the limitations of current equipment and techniques.
Furthermore, by modifying sampling characteristics of the system it may be possible to improve spatial resolution.

2.1.5 Application to Dose Delivery Systems

LST is a well established mathematical tool used to describe the spatial resolution capabilities of imaging systems; [50, 51] the benefit being that it permits the quantification of resolution degradation at each stage in the imaging process independently of the object being imaged. Recently, there have been several investigations into quantifying the capabilities of dose delivery systems using similar models, [52, 53] some of which have employed methods that can be used when assuming a linear, spatially invariant system. [37, 54] In particular, Bortfeld et al. have described an investigation into the optimal leaf width of a multileaf collimator [37]. Using sampling theory they conclude that there is no benefit in decreasing the leaf width below a specific value given the range of photon scatter and electron transport in tissue. This implies an optimal leaf width on the order of 1.5 to 2 mm. In practice, however, MLC leaf widths typically vary from 3 to 10 mm. In the following investigation a complete model of resolution degradation is presented. By understanding the processes involved in reducing spatial resolution it will be possible to evaluate the limits imposed by current devices and techniques. Also, and perhaps more importantly, this investigation provides insight into a method to circumvent these limits and improve spatial resolution of the system.

2.2 Theory: Dose Transfer Function of Spatial Resolution Degradation

A three-dimensional dose distribution is formed by the superposition of multiple beams where the shape and the intensity of each beam is defined by a stationary or dynamically moving two-dimensional aperture. For conformal radiotherapy, the
optimal dose distribution is one where there is 100% of the desired dose encompassing the greatest extent of the PTV and 0% everywhere else. The ideal dose distribution for any given beam is defined as a two-dimensional function perpendicular to beam central axis \( D_{\text{ideal}}(x, y) \). This is equivalent to a binary mask with value 1 inside the PTV and 0 outside. For IMRT the ideal fluence distribution is defined by intensity modulation and can have dose levels ranging from 0% to 100%.

The dose delivery system has inherent spatial resolution limitations that inevitably cause a degradation of the ideal distribution, which may or may not be clinically relevant. These limitations can be separated into different components. Effects due to a finite source size, the MLC leaf dimensions and scattering and electron transport in the patient are considered here.

### 2.2.1 MLC Sampling

The MLC reduces the spatial resolution by forcing a sampling at width \( w \) equivalent to the spacing of MLC leaves in one dimension and a sampling at the incremental position of the leaves in the direction of motion in the orthogonal dimension. The sampling interval in the direction of leaf motion is much less than the interleaf sampling interval which allows the overall sampling to be described by the one-dimensional “comb” function \( \Pi(y/w) \), where \( w \) is the distance between leaf centers. The sampled dose function is then convolved with the function describing the two-dimensional transmission of an MLC leaf \( MLC(x, y) \), further modifying the dose distribution.

### 2.2.2 MLC Function

In this thesis a simplified model of the MLC is used. The MLC leaf is considered to be a one-dimensional square “rect” function [37] (see Figure 2.3(a) on page 40). The
actual MLC leaf function (i.e. to include tongue and groove effects and rounded leaf ends) may be substituted in future work.

2.2.3 Dose Spread Kernel

Mohan et al. have shown that an infinitely thin pencil beam of high energy photons (effectively an impulse function) will generate a point spread function of energy deposition (dose) when interacting with matter [55]. This impulse response to photon fluence allows calculation of dose using a point spread function, which is referred to as the dose spread kernel $DSK(x, y)$. The previous result of spatial resolution degradation due to the MLC is therefore convolved with the DSK, providing a further degradation in spatial resolution. The DSK is approximated by a two dimensional Gaussian.

$$DSK(x, y) = \frac{1}{2\pi\sigma^2} \exp\left(-\frac{x^2 + y^2}{2\sigma^2}\right) \tag{2.11}$$

This approximation has been shown to be an adequate model for dose calculation [56] and combines the effects of dose deposition at a given depth and finite source size into one kernel. Experimentally derived DSKs could be used to model a specific linac and beam energy in future work.

2.2.4 Dose Transfer Function

Combining all of these terms, the resulting dose distribution $D(x, y)$ is calculated from:

$$D(x, y) = \left[D_{\text{ideal}}(x, y)\Pi\left(\frac{y}{w}\right)\right] \otimes MLC(x, y) \otimes DSK(x, y) \tag{2.12}$$
In order to approximate a spatially invariant system, it is assumed that the fluence and energy spectrum of the open beam is the same everywhere perpendicular to beam central axis. Linacs are designed with flattening filters that minimize variations in photon fluence across the beam. Small perturbations due to non-uniformity of the radiation beam do not affect the ability of the model to evaluate spatial resolution capabilities of the dose delivery device. These approximations facilitate an understanding of the underlying processes in applying LST using Fourier analysis. Obtaining the Fourier transform of Equation 2.12, we define the "Dose Transfer Function" (DTF) where:

\[ D(\mu, \nu) = [D_{\text{ideal}}(\mu, \nu) \otimes \text{III}(\nu w)] MLC(\mu, \nu) DSK(\mu, \nu) \]  

The arguments \((\mu, \nu)\) signify the function is the Fourier transform of its spatial domain function from Equation 2.12.

2.2.5 Spatial Frequency Representation Of Dose

Figure 2.3 illustrates component functions of the DTF for a one-dimensional system and provides a useful example to aid the conceptual step to a two-dimensional system. Dose degradation functions for a 1 cm wide square MLC leaf and a Gaussian DSK with \(\sigma = 2\) mm are displayed in the spatial and spatial frequency domains in Figure 2.3(a) and (b) respectively. Figure 2.3(c) shows a desired simplified dose profile resembling a saw tooth function. MLC leaf motion is perpendicular to the y-axis. After sampling and convolution by the leaf function and DSK, it is apparent that this system is unable to match the desired dose profile. The sampling is too coarse and, as a result, misses the first valley and the third peak almost entirely. Also, limitations imposed by the leaf width and DSK cause a further blurring, degrading even the more adequately sampled portions and extending dose beyond the edge of
Figure 2.3: Degradation of a desired dose distribution due to the MLC and DSK. An MLC sampling function for a 10 mm MLC leaf and a Gaussian Dose Spread Kernel with $\sigma=2$ mm are shown in the spatial domain (a) and the frequency domain (b). The ability of this MLC and DSK to deliver a "sawtooth" dose distribution is shown in the spatial domain (c) and the frequency domain (d). Greater detail in the spatial frequency spectrum degradation can be seen in (e).
the desired profile. The frequency spectrum of the same process is shown in Figure 2.3(d) with an expanded scale in Figure 2.3(e).

Aliasing occurs when the intensity profile is sampled at less than the optimal frequency required to completely describe it. Due to the finite sampling interval, the frequency spectrum of the profile is repeated in the Fourier domain at integer multiples of the sampling frequency (Figure 2.3(d)). Therefore, if the sampling frequency is less than twice the maximum spatial frequency of the object (Nyquist criterion), adjacent spectra will overlap and degrade the optimal profile. Aliasing of the desired dose profile due to insufficient sampling is shown in Figure 2.3(e). Multiplication by the leaf function dampens the higher spatial frequencies, removing most of the aliased peak. Finally, multiplication by the DSK acts as a low pass filter, preferentially dampening higher frequency components.

2.3 Method: Application to 2-dimensional Dose Distributions

Equations 2.12 and 2.13 are separated into a series of two-dimensional convolution and multiplication operations that are applied to an ideal distribution. The degradation in dose conformity resulting from each operation can then be qualitatively interpreted in the spatial and spatial frequency domains using two-dimensional intensity maps. For visualization of the higher frequencies the logarithm of the frequency spectrum \( F(\mu, \nu)_{\text{intensity}} \) is assigned to each intensity value.

\[
F(\mu, \nu)_{\text{intensity}} = A \log[1 + F(\mu, \nu)]
\]  

(2.14)
Figure 2.4: Illustration of MLC sampling in the degradation of an ideal dose distribution in the spatial and frequency domains using the Linear Systems model. $\times$ denotes a multiplication and $\otimes$ denotes a convolution. In the spatial domain, the ideal dose distribution (a) is multiplied by (b), sampling due to the MLC leaf spacing. The Fourier spectrum of (a) is convolved by the Fourier transformed sampling function of (b). The sampled distribution is shown in (c).
A is a normalization factor that exploits the full range of grayscale values and 1 is added to \( F(\mu, \nu) \) so that \( F(\mu, \nu)_{\text{Intensity}} \) does not go below 0. Figure 2.4, Figure 2.5 and Figure 2.6 show the effect of applying each term of Equations 2.12 and 2.13 to a head and neck PTV shape in the spatial domain and in the frequency domain respectively.

Observing modifications to the spatial frequency spectrum allows a direct visualization of the degradation in spatial resolution resulting from each term of the DTF. The first step is a multiplication of the ideal dose \( D_{\text{ideal}}(x, y) \) shown in Figure 2.4(a) by the one-dimensional comb function, Figure 2.4(b). In the frequency domain this corresponds to a convolution by a series of delta functions positioned along the \( y \) frequency axis. The result of these operations is shown in Figure 2.4(c). It is apparent that at this leaf spacing there is significant aliasing causing an addition of erroneous high frequency components to the spectrum.

The next step is to convolve (or multiply in the frequency domain) the result by the MLC leaf function shown in Figure 2.5(b). This causes a further modification of the frequency spectrum and dampens the higher frequency components from the previous step preferentially in the \( y \) direction (Figure 2.5(b)). The MLC shaped distribution is shown in Figure 2.5(c). Aliasing effects are apparent from the increase in higher spatial frequencies when the result is compared to the initial spectrum of Figure 2.4(a).

The final step is to convolve (multiply in the frequency domain) this result by the DSK (Figure 2.6(b)). This produces a blurring of the MLC leaf fluence as shown in Figure 2.6(c), which is similar to a low pass filter, corresponding to a further symmetric dampening of higher frequency components.
Figure 2.5: The sampled dose distribution (a) from the result of Figure 2.4 is convolved by the MLC leaf function (b). In the Frequency domain the sampling spectrum is multiplied by the Fourier transformed leaf function of (b). The resulting MLC shape is shown in (c). It is apparent that aliasing has occurred from the increased high frequency component that is not present in the initial spectrum of Figure 2.4(a)
In conformal radiation therapy, the goal of the MLC is to block all healthy tissue around the greatest extent of the projection of the PTV. The beams-eye-view is used to derive the leaf positions that provide the maximum conformity. The degree of conformity is related to the shape of the PTV and the characteristics of the MLC. Conformity error consists of two elements, irradiation of healthy tissue and underdosing of the treatment volume.

For the purposes of evaluating the model, geometric conformity is defined as:

$$\text{%conformity} = 100 \times \frac{\text{Area of PTV} - (\text{Area of PTV underdose} + \text{Area of healthy tissue dose})}{\text{Area of PTV}}$$

(2.15)

The area boundaries are defined to be the 50% isodose line. MLC leaf positions are chosen so that the center of the leaf edge intersects the edge of the PTV. This method is used because it does not bias the results toward under-dosing of the PTV or toward greater healthy tissue dose. Also, the definition of conformity in Equation 2.15 encompasses both effects and is therefore not sensitive to the leaf positioning method.

### 2.3.1 Study of Circular PTV Shaping

Before embarking on a study using clinical PTV shapes, a baseline was needed to interpret results effectively. For that purpose a study was undertaken to evaluate the conformity of circular PTV shapes of varying radius. A circle is an ideal shape because it is independent of collimator rotation and has a wide range of spatial resolution requirements [57]. For example, the central MLC leaf edges are almost parallel to the circle edge and provide a very close fit (see Figure 2.7). The conformity degrades for leaves farther away from the center until eventually the leaf edges and circle edges are perpendicular. The final advantage of using a circle is that it is possible to find an analytic solution to the conformity as a function of the circle radius $r$ and the leaf width $w$ [57].
Figure 2.6: The resulting MLC shape (a) is convolved by the dose spread kernel (b) to give the deliverable dose distribution (c). In the frequency domain the MLC shape spectrum is multiplied by the Fourier transform of the dose spread kernel in (b). The DSK acts as a symmetric low pass filter, preferentially reducing high frequency components as shown in (c).
2.3.2 Study Of PTV Shapes

A PTV that has a shape with many protrusions of small diameter will have more high frequency component than that of a uniform shape of similar dimensions [58]. It is therefore important to also evaluate the resolution degradation of varying shaped PTVs. For a qualitative investigation, typical clinical shapes were chosen for their diversity in shape, size and planning complexity [59]. These shapes were: a prostate, a head and neck lesion (nasopharyngeal carcinoma), and a central nervous system (CNS) lesion (an arteriovenous malformation). Table 2.1 lists their largest diameter and average spatial frequency at one tenth of the zero frequency peak. The prostate, head and neck, and CNS shapes show an increasingly larger component of high spatial frequencies. Therefore, for conciseness, these shapes will henceforth be referred to as the Low Frequency Shape (LFS), Moderate Frequency Shape (MFS) and High Frequency Shape (HFS) respectively. Two-dimensional dose distributions were calculated using Equation 2.12 with leaf width equal to 5 mm. A DSK with $\sigma = 2$ mm (comparable to a 6 MV beam at moderate depths of approx. 10 cm) was used in all cases [37]. It should be noted that the DSK is a function of depth. The DSKs from a commercial treatment planning system (CadPlan, Varian, Palo Alto, CA) at depths ranging from 5 to 20 cm for a 6 MV photon beam have been evaluated. The $\sigma$ used when fitting Gaussians to these DSKs were found to increase slightly with depth.

The DTF is also applicable to IMRT. The intensity levels required over the two-dimensional shape complicate resolution requirements. In order to provide a comparison with the resolution degradation of conformal treatments, an example of an IMRT prostate distribution was also calculated using a 10 mm MLC leaf width and $\sigma = 2$ mm DSK.

Finally, conformity was calculated as a function of MLC leaf width and DSK size for the 3 shapes listed in Table 2.1. Due to the irregularity of the shapes, results are
dependent on the collimator rotation angle. To remove this dependence, only the optimal collimator angles were used. The optimal angles were derived by calculating the conformity (equation 2.15) at 5 degree collimator angle increments and selecting the one with the greatest conformity.

Table 2.1: A summary of the PTV shapes used in the qualitative and quantitative investigations of the model. Each shape is referred to using the acronym describing its spatial frequency characteristics.

<table>
<thead>
<tr>
<th>Site</th>
<th>Largest Diameter (cm)</th>
<th>Mean Spatial Frequency at 10% of Zero Frequency Peak (cm⁻¹)</th>
<th>Descriptor</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prostate</td>
<td>8.3</td>
<td>0.14</td>
<td>Low Frequency Shape (LFS)</td>
</tr>
<tr>
<td>Head and Neck</td>
<td>6.3</td>
<td>0.38</td>
<td>Medium Frequency Shape (MFS)</td>
</tr>
<tr>
<td>CNS</td>
<td>3.0</td>
<td>0.62</td>
<td>High Frequency Shape (HFS)</td>
</tr>
</tbody>
</table>

2.4 Results: Circular PTV Shaping

Derivation of the analytic solution of conformity resulting from MLC shaping of circular PTVs is given in Appendix A. Presented here are specific results of that derivation given in the context of the LST model.

2.4.1 MLC Effects – Analytic Representation

The analytic solution to the conformity of circular PTV shapes involves the sum of a series of area integrals with limits of integration defined by the intersection points of the MLC with the circle, as indicated in Figure 2.7. The simplified equation is:
\[
\% \text{conformity} = 100 \times \frac{\pi r^2 - 4 \sum_{k=0}^{N} (\text{Area } A_k + \text{Area } B_k) + \text{Remainder Area}}{\pi r^2}
\] (2.16)

Figure 2.7: MLC conformity for circular PTVs. Dark shaded areas indicate underdosed regions of the PTV. Light shaded areas indicate where healthy tissue is receiving dose. Poor coverage is observed for the peripheral leaves, i.e. when the leaf edges are oblique to the circle edge. A full derivation of the analytic solution is given in Appendix A.

2.4.2 Dependence On Leaf Width and Circle radius

The derived circular shape conformity function is plotted in Figure 2.8 as a function of leaf width for 2.5 cm and 10 cm diameter circles. The conformity is periodic and, in general, decreases with increasing leaf size. The periodic nature of the curves is primarily due to the peripheral leaves. When the circle diameter is an integer multiple of the leaf width, the leaves furthest from the center coincide exactly with the edge of the circle. At this point the conformity is at a maximum. If the leaf width is increased or decreased the condition is no longer satisfied and the conformity
deteriorates until it becomes a minimum when the leaf width is halfway to being an integer multiple of the circle diameter. Ignoring the periodic characteristics of Equation 2.16, the overall decline in conformity with leaf width can be written as:

\[
\%\text{conformity} = 100 \times \left(1 - \frac{w}{\pi r}\right)
\]  

(2.17)

Figure 2.8: Conformity vs leaf width for circular PTV shapes with and without a σ=2 mm dose spread kernel. The decrease in conformity is a periodic function of leaf width. With the periodicity removed the decline in conformity is a linear function of leaf width, w, with slope equal to \(-\frac{w}{\pi r}\).
Equation 2.17 is a linearly decreasing function with respect to the leaf width $w$ and linearly increasing with respect to the circle radius $r$. It should be noted that the amount of high spatial frequency component contained in a two-dimensional shape is inversely proportional to its size. Therefore, a given resolution-degrading component will have a more severe effect on a smaller target. This property is preserved in equation 2.17. As the circle gets smaller, spatial resolution requirements are increased proportionately, which results in poorer conformity for a given MLC leaf width. The same argument holds for the spatial filtering effects of larger MLC leaves. As the leaves get wider they allow less high frequency component to pass through the system, resulting in poorer conformity for a circle of given radius.

2.4.3 Dose Spread Kernel

Also plotted in Figure 2.8 are results of conformity versus leaf width for the same circular PTV shapes obtained with the complete degradation function (see equation 2.12). The difference in the curves is caused by convolution with the DSK, which is not taken into consideration in the previous analytic result. When the DSK is applied, the area boundaries are defined to be the 50% isodose line. Although the periodic nature of the curves is preserved, the inclusion of the DSK provides a higher degree of conformity at larger leaf widths. This effect seems counter intuitive at first but is explained using the LST formulation. At larger leaf widths the sampling interval increases. Aliasing effects due to finite sampling become more severe with increased leaf spacing. Recall that aliasing introduces erroneous high frequency component to the spatial frequency spectrum (see section 2.1.4.1 and Figure 2.2). The DSK filters out high frequency components and is therefore competing against the effects of aliasing. When the leaf spacing gets large enough, the DSK filters out more incorrect high frequency than true frequency information. Therefore, the DSK will actually improve conformity at larger leaf widths. At smaller leaf widths, the DSK becomes the dominant factor degrading the dose distribution and results in lower conformity.
2.5 Results: Clinical Simulation Of Resolution Degradation

The LFS and HFS ideal dose distributions are displayed with their spatial frequency spectra in Figure 2.9(a) and (c) respectively. The HFS exhibits a greater proportion of high frequency components than the LFS. Also, the HFS frequency spectrum is not symmetric, showing higher spatial frequency along the axis perpendicular to the location of the thinnest part of the PTV in the spatial domain. Figure 2.9(b) and (d) show the LFS and HFS results when using a 5 mm leaf width to generate dose distributions. The LFS result shows a superior level of conformity when compared to the HFS. This is evidenced by the degree of degradation of spatial frequencies in each case. Because the LFS has a smaller high frequency component than the HFS it will be less sensitive to spatial frequency filtering caused by the MLC and DSK. Detailed results for the MFS were displayed in Figure 2.4 to Figure 2.6.
Figure 2.9: Degradation of the ideal dose distribution in the spatial and spatial frequency domain for (a), the LFS and (c) the HFS. Resulting degraded distributions are shown in (b) and (d). A 5 mm leaf width was used for both PTV shapes. Dose distributions were calculated using Equation 2.12. A Gaussian dose spread kernel with a $\sigma = 2 \text{ mm}$ was used in all cases.
The same effect is observed when using a 1 cm leaf width to generate the IMRT dose distribution as shown in Figure 2.10(b). Higher spatial frequencies have been modified, causing an overall blurring and "pixelation" of the ideal distribution. The aliasing phenomenon can be seen at the top and bottom of the resulting frequency intensity map. It is more severe than the conformal cases due to a larger MLC leaf width.

\[ \text{SPATIAL DOMAIN} \quad \text{FREQUENCY DOMAIN} \]

![Spatial and Frequency Domains](image)

Figure 2.10: An ideal IMRT prostate field dose distribution is shown in (a). Degradation of the ideal dose distribution in the spatial and spatial frequency domain for the resulting distribution was calculated using Equation 2.12 and is shown in (b). A 10 mm leaf width MLC and a Gaussian dose spread kernel with \( \sigma = 2 \text{ mm} \) was used in the calculation.

### 2.5.1 Collimator Angle Dependence

The collimator angle that provides optimal conformity was used for all results in the quantitative analysis of clinical PTV shapes. Aliasing effects due to the MLC leaf spacing and multiplication by the leaf function will cause a more severe deterioration
in the frequency spectrum for certain collimator angles. A plot of optimal collimator angle vs. leaf width for the clinical shapes is displayed in Figure 2.11. The LFS is fairly circular and therefore does not exhibit a preferred angle. The MFS has a preferential collimator angle at approximately 90 degrees. This same collimator angle was used in Figure 2.4. The frequency spectrum of the MFS is asymmetric (Figure 2.4(a)). The optimal collimator angle places the shape with the higher frequency components in the direction of leaf motion, which is the direction suffering the least amount of high spatial frequency degradation. The HFS shows a wide range of optimal collimator angles. Over certain ranges of leaf widths there is a preference but, in general, there is no distinct orientation that is obtained consistently.
Figure 2.11: The collimator angle providing optimal conformity is plotted versus leaf width for the LFS, MFS and HFS.

2.5.2 MLC Leaf Width – Aperture Shaping

Results of conformity versus leaf width for the three clinical shapes are shown in Figure 2.12. The LFS conformity is a slowly decreasing function of MLC leaf width. The MFS is slightly offset, shows a steeper slope and becomes more chaotic after approximately 7 mm leaf width. The HFS has a much more significant offset on the conformity axis than the previous 2 examples. Also, the rate of decrease and amount of variation increases more significantly for leaf widths greater than 4 to 5 mm. It is intuitive that it should be more difficult to conform to the smaller, more complex
volumes. It was shown in Figure 2.9 that the spatial frequency spectrum of the HFS contains a greater proportion of higher frequency information than the LFS and MFS. As a result, the degradation in the frequency spectrum for a given MLC leaf width is more severe for the HFS. The offset toward lower conformity at very small leaf sizes for the MFS and HFS is caused by spatial resolution degradation due to the DSK. Degradation in spatial frequency resulting from the MLC leaves is significantly less and the dominant effect becomes the filtering of high spatial frequencies caused by the degradation due to the DSK. Again, this effect is more severe for the HFS than the other two shapes.

![Graph showing conformity vs leaf width for LFS, MFS, and HFS.](image)

*Figure 2.12: Conformity vs leaf width for the LFS, MFS, and HFS. A gaussian dose spread kernel with $\sigma=2$ mm was used in all cases.*
2.5.3 Dose Spread Kernel – Blurring

The dependence of conformity on the DSK for varying leaf width is shown for the HFS in Figure 2.13. As expected, the conformity with no DSK approaches 100% for small leaf widths. Increasing the width of the DSK increases the offset of the curves on the conformity axis. For larger leaf widths the same effect observed in the circular PTV shapes in section 2.4.3 is observed here. Instead of the DSK consistently reducing the conformity at greater leaf widths there is actually an increase, as shown by the crossover in the range 3 to 5 mm. This can also be explained using the LST formulation in Equation 2.12 and 2.13 and was also observed in the circular PTV shapes. The aliasing effect due to the sampling interval of the MLC leaves causes an addition of incorrect high frequency information that degrades the frequency spectrum. The effect is more severe for greater leaf spacing. The DSK counteracts this effect by acting as a low pass filter. The aliased high frequency information is therefore suppressed, providing better conformity in the final result. The leaf width at which these two effects are in equilibrium is located at the intersection of the curves with and without the DSK. This lies at approximately 3.5 mm and 5 mm for the $\sigma=2$ mm and $\sigma=3$ mm DSKs respectively.
2.6 Discussion: Implications of Resolution Degradation

The application of LST to dose delivery systems provides a model for quantifying inherent limitations in obtaining desired dose distributions. The same theory has been used on imaging systems to quantify spatial resolution degradation. In imaging, the modulation transfer function (MTF) is used to describe the degradation of spatial resolution [44]. The MTF quantifies the transfer of spatial frequencies from the object to the image and is the most commonly used comparator when evaluating different devices. This concept can be borrowed from imaging to describe an

Figure 2.13: Conformity vs leaf width for the HFS shapes using a gaussian dose spread kernels with $\sigma=0$ mm, 2 mm and 3 mm. The $\sigma=2$ mm HFS curve is also plotted in Figure 2.12.
effective MTF [54] which we refer to as the DTF, for a dose delivery system. In a practical case, this would involve the experimental determination of the transfer of spatial frequencies for each component that affects spatial resolution. Again, methods similar to those used in imaging could be employed. It would then be possible to compare various linacs (x-ray source), energies and MLCs (leaf width, tongue and groove, rounded leaf ends) in a simple, standardized way by comparing their DTFs. Finally, by including a depth varying DSK it will be possible to extend the model to the third dimension and allow the evaluation of spatial resolution degradation resulting from a multiple beam geometry.

The principle advantage in modeling dose delivery as a linear system is that it allows each spatially degrading component to be evaluated separately. The Fourier transform of each component function in the system describes the spatial frequency transfer for that component. It is therefore possible to investigate the spatial resolution degradation effects from each component independently of the volume being treated. This provides a significant benefit in that improvements in dose delivery devices no longer have to be evaluated on their ability to treat an arbitrarily chosen treatment volume.

Another benefit of using this technique for evaluation is that the desired characteristics of the linac can be tailored to the treatment site. Frequency analysis of the PTV shapes showed that smaller, more complex structures have a greater high frequency component, making them more sensitive to spatial filtering caused by the DTF. This prediction was validated in the quantitative investigation of conformity where it was shown that there is only a small benefit in using a 5 mm leaf width when treating shapes with lower spatial frequencies while shapes with higher spatial frequencies benefit significantly from the smaller leaf width. Information on the characteristics of the delivery system that are required to provide adequate conformity
can be obtained by comparing the spatial frequency spectrum of the PTV and the DTF defined by the delivery system.

Several authors have proposed methods of improving existing dose delivery systems [60-64]. For example, Bortfeld et al. have used sampling theory to investigate the advantage of shifting the patient half a leaf width during treatment [37]. The advantages of these methods can be evaluated by calculating the effective DTF of the modified system.

The LST model presented here is focused entirely on the dose delivery system and static patient. In a real life scenario, some notable limitations have to be considered. These include inaccuracies due to patient motion and PTV delineation using images that have their own specific resolution. It is possible to incorporate these effects into the model. For example, patient motion causes a blurring of the dose distribution. If the motion is well defined it could be described by a low pass filter, adding another term to the DTF. To account for limited imaging resolution, the imaging MTF could be incorporated directly into the DTF, again by adding an appropriate term.

2.7 Discussion: Application Of The DTF To IMRT

In the previous discussion the benefits of using the LST formulation to describe spatial resolution degradation in dose delivery were presented. The focus has been, for the most part, on aperture shaping of the MLC and scatter and electron transport when photons interact with the patient. In IMRT, complex fluence maps are generated by superimposing multiple MLC defined apertures. The cumulative sum of these apertures is a 2-dimensionally varying fluence. The additivity property (equation 2.2) of linear systems permits an extension of the model beyond simple shaping of conformal fields. The result of applying the LST formulation to IMRT was shown in Figure 2.10.
Revisiting the DTF, there are three basic terms in the spatial domain mathematical expression. Sampling of the ideal dose distribution, convolution by the MLC and convolution by the DSK. The MLC function is given by the design of MLC leaves of the given linac. The DSK is a function of the photon energy and the size of the x-ray source. Both of these functions are therefore fixed parameters for any given linac. The only remaining parameter is the MLC sampling. Although the sampling interval is fixed for any single aperture, rotation of the collimator allows flexibility in choosing the sampling direction for the MLC leaves. The benefits of selecting the optimal collimator angle were observed in the results of collimator angle versus conformity shown in Figure 2.11.

These results indicate that some collimator angles provide better sampling directions than others. Extending this effect to IMRT, it should be possible to generate higher resolution fields by modifying the collimator angle for each sub-field aperture. This observation is the key to improving spatial resolution for intensity modulated fields. The remainder of this thesis is devoted to developing, testing and evaluating a new method of generating intensity modulated fields. The novelty of the method is based on rotating the collimator by an angle increment between each sub-field. Deriving the apertures necessary to generate arbitrary IMRT fields in a rotating geometry is significantly more complex due to the additional degree of freedom. A new method of calculating rotated MLC apertures has been developed as part of this thesis work. The following chapters will describe the apparatus, leaf motion algorithms and experimental evaluation of this new technique.
Chapter 3

IMRT DELIVERY WITH MLC ROTATION

MLC based IMRT techniques are well established but suffer several physical limitations. Dosimetric spatial resolution is limited by the MLC leaf width; interleaf leakage and tongue-and-groove effects degrade dosimetric accuracy and the range of leaf motion limits the maximum deliverable field size.

Collimator rotation is used in standard radiation therapy to improve the conformity of the MLC shape to the target volume. Except for opposed orthogonal fields, collimator rotation has not been exploited in IMRT due to the complexity of deriving the MLC leaf configurations for rotated sub-fields. Here, a new way for MLC based IMRT delivery is proposed which incorporates collimator rotation, providing an extra degree of freedom in deriving leaf sequences for a desired fluence map. Specifically, a series of unique algorithms were developed that are capable of determining rotated MLC segments. These IMRT fields may be delivered statically (with the collimator rotating to a new position in between sub-fields) or dynamically (with the collimator rotating and leaves moving simultaneously during irradiation).

In this chapter a detailed description of potential improvements in spatial resolution, reduction of systematic dosimetric error and improved field size capability with collimator rotation is provided [65, 66]. Linear accelerators were not designed to produce fluence maps with combined MLC motions and collimator rotations. For that reason the mechanical characteristics of the linac that control MLC motion, collimator rotation and radiation production are described as they apply to rotational IMRT delivery. A series of experiments were performed to evaluate the magnitude of
errors that are inherent in rotating the collimator during radiation production. Mechanical properties include accuracy of the center of collimator rotation and collimator angle reproducibility. Radiation production characteristics were also investigated. Low dose sub-fields were investigated for static delivery. In dynamic delivery the uniformity of collimator rotation speed and constancy of the dose rate were evaluated. Finally, the feasibility of delivering IMRT with collimator rotation is discussed in the context of results provided by the mechanical and dosimetric experiments.

3.1 Rotational Delivery Method

A field of non-uniform intensity can be divided into a series of uniform sub-fields each having a different multileaf collimator configuration, as described in section 1.5.1.1. By delivering each one of these constant intensity sub-fields it is possible to build up a field of non-uniform intensity [17]. The technique for intensity modulation developed in this thesis is different from conventional methods in that the entire collimator, including the MLC, is rotated between each sub-field. A full description of conventional IMRT delivery methods is provided in Section 4.1. Displayed in Figure 3.1 is an example of a linear accelerator equipped with an MLC and rotating collimator. The patient is placed on a moveable table used to position the treatment site with respect to the linac isocenter. The linac gantry is rotated so that the beam enters the patient from a desired direction. The collimator rotates about an axis that passes through the isocenter and the MLC leaves move perpendicular to that axis.
Figure 3.1: Simplified diagram of a linear accelerator equipped with an MLC and rotating collimator. The collimator rotates about an axis that passes through the isocenter. The orientation of that axis is always perpendicular to the direction of leaf motion.

Shown in Figure 3.2 is a diagram of the collimator at several distinct rotation angles. The fluence contribution from the MLC aperture at each angle as well as the progressive cumulative fluence distribution from each one of those apertures is also shown. The MLC leaves are used to form a uniquely shaped aperture as shown in Figure 3.2(a). Each leaf is adjacent to its neighbour and may only move linearly in and out of the radiation field. A predetermined quantity of radiation is delivered with the first sub-field configuration before rotating the collimator and changing the leaves to the configuration of Figure 3.2(b). Again, a predetermined quantity of radiation is delivered. This aperture defines a different shape than the previous aperture. Part of the areas overlap as shown in Figure 3.2(c). The resulting fluence distribution in those
Figure 3.2: Multiple MLC apertures at varying collimator angles contribute to the final fluence distribution. Two different MLC apertures are used to generate uniform fluence distributions in (a) and (b). The sum of (a) and (b) are displayed as a surface map in (c). Complex fluence distributions as shown in (d) are formed using several sub-fields each having a different rotation angle and MLC configuration.
areas is the sum of both sub-field contributions. Conversely, where they do not overlap the contribution is only from the sub-field that is open at that point (ignoring leakage and scattered radiation). In this fashion, multiple apertures at varying rotation angles are generated and radiation is delivered. The fluence resulting from each sub-field is added to the previous sub-field contributions. The final distribution of fluence is therefore the cumulative sum of the contributions from each individual sub-field as shown in Figure 3.2(d).

3.2 Enhancements To IMRT With MLC Rotation

The MLC has certain characteristics that limit its ability to conform to a target and reduce dose to healthy tissue. Because each leaf has a finite width, the spatial resolution of fluence maps perpendicular to the direction of leaf travel is limited to that width [37, 57]. The MLC is constructed with a tongue-and-groove shape on the side of each leaf in order to minimize interleaf leakage. However, transmission through the leaves is still non-uniform and cannot be compensated for entirely [67, 68]. Also, the tongue-and-groove creates unwanted under-dosing effects for some intensity modulated fields [69]. MLCs are constructed with enough leaves to cover a given length (e.g. 40 cm). Due to mechanical and physical limitations described in section 3.2.3, the range of leaf motion is restricted to a portion of that length [70]. The maximum intensity modulated field size is therefore limited to a rectangle whose width is given by the range of leaf motion. Several authors have reported variations on standard MLC based techniques that attempt to improve the spatial resolution of fluence maps [61, 71, 72], reduce interleaf leakage and tongue-and-groove effects [52, 73, 74], or increase the maximum deliverable field size [75].

Collimator rotation adds a degree of freedom to the sub-fields used to build up an intensity modulated field. As a result, there are several potential advantages over conventional techniques. For example, the spatial resolution of the desired field is
not limited to the leaf width in the direction perpendicular to leaf travel because the leaves are rotated to a different location with each sub-field. The leaf edges are also in a different location with each rotation, reducing the effects of unwanted interleaf leakage and tongue-and-groove under-dosage. Finally, because the rotational method produces sub-fields at multiple collimator angles, the maximum possible field size is given by the superposition of all fields which removes restrictions imposed by the range of leaf motion.

3.2.1 Spatial Resolution

In Chapter 2 it was determined that by rotating the sampling geometry of the MLC leaves it should be possible to increase the spatial resolution of fluence maps. Another way of conceptualizing this improvement is to look at the minimum size of a fluence "pixel" that can be generated in a sequence of MLC defined sub-fields. Referring to Figure 3.3(a), in conventional delivery each leaf moves linearly in and out of the radiation field. The leaves can have displacements that are less than 1 mm, providing high resolution in the direction of leaf motion [37, 76]. In the orthogonal direction there is no leaf movement. The minimum size of a fluence pixel in conventional delivery is therefore a rectangle with width given by the minimum leaf displacement and length given by the width of the MLC leaf.

In rotational delivery the direction of leaf motion changes at each collimator angle as seen in Figure 3.3(b). The minimum size of intensity pixel is therefore no longer limited to a rectangle but is more closely approximated by a circularly symmetric point consisting of a plateau region in the center with diameter given by the minimum leaf displacement which then gradually decreases radially outwards.
Figure 3.3: Potential improvements in spatial resolution between conventional and rotational delivery methods are shown. (a) In conventional delivery the MLC leaves move linearly in and out of the radiation field providing a rectangular pixel size whose length is limited by the leaf width. (b) With rotation the direction of leaf motion changes, allowing for a smaller fluence pixel that is not limited by the leaf width.

3.2.2 Interleaf Effects

The MLC leaves used in this thesis are composed of a Tungsten Alloy, are 6 cm deep and are designed to block greater than 95% of incident photons [77, 78]. For the MLC leaves to move independently, adjacent leaves are not physically connected. Also, a small gap is provided between adjacent leaves so that they may move smoothly in between sub-field irradiations. The interleaf gap allows more photons to leak through the MLC than at intraleaf locations [79]. The effect is reduced by using
a modified leaf edge design as described in Figure 3.4. Instead of a flat edge a tongue-and-groove shape is machined into the side of each leaf. With this design there is no location that will let photons pass through unhindered because either the tongue or the groove will block incident photons. Still, the thickness of metal that the photons have to pass through is less than the full thickness of the leaf, providing greater photon fluence in interleaf locations. Interleaf leakage causes non-uniformity in the transmitted fluence where the MLC is closed. Conventional IMRT delivery methods are unable to compensate for this effect and can result in overdosing at some interleaf locations.

Figure 3.4: MLC leaves are designed with an interlocking tongue-and-groove shape on the side of each leaf. Although interleaf leakage is reduced significantly with this design there are still more photons transmitted through the interleaf gaps, causing non-uniformity in the transmitted fluence where the MLC is closed. Conventional IMRT delivery methods are unable to compensate for this effect and can result in overdosing at some interleaf locations.
in delivered dose distributions that cannot be compensated for in conventional IMRT delivery [80, 81].

A consequence of introducing the tongue-and-groove shape is that for some intensity modulated fields there are underdosing effects along the leaf edges [82]. The fluence profile generated at an open leaf position is shown in Figure 3.5. Because the tongue or the groove protrudes into the field, the fluence at the leaf edge does not increase abruptly. Instead, there is a step of lower fluence before reaching the maximum. Now, consider the case where the leaf that was initially closed is now open and the adjacent leaf is closed. This situation arises frequently during IMRT delivery due to the different sub-field shapes that are required. The desired fluence for such a situation is a constant fluence profile between both leaves. Unfortunately this is not the case and there is a reduction in fluence under the tongue and the groove as shown in Figure 3.5.

The physical mechanism for this discrepancy results from the exponential attenuation of photons as they pass through matter. It has not been characterized in the literature and will therefore be described here. For incident intensity $I_0$ the transmitted intensity through an MLC leaf of thickness $t_{\text{leaf}}$ is given by

$$I_{\text{leaf}} = I_0 e^{-\lambda t_{\text{leaf}}} \quad (3.1)$$

where $\lambda$ is the linear attenuation coefficient of the leaf material. For simplicity, consider a tongue and groove design where the tongue and the groove are both equal to $\frac{1}{2}$ the total leaf thickness.
Figure 3.5: The tongue-and-groove effect is illustrated by considering the transmitted photon fluence at an open leaf edge. The dotted arrows indicate the direction and amount of photon transmission. The tongue causes an intermediate step in the fluence profile shown in (1). Closing this leaf and opening the adjacent leaf results in the same effect although this time caused by the groove. The sum of both profiles should ideally result in a constant fluence across both leaves. Due to the exponential nature of photon attenuation, the sum of both (1) and (2) results in a fluence reduction error at the tongue-and-groove interface.

\[ t_{\text{tongue}} = t_{\text{groove}} = \frac{t_{\text{leaf}}}{2} \]  \hspace{1cm} (3.2)

The total transmitted intensity through the center of the leaf in the open and then the closed position is

\[ I_0 + I_{\text{leaf}} = I_0 \left(1 + e^{-3t_{\text{leaf}}} \right), \]  \hspace{1cm} (3.3)

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and through the tongue and then the groove it is

\[ I_{\text{tongue}} + I_{\text{groove}} = 2I_0 e^{-\frac{\mu_{\text{tot}}}{2}}. \]  
(3.4)

The magnitude of tongue-and-groove error is a function of the attenuation coefficient and leaf thickness. As the leaf transmission is decreased the effect becomes more pronounced. The transmission for a typical MLC is on the order of 2%, which corresponds to a tongue-and-groove fluence reduction of approximately 70%.

In conventional IMRT delivery the location of leaf edges is fixed. Dosimetric error resulting from interleaf leakage and the tongue-and-groove effect are concentrated along the lines described by these edges. With collimator rotation the location of leaf edges change throughout delivery. Each sub-field is rotated with respect to the previous sub-field as seen in Figure 3.6. With each rotation the leaf edges fall along a different line so that any leakage or tongue-and-groove effects are spread over the 2-dimensional field area. Therefore, the maximum dosimetric error due to these effects will be a fraction of that observed in conventional IMRT delivery.
Figure 3.6: An example of 2 rotated sub-fields is shown. Because of rotation the leaf edges of sub-field 2 are not coincident with those of sub-field 1. With several sub-fields the position of leaf edges is blurred over the entire field area. Interleaf leakage and tongue-and-groove underdosing is therefore reduced with collimator rotation. In conventional delivery the leaf edges are fixed, amplifying systematic underdosing and overdosing error at the leaf edges.

3.2.3 Maximum Deliverable Field Size

The conventional dynamic method of delivering intensity modulated fields uses a 'sliding window' technique. Treatment begins with the MLC leaves forming an aperture at one side of the field. Once the beam is activated each leaf pair starts moving towards the other side of the field. Varying the speed of each leaf throughout delivery generates the desired intensity modulated field.

With this method each leaf pair must travel from one field boundary to the other. The maximum field width that can be generated is therefore limited to the length of the MLC leaf. Otherwise, the far edge of the leaf would pass beyond the edge of secondary jaws. Commercially available models have leaf lengths on the
order of 15 cm projected at isocenter. The maximum field length however is only limited to the length of the leaf bank (number of leaves multiplied by their width) and is typically 40 cm. In order to deliver wider field sizes it is necessary to deliver multiple adjacent IMRT fields [75].

\[ \text{Maximum Field Size} \]

\[ \text{Leaf Length} \]

\[ \text{Length of Leaf Bank} \]

**Conventional**

**Rotational**

*Figure 3.7: The maximum field size for conventional and rotational delivery methods in dynamic mode are shown. The field width is limited to the length of the MLC leaves in conventional delivery. With rotation the maximum field size is larger because (1) the leaves are not required to span the entire field width and (2), because the direction of leaf travel is rotated with each sub-field. The upper limit on rotational IMRT field sizes is a circle whose diameter is equal to the length of the leaf bank.*

The maximum field size of intensity modulated fields that are generated with collimator rotation will not necessarily be subject to the same restrictions (see Figure 3.7). First, because delivery is not performed using a sliding window the leaves are not required to move from one field edge to the other. Therefore, the maximum field
width may be extended to twice the leaf length. Secondly, the orientation of the leaf bank, which has a maximum length of 40 cm, is constantly rotating with respect to the desired field. The upper limit on field size is therefore a 40 cm diameter circle.

3.3 Mechanical Characteristics

Linear accelerators that use an MLC to generate arbitrary field shapes are designed with collimator rotation capabilities. In standard radiation therapy it is sometimes desirable to rotate the MLC leaves so that they may fit more conformally to the target. Although this capability exists on the linacs used in this thesis, it was never intended for intensity modulation. The accuracy of MLC leaf positioning combined with collimator rotation must be assessed before embarking on the actual delivery of rotational IMRT fields. The following subsections describe the mechanisms of the MLC to accurately position leaves in dynamic as well as static modes. The collimator rotation mechanism is also described. Finally, the results of a series of experiments used to evaluate the accuracy and stability of the collimation system are presented.

3.3.1 Multileaf Collimator

Individual DC motors drive each MLC leaf. The MLC leaves consist of two opposing sets of banks that are mounted to a carriage. The carriage can be translated in the direction of leaf motion in order to move all leaves simultaneously. The MLC head assembly includes transceiver cards that convert leaf position data for the controller computer [16]. It also includes the motor driver cards that power the leaf and carriage motors. Finally, it includes secondary feedback circuitry from the leaves and carriages that are used to independently verify leaf position.

Primary feedback of leaf position is derived from each leaf motor through the leaf position encoder. The encoder divides each millimeter of leaf travel into 600
parts. Each encoder signals a relative position for its leaf. When the MLC is powered up a calibration procedure is used to initialize each encoder at a known leaf position.

Secondary feedback for each leaf is provided from two mechanical brushes mounted on and along each leaf in the direction of travel. As leaves move in and out of the radiation field, the brushes complete a circuit across contact points. This provides a signal that translates to leaf position. These feedback mechanisms are independent from each other. If either mechanism indicates an incorrect leaf position or if they become unsynchronized the delivery will terminate. This provides for accurate and reliable leaf positioning throughout dynamic and static delivery to within 1 mm.

The leaf calibration procedure is performed using an optical receiver and an infrared LED [16] as shown in Figure 3.8. First, all leaves are completely retracted so that the optical beam may pass between the emitter and receiver. The position of the straight line described by the sensor and the infrared LED is known. A given leaf is selected and slowly translated into the optical path. Once the leaf blocks the infrared light the optical sensor signals that the leaf has reached the calibration position. The leaf position is stored and the leaf is retracted. The next leaf is selected and the process continues until all leaves are calibrated on both banks.
Figure 3.8: MLC leaf positions are calibrated individually using an infrared LED emitter and optical sensor. The position of the optical path described by the sensor and the infrared LED is known. Each leaf is translated into the optical path. Once the leaf blocks the infrared light the optical sensor signals that the leaf has reached the calibration position. The leaf position is stored and the leaf is retracted.

3.3.2 Collimator Rotation

The collimator motor assembly is capable of driving the collimator through a rotation range of 330 degrees. The collimator is rotated with a DC motor attached to a mechanical chain that circles the outer perimeter of the collimator. Applying a signal to the motor translates the chain, which rotates the collimator system. The rotation angle is verified by a potentiometer attached to one of the gears that guide the chain around the collimator. As the chain translates, the signal from the potentiometer changes by an amount proportional to the angle of rotation. The linac control computer receives this signal and uses it to indicate when the collimator has reached a desired angle.
3.3.3 Collimator Angle Reproducibility

Varian models CI2100EX and CI2100C/D (Linac 1 and Linac 2 respectively) were used throughout this thesis. The main difference between these two linacs is their MLC leaf designs. A full description of the MLC design differences is presented in section 5.2.1 and is not relevant to the following study. The feedback and control mechanisms of the collimation system are virtually identical. In order to evaluate the feasibility of using collimator rotation in IMRT, measurement data from the quality assurance program already in place was obtained to evaluate the stability and accuracy of the collimator for both linacs. These measurements are obtained with a high level of precision because their quality will affect the quality of actual patient treatments.

The reproducibility of collimator rotation was evaluated monthly over a period of 45 months. An established quality assurance procedure was used to verify collimator angle position. First, the gantry was rotated to 90 degrees so that the collimator rotation axis was oriented perpendicular to the direction of the force of gravity. Because part of the collimator housing is shaped like a box with four flat sides, the top edge of each one of those sides should be oriented perpendicular to gravity at increments of 90 degrees. The collimator angle was then verified at each increment using an electronic spirit level placed on the upper edge of the collimator housing. Angles of 0, 90 and 270 degrees were evaluated in turn by selecting them from the control computer and allowing the collimator to rotate freely to the desired position.

Displayed in Figure 3.9 are histograms of the measured collimator angle for Linac 1 and 2 at 0, 90 and 270 degrees. The majority of the results are distributed close to the desired angle. These results show that the collimator rotation angle is accurate to within a maximum variation of 0.5 degrees. Considering that measurements were performed over a period of 45 months, the reproducibility of
collimator rotation is not an impeding factor to implementing rotational IMRT in a clinical setting.

3.3.4 Center of Rotation

In addition to collimator angle the accuracy and reproducibility of the collimator rotation axis was also assessed. A light field produced in the head of the linac and mimicking the radiation field is used to position patients for treatment. A cross hair suspended inside the collimator is used to indicate the central axis of the radiation field, which is also the center of rotation of the collimator. The light field projects through the cross hair and can be seen on the couch surface. When the collimator rotates the cross-hair projection also rotates.

The maximum displacement of the cross hair center as the collimator rotates through 180-degree was recorded weekly over a period of 1 year. Measurements were obtained using gantry angle of 0, 90 and 270 degrees. The results for Linac 1 and 2 are shown in Figure 3.10. The collimator rotation axis remained stable to within 0.5 mm on every occasion, displaying a high level of reproducibility. The error tolerances of linear accelerators are typically on the order of 1 to 2 mm. The collimator rotation axis exhibits a superior level of accuracy (< 1mm) than that required and will therefore not contribute significantly to dosimetric error when delivering intensity modulated fields.
Figure 3.9: Plotted are histograms of collimator angle reproducibility. Collimator angle was verified once per month over a period of 45 months. Measurements were performed at 0, 90 and 270 degrees for Linac 1 and Linac 2. Results show that the collimator angle is reproducible to within 0.5 degrees over a period of almost 4 years.
Linac 1

Linac 2

Figure 3.10: Histograms of the variation in the center of rotation over a 180-degree rotation. Measurements were obtained weekly for a period of 1 year. Results are shown for Linac 1 and Linac 2.

3.4 Linac Control

In addition to the mechanical characteristics of the collimation systems it is necessary to consider the x-ray production properties of the accelerator. Static and Dynamic delivery modes require that various components of the linac hardware function accurately and dependably. The base fluence generated by the source must be consistent and symmetric about the rotation axis. The monitor ionization chamber located in the head of the linac is divided into 4 quadrants. Each quadrant is independently verified throughout delivery. If any one of the quadrants shows a variation outside an accepted tolerance, the electron beam steering mechanism is adjusted to compensate. In this fashion the 'flatness' as well as the beam symmetry is assured to remain constant throughout delivery.
3.4.1 Static Delivery

Intensity modulation by multiple static segments requires that some sub-fields be delivered with relatively lower dose when compared to conventional radiation therapy. When the monitor chamber signals that the programmed delivery is complete the production of radiation is terminated. Radiation is produced in discrete pulses. Because the beam cannot be terminated within an individual pulse there is always an error of up to 1 pulse for each sub-field. For very short irradiation periods the percentage error resulting from an additional pulse may become significant [83, 84]. Also, there is a time lag between the signaling of the monitor chamber and the complete termination of x-ray production.

3.4.1.1 Dose Linearity

An experiment was devised to evaluate the dosimetric error introduced by lower monitor unit segments. A monitor unit is defined as the amount of charge collected in the monitor ionization chamber located in the head of the linac that will provide a specific dose under a specific set of delivery conditions. For the linacs used in this thesis 1 monitor unit will provide 1 cGy of dose to water for a 10 cm x 10 cm field size at the depth of dose maximum with the point of measurement located at 100 cm from the source. An ionization chamber was placed in a water equivalent phantom at this reference depth and field size. MU settings ranging from 1 to 10 were programmed and delivered using a dose rate of 300 MU/min. The results were then evaluated in terms of the linearity of radiation production with MU setting. If the delivery system were ideal, the dose delivered with 10 MU would be exactly 10 times the dose delivered with a setting of 1 MU. Measurements were also obtained at dose rates of 100 and 600 MU/minute to evaluate whether the lag time of the monitor chamber was a function of dose rate.
Figure 3.11: Measured dose versus Monitor Unit setting is plotted at dose rates of 100, 300 and 600 MU/min. Also plotted is a linear fit of the 100 MU/min results. Results consistently show that dose is linear with MU setting within experimental error. Shown in the bottom right hand corner is a magnified section of the dose axis intercept. The dose offset indicates a monitor chamber lag time resulting in an overdose of approximately 0.06 cGy and causing significant dosimetric error of over 2% for sub-fields of less than 4 MU.

Plotted in Figure 3.11 is measured dose versus MU setting for 100, 300 and 600 MU/min dose rates. Results show that dose delivered is linear with Monitor Unit setting for all dose rates. Also plotted in Figure 3.11 is a linear fit for the 100 MU/min results with an extrapolation of the straight line through the dose axis. A
blow-up of the plot near the origin shows that the intercept is located at an offset of 0.06 cGy. If the monitor chamber feedback process were perfect the amount of dose delivered at 0 MU would equal zero. The fact that the intercept is greater than zero indicates a lag in monitor chamber feedback. The lag time between the monitor chamber and termination of the radiation beam results in an overdose of 0.06 cGy at reference depth and field size. Lag times for the other dose rates tested were identical to within experimental error.

It can be concluded from these results that individual sub-field contributions will not be effected by dose non-linearity. The absolute number of monitor units for each sub-field can therefore be scaled according to the prescribed dose. Due to the monitor chamber lag time, there will be a large percentage error for very low dose sub-fields. Therefore, in order to reduce dosimetric error to less than 2% it is required that a minimum of 4 MU is delivered per sub-field.

3.4.2 Dynamic Delivery

The ability of the accelerator to accurately control dose rate and collimator rotation speed in real time was evaluated. The collimator rotation speed must be constant throughout the delivery. Also, the speed of rotation must be the same for each individual delivery to avoid constant recalibration. Finally, the dose rate must also remain constant throughout each delivery. Modification of either the dose rate or the collimator rotation speed during delivery will cause a phase shift between the desired location of each field and the field that is actually delivered. Following are a series of experiments that were designed to evaluate each of these potential sources of error.

3.4.2.1 Rotation Speed Reproducibility

The time for a 180-degree rotation was measured over 14 months for Linac 1 and Linac 2. A timer was used to measure the interval between when rotation was started
at the control workstation and when the collimator came to a full stop. Linac 1 and 2 showed consistent rotation times of 48 +/- 0.2 s and 45 +/- 0.2 s respectively. Over the 14-month period no significant deviation from these times was noted, proving that the period of collimator rotation is highly reproducible.

3.4.2.2 Dose Rate Stability

In addition to cumulative dose, the monitor chambers also verify that the accelerator maintains the desired dose rate. If the dose rate is lower than desired the feedback mechanism of the monitor chamber will allow more dose pulses to be delivered. Conversely, if the dose rate is too high some dose pulses will be removed. In the event that the dose rate changes by greater than 1% the beam is terminated. The delivery must then be restarted with MLC and collimator set to their positions when the beam was terminated. These mechanisms insure that no treatment field is delivered with an incorrect dose rate, making the chance of rotational delivery phase shift error due to dose rate variations unlikely.

In order for the linac to maintain a constant dose rate it is manually 'tuned' at regular intervals. Still, it is possible for the dose rate to become unstable which would cause errors in the delivery of intensity modulated fields.

3.4.2.3 Collimator Rotation Stability

The previous sections described the method of dose rate control and collimator rotation monitoring that is present with the current linac hardware. The period of rotation was found to be consistent and dose rate control is robust. However, to ensure IMRT with dynamic collimator rotation is realizable within a small enough margin of error further testing is required.

A testing procedure was developed to evaluate the combined effect of collimator rotation instability and dose rate fluctuations. Although it has been found
that the period of rotation of the collimator is consistent, there is no reason to believe that the speed remains constant throughout the motion. This could result in an unacceptable level of error when combined with dose rate fluctuations that exist during the beginning and end of the delivery.

Displayed in Figure 3.12 are a series of four sections in the dynamic irradiation of a small opening in the MLC located at radial distance $r$ from the isocenter. Radiographic film (X-OMAT V2) was placed in a water equivalent phantom at an arbitrary depth of 5 cm. The film was placed perpendicular to the beam central axis and irradiated with a 6 MV beam. Dose was deposited onto the film during the delivery through the small MLC aperture using Linac 1. A small opening in the MLC rotated by 180 degrees will describe an arc of dose on the film. If the rotation speed and dose rate is constant the dose measured along the arc should also be constant. Any variations indicate that the dose rate or collimator rotation speed was fluctuating during delivery.

In order to expose the film within its dynamic range the approximate radial distance $r$ and width of each MLC aperture $w$ for each dose rate was determined. The aperture width was fixed at 2 cm. The fraction of dose delivered at each point along the arc throughout rotation is given by the dose that would be delivered without rotation scaled by the ratio of the aperture width $w$ to the total path length $\pi r$.

$$D_{\text{rot}} = D \frac{w}{\pi r}$$ \hspace{1cm} (3.5)

The total dose delivered, $D$, is a function of the period of rotation $\omega$ (fixed) and the dose rate $\dot{D}$. For a rotation through 180 degrees (one half period) we have.
Figure 3.12: A procedure to test the combined effects of collimator rotation instability and dose rate variations. A small aperture is opened at radius $r$ from the central axis and rotated through 180 degrees during irradiation. The resulting profile of dose describes an arc and is measured with radiographic film. Dose along the profile should ideally be constant. Larger dose rates are evaluated by repeating the rotation with a larger radial distance to the MLC opening given by equation 3.7.
Solving for $r$, the radial distance from the center of rotation of each MLC aperture can be calculated with

$$D = \frac{\dot{D}\omega}{2}$$  \hspace{1cm} (3.6)

For a desired exposure $D_{rot}$ of 15 cGy, which is within the dynamic range of the film, the radii $r$ were calculated to be 3, 6, 9, 12, 15 and 18 cm for the 100, 200, 300, 400, 500 and 600 MU/min dose rate settings respectively. The film pixel values were converted to dose using the calibration procedure described in section 5.3.1

Displayed in Figure 3.13 is a grayscale image of a radiographic film exposed at all 6 dose rates. The dark lines of the 'rainbow' arcs show the dose delivered by dynamic rotation. Radius is increased with increasing dose rate. Visually, the arcs look uniform except for the beginning and the end of each trajectory. A decrease in dose is expected at those positions due to part of the MLC aperture exposing the arc ends for a smaller fraction of the delivery. A sharp increase in dose can also be seen at the ends of some trajectories. These increases indicate a deceleration of the collimator that does not coincide with the dose rate.
Figure 3.13: A grayscale image of the 2-dimensional dose distribution generated in the collimator rotation and dose rate stability test. Dark lines making up the arc of the 'rainbow' correspond to exposures at each dose rate. Gradients at the beginning and end of each path are expected and are caused by the edge of each aperture exposing the tips for a smaller period than the rest of the trajectory. Slightly higher doses are seen near the end of each trajectory and are likely due to a deceleration of the collimator in advance of dose rate termination.

Plotted in Figure 3.14 are dose profiles obtained along the aperture trajectory for all six dose rates. All plots have been normalized to the 90° position. Results show that overall uniformity is good although there are some dose spikes at the beginning and end of certain profiles. At the beginning of the 200 MU/min profile there is a sharp increase which is likely due to the collimator accelerating at a slower rate than the dose rate. It is only observed in this one case and is therefore most likely not a systematic error. For 400 MU/min and greater deliveries there is a dose spike at the end of the trajectory, as was visually noted in Figure 3.13. This is most likely due to the collimator decelerating at a slower rate than the dose rate. When the
Figure 3.14: Dose vs collimator rotation angle is plotted for dose rates ranging from 100 to 600 MU/min. Dose was measured along each trajectory of the open MLC apertures from the radiographic film of Figure 3.13. Non-uniformity in the dose profile indicates that the collimator rotation speed and dose rates are unsynchronized. Good overall uniformity is observed although there are some collimator deceleration artifacts and an increase in the standard deviation at 400, 500 and 600 MU/min.
monitor chamber records that the desired dose has been delivered the linac will stop producing radiation virtually instantaneously. The collimator cannot decelerate as quickly. There is therefore a period at the end of the trajectory where the collimator is moving slower while the linac is still at the maximum dose rate.

Note that the 100 MU/min plot shows a slow increase over the first 20° and then a decrease after 160°. This effect is expected and was described earlier on page 89. It is more pronounced for the 100 MU/min plot because the overall path length relative to the aperture size is much shorter.

The standard deviation $\sigma$ [85] was obtained over the central 120 degrees of each dose profile. The results are displayed under each respective plot in Figure 3.14. The standard deviation increases with increasing dose rate indicating that variations in dose rate and collimator speed increase as the dose rate setting is augmented. Furthermore, at higher dose rates it is evident that there is a lower dose in the center of the profile than at the outer edges, indicating a systematic difference between collimator rotation speed and dose rate. There is therefore an additional systematic error that must be considered when delivering dynamic rotational treatments at higher dose rates. The positional uncertainty resulting from the rotational instability will be greater with increasing radius and will therefore depend on the position and extent of the fluence map with respect to the center of rotation.

3.5 Feasibility of IMRT with Collimator Rotation

In section 3.2 the potential advantages of using collimator rotation in IMRT delivery were presented. Benefits resulting from increased spatial resolution, decreased interleaf effects and larger maximum field size can only be realized if the delivery apparatus is capable of controlling the collimation mechanism and x-ray production system accurately. In particular, rotation of the collimator adds a degree of complexity that has not previously been investigated in the context of IMRT delivery.
For this reason, a full mechanical and dosimetric study was performed to evaluate the feasibility of the proposed technique. In sections 3.3.3 and 3.3.4 it was shown that collimator angle and center of rotation are both accurate and reproducible. In static delivery the production of low dose segments must be considered. The results provided in section 3.4.1.1 showed that dose delivered was linear with MU setting. A dose offset due to the speed of the feedback mechanism was observed which can be avoided by only using MU settings greater than 4. These results indicate that for static rotational delivery there should be no significant increase in delivery error due to limitations of the linac.

With dynamic delivery there is the additional complication of ensuring that the dose rate and collimator speed are synchronized throughout delivery. MLC leaf motion and dose rate is synchronized through a feedback mechanism that has already been established for conventional IMRT delivery. In section 3.4.2.3 the results of an investigation into dose rate and collimator rotation synchronization were presented. Although the overall results show uniform rotation speed and dose rate there is a distinct trend toward a greater systematic error at higher dose rates. Also, acceleration and deceleration of the collimator are not matched to the increasing and decreasing dose rate at the beginning and end of each delivery. The errors described in section 3.4.2.3 could substantially affect the delivery of rotational IMRT fields. By taking into consideration these discrepancies when delivering actual IMRT fields it will be possible to minimize their effect.
Chapter 4

Leaf Motion Derivation

In the previous chapter the potential advantages of using MLC rotation in IMRT delivery were described. The viability of delivering fluence maps in this way with conventional linear accelerators was also investigated. It was determined that although they were not designed for this purpose, they generally perform within an acceptable margin of error. The results of this investigation will provide guidance as to how the linac should be controlled to reduce any additional uncertainties associated with collimator rotation.

Although rotation of the MLC provides an additional degree of flexibility when delivering a desired fluence map, it forces an increase in complexity when deriving the necessary leaf motions. To generate an arbitrary fluence map using a series of MLC defined sub-fields a derivation of the exact position of each leaf for each sub-field is required. With rotation, the collimator angle must also be calculated. It was therefore necessary to develop a series of specialized algorithms that are capable of determining rotated MLC segments as part of this thesis.

In this chapter a detailed description of the algorithms is presented (also published in Physics in Medicine and Biology [65]). Included is a brief review of conventional leaf motion calculation methods as well as a geometric analysis of the increase in complexity with collimator rotation. Next, the mathematical formulation of the leaf position problem is derived. Finally, the bulk of the chapter contains a detailed description of the optimization-based method capable of obtaining a solution.
4.1 Conventional Techniques - Review

Various techniques have been described to deliver IMRT. MLC based techniques are the most common and may be implemented on the majority of manufacturer's linacs. The MLC is used to define multiple uniquely shaped sub-fields at a fixed gantry angle, the sum of which result in a complex 2-dimensional fluence map. There are two basic variations on MLC based delivery techniques: static (step-and-shoot) and dynamic (sliding window). The former is a stationary method where each sub-field is shaped while the radiation beam is off and then a portion of radiation is delivered once the leaves are in position [17, 18, 25]. The latter involves moving the leaves while the beam is on and is similar to the first method but with a large number of time varying sub-fields [19, 86, 87].

4.1.1 Step And Shoot (Static)

Many static leaf motion calculation methods have been reported in the literature [19, 24, 25, 86-89]. The majority of these techniques are focused on reducing the total number of sub-fields required to deliver the desired fluence map. The step-and-shoot method is employed by the Varian Helios v.6.2 IMRT planning system which is used in the comparison study between conventional and rotation methods in Chapter 5.

Consider the fluence map of Figure 4.1(a) where the MLC leaves are oriented to travel horizontally between A and B. A profile obtained between an arbitrary leaf pair is shown in Figure 4.1(b). Each leaf must move along a specific trajectory to produce the desired fluence. To deliver the fluence in a finite number of sub-fields, the continuous distribution is quantized into a series of discrete levels as shown by the shaded area in Figure 4.1(b). Choosing more levels in the quantization procedure will result in a more accurate reproduction of the desired fluence but will necessitate a larger number of segments for delivery. The leaf positions are then derived using this discrete representation of the fluence profile. Figure 4.1(c) shows a plot of fluence
Figure 4.1: Method of generating intensity modulated fields using the 'step-and shoot' technique. The 2-dimensional fluence map in (a) is sampled along the leaf trajectory AB and is plotted in (b). The continuous fluence is then modified to produce regions of constant fluence as shown in the shaded area of (b). The leaf trajectories for leaf A and B are derived as a function of fluence index and are plotted in (c). The difference in fluence index between each trajectory is proportional to the fluence delivered at that point. The area between the two trajectories plotted as a function of leaf position is therefore equal to the shaded area in (b).
index versus leaf position for Leaf A and Leaf B. The fluence index represents the relative quantity of radiation that has been produced by the linac. The location of the open MLC area at any moment in the delivery is therefore given by the difference in leaf positions at that fluence index. Also, the total fluence delivered to a point in the field is given by the difference between the fluence indices of leaf A and leaf B at that point. Therefore, the fluence distribution of Figure 4.1(b) is equal to the area between the two leaf trajectories of Figure 4.1(c) plotted as a function of leaf position.

The trajectories for each leaf pair may therefore be solved in turn by the method described above. At the beginning of each index the leaves are moved to the derived position (stepped). The beam is activated and the desired quantity is delivered (shoot). The process continues in this manner until the desired fluence has been delivered.

4.1.2 Sliding Window (Dynamic)

Dynamic delivery of IMRT fields is performed in a similar fashion to the step-and-shoot method described above. The fluence profile between a leaf pair is quantized into a large number of discrete levels as shown in Figure 4.2(a). Unlike the static method, MLC leaves move continuously throughout delivery [19, 86, 87]. The position of each leaf is a function of the fluence index as plotted in Figure 4.2(b). As the fluence index increases the position and length of the gap (window) between the two leaves is translated (sliding) across the area of desired fluence. As with the step-and-shoot method the fluence accumulation is given by the area between the two leaf trajectories from Figure 4.2(b) plotted as a function of leaf position.
Figure 4.2: The sliding window technique is similar to the Varian step-and-shoot except that the fluence is divided into a large number of fluence levels as shown in (a). Also, each leaf pair moves continuously during the delivery as seen in (b), creating an opening in the MLC that slides from one side of the field to the other. The size and position of the window is calculated as a function of the fluence index. The fluence shown in (a) is generated by modifying the window width and position as plotted in (b).

4.2 Rotational Technique Leaf Motion Derivation

4.2.1 Increased Complexity

Conventional IMRT leaf motion calculations are effectively a one-dimensional problem. A single leaf and its opposing pair are the only two leaves that directly affect the fluence at any given point. Therefore, the necessary leaf motions required to deliver the correct fluence at that point are almost entirely dependent on those two leaves. The derivation of all leaf motions can therefore be separated into a series of one-dimensional equations involving each leaf and its opposing pair.

With MLC rotation this approach is no longer possible. Different leaves will affect different points in the desired fluence map depending on their location and the
collimator rotation angle. This is shown in Figure 4.3 which illustrates the fluence and MLC apertures at nine collimator angles in the dynamic delivery of a wedge shaped intensity distribution.

Figure 4.3: Fluence maps are generated dynamically by rotating the collimator while the MLC leaves are in motion. Each frame in (a) to (i) shows the progressive build-up of a wedge shaped fluence with the rotated MLC aperture at that instant.

Referring to Figure 4.4(a), for a desired fluence pixel at a point \((x,y)\) and the collimator rotated to angle \(\theta\), the MLC leaf position \(P\) and leaf pair \(L\) that intersect that pixel are given by:

\[
P = x' = x\cos\theta - y\sin\theta
\]
\[
L = y' = x\sin\theta + y\cos\theta
\]

(4.1)
where the coordinate system \((P, L) = (x', y')\) is fixed to the frame of reference of the MLC.

![Diagram](image)

**Figure 4.4:** (a) The rotating coordinate system of the MLC. (b) A sinogram of the trajectory of points A, B and C as they move through opposing leaf pairs. The points follow a sinusoidal path through the MLC leaf pairs with each point having a different phase and amplitude. The amplitude is equal to the radial distance of the point from the isocenter and the phase is a function of its initial position along the leaf bank.

As the collimator rotates, the MLC leaf pairs capable of modulating the fluence at a specific point change. A sinogram of the trajectory of individual fluence pixels through the MLC leaves as a function of collimator angle is shown in Figure 4.4(b). The level of complexity of the leaf motion calculation is appreciated by observing the trajectory of points A, B and C shown in Figure 4.4(a) through the MLC sinogram. As the collimator rotates, each point follows a sinusoidal trajectory through the leaf pairs. Points A and B are located at the same distance from the isocenter and have the same amplitude whereas Point C is closer and has a smaller amplitude. The amplitude of the sinusoid is equal to the radial distance of that point from the center of rotation. Points that are located further from the isocenter will
therefore intersect a larger number of leaf pairs throughout the rotation. Although Points A and B have the same amplitude in the sinogram their trajectories are out of phase. The phase of the sinusoid is a function of the initial position of the fluence point along the MLC leaf bank. The interdependence of leaf position and collimator angle is inherently complex and must be incorporated into the leaf motion calculation algorithms.

4.2.2 Analytic Model

To develop a method of calculating the leaf positions and collimator angles for arbitrary fluence maps, it was necessary to derive a mathematical model of the system. A 2-dimensional distribution of fluence $\Phi(x, y)$ generated by an MLC aperture $\Omega(x, y)$ with primary base fluence originating from the source $\Phi_p(x, y)$ is given by

$$\Phi(x, y) = \Omega(x, y)\Phi_p(x, y)$$  \hspace{1cm} (4.2)

where $\Phi_p(x, y) \equiv 1$ for a clinical linear accelerator ($\Phi_p(x, y)$ is compensated for in the treatment planning phase before the desired fluence maps are generated and is therefore not included in the following calculation). A complex fluence generated by a series of $M$ MLC apertures is

$$\Phi(x, y) = \sum_{m=1}^{M} \Omega_m(x, y).$$  \hspace{1cm} (4.3)

For the purposes of fluence generation an MLC with total number of leaves $N$, for each leaf $n$, the aperture generated by the left leaf $L$ and the right leaf $R$ are defined by the step functions
\[ L_n(x, y) = \begin{cases} 0 & x \leq 0, (n-1)w < y < nw \\ 1 & x > 0, nw < y < (n-1)w \end{cases} \tag{4.4} \]

\[ R_n(x, y) = \begin{cases} 0 & x \geq 0, (n-1)w < y < nw \\ 1 & x < 0, nw < y < (n-1)w \end{cases} \tag{4.5} \]

respectively where \( w \) is the leaf width. In one-dimension, the aperture function for any leaf pair may be written

\[ \Omega(x) = L_n(x - x_L)R_n(x - x_R) \tag{4.6} \]

where \( x_L \) and \( x_R \) are the left and right leaf end positions. Equations 4.4 and 4.5 are plotted in Figure 4.5(a) with an example of a 1-dimensional aperture function. A full 2-dimensional aperture is described by

\[ \Omega(x, y) = \prod_{n=1}^{N} L_n(x - x_L(n), y)R_n(x - x_R(n), y) \tag{4.7} \]

an example of which is plotted in Figure 4.5(b).
Figure 4.5: One-dimensional representation of the left and right leaf functions $L_n(x, y)$ and $R_n(x, y)$ from equation 4.4 and 4.5 respectively are shown in (a). Also shown is an example of a 1-dimensional aperture function $\Omega(x)$ from equation 4.6 where $x_L$ and $x_R$ are the left and right leaf positions for a single leaf pair respectively. A full 2-dimensional aperture function using multiple leaf pairs is shown in (b).

These equations define the aperture function for a non-rotating MLC. With collimator rotation, the angle of rotation $\theta$ is introduced. Using the coordinate transformation defined in equation 4.1 and substituting $x$ and $y$ for $x'$ and $y'$ respectively, the aperture function is

$$
\Omega_\theta(x, y) = \prod_{n=1}^{N} L_n \left( x\cos\theta - y\sin\theta - x_L(n), x\sin\theta + y\cos\theta \right) \times R_n \left( x\cos\theta - y\sin\theta - x_R(n), x\sin\theta + y\cos\theta \right). \quad (4.8)
$$

The fluence generated for a series of apertures through a rotation of 180 degrees is

$$
\Phi(x, y) = \sum_{\theta=0}^{\pi} \Omega_\theta(x, y) \quad (4.9)
$$
which, when expanded, is equal to

\[
\Phi(x, y) = \sum_{\theta=0}^{N} \prod_{n=1}^{N} L_n \left( x \cos \theta - y \sin \theta - x_L(n), x \sin \theta + y \cos \theta \right) \\
\times R_n \left( x \cos \theta - y \sin \theta - x_R(n), x \sin \theta + y \cos \theta \right).
\]

Therefore, given an arbitrary desired fluence \( \Phi(x, y) \), it may be reproduced using a series of MLC defined sub-fields by solving equation 4.10 for the leaf positions \( x_L(n, \theta) \) and \( x_R(n, \theta) \) for all leaf pairs \( n \) and over all collimator angles \( \theta \).

There is no method of solving equation 4.10 directly. Also, due to the discrete representation of fluence maps as a matrix with finite pixel sizes, a method to describe them as a continuous function is not obvious [90]. Still, the above formulation is an accurate description of the system that must be evaluated. The following sections describe alternative solutions to equation 4.10 using optimization techniques.

### 4.2.3 Optimization Methods

Optimization techniques have been used in IMRT since its inception although not in the context of generating leaf sequences. Derivation of the desired fluence map for a sequence of beams is accomplished using gradient and stochastic based optimization methods [21, 22, 91, 92]. The problem, in simple terms, is to arrive at a series of 2-dimensionally varying fluence maps such that a uniform dose is delivered to the tumour while avoiding dose to the surrounding healthy tissue. See section 1.5.2 for a full description. A general discussion of optimization methods is provided here as an introduction to the rotational leaf motion calculation algorithms.

Firstly, the optimization problem is described by an objective function. This function serves to define the goal of the optimization. It also quantifies the progress
of the optimization as it moves towards that goal. The objective function for the leaf motion derivation problem is defined as the absolute difference of the desired and calculated fluence maps summed over all pixels.

\[
Obj = \sum_{x,y} |\Phi_{\text{desired}}(x,y) - \Phi_{\text{calculated}}(x,y)|
\]  

(4.11)

As \( Obj \) approaches 0 the difference between desired and calculated fluence maps approaches 0. The goal of the optimization is to calculate the leaf positions \( x_L(n,\theta) \) and \( x_R(n,\theta) \) such that equation 4.11 is a minimum. Because equation 4.11 contains equation 4.10 it is not feasible to obtain an analytical solution. Instead, equation 4.11 is evaluated at different values of \( x_L(n,\theta) \) and \( x_R(n,\theta) \) until a solution is obtained. After each attempt (iteration) \( Obj \) is evaluated to determine how the new values have changed the result. The aim is to consistently reduce the objective over several iterations until it no longer improves as shown in Figure 4.6(a). How the values of \( x_L(n,\theta) \) and \( x_R(n,\theta) \) are determined after each iteration is given by the optimization method.

4.2.3.1 Gradient Based Methods

A minimum of equation 4.11 is obtained when the gradient of \( Obj \) with respect to \( x_L(n,\theta) \) and \( x_R(n,\theta) \) is equal to 0.

\[
\frac{\partial Obj}{\partial x_L(n,\theta)} = 0 \quad \frac{\partial Obj}{\partial x_R(n,\theta)} = 0
\]  

(4.12)

Gradient based optimization methods attempt to arrive at a minimum by evaluating the local gradient at some \( x_L(n,\theta)_m \) and \( x_R(n,\theta)_m \). The values of \( x_L(n,\theta) \) and
$x_R(n, \theta)$ for the next iteration $m+1$ are chosen by incrementing their value from the previous iteration $m$ in the negative direction of the gradient

$$x_L(n, \theta)_{m+1} = x_L(n, \theta)_m - k \nabla \text{Obj} \quad x_R(n, \theta)_{m+1} = x_R(n, \theta)_m - k \nabla \text{Obj} \quad (4.13)$$

which is the downward slope of the function at that point. The magnitude of the step $k$ is, in general, reduced as the gradient approaches zero. Variations exist on the basic methodology described here that attempt to determine the minimum more efficiently and/or more accurately [93].

Figure 4.6: Example of the minimization of an objective function $\text{Obj}$ where $x_L(n, \theta)$ and $x_R(n, \theta)$ are modified after each iteration $m$. Gradient based methods fail to locate the global minimum when there is a high density of local minima as shown in (b).

To find the global minimum with gradient based methods, it is necessary that either the function have only one minimum or the first iteration be chosen at a value already in the valley of the global minimum. Otherwise, the optimization will become trapped in a local minimum as shown in Figure 4.6(b). Gradient based methods will therefore be unsuccessful for objective functions that have a topography with many local minima.
An investigation into using gradient-based methods for rotational leaf motion derivation was performed as part of this thesis work. Preliminary results revealed that the topography of the objective function defined in equation 4.11 consists of a high density of local minima. The optimization consistently terminated after a small number of iterations producing a calculated fluence map with errors of greater than 50% over the majority of its area. Based on these results no further investigation was performed using these methods.

4.2.3.2 Stochastic Methods

Certain optimization problems have been identified that benefit from a random selection of the argument values [94]. Problems with many shallow local minima can be solved more accurately because, through random processes, the values of the arguments are able to "jump" from one valley to another. The objective function is still used to evaluate the progress of the optimization but is more limited in determining the argument values for the next iteration. The degree of randomness is typically a parameter that is adjusted as the number of iterations increases. For example, the arguments may have values that vary wildly from one iteration to the next at the beginning of the calculation and then slowly become less erratic towards the end. This permits an overall evaluation of the objective function topography at the beginning with a more local search for the minimum at the end.

4.3 Rotational Leaf Motion Optimization

The collimator rotations and leaf sequences used in this paper are derived using an optimization technique developed in-house\footnote{Patent Pending.} [65, 66]. The basic method of the algorithm is stochastic in nature and it can, in some respects, be likened to simulated annealing [94]. In the calculation, leaf positions are randomly varied at prespecified

\footnote{Patent Pending.}
collimator angles. With each variation the resulting primary fluence is updated. The absolute difference between calculated and desired fluence maps is then evaluated. Whether the leaf position variation is accepted or not is based on two separate criteria. First, if the variation brings the calculated fluence closer to the desired fluence then it is accepted. If there is no improvement an examination of those particular fluence pixels that are affected is performed. If those pixels fall within a dynamically controlled range of acceptable error then the variation is also accepted. If neither criterion is satisfied then the variation is discarded and a new variation is attempted. The following sections provide a detailed description of the algorithms.

4.3.1 Preprocessing

Desired fluence maps are generated with a pixel size of 2.5 x 2.5 mm². All deliveries are planned with a predetermined collimator rotation range that is divided into equally spaced segments. The rotated position of MLC leaves with respect to pixels in the fluence map are precalculated for all possible leaf positions at each rotated segment. Only a fraction of the precalculated segment angles are used with the static delivery method while all segments are used when delivery is performed dynamically. This calculation is performed only once. The precalculated rotated positions are loaded into memory each time the optimization software is started. This reduces calculation times significantly because the location of changes to the calculated fluence from varying leaf positions are obtained directly from memory instead of calculated with each leaf position change using equation 4.1.

4.3.2 Constraints

MLC leaves are constrained so that the maximum leaf span does not exceed the mechanical limits of the given multileaf collimator. Overlapping of opposing MLC leaves is also forbidden. In dynamic delivery there is the additional constraint that
the leaves can only move by a small amount between segments. This constraint is a direct result of the (currently) fixed collimator rotation speed, dose rate and maximum leaf velocity. The maximum intersegment leaf displacement $s_{\text{max}}$ is limited by the maximum leaf velocity $v_{\text{max}}$, the number of segments per collimator angle $N$ and the collimator rotation rate $\omega$ where

$$s_{\text{max}} = \frac{v_{\text{max}}}{N\omega}.$$ (4.14)

I.e., at the maximum leaf speed there will be a maximum leaf displacement between any two segments. If a greater leaf speed was required the collimator would have to slow down or the dose rate would have to be reduced in order to compensate, which is currently not an option.

Finally, there is a minimum gap required between opposing leaves in order to avoid collisions when both leaves are moving. A width of one pixel (2.5 mm) is used for this minimum gap due to the finite pixel size of calculated and desired fluence maps.

### 4.3.3 Fixed Parameters

In addition to the rotation range and maximum number of segments, there are some additional fixed parameters that are defined prior to optimization. The desired fluence map is normalized to a value that represents the radiation efficiency. The radiation efficiency is defined as the fraction of the total fluence produced by the source that contributes to the point of maximum fluence. Equivalently, it is the percentage of the delivery that the multileaf collimator exposes the point of maximum fluence. E.g. a value of 100% would require that the point of maximum intensity in the desired fluence map be open to the source throughout delivery (i.e. no MLC leaf
will cover that point). A lower value will force the location of maximum intensity to be covered by the MLC for a certain portion of the delivery. Typical values range from 60% to 90% and are dependent on the complexity of the desired fluence map. Another parameter that is defined beforehand is whether the delivery will consist of multiple static rotated segments (static mode) or if the collimator will rotate throughout delivery (dynamic mode). Finally, the segment weights are also fixed, which is required to maintain a constant dose rate in dynamic delivery, a current limitation of the linac control software when collimator rotation is used.

4.3.4 Initialization

Before the optimization can begin it is necessary to define a starting position for all MLC leaves at each segment. With dynamic delivery the leaves are initialized flush with each other and closed so that the central leaf pair closing point is at the central axis. Only a fraction of the total number of possible segments are used at the beginning of the optimization procedure. Typically, 10 out of the 160 possible segments are initialized. These 10 segments span the entire rotation range and are equally spaced. The other segments are introduced successively throughout the optimization.

The initialization procedure is slightly different when multiple static segments are used. Unlike the dynamic delivery initialization, the closing point of leaf pairs for each segment is chosen to coincide with the point of maximum fluence along the trajectory of those leaves. Therefore for each leaf pair and at each rotated segment, the leaves are closed over the points of maximum fluence. This technique has advantages in that when the optimization starts, the leaves open at a location of maximum fluence, which is where the MLC will be open for the greatest portion of the treatment. Initialization in this manner is limited to the multiple static segment delivery method because it is possible that the same leaf pair might be initialized to
dramatically different locations between consecutive sub-fields, which would cause the maximum leaf displacement restriction of equation 4.14 to be violated.

4.3.5 Optimization

A flowchart of the basic optimization process is shown in Figure 4.7. A segment, MLC leaf and displacement direction are randomly selected. Displacements are made in 2.5 mm increments. The proposed change in the MLC configuration is verified with the mechanical and physical constraints imposed by the MLC. If the constraints are violated then a new randomly selected modification is attempted. Otherwise, the objective function is updated. The objective function is the absolute difference of the desired and calculated fluence maps summed over all pixels as described by equation 4.11.

If the objective function has decreased then the modification is accepted and another leaf motion is attempted. Otherwise, a second test is performed. If the mean difference between desired and calculated fluence of the pixels in the area where the modification has taken place \((x,y)\) is within a margin of error \(\varepsilon\) then the iteration is also accepted.

\[
\text{mean}\left|\Phi_{\text{desired}}(x, y) - \Phi_{\text{calculated}}(x, y)\right| \leq \varepsilon
\]  

(4.15)

If this second criteria is not satisfied then the modification is rejected and a new one is attempted. Due to the similarities between equations 4.11 and 4.15 only the left-hand side of equation 4.15 needs to be calculated during optimization. The calculation result is used to update equation 4.11 after which both criteria can be tested. The optimization proceeds in this fashion until the objective function no longer improves over a pre-specified number of iterations.
4.3.6 Segment Doubling

With dynamic delivery, the optimization procedure starts with 10 possible segments from which the algorithm randomly chooses. Eventually, after several iterations, variations in the MLC configurations no longer reduce the objective function significantly. At this point the number of segments are doubled. The new segments are placed at collimator angles half way between the previous segments. The leaf
positions of new segments are derived by linearly interpolating between the leaf positions of the adjacent segments. The optimization then continues as described above. The doubling of segments continues until all 160 segments are optimized.

4.3.7 Dynamic Error Margin Control

Throughout the optimization, the error margin $\epsilon$ from equation 4.15 is modified. Initially, the error margin is set to a maximum and, as the objective function decreases, the error margin is decreased. After each segment doubling, the error margin is reduced by a factor of 2. After the final doubling has occurred the error margin is reduced further until the objective function no longer decreases.

4.3.8 Rotational Leaf Motion Software

Using Matlab\textsuperscript{4} software, a series of computer programs were developed as part of this thesis to perform the leaf motion calculation, evaluate the resulting calculated fluence and output the MLC leaf sequence for delivery on the linear accelerator. A ‘screenshot’ of the graphical user interface used to perform these tasks is shown in Appendix B (Figure B.1). The software also facilitated the manipulation and analysis of various fluence maps used in the experimental evaluation presented in Chapter 5.

4.4 Algorithm Considerations

Derivation of the rotational leaf positions is accomplished through an optimization process. Several parameters are fixed prior to optimization that will affect the resulting fluence map. Range of rotation, number of segments, radiation efficiency and the initialization procedure are not included as optimized variables. Due to the stochastic nature of the algorithms a repetition of the leaf sequence calculation may

\textsuperscript{4} The Mathworks, Natick, MA
change the result depending on the random number generation seed and sequence used in the optimization. Any variation will have to be minimized if the algorithms are to be considered robust. Finally, the actual fluence generated by the linac may not be identical to the predictions of the leaf motion calculator.

The following chapter addresses these issues by including a section where desired and measured dose distributions are compared. Also, a series of simulations are performed to evaluate the dependence of the resulting fluence maps on each one of the fixed optimization parameters.
Chapter 5

ROTATING MLC EVALUATION

This chapter describes a series of experiments devoted to a thorough evaluation of the rotational technique (also published in Physics in Medicine and Biology [65]). Included is a section describing the results of these experiments followed by a discussion of their significance focusing on clinical relevance and improvements from conventional techniques.

The first series of experiments involve a thorough characterization of the algorithms. Included is a series of simulations focusing on the algorithm reproducibility. Dependence on radiation efficiency, the range of collimator rotation and number of delivery segments is also evaluated.

Next, a series of clinically relevant fluence maps are used to verify the algorithms under a variety of conditions. The leaf motions for each clinical fluence map are generated for both static and dynamic rotational deliveries. Furthermore, the calculation is repeated using typical clinical MLC leaf widths of 5 mm and 1 cm for each fluence map. To interpret the results effectively comparisons are made between rotational and conventional IMRT methods. For this purpose, the entire series of leaf motion derivations were repeated for each fluence map and MLC design using the conventional step-and-shoot and sliding window methods.

A dosimetric evaluation of the technique is also presented. Measurements of varying complexity fluence maps are performed using a high-resolution film dosimetry technique. In order to properly evaluate the resolution capabilities of
rotational IMRT delivery a robust film dosimetry system was developed as part of the thesis work and is described in this chapter. Measurements are performed for 5 mm and 1 cm leaf width designs for both static and dynamic rotational delivery methods. Again, for comparison purposes, the entire series of measurements are repeated using leaf motions derived from conventional step-and-shoot and sliding window methods.

Finally, the remainder of the chapter is devoted to evaluating spatial resolution capabilities, interleaf effects and maximum deliverable field size. Specific experiments are designed to evaluate each characteristic. Experiments are repeated for both MLC designs and results are compared to conventional techniques.

5.1 Method

5.1.1 Algorithm Assessment

The ability of the algorithms to accurately derive rotational leaf motions was evaluated using a 5 field thyroid treatment plan shown in Figure 5.1. IMRT was indicated for this patient due to the close proximity of the spinal cord to the thyroid planning target volume. The fields were centered on the target volume (thyroid) and oriented coplanar to the transverse plane with equally spaced gantry angles (0, 72, 144, 216 and 288 degrees). A uniform dose to the target volume with minimal dose to the spinal cord was selected for plan optimization. The fluence maps were then optimized using inverse planning software (Varian Helios v 6.2).

Desired fluence matrices were calculated on a 2.5 x 2.5 mm$^2$ grid and are displayed in Figure 5.2(a) to (e). A 2 dimensionally varying sinusoid was also used in the evaluation and is shown in Figure 5.2(f).
Figure 5.1: A 3-dimensional view of the thyroid treatment geometry. IMRT was indicated in this case due to the close proximity of the spinal cord to the PTV. A dose distribution providing minimal dose to the spinal cord and a uniform dose to the thyroid was obtained by optimizing the fluence maps of the 5 fields that are shown.
Figure 5.2: Five clinical fluence maps as well as a 2-dimensionally varying sinusoidal fluence were used to evaluate the rotational leaf motion algorithm. (a) to (e) are optimized fluence maps for a thyroid treatment at gantry angles of 0, 72, 144, 216 and 288 degrees respectively. The 2 dimensional sinusoid is shown in (f).
5.1.2 Fluence Generation Parameters
Rotational leaf motions were calculated using a radiation efficiency of 80% and a rotation range of 180 degrees. Dynamic as well as static delivery methods were tested with 20 equally spaced segments used for static delivery. The effect of different MLC leaf widths was evaluated by calculating leaf motions for 5 mm as well as 1 cm width leaves. Resulting fluence maps for each combination of delivery parameters were then compared to the desired fluence maps.

5.1.3 Conventional IMRT Fluence Generation
Results of the rotational leaf motion calculation were also compared to the fluence generating capabilities of a conventional IMRT planning system leaf motion calculator (Varian Helios v 6.2). Sliding window as well as step-and-shoot leaf motion calculations were performed for both 5 mm and 1 cm leaves. The same number of segments were used for static rotational and conventional step-and-shoot methods.

5.2 Dosimetric Evaluation
A series of experiments were also performed to: (1) Evaluate the capability of linac and MLC hardware to generate accurate dose distributions with the rotational method and (2) Investigate the potential dosimetric advantages with regards to spatial resolution and reduced interleaf effects. Several fluence maps of varying complexity were used and are shown in Figure 5.3. They are: (a) a Gaussian, (b) a constant “wedge” gradient, (c) the complex 2–dimensional sinusoid and (d) a C-shape of constant intensity used to evaluate edge conformity. Other investigators have reported similar test fluences and shapes [61, 95]. Dose distributions were measured using Kodak EDR-2 film placed in a solid water cassette. The film cassette was oriented perpendicular to beam central axis and positioned at the isocenter in a square
phantom at 5cm depth. The film measurements provide a dose distribution resulting from the fluence given by the calculated leaf motions and collimator rotations. Measured dose distributions were compared with the desired dose at 5cm depth in water calculated from the desired fluence maps. Our technique was also compared to conventional step-and-shoot and sliding window methods using an IMRT planning system (Varian Helios v 6.2). The number of segments was forced to be equal between the “step-and-shoot” and static rotational techniques in order to provide an accurate comparison.

Figure 5.3: Test fluence maps of varying complexity were used to evaluate the rotational delivery, (a) a Gaussian, (b) a wedge and (c) the 2-dimensionally varying sinusoid. (d) A constant intensity C-shape was used to evaluate the ability of the technique to conform to an irregular shape.
5.2.1 Delivery

Rotating IMRT fields may be delivered statically (with the collimator not rotating and the leaves stationary during irradiation) or dynamically (with the collimator rotating and leaves moving simultaneously during irradiation) with no modifications to existing linac hardware required. The fields are delivered using linacs 1 and 2 (Varian 2100EX and 2100C/D, Palo Alto, CA) with different MLC models. The first has a 52 leaf model (Mark2) with 1.0 cm width leaves and the second has a 120 leaf model (Millennium) with 0.5 cm leaves over the central 20 cm and 1.0 cm leaves outside. MLC sub-field information is converted from the leaf motion and collimator rotation calculation algorithms into a format compatible with the MLC workstation software. During static delivery, individual fields are transferred to the MLC workstation and each collimator angle is programmed separately into the linac control interface. Once the collimator and MLC are in position the desired Monitor Units (MU) for that sub-field are delivered. The MU values for each sub-field were calculated to give a dose distribution that lied within the range of the film. During dynamic delivery all sub-fields are downloaded into the MLC workstation at once with MU values and dose rates chosen to generate a dose distribution that, again, is within the dynamic range of the film. Results were compared in terms of relative dose only. The collimator is rotated to the starting angle and the termination angle is programmed. At the same time that the collimator starts rotating, the beam is activated and the dynamic treatment starts on the MLC workstation. The collimator rotates with the MLC leaves moving until it reaches the termination angle.

5.3 Film Dosimetry

In evaluating 2-dimensionally varying high-resolution dose distributions it is necessary that the measurement technique have resolution characteristics that do not increase the uncertainty of the result. An ionization chamber, for example, measures
only at a point. Furthermore, the sensitive volume of most point detectors is greater than 2 mm [96] causing a blurring of the true value at that point. Smaller detectors, although having adequate resolution, still require multiple measurements to acquire a 2-dimensional distribution. A 10x10 cm² distribution measured at 1 mm intervals would require 10000 individual measurements.

For the purposes of this study a radiation measurement procedure using radiographic film was developed. Used in diagnostic radiology as a high-resolution image receptor, film is capable of measuring dose deposition at a spatial resolution of < 1mm. Several investigators have evaluated the properties of film as a radiation dosimeter under various conditions [97-99]. Their results indicate that, above all, consistency throughout the measurement and development process is required to obtain accurate and reproducible results. Other considerations include increased sensitivity to low energy photons and variations in sensitivity between film batches [100-102]. The following is a description of the film measurement and calibration procedure developed specifically for planar IMRT dose measurements as part of this thesis.

5.3.1 Conversion To Dose

After the film has been exposed and processed, the amount of dose deposited at a point is proportional to the optical density at that point. Optical density is determined by measuring the transmission of a collimated light source through the film (KODAK EDR2). Using a VIDAR Dosimetry Pro film digitizer each film is scanned and then output as a 2-dimensional matrix of pixel values that are directly proportional to optical density. Next, the pixel values are converted to dose. In order to establish the relationship between optical density and dose a separate irradiation is performed using known doses. By obtaining the optical density values for a range of known
doses a calibration curve is generated. The 2-dimensional matrix of optical density values is converted to dose using this calibration curve.

5.3.2 Film Calibration Technique

Due to the higher atomic number of the radiographic film emulsion there is a higher number of photoelectric interactions for low energy photons in film than for tissue. Film is therefore more sensitive to lower energy photons than tissue (effectively water). If the spectrum of incident photon energies was the same over the area where dose is deposited there would be a simple scaling of the film measurement that would be accounted for in the calibration procedure. Unfortunately, due to scatter of low energy photons the incident spectrum varies slightly over the field area. Non-uniformity of the energy spectrum becomes more significant at larger field sizes and is also a function of the depth of measurement [99]. A film calibration procedure was developed to minimize these effects. The calibration measurement is performed using a simple form of intensity modulation. This provides a calibration relationship that is more representative of IMRT scattering conditions resulting in a more accurate conversion of optical density to dose.

5.3.2.1 Enhanced Dynamic Wedge (EDW)

A simple form of intensity modulation involves moving one of the secondary collimator jaws across a radiation field to produce a 1-dimensional ‘wedge’ gradient as shown in Figure 5.4(a). The ‘Enhanced Dynamic Wedge’ (EDW) is the most basic form of IMRT using multiple sub-fields to generate a field of non-uniform intensity. Photon scatter characteristics of EDW fields will be similar to those of complex IMRT fields.
Figure 5.4: A diagram depicting the generation of a one-dimensional gradient using an Enhanced Dynamic Wedge (EDW) is shown in (a). The collimator jaw is continuously translated across the field of incident photons during delivery. The transmitted fluence at a point is given by the amount of radiation delivered before the jaw passes over that point. Results of water tank ionization chamber measurements shown in (b) were obtained for two EDWs with different MU settings. In order to generate an adequate calibration curve, the number of MU was chosen so that the dose from the two deliveries would span the range of doses of each IMRT field.

5.3.2.2 Calibration Curve

The optical density to dose calibration is measured using 2 EDW fields delivered side by side as shown in Figure 5.5(a). The two fields have different MU settings so that each one spans a different dose range in the calibration curve. The true dose delivered by these fields was determined in a water tank fitted with a small mechanically positioned ionization chamber (Welhöffer IC10). All measurements were performed at 5 cm depth and at 1 cm intervals along the gradient of each wedge.
The scatter contribution from the adjacent wedge was also included in each measurement. Results are displayed in Figure 5.4(b).

Figure 5.5: A scanned image of a typical calibration film is shown in (a). Pixel values are obtained at 1 cm intervals along the gradient of each field. The dose at these points was previously measured with an ion chamber as shown in Figure 5.4(b). Plotted in (b) are the pixel values versus dose values for the measurement points of (a). The resulting curve is used to convert each measurement film to dose. A 3rd order polynomial is fit to the plotted values to simplify the conversion. Finally, the calibration is verified by observing the overlapping dose values from the high and low doses of Field 1 and Field 2 respectively. Any discontinuity between the two curves will indicate an error in the calibration.

A calibration film is generated for each set of IMRT measurements. Optical density values are obtained at the same points used in the ion chamber measurements. The optical density to dose calibration curve is generated by plotting the optical density values on the calibration film versus the dose obtained from the ion chamber. Plotted in Figure 5.5(b) is a typical calibration curve. The Monitor Unit (MU) settings of each EDW field are chosen such that dose points at the low end of the high
MU EDW overlap with dose values at the high end of the low MU EDW as seen in Figure 5.5(b). Any discrepancies due to an error in calibration, scattering characteristics, film emulsion non-uniformity or processing variations will be apparent if there is a discontinuity between the two curves at the overlapping points.

Finally, for convenience in the conversion of IMRT film pixel values to dose, the data is fit to a 3rd order polynomial. Measurement uncertainties for film dosimetry using Kodak EDR2 film have been reported by several investigators [100-102]. From their results it is estimated that the uncertainty in measurement using the technique described here is less than 2%. Therefore, unless otherwise indicated, the results presented in the remainder of this thesis have a measurement error of 2%.

5.4 Results

5.4.1 Algorithm Characteristics

Plotted in Figure 5.6 is a graph of the objective function versus number of iterations for a typical dynamic delivery (Fluence map of Figure 5.6(d)). Each successful modification of the MLC configuration is defined as one iteration. Roughly 200,000 iterations are required before the objective function ceases to improve. Convergence is achieved in less than one minute on a 1 gigahertz Pentium III (Intel Corporation) computer. Also plotted in Figure 5.6 are the error margin and the number of segments as a function of the number of iterations. With each segment doubling, the objective function is temporarily increased but quickly reduces to a value lower than that prior to the doubling. Segment doubling occurs after 10,000 iterations have passed with < 0.1 % decrease in the objective function. As the number of segments is increased the error margin is also reduced. After the last segment doubling there is a final reduction in the error margin which is used to force any points that are close to their desired value to the smallest possible difference. Increasing the number of
segments from 80 to 160 yields only a small benefit and there is no significant benefit with further increases.

Figure 5.6: A typical dynamic rotating leaf motion optimization history. The mean difference between desired and calculated fluence maps decreases rapidly over the first few thousand iterations. Once the mean difference begins to converge the number of segments is increased and the margin of acceptable error is decreased. The optimization proceeds in this fashion until no further improvements are observed.
5.4.2 Reproducibility

The algorithm is stochastic in nature and will therefore potentially yield different results each time the calculation is performed. To evaluate this effect the optimization was repeated 100 times for a series of cases with each calculation having a different random number generation seed. Displayed in Figure 5.7 is a typical histogram of the resulting mean difference between desired and calculated fluence from 100 repeated calculations for the same fluence used in Figure 5.6 (shown in Figure 5.2(d)). The mean difference ranges from 2.30% to 2.77% with an average value of 2.50%.

![Figure 5.7: A histogram showing the reproducibility of the resulting mean difference between desired and calculated fluence maps using the rotating leaf motion algorithm. Leaf motions were calculated for the same fluence with the same parameters 100 times. Although the algorithms are stochastic in nature the resulting accuracy is highly reproducible.](image-url)
5.4.3 Radiation Efficiency

Plotted in Figure 5.8 is the result of the leaf motion calculation for the 2D sinusoidally varying fluence map (Figure 5.2(f)) as a function of radiation efficiency. Setting the efficiency parameter to values lower than 50% does not show a significant benefit. As the efficiency is increased beyond the 50% level the discrepancies between desired and calculated fluence maps increase. Although the magnitude of

![Graph showing radiation efficiency effect]

Figure 5.8: Effect of increasing the radiation efficiency. As the efficiency is increased the quality of the calculated fluence map degrades and the mean difference increases. There is no benefit to decreasing the efficiency to values less than 50% in this example.
these effects differ depending on the complexity of the desired fluence map the same general trends are observed in other cases.

5.4.4 Rotation Range

The extra degree of freedom introduced by rotating the collimator will provide varying results depending on the range of collimator rotation that is used. Plotted in Figure 5.9 are the calculation results for the fluence map used in Figure 5.8 and shown in Figure 5.2(d) as a function of the rotation range. The number of segments is

![Figure 5.9: Illustration of the effect of collimator rotation range on the algorithm result. The mean difference decreases rapidly as the rotation range is increased from 90 degrees to 180 degrees. Increasing the range further provides only minor benefit.](image)

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the same in each case. The results presented in Figure 5.9 are typical for the fluence maps tested and a reasonable solution is usually not attainable below 90 degrees of rotation. As the rotation range is increased the mean difference decreases until approximately 180 degrees. Increasing the range further results in only minor improvements. Virtually no benefit is observed when the rotation range is increased beyond 270 degrees. Although spatial resolution considerations indicate that a rotation range of 90 degrees should be sufficient, due to the inability of the MLC to generate concave apertures in the direction opposite to leaf travel the capabilities of the MLC are only exploited when a minimum 180 degree rotation is used.

5.4.5 Number of Segments

When generating fluence maps for delivery by multiple static segments, the accuracy of the fluence will be a function of the total number of segments. Plotted in Figure 5.10 is the calculation result for the sinusoidal fluence shown in Figure 5.2(f) as a function of the total number of static segments. As the number of segments increases, the difference between calculated and desired fluences is reduced. The degree of improvement becomes less significant as the number of segments is increased. The mean difference asymptotically approaches a minimum after which any further increases in the number of segments provides no significant benefit. The same trend is observed in other cases. For lower complexity fluence maps a smaller number of segments (e.g. 10 segments) may generate a result that is considered clinically acceptable.
Figure 5.10: In static mode the algorithm result is a function of the number of segments used for delivery. Accuracy of the algorithm improves rapidly as the number of segments is increased until approximately 40 segments after which adding more segments has only minimal benefit.

5.4.6 Algorithm Results

Plotted in Figure 5.11 is the absolute difference between desired and calculated fluence as a percentage of the maximum for each of the five thyroid fields shown in Figure 5.2. Results of leaf motion calculation for both rotational and conventional delivery methods in the static delivery mode are shown. Results of leaf motion
calculation for both rotational and conventional delivery methods in the dynamic delivery mode are shown in Figure 5.12. For all fields, the 5 mm leaf rotational method shows superior accuracy over the 5 mm leaf conventional method in both static and dynamic modes. For 1 cm leaves the rotational delivery method provides an

Figure 5.11: Displayed is the mean difference between calculated and desired fluence maps for the 5 field thyroid IMRT plan. Results for both the rotational and conventional methods are plotted for the static delivery mode. Results are plotted for the 5 mm leaf MLC as well as the 1 cm leaf MLC.
even more significant improvement in accuracy when compared to conventional delivery methods in both static and dynamic modes.

Thyroid Dynamic Calculation Results

Figure 5.12: Displayed is the mean difference between calculated and desired fluence maps for the 5 field thyroid IMRT plan. Results for both the rotational and conventional methods are plotted for the dynamic delivery mode. Results are plotted for the 5 mm leaf MLC as well as the 1 cm leaf MLC.

Another method of quantifying the conformity between desired and calculated fluence is to evaluate the distance to agreement between the desired fluence and the actual fluence maps [103, 104]. The thyroid fluence maps were assessed based on a
similar criterion. The area of the fluence maps where a difference in fluence of greater than 5% evaluated over a 5 mm diameter region centered on each fluence pixel was determined. This quantifier provides the area of agreement between desired and calculated fluence maps and offers additional information on spatial resolution. For the 5 mm MLC deliveries the area of agreement was greater than 95% in all cases for conventional and rotational delivery methods in both static and dynamic mode. With the 1 cm MLC rotational deliveries the area of agreement was greater than 98% in dynamic mode and was 94% in static mode. The conventional method resulted in an area of agreement of only 88% in both step-and-shoot and sliding window modes with the 1cm leaf width MLC. These results correlate well with the results presented in Figure 5.11 and Figure 5.12.

5.4.7 Dosimetric Characteristics

Shown in Figure 5.13 are measured profiles for the rotational technique using the 5mm leaf MLC. Excellent agreement is observed for the Gaussian fluence shown in Figure 5.13(a) using both the dynamic and static delivery techniques, with all points within 3% or 3mm. Results from the static and dynamic rotational deliveries of the sinusoid intensity map shown in Figure 5.13(b) exhibit good agreement throughout most of the profile except for the third peak where there is a discrepancy of 4% over 3 mm. Results from rotational delivery also agreed well for the wedge profile. The maximum deviations were from 2% to 3% and occurred near the isocenter. Static rotational delivery results showed good agreement but were slightly greater than 3% at the lower end and near the center of the wedge.

Displayed in Figure 5.14 is a plot of the mean difference between desired and measured dose distributions for the sinusoidal test fluence map. This test fluence has a large number of "peaks" and "valleys" that translate into local minima in the leaf motion optimization. It therefore presents a particularly challenging test of the
Figure 5.13: Displayed are measured dose profiles for the rotational technique plotted against the desired calculated dose. Dynamic and static delivery results are shown for the (a) Gaussian and (b) Sinusoid fluence maps. All fluence maps were delivered with the 5mm leaf width MLC.
Results for rotational as well as conventional delivery methods are shown for static and dynamic delivery with both 5mm and 1cm leaf width designs. The 5mm leaf static deliveries show similar results for both conventional and rotational delivery methods and the 1cm leaf delivery shows superior agreement for the rotational method. For dynamic delivery the rotational method shows discrepancies that are larger than those seen with the conventional method in this case.

Figure 5.14: Plotted is the mean difference between measured and desired dose distributions for the 2 dimensional sinusoidal test fluence. Results for rotational and conventional delivery methods in static as well as dynamic mode with both the 5mm Millenium MLC and the 1cm Standard MLC are shown.
5.4.8 Spatial resolution

Improvements in dosimetric spatial resolution with the rotational technique are more significant at larger leaf widths. The constant intensity C-shape distributions generated using the 5mm leaf MLC are shown in Figure 5.15(a) and (b) with good conformity for the (a) rotational and (b) conventional techniques. Displayed in Figure 5.15(c) and Figure 5.15(d) are the resulting constant fluence C-shape distributions for the 1cm leaf MLC. Jagged field edges observed in the conventional technique are almost eliminated when dynamic rotation is used. Improvements in fluence map spatial resolution are most apparent perpendicular to the direction of leaf

Figure 5.15: Dose conformity for the C-shape fluence maps are shown for the 5mm leaf MLC using (a) rotational and (b) sliding window techniques. 1cm leaf MLC conformity results are displayed in (c) and (d).
motion. In Figure 5.16 this effect is demonstrated when delivering a high frequency version of the sinusoidal fluence map using the 1cm leaf MLC. A maximum deviation of 4% with good overall agreement is observed for the rotational technique. The sliding window technique is inherently incapable of generating the fluence accurately and differences of up to 10% can be observed throughout the profile.

Figure 5.16: Dose profiles showing the spatial resolution capabilities of the rotational technique for a high frequency version of the sinusoidal fluence map using the 1cm leaf MLC. Significant error (10%) is observed for the conventional delivery profile obtained perpendicular to the direction of leaf motion, where the limitations of leaf width are most apparent. The rotational technique profile shows only minor discrepancies.
5.4.9 Interleaf effects

Leakage was determined by exposing Kodak X-OMAT V film with 800 MU with the MLC completely blocking a square field defined by the secondary collimator jaws. X-OMAT V film was used in this case due to the higher sensitivity required to accurately measure low doses (< 5 cGy). Displayed in Figure 5.17(a) and (b) are the MLC leakage patterns for the 1cm leaf MLC using the rotational (collimator rotated through 180°) and conventional (collimator stationary) techniques respectively. With the rotational technique the majority of interleaf leakage has been spread out over the field area although some circular rings of higher dose can still be observed. Conventional leaf sequencing methods that optimize delivery efficiency reduce interleaf leakage [24, 105]. Still, non-uniform leakage cannot be compensated for entirely in conventional IMRT delivery and is a significant source of error for some dynamic fields [52, 67, 68].
Improvements in base leakage uniformity with the rotational technique are apparent in relative dose profiles displayed for the 5mm and 1cm leaves in Figure 5.18(a) and Figure 5.18(b) respectively.
Figure 5.18: Relative dose leakage profiles across the leaves and through the center of rotation are shown for the 5mm leaf MLC and 1cm leaf MLC in (a) and (b) respectively. Leakage decreases gradually at the edges of the rotational technique profiles due to the non-uniform contribution from the corners of the square aperture defined by the collimator jaws.
Tongue-and-groove effects are also minimized by the rotational technique as observed in a magnified section of the wedge profile shown in Figure 5.19. Tongue-and-groove underdosing errors as high as 29% were observed in a previous study [106].

Figure 5.19: An example of the tongue-and-groove effect observed in the conventional technique delivery of the wedge shaped fluence distribution. Tongue-and-groove effects are not present when the same fluence is delivered with the rotational technique.

5.4.10 Maximum field size

The lengths of the MLC leaves for the MLC models in this study are 14.5 cm. Therefore, the difference between the maximum and minimum leaf extension on each
bank may not exceed 14.5 cm. In the conventional sliding window technique each leaf pair must travel from one field edge to the other. The maximum field width that can be accommodated is therefore 14.5 cm. Techniques for combining multiple fields must be used to generate wider fields [75]. With collimator rotation the leaf length does not impose the same limit on maximum field size due to inherently different delivery geometry as described in section 3.2.3. The rotational method does not use a sliding window so individual leaves are not required to span the entire length of the field, allowing for gaps between individual leaves up to 29 cm wide. Furthermore, because the direction of leaf travel changes with respect to the desired fluence map at each collimator angle, only the length of the leaf bank dictates the upper limit of maximum field size. The maximum IMRT field size that may be delivered in one irradiation with the rotational method is therefore a 40 cm and a 29 cm diameter circle for the 120 leaf and 52 leaf MLCs respectively.

5.5 Discussion

Results for the fluence generation of the 5 field thyroid treatment show a consistent improvement in fluence generation capabilities over conventional delivery methods and, for both dynamic and static deliveries, the 1 cm leaf rotational delivery is on average as accurate as the 5 mm leaf conventional method. Advantages of the rotational method over conventional methods are more substantial for the 1 cm leaf than for the 5 mm leaf MLC. These results correlate with the findings of section 2.5 where it was shown that resolution degradation due to dose dispersion (DSK) limits the improvements in spatial resolution of smaller leaf widths. The implication of these results will depend on the threshold considered acceptable by the clinician. Inaccuracy of the prescription over greater than 10% of the field (as seen with the 1 cm conventional but not the 1 cm rotational fluence maps) may be considered
unacceptable in some cases but will also depend on patient motion and set-up reproducibility as described in section 1.6.2.

The histogram plotted in Figure 5.7 shows that the leaf motion calculation algorithms are highly reproducible, with any stochastic effects due to the optimization causing only minor variations in the resulting fluence maps. The plot in Figure 5.8 shows that radiation efficiency affects the resulting fluence map, with lower efficiency improving overall accuracy. The rotation range also affects the resulting fluence map, with rotation ranges of less than 180 degrees causing a significant reduction in accuracy as seen in Figure 5.9. Although spatial resolution considerations indicate that a rotation range of 90 degrees should be sufficient, the aperture forming capabilities of the MLC are only exploited when a minimum 180 degree rotation is used. Finally, it is shown in Figure 5.10 that increasing the total number of segments in static rotational delivery will improve the resulting fluence maps until approximately 80 segments, after which only minimal benefit is observed.

Results of the dosimetric investigation seen in Figure 5.13 show that the rotational algorithms and linac delivery are able to accurately reproduce desired fluences. In a comparison between the rotational method and conventional method shown in Figure 5.14 a similar level of dosimetric accuracy is observed, with the rotational technique having slightly larger discrepancies for dynamic delivery.

The rotational leaf motion calculation algorithms are based on a simplified model of the MLC. Various refinements to conventional dynamic leaf motion calculation have been reported by several investigators that have subsequently been incorporated into conventional leaf motion calculation algorithms. These include compensation for round leaf ends, MLC transmission [19] and extra-focal scattered radiation [52, 107, 108]. The above investigators and others [109] have observed discrepancies reduced to less than 3% once these effects have been taken into
consideration. Modifying these established refinement techniques to account for MLC rotation may produce a similar degree of improvement.

In rotational delivery the stability of the collimator can affect the accuracy of fluence maps. The axis of rotation of the collimation system is highly reproducible to within 1 mm of the isocenter as shown in section 3.3.4 and is therefore not a significant source of error. When fields are delivered dynamically, the speed of the collimator and the dose rate must be synchronized or there will be a collimator phase shift error. In section 3.4.2.1 it was found that the collimator speed is highly reproducible. The time for a 180° rotation changed by less than 1% over a period of six months. At the beginning of delivery there is a period of approximately 1s where the dose rate is increasing. During this time the collimator is accelerating. Errors due to a lack of synchronization were detected in an investigation of collimator rotation stability described in section 3.4.2.3, which would account for some of the delivery errors encountered in dynamic mode.

Improvements in spatial resolution over conventional delivery methods are seen in Figure 5.15 and Figure 5.16. Techniques to improve the spatial resolution of intensity modulated fields have been reported by others. One method consists of indexing the couch position perpendicular to the direction of leaf motion between multiple intensity modulated fields [72]. Another uses 2 orthogonal (collimator rotated by 90°) fields [61, 71]. The principle of moving MLC leaves outside the direction of leaf travel is used in these techniques as well as the rotational technique and similar improvements in spatial resolution over conventional methods are observed. In the orthogonal delivery technique described by Siochi as well as Evans and Partridge, the high resolution fluence maps must go through a filtering process in order to separate them into two orthogonal deliveries. The amount of degradation that results from this filtering may, in some cases, cause an unacceptable modification of the desired fluence. Also, the junctioning of orthogonal leaves can cause effects
similar to interleaf leakage and tongue-and-groove effects. The main disadvantage of other techniques is that they use multiple intensity modulated fields and therefore generally require a larger total number of MU. Finally, although further investigation is required before the relative benefits of these techniques can be properly assessed, the fact that the rotational technique is not limited to two fixed collimator angles inherently offers more degrees of freedom for generating a desired fluence map.

The amplitudes of the interleaf leakage spikes with the rotational technique are a small fraction of the interleaf leakage spikes observed in the conventional technique as seen in Figure 5.18. Although the actual leakage contribution in a clinical case will depend on the characteristics of the desired fluence map, the results show that the rotational method inherently provides a more uniform base leakage than conventional delivery methods. Also, no discrepancies due to the tongue-and-groove MLC leaf design have been observed with the rotational technique. The current dosimetric accuracy is not sensitive enough to discern any errors resulting from this effect. Finally, in conventional delivery there is no way to fully compensate for interleaf effects because the leaf edge positions are fixed and oriented parallel to the direction of leaf motion. This is not the case when rotation is used and it should therefore be possible to compensate for them in the rotational leaf motion calculation.

It was shown that the rotational method allows an increase in maximum field size from a 14.5 cm wide field to a 29 cm and 40 cm diameter circle for the 52 leaf and 120 leaf MLCs respectively. Theoretically, increasing the field size to these upper limits could cause a reduction in the spatial resolution advantages of the rotational technique (field sizes greater that 29 cm diameter), particularly when high spatial resolution is required at the periphery of the field. Such effects have not yet been observed and, in any event, clinical IMRT field sizes exceeding 29 cm diameter are rare.
Chapter 6

CONCLUSION

6.1 Conclusion

Currently available fluence generation techniques suffer from limitations in spatial resolution and dosimetric accuracy imposed by the multileaf collimator (MLC). In this thesis a novel method of controlling the MLC was developed that is capable of generating fluence maps with higher spatial resolution and less systematic error than conventional methods. In the first part of the thesis an investigation into the factors that reduce spatial resolution is presented. Using a linear systems theory model of the dose delivery process, a new technique was developed to evaluate the factors that limit a radiation therapy device from delivering an ideal dose distribution. The model allows the effect of the MLC and Dose Spread Kernel (DSK) to be evaluated separately, providing a greater understanding of the individual processes that modify the optimal dose distribution. Fourier analysis was used to provide insight into spatial resolution limitations at each step in the formulation. Effects of varying leaf width and DSK for treatment shapes of varying complexity were investigated. The model provided an accurate prediction of the dependence of geometric conformity on MLC leaf width and DSK size. Also, smaller more complex PTV shapes were shown to have a larger high spatial frequency component making them more sensitive to leaf width and DSK. The DTF can be used as a tool for comparing different dose delivery devices. Frequency analysis of the PTV can aid in deciding on resolution capabilities required to treat a site with a known complexity. By choosing the appropriate
apparatus and technique for a given treatment, the resources available to a clinician will be used more efficiently and greater overall patient care can be achieved in the clinic.

The model was also applied to IMRT delivery and similar results were observed. Through this study a method of improving the spatial resolution of conventional IMRT delivery methods was identified. One-dimensional sampling of the MLC causes a degradation of the deliverable distribution in the direction perpendicular to leaf motion. It was hypothesized that an improvement in spatial resolution could be obtained by modifying the sampling geometry for each sub-field. In particular, by rotating the collimator between each sub-field it should be possible to obtain high resolution in the entire plane of fluence delivery. The mechanical and radiation producing capabilities of conventional linacs under collimator rotation conditions were evaluated. Results showed that accurate delivery is feasible with the current control mechanism although synchronization errors between collimator rotation speed and dose rate were identified in dynamic mode.

Based on these findings, a new technique for the delivery of IMRT was developed that, by including collimator rotation exploits all degrees of freedom of the collimation system. A set of novel algorithms was developed to calculate the leaf positions at each collimator angle in the delivery. The algorithms were presented and analyzed in terms of their dependence on radiation efficiency, the range of collimator rotation and number of segments. A thyroid test case was used to evaluate the fluence generation capabilities of the algorithms. Advantages over conventional deliveries in generating high-resolution fluence maps were observed for 5mm and 1cm MLC leaf widths in static and dynamic delivery modes.

The potential dosimetric advantages of incorporating rotation were also investigated. Results showed that higher spatial resolution dose distributions are attainable with the rotational technique, allowing for superior target coverage and
healthy tissue sparing. It was also shown that interleaf leakage and tongue-and-groove effects are substantially reduced, decreasing the degree of systematic overdosing and underdosing observed in conventional IMRT delivery. Finally, the rotational technique removes restrictions on field size, allowing larger individual IMRT fields to be delivered.

Dosimetric results show that the rotation technique provides satisfactory agreement between measured and desired dose distributions. Advantages over conventional methods are most apparent with 1 cm MLC leaves in the static delivery mode. Currently, the accuracy of rotational dynamic delivery is limited by the control mechanism of the linac. Improved linac control may provide improvements in overall accuracy and reproducibility, permitting advantages in spatial resolution, leaf edge effects and field size capabilities to be clinically realized in dynamic delivery.

In summary, the rotational method is capable of generating fluence maps that have higher spatial resolution and less systematic error than those generated by conventional delivery methods. Dose distributions may be delivered that conform more closely to the target volume [38, 39], reducing the dose received by surrounding healthy tissue and decreasing the probability of negative side effects. Also, the rotational method provides physicians with the option of increasing the prescribed tumour dose and improving the probability of tumour control while maintaining the same level of healthy tissue damage.

6.2 Future Work

Changes in dosimetric accuracy due to increasing or decreasing the efficiency parameter are currently under investigation. Preliminary results show that if the efficiency is set too high there is a loss of dosimetric accuracy. Conversely, by setting it too low the sub-field MLC apertures become smaller, making leaf end and
transmission effects more important. Although further investigation is required, preliminary results are encouraging in that for more complex fluence maps the rotational technique attains a superior level of efficiency over conventional methods.

During the optimization there is no modification of the individual sub-field weights. In dynamic delivery field weight modification is analogous to dose rate modification. By optimizing the field weights it may be possible to arrive at solutions that further improve the accuracy of the delivered fluences. Also, improvements in dosimetric accuracy may be attainable by refining the physical model of the MLC to include tongue-and-groove, interleaf leakage, non-uniform collimator scatter and rounded MLC leaf ends.

The MLC and linac control software is not designed for combined automatic collimator rotation and leaf sequencing. Currently, the delivery time is only indicative of how fast the particular operator performing the delivery is able to access various linac functions and override software interlocks. It is therefore not possible to assess the delivery time with any degree of accuracy until full automatic control is implemented. Also, the possibility of increasing the maximum angular speed of the collimator is being investigated. These enhancements will allow for a more realistic evaluation of clinical delivery times.

Although Varian linacs are used to deliver rotational IMRT fields in this study, other linacs and MLCs could also be used. Modifications to the rotational leaf motion calculation that incorporate Elekta MLC leaf constraints and allow rotational delivery on Elekta linacs are currently under investigation.

Finally, testing of multiple 3-dimensional dose distributions with simulated patient motion and set-up reproducibility will provide a more extensive clinical evaluation.
Bibliography


77. Pasquino, M., V. Casanova Borca, and S. Tofani, [Physical-dosimetric characterization of a multi-leaf collimator system for clinical implementation


Appendix A
Circular PTV Conformity Derivation

N = # of MLC leaves in one quadrant = \( \text{round} \left( \frac{r}{w} \right) \)

In cartesian coordinates the leaf centers intersect the circle at:

\[
y = \left( k + \frac{1}{2} \right)w
\]

\[
x = \left( r^2 - \left( k + \frac{1}{2} \right)^2 w^2 \right)^{\frac{1}{2}}
\]

\[
k = \{0,1,2,\ldots,N\}
\]

The edges of the leaves intersect the circle points shown by a, b, c, e on the diagram which have values x given by:
\[ x_a = \left( r^2 - k w^2 \right)^{\frac{1}{2}} \]
\[ x_b = \left( r^2 - \left( k + \frac{1}{2} \right)^2 w^2 \right)^{\frac{1}{2}} \]
\[ x_c = \left( r^2 - (k + 1)^2 w^2 \right)^{\frac{1}{2}} \]
\[ x_e = 0 \]

Area \( A_k \) is calculated in the following integral:

\[
\text{Area } A_k = \int_b^a \left( r^2 - x^2 \right)^{\frac{1}{2}} - kw \, dx
\]
\[ = \frac{x_a}{2} \left( r^2 - x^2 \right)^{\frac{1}{2}} + \frac{r^2}{2} \sin^{-1} \left( \frac{x_a}{r} \right) - \frac{x_a}{2} \left( r^2 - x^2 \right)^{\frac{1}{2}} + \frac{r^2}{2} \sin^{-1} \left( \frac{x_b}{r} \right) - kw(x_e - x_b) \]

Area \( B_k \) is calculated in the following integral:

\[
\text{Area } B_k = \int_c^b \left( k + 1 \right) w - \left( r^2 - x^2 \right)^{\frac{1}{2}} \, dx
\]
\[ = kw(x_b - x_e) - \frac{x_b}{2} \left( r^2 - x_b^2 \right)^{\frac{1}{2}} + \frac{r^2}{2} \sin^{-1} \left( \frac{x_b}{r} \right) + \frac{x_c}{2} \left( r^2 - x_c^2 \right)^{\frac{1}{2}} + \frac{r^2}{2} \sin^{-1} \left( \frac{x_e}{r} \right) \]

The Remainder Area is calculated using the following integral:

\[ k = N = \text{round} \left( \frac{r}{w} \right) \]
if the remainder of $\frac{r}{w} < 0.5$ then:

$$\text{Remainder Area} = \int_e^a \left( (r^2 - x^2)^{1/2} - Nw \right) dx$$

$$= \frac{x_a}{2} \left( r^2 - x_a^2 \right)^{1/2} + \frac{r^2}{2} \sin^{-1} \left( \frac{x_a}{r} \right) - Nwx_a$$

if the remainder of $\frac{r}{w} > 0.5$ then:

$$\text{Remainder Area} = \int_e^b \left( (N+1)w - (r^2 - x^2)^{1/2} \right) dx$$

$$= (N+1)wx_b - \frac{x_b}{2} \left( r^2 - x_b^2 \right)^{1/2} - \frac{r^2}{2} \sin^{-1} \left( \frac{x_b}{r} \right)$$

The % conformity is defined as:

$$\%\text{conformity} = 100 \times \frac{\text{Area of PTV} - (\text{Area of PTV underdose} + \text{Area of healthy tissue dose})}{\text{Area of PTV}}$$

$$= 100 \times \frac{\pi r^2 - 4 \left( \sum_{k=0}^N \right) \left( \text{Area A}_k + \text{Area B}_k \right) + \text{Remainder Area}}{\pi r^2}$$
Appendix B
Rotational Leaf Motion Calculation Engine

Figure B.1: Software was developed to derive the rotating leaf motions. A graphical user interface facilitated the calculation, analysis and verification of different fluence maps. Functions of the software include: Reading in desired fluence maps, selecting fixed parameters (MLC type, dynamic vs. static delivery, radiation efficiency, initialization method etc...), displaying calculated fluence maps, evaluation of the calculated fluence maps and supplying MLC file output for delivery.