DOSE VERIFICATION OF ROTATING COLLIMATOR
INTENSITY MODULATED RADIATION THERAPY

by

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Abstract

A novel method of delivering Intensity Modulated Radiation Therapy (IMRT) has been proposed in which the entire multileaf collimator (MLC) is rotated between each segment. Current linear accelerators (linacs) are not designed for IMRT delivery with collimator rotation therefore extensive quality assurance testing must be done before this technique can be used clinically. This thesis has two main objectives. The first is to evaluate whether IMRT plans incorporating collimator rotation can be delivered on a standard radiotherapy linac. The second is to evaluate whether the dose distributions produced with rotating MLC (RMLC) IMRT are comparable to IMRT plans conventionally delivered, without collimator rotation.

First, the Dynamic Beam Delivery (DBD) toolbox, the interface required to independently control collimator rotation, was tested. It was verified that the DBD toolbox could rotate the collimator accurately to its intended angle within 0.5°. Also, fluence distributions of varying complexity were generated using the in-house RMLC algorithm and validated using film-based verification methods. Dosimetric analysis provided information regarding the MLC modeling and resolution limitations of the algorithm. Clinical RMLC IMRT treatment plans were generated and delivered and, after incorporating a 1.4 mm leaf-gap requirement, good agreement between the measured and calculated distributions for single fields were obtained. These test results indicated that RMLC IMRT is viable on radiotherapy linacs.

To evaluate whether treatment plans optimized for RMLC IMRT delivery were comparable to plans optimized for conventional IMRT delivery, composite multi-field dose distributions delivered to a water-equivalent phantom were analyzed. Prostate, nasopharynx and c-shape structures were used as the target volumes and the treat-
ment sites were optimized for RMLC, conventional, and rotating aperture optimization (RAO) IMRT. The measured and calculated (Varian CadPlan treatment planning system) dose distributions were compared using orthogonal profiles and a gamma factor analysis. Results indicate that RMLC and conventional IMRT plans are comparable when using a 5.0 mm width beamlet (versus the 2.5 mm width beamlet) during the optimization. In addition, it was found that the RAO IMRT plans, which also incorporated collimator rotation, yielded the best results.
# Contents

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Abstract</td>
<td>ii</td>
</tr>
<tr>
<td>Contents</td>
<td>iv</td>
</tr>
<tr>
<td>List of Tables</td>
<td>vii</td>
</tr>
<tr>
<td>List of Figures</td>
<td>viii</td>
</tr>
<tr>
<td>Acknowledgements</td>
<td>xi</td>
</tr>
<tr>
<td>1 Introduction</td>
<td>1</td>
</tr>
<tr>
<td>1.1 Radiation Therapy Overview</td>
<td>1</td>
</tr>
<tr>
<td>1.1.1 Historical Background</td>
<td>1</td>
</tr>
<tr>
<td>1.1.2 Tumour and Healthy Tissue Response to Dose</td>
<td>2</td>
</tr>
<tr>
<td>1.2 Basic Concepts</td>
<td>3</td>
</tr>
<tr>
<td>1.2.1 Linear Accelerator Design</td>
<td>3</td>
</tr>
<tr>
<td>1.2.2 Dose Deposition and Photon Interactions in Tissue</td>
<td>8</td>
</tr>
<tr>
<td>1.2.3 Fluence and Absorbed Dose</td>
<td>10</td>
</tr>
<tr>
<td>1.2.4 Modeling Dose Deposition</td>
<td>11</td>
</tr>
<tr>
<td>1.3 Radiation Therapy Delivery</td>
<td>12</td>
</tr>
<tr>
<td>1.3.1 Treatment Planning</td>
<td>12</td>
</tr>
<tr>
<td>1.3.2 Conformal Radiation Therapy</td>
<td>13</td>
</tr>
<tr>
<td>1.3.3 Intensity Modulated Radiation Therapy</td>
<td>15</td>
</tr>
<tr>
<td>1.4 Dosimeters Used For IMRT</td>
<td>17</td>
</tr>
</tbody>
</table>
### Contents

1.4.1 Point Dose Detectors ........................................ 17  
1.4.2 Two Dimensional Dose Detectors .......................... 21  
1.4.3 Three dimensional Dose Detectors .......................... 25  
1.5 Thesis Objectives .............................................. 25  

2 Intensity Modulated Radiation Therapy ....................... 27  
2.1 Conventional IMRT Overview .................................. 27  
2.2 Treatment Plan Optimization ................................ 28  
2.2.1 Forward Treatment Planning ............................... 28  
2.2.2 Inverse Treatment Planning ............................... 29  
2.3 Leaf Sequencing .............................................. 32  
2.3.1 Dynamic (Sliding Window) IMRT ......................... 33  
2.3.2 Static (Step-and-Shoot) IMRT ........................... 34  
2.4 Delivery .................................................... 35  

3 Rotating Multileaf Collimator IMRT ............................ 36  
3.1 Advantages of Collimator Rotation .......................... 36  
3.2 Rotating Leaf Motion Derivation ............................ 38  
3.2.1 Rotating Leaf Motion Optimization ....................... 39  
3.2.2 Rotated Leaf Motion Parameters ......................... 40  
3.3 IMRT Delivery With a Rotating MLC ......................... 40  

4 Testing of the Dynamic Beam Delivery Interface ............. 41  
4.1 Collimator Angle Accuracy .................................. 41  
4.2 Monitor Unit Accuracy ...................................... 43  
4.3 Conclusion .................................................. 47  

5 Fluence Modeling and Dose Calculation ....................... 48  
5.1 Fluence and Dose Calculation ............................... 48  
5.2 Modeling of Non-Rotating Fields ............................ 50  
5.2.1 Square Aperture (Open Field) ........................... 50  
5.2.2 Bar Pattern .............................................. 50  
5.3 Modeling of Rotated Fields ................................ 54  
5.3.1 Sub-sampling ............................................ 54  
5.3.2 Rotating Square ......................................... 55  
5.3.3 Rotating IMRT Results of Single Fields ................. 56
## Contents

5.4 Dosimetric Leaf Gap ................................................. 57
   5.4.1 Determining the Leaf Gap ................................. 60

6 Dosimetric Treatment Verification Using RMLC IMRT .......... 68
   6.1 Generating Deliverable IMRT Plans with RMLC ............. 68
      6.1.1 Treatment Plan Generation ............................ 68
      6.1.2 Treatment Plan Delivery .............................. 70
   6.2 AVID IMRT Verification Phantom ............................ 70
      6.2.1 Phantom Measurements ................................. 70
      6.2.2 Dose to Phantom: TPS Calculation .................. 71
   6.3 Film Calibration Technique ................................. 72
   6.4 Dosimetric Verification Methods ............................ 73
      6.4.1 Orthogonal Profiles .................................. 73
      6.4.2 Gamma Factor Analysis ................................. 73
   6.5 Treatment Plans and Results ................................. 76
      6.5.1 Prostate Carcinoma ................................... 77
      6.5.2 Nasopharynx Carcinoma ............................... 92
      6.5.3 Geometric C-shape ..................................... 105

7 Conclusion .................................................................... 108
   7.1 Summary and Conclusion .................................... 108
   7.2 Future Work .................................................... 110

Bibliography ............................................................... 112
List of Tables

4.1 Statistical results for the MU accuracy test. 47
5.1 Summary of leaf gap test results. 62
5.2 Point dose measurements within a prostate carcinoma plan. 63
5.3 The average target doses for the prostate carcinoma distributions. 67
6.1 Summary of the prostate carcinoma measurements. 92
6.2 Ion chamber measurements for the prostate carcinoma plans. 93
6.3 Summary of the nasopharynx carcinoma measurements. 104
6.4 Ion chamber measurements for the nasopharynx carcinoma plans. 104
6.5 Summary of the geometric c-shape measurements. 106
List of Figures

1.1 Classical and actual dose-response curves. ........................................ 3
1.2 The internal and external components of a linear accelerator. ............ 5
1.3 The relative importance of three major types of photon interactions in
tissue, as a function of photon energy. .................................................. 9
1.4 The photoelectric, Compton and pair production interactions. .......... 10
1.5 The dose spread kernel, convolution of dose kernels and the dose profile. 12
1.6 The percent depth-dose curve for a 6 MV photon beam. ................. 14
1.7 Relative dose profiles for one and two sets of parallel-opposed field pairs. 15
1.8 A photograph of an MLC assembly. .................................................... 16
1.9 A typical thimble ionization chamber. ............................................... 17
1.10 A typical silicon p-n junction diode. ............................................... 18
1.11 The TLD cycle and glow curve. ....................................................... 19
1.12 The cross-section of double emulsion film. ...................................... 21
1.13 The latent image formation process for radiographic film. ............... 22

2.1 A flow chart of the IMRT treatment planning process. ...................... 29
2.2 The ideal and actual DVHs for a target and OAR. .............................. 31
2.3 The fluence and aperture generation for dynamic MLC delivery. ........ 33
2.4 The fluence and aperture generation for static MLC delivery. ............ 34

3.1 The tongue-and-groove effect. ......................................................... 37
3.2 Interleaf effects for rotated and conventional IMRT. ......................... 38
3.3 The inherent resolution in conventional and rotational IMRT. ............ 39
List of Figures

<table>
<thead>
<tr>
<th>Number</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.1</td>
<td>The collimator-angle accuracy test pattern.</td>
<td>42</td>
</tr>
<tr>
<td>4.2</td>
<td>Results from the collimator-angle accuracy test.</td>
<td>43</td>
</tr>
<tr>
<td>4.3</td>
<td>Collimator-angle reproducibility histograms.</td>
<td>44</td>
</tr>
<tr>
<td>4.4</td>
<td>Results for the monitor unit output test.</td>
<td>46</td>
</tr>
<tr>
<td>5.1</td>
<td>The implications of using large and narrow dose spread kernels.</td>
<td>49</td>
</tr>
<tr>
<td>5.2</td>
<td>Dosimetric results for an open field pattern.</td>
<td>51</td>
</tr>
<tr>
<td>5.3</td>
<td>Bar pattern results with the MLC rotated 90°.</td>
<td>52</td>
</tr>
<tr>
<td>5.4</td>
<td>Bar pattern results with the MLC rotated 180°.</td>
<td>53</td>
</tr>
<tr>
<td>5.5</td>
<td>The conventional and sub-sampled rotational beamlets.</td>
<td>54</td>
</tr>
<tr>
<td>5.6</td>
<td>The rotating square pattern.</td>
<td>55</td>
</tr>
<tr>
<td>5.7</td>
<td>Dosimetric results for the rotating square pattern.</td>
<td>56</td>
</tr>
<tr>
<td>5.8</td>
<td>Single-field measurements for a five-field prostate carcinoma treatment.</td>
<td>58</td>
</tr>
<tr>
<td>5.9</td>
<td>The penumbral effects due to rounded MLC leaf tips.</td>
<td>59</td>
</tr>
<tr>
<td>5.10</td>
<td>Determining the leaf gap using the match-line method.</td>
<td>60</td>
</tr>
<tr>
<td>5.11</td>
<td>The leaf gap test results.</td>
<td>61</td>
</tr>
<tr>
<td>5.12</td>
<td>Prostate carcinoma, field 3: RMLC IMRT using a 0.0 mm leaf gap.</td>
<td>64</td>
</tr>
<tr>
<td>5.13</td>
<td>Prostate carcinoma, field 3: RMLC IMRT using a 1.4 mm leaf gap.</td>
<td>65</td>
</tr>
<tr>
<td>5.14</td>
<td>Prostate carcinoma, field 3: RMLC IMRT using a 2.0 mm leaf gap.</td>
<td>66</td>
</tr>
<tr>
<td>6.1</td>
<td>The AVID IMRT verification phantom.</td>
<td>71</td>
</tr>
<tr>
<td>6.2</td>
<td>Percent depth dose and calibration curves.</td>
<td>72</td>
</tr>
<tr>
<td>6.3</td>
<td>The 1D and 2D representations of the factor criteria.</td>
<td>75</td>
</tr>
<tr>
<td>6.4</td>
<td>Prostate carcinoma dose distributions in the patient and phantom.</td>
<td>77</td>
</tr>
<tr>
<td>6.5</td>
<td>Results for the prostate carcinoma plan delivered with RMLC IMRT (2.5 mm width beamlet) for target PTV 1.</td>
<td>78</td>
</tr>
<tr>
<td>6.6</td>
<td>Results for the prostate carcinoma plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 1.</td>
<td>80</td>
</tr>
<tr>
<td>6.7</td>
<td>Typical dose profiles resulting from 5.0 mm and 2.5 mm width beamlets.</td>
<td>82</td>
</tr>
<tr>
<td>6.8</td>
<td>The beamlet sizes used in conventional and RMLC IMRT optimizations.</td>
<td>83</td>
</tr>
<tr>
<td>6.9</td>
<td>Results for the prostate carcinoma plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 1.</td>
<td>85</td>
</tr>
<tr>
<td>6.10</td>
<td>Results for the prostate carcinoma plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 2.</td>
<td>87</td>
</tr>
<tr>
<td>6.11</td>
<td>Results for the prostate carcinoma plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 2.</td>
<td>89</td>
</tr>
</tbody>
</table>
List of Figures

6.12 Results for the prostate carcinoma plan delivered with RAO IMRT (2.5 mm width beamlet) for target PTV 1 .......................... 91
6.13 Nasopharynx carcinoma dose distributions in the patient and phantom. .......................... 93
6.14 Results for the nasopharynx plan delivered with RMLC IMRT (2.5 mm width beamlet) for target PTV 1 .................................. 94
6.15 Results for the nasopharynx plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 1 .................................. 96
6.16 Results for the nasopharynx plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 1 .................................. 98
6.17 Results for the nasopharynx plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 2 .................................. 100
6.18 Results for the nasopharynx plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 2 .................................. 101
6.19 Results for the nasopharynx plan delivered with RAO IMRT (2.5 mm width beamlet) for target PTV 1 .................................. 103
6.20 A contoured complex geometric c-shape target and the resulting dose distribution .................................. 105
6.21 Results for the geometric c-shape plan delivered with RAO IMRT (2.5 mm width beamlet). .................................. 107
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CHAPTER 1

Introduction

1.1 Radiation Therapy Overview

1.1.1 Historical Background

Following the discovery of x-rays in 1895 by Wilhelm Röentgen and the development of the medicinal use of radioactivity in 1896 by Henri Becquerel and Marie Curie, a new branch of medicine and physics research began. It was soon realized that the penetrating and non-evasive characteristics of x-rays could be exploited to acquire images of internal structures within the body and that higher-energy x-rays could be used to treat malignant tumours. The first patient cured by radiation therapy was reported in 1899 [1, 2].

In its infancy, radiation therapy was used to treat superficial malignancies, but the appearance of normal-tissue complications and cancer recurrences led to the development of a treatment technique known as brachytherapy. With this technique, radium needles were inserted directly into the diseased tissue to irradiate the malignancy. In 1913, Coolidge developed the first x-ray tube and by 1922 the tubes were able to generate a photon spectrum with a maximum energy of 200 keV, making it easier to treat deep-seated lesions. In the 1940’s higher-energy particle accelerators were introduced into megavoltage x-ray treatments and in 1951 the first patient in the world was treated, in Canada, with $^{60}$Co gamma-rays. With a high specific activity, a half-life of 5.26 years and an average photon energy of 1.25 MeV, $^{60}$Co became the most popular method of radiation treatment and is still found in many cancer centres worldwide [1].
Chapter 1. Introduction

The most important development in modern radiation therapy has been the introduction of linear accelerators in the 1960's. The linear accelerator (linac) is the most common tool used to generate megavoltage photon and electron beams for cancer treatment.

1.1.2 Tumour and Healthy Tissue Response to Dose

The goal of radiation therapy is to irradiate and kill cancerous cells whilst sparing surrounding healthy tissue. A balance must exist between cell death and cell survival as it is unfeasible to irradiate cancerous tissue only. This balance depends on the response of tumours and healthy tissues to dose.

The response of cells and tissue to radiation is complex. Different cells within different tissues may not respond to radiation in the same way and malignant cells will respond differently than healthy cells within an organ. Although cancerous tissue is the target, the radiation beam will pass through normal healthy tissue before and after reaching the desired destination. Figure 1.1(a) shows, in its simplest form, the classical response of tumours and healthy tissue to dose [1, 3]. At low doses very little tumour response is seen and significant tumour destruction occurs upon delivering a specific high-dose threshold. Beyond that dose level no further response is seen in the tissue. The same sigmoidal dose-response is achieved in normal tissue but the damaging effects do not occur until higher doses are delivered, implying that normal tissue is less sensitive to radiation. A treatment dose is prescribed such that the maximal amount of tumour will be killed with minimal adverse effects to normal tissue.

Actual dose-response curves generated from limited clinical data demonstrate that classical dose-response curves are misleading; the curves shown in Figure 1.1(b) depict a more accurate representation of the effects of radiation on malignant and healthy tissues [1]. The primary difference between the two models is that normal tissue dose-response is less sensitive than for healthy tissue. Considering the gradients of the two curves, the normal tissue curve is much steeper than the tumour curve, therefore normal tissue is more sensitive to lower doses than the tumour. The shallower slope of the tumour curve is due to the heterogeneity of cancerous cells throughout the tumour volume. Also, in some cases, the tumour control curve never reaches 100% response. It has been argued that this may arise from the metastatic or microscopic spread of disease beyond the tumour site [1].

From Figure 1.1(b) it is apparent that the dose to healthy tissue should be kept to
a minimum; this can be achieved through sophisticated treatment-planning and beam arrangements. The dose-response curve of healthy tissue is steep and a small increase in dose can increase the unwanted destruction of healthy tissue substantially. For this reason, the need for improved delivery efficiency and increased spatial resolution and accuracy in the generation of treatment plans is crucial.

1.2 Basic Concepts

1.2.1 Linear Accelerator Design

The linac was first introduced in the 1960’s and quickly became the most common tool used to generate megavoltage photon and electron beams for cancer treatment. The linac accelerates electrons to high kinetic energies ranging from 4 to 25 MeV using microwave radio-frequency (RF) fields in the range of 10^3 MHz (L-band) to 10^4 MHz (X-band); most linacs run at 2856 MHz (S-band). Electrons can be scattered through a thin foil to produce useful clinical beams or can pass through a tungsten target to produce bremsstrahlung photons [4].

Externally, a linac is composed of a gantry, gantry stand, patient couch, modulator cabinet and a control console which is located outside the linac vault. The internal components of the linac include the injection system, RF power generation system, accelerating waveguide, auxiliary system, beam transport system and beam collimation
and monitoring system. Schematic diagrams of the internal and external components of the linac are shown in Figures 1.2(a) and 1.2(b), respectively. A brief description of these components will be provided for completeness.

Injection System

An electron gun provides the source of electrons for the accelerating waveguide in pulses of 1 to 80 keV. Electrons are emitted thermionically from a heated cathode and focussed into a pencil-shaped beam by a curved focusing electrode. The electron beam is accelerated towards a perforated anode and then drifts toward the accelerating waveguide [4].

RF Power Generation System

The RF power generation system is composed of a power source and a pulsed modulator. The modulator emits high-voltage pulses at rates between 30 and 300 Hz. These short-duration pulses (~ 1μs), required by both the RF power source (magnetron or klystron) and the electron gun, are synchronized such that the gun emits electrons at the same time as the microwaves reach the accelerating waveguide.

A magnetron is used primarily in low energy (4 – 8 MeV) linacs as a source of high power RF required for electron acceleration. It is a self-contained system in that it is both a source and an amplifier of RF waves. A klystron is a RF power amplifier only and thus requires a low-power RF driver as input [4].

Accelerating Waveguide

The waveguide consists of a series of hollow or gas-filled metallic pipes that transmit microwaves from the magnetron or klystron to the accelerating structure. The accelerating structure is an evacuated copper tube that transfers energy from the high-power RF fields to the moving electrons. Two types of accelerating waveguides are used to accelerate electrons: a traveling-wave structure and a standing-wave structure.

In the traveling-wave design, the accelerating cavity is divided into sections by a series of unevenly-spaced apertures which control the phase velocity of the wave. The apertures are spaced closely at the start of the accelerating waveguide, near the electron gun, gradually becoming more widely spaced. The spacings are designed so that the electrons entering the cavity are bunched together into groups and accelerated by the electric field crest of the traveling wave. The velocities of the electric field and electron
Figure 1.2: (a) A schematic diagram of the internal components of a linac. (b) A schematic diagram of the external view of a linac, with the path of rotation for the gantry and couch outlined.
Chapter 1. Introduction

bunches are matched so that the electrons stay on the crest of the wave as it moves down the tube. The microwaves entering the waveguide propagate toward its end where they can be either absorbed without reflection, exit the waveguide and be absorbed in a resistive load or fed back to the input end of the waveguide. To obtain energies of more than ~8 MeV, the accelerating waveguide becomes impractically long (>1m) [4, 5].

In the standing-wave design, the microwaves are reflected at both ends of the accelerating waveguide, facilitating wave travel in both directions. The waveguide tube is sectioned into equally-spaced apertures; in this configuration every second cavity has no electric field and so there is no energy gain for the electrons. These cavities serve only as coupling cavities and can be moved out to the ends of the waveguide structure, effectively shortening the waveguide by 50% [4]. In this design a circulator is placed between the RF generator and the accelerating waveguide. The circulator transmits RF power from the RF generator to the waveguide but is immune to reflected waves, thereby protecting the power source from reflections [4].

Auxiliary System

The auxiliary-system components are not involved directly with beam production but are required to keep the linac functioning. This system is comprised of four separate systems: (1) a vacuum pumping system to maintain vacuum pressure in the accelerating waveguide and RF generator, (2) a water cooling system for the accelerating waveguide, target, circulator and RF generator, (3) an optional air pressure system for pneumatic movement of the target and beam shaping components and (4) shielding against radiation leakage [4].

Electron Beam Transport

The electron beam transport system comprises steering coils, focusing coils and bending magnets. The steering coils keep the electron beam as close to the central axis of the accelerating waveguide as possible and steer the beam directly onto the target (straight-through linacs) or toward the opening which connects the accelerating waveguide with the beam transport system. The role of the focusing solenoid coils is to minimize beam divergence and cross-section. For linacs operating at energies above 6 MeV, the accelerating waveguide becomes too long for straight-through mounting and bending magnets are required to transport the electron beam from the accelerating waveguide to the target. Bending magnets can steer the beams through 90°, 270° and 112.5° [4].
Linac Treatment Head

The linac treatment-head houses components that affect the production, shaping, localization and monitoring of clinical photon and electron beams.

Once electrons from the electron gun have been accelerated through the waveguide to the appropriate kinetic energy and brought through the beam transport system to the treatment head, the narrow pencil beam either passes through an x-ray target/flattening filter combination to produce bremsstrahlung photons or passes through a scattering foil to produce a clinically-useable electron beam. Each clinical photon beam has its own target/flattening filter combination. The flattening filters and scattering foil are mounted on a carousel which can be rotated in and out of the beam, depending on the treatment beam to be used.

Beam Collimation

The primary collimator, located just above the flattening filter, is composed of a tungsten shielding block and defines the largest available circular field size. The shielding block is thick enough to attenuate the average primary x-ray beam-intensity to less than 0.1% of its initial value [4].

The secondary collimators consist of four blocks, two forming the upper and two forming the lower jaws of the collimator. The jaws are moveable and define rectangular and square fields up to a maximum dimension of $40 \times 40 \text{ cm}^2$ at the linac isocentre. The collimators can also rotate about their axis, a process known as collimator rotation, to provide an additional degree of freedom [4]. The primary and secondary collimators are constructed using high density, high atomic number materials such as lead, tungsten and tungsten-copper alloy, to maximize the attenuation per given thickness of material [4].

Tertiary collimators, located below the secondary jaws, constitute the final collimating structures before the beam enters the patient. The purpose of tertiary collimators is to tailor the radiation beam to the shape of the target volume. Blocks, wedges and multileaf collimators (MLC) are examples of tertiary collimators. Blocks are custom-designed for each patient and are made out of a highly-attenuating alloy. These devices must be attached to the linac prior to a treatment. The MLC is a permanently installed device consisting of a number of individually-moving leaves (some models have up to 120 leaves or 60 leaf pairs) covering fields up to $40 \times 40 \text{ cm}^2$ in size. Each leaf is controlled by its own motor enabling arbitrarily-shaped fields to be produced.
Chapter 1. Introduction

Dose Monitoring System

The treatment beam is constantly monitored for deviations in dose delivery, beam flatness and dose rate. The dose monitoring systems commonly used are transmission ionization, or ion, chambers permanently mounted above the secondary collimator that continuously monitor the photon and electron beam output during patient treatment. The dose monitoring system usually consists of two independent ion chambers with completely separate biasing power supplies and readout electrometers. Failure of the primary chamber during patient treatment will cause the secondary chamber to terminate the irradiation [4].

The primary ion chamber measures monitor units (MU). At most clinics, the chamber circuitry is configured such that 1 monitor unit corresponds to a dose of 1 cGy delivered to a water phantom at a depth of dose maximum on the central beam when irradiated with a $10 \times 10$ cm$^2$ field at a source-surface distance (SSD) of 100.0 cm. At the cancer centre in Vancouver, BC, the chamber circuitry is calibrated using a source-axis distance (SAD) of 100.0 cm. The MU are entered on the linac control console and once the desired number of MU have been reached the primary ionization chamber circuitry stops the radiation beam delivery to the patient [4].

The dose monitoring system also checks other beam parameters including the energy, flatness and symmetry. To monitor these parameters the ion chambers are sectioned into four separate quadrants. Collecting electrodes for each quadrant are used to control the currents in the beam bending and steering magnets so that a symmetric beam is produced at all times [4].

1.2.2 Dose Deposition and Photon Interactions in Tissue

When a photon interacts with tissue, energy is transferred to the surrounding medium and dose is deposited. Five basic types of interactions can occur for the range of clinically-produced energies of 0 - 25 MeV. These include Rayleigh scattering, the photoelectric effect, the Compton effect, pair production and nuclear disintegration. Rayleigh, or coherent, scattering is elastic and does not impart any energy to the tissue. Nuclear disintegration is significant only for photon energies above 10 MeV and, compared to other photon interactions, has a small probability of occurring. The interactions relevant for the energies used in radiation therapy include only the photoelectric effect, Compton effect and pair production. The relative importance of these three interactions is depicted in Figure 1.3 [5, 6].
Chapter 1. Introduction

Figure 1.3: The relative importance of three major types of photon interactions in tissue, as a function of photon energy.

**Photoelectric Effect**

The photoelectric effect is the most important interaction with matter for low-energy photons. A photon of energy $h\nu$ interacts with an electron which is tightly bound to an inner shell of an atom with binding energy $E_B$, as shown in Figure 1.4(a). The electron is ejected at an angle $\theta$ relative to the incident photon direction and the recoil atom departs at an angle $\phi$ with virtually no kinetic energy (to conserve momentum). The photon is absorbed in this interaction. The energy transferred to the electron, $E_{ir}$, is given by

$$E_{ir} = h\nu - E_B. \quad (1.1)$$

The probability of the photoelectric effect occurring is greatest when the energy of the incident photon is slightly greater than the binding energy of the electron.

**Compton Effect**

In Compton interactions an incoming photon of energy $h\nu$ interacts with a loosely bound or free electron. After the collision the electron leaves at an angle $\theta$ and the photon is scattered at an angle $\phi$ with a lower energy $h\nu'$, as shown in Figure 1.4(b). The energy transferred to the electron is given by

$$E_{ir} = h\nu - h\nu'. \quad (1.2)$$

Compton scattering is almost elastic for low energy photons.
Chapter 1. Introduction

Figure 1.4: (a) The photoelectric effect. (b) The Compton effect. (c) Pair production.

Pair Production

Pair production occurs when a photon is absorbed, giving rise to an electron and a positron, as illustrated in Figure 1.4(c). This interaction can occur only in the electromagnetic field of an atomic nucleus. The energy transferred to the electron-positron pair is governed by conservation of energy,

\[ E_{tr} = h\nu - 2m_e, \]  

where \( m_e \) is the rest mass of both the electron and positron.

1.2.3 Fluence and Absorbed Dose

The transfer of energy from a high-energy photon beam to the medium takes place in two stages [6, 7]. In the first stage, the high-energy photon interacts with the medium and causes an electron to be ejected. This initial transfer of energy to a point in the medium, as done through the interaction processes described previously, is known as kerma, \( K \), (kinetic energy released in the medium) and is defined as

\[ K = \frac{dE_{tr}}{dm} = \Phi \left( \frac{\mu}{\rho} \right) E_{tr} \]  

where \( E_{tr} \) is the energy transferred to the electrons at each interaction in the mass \( m \) of the material, \( \Phi \) is the photon fluence and \( \mu/\rho \) is the mass attenuation coefficient for the medium. The units of kerma are joules per kilogram. In the second stage, energy is transferred from the ejected electron to the atoms in the surrounding medium through excitation and ionization. It is not until the energy is absorbed that there is a concept of dose [5, 6]. When an equilibrium between kerma and absorbed dose exists (i.e. when
Chapter 1. Introduction

the number of electrons entering the point of interest is equal to the number of electrons set in motion) the relationship between fluence and absorbed dose, $D$, can be written as

$$ D = \frac{dE_{ab}}{dm} = \Phi \left( \frac{\mu}{\rho} \right) E_{ab} \quad \text{(1.5)} $$

where $E_{ab}$ is the energy absorbed in the medium in a mass $m$ of the material. In most cases the equilibrium requirement is not satisfied and the relationship between dose and fluence becomes much more complex.

Fluence cannot be measured directly using standard radiotherapy dosimeters (e.g. ionization chambers, film) but the dose imparted to the medium can be. In order to obtain an accurate and meaningful relationship between fluence and dose, it is essential to have accurate dose-calculation methods. Currently, Monte Carlo calculation techniques for dose calculations are the accepted standard in radiation therapy physics because individual interactions are modeled for large numbers of photons and electrons. Due to computational-power limitations, the use of Monte Carlo techniques for dose calculations in the clinical setting is not yet practical. For this reason generalized models are used to simplify the dose calculation in standard treatment planning systems.

1.2.4 Modeling Dose Deposition

The most commonly used model for dose deposition in tissue is known as the pencil-beam model. In this model the photon beam is considered to be composed of a number of infinitely-thin pencil beams (Figure 1.5(a)). When the pencil beam interacts with tissue, dose is deposited in and around the point of interaction due to photon scatter and electron transport. From each single pencil beam the distribution of radiation is a symmetric function referred to as the dose spread kernel (DSK) [8, 9, 10], or simply the dose kernel (Figure 1.5(b)). Each radiation field is composed of many pencil beams, cumulatively contributing to the final dose distribution. The final dose distribution is therefore the sum of the individual DSKs for all pencil beams in the field (Figure 1.5(c)). As the number of pencil beams approaches infinity, the limit of the summation operation is a convolution,

$$ D(x,y) = \Phi(x,y) \otimes DSK(x,y) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \Phi(\alpha,\beta) DSK(x-\alpha,y-\beta) \, d\alpha \, d\beta. \quad \text{(1.6)} $$

In this simplified model, when the medium is homogeneous and the photon energy spectrum is spatially invariant, it is possible to approximate the final dose distribution
Chapter 1. Introduction

Photon pencil beam Multiple pencil beams Photon beam

(a) (b) (c)

Figure 1.5: (a) A single photon pencil-beam produces a dose spread kernel (DSK). (b) Merging a number of pencil beams results in a convolution of the dose kernel. (c) A photon beam, an extension of the cluster of pencil beams, produces a dose profile.

by convolving the incident photon fluence with the DSK [11, 12, 13]. Commercial Treatment Planning Systems (TPS) have modified the model to better account for these limitations [14]. It has been shown that higher resolution dosimeters (e.g. film, diodes) are necessary to derive more accurate dose kernels and that sharper dose kernels will improve the dose calculation accuracy for complex dose distributions [10, 15, 16]. Most TPS models are too basic for high resolution dose calculations. Problems concerning the resolution limitations will be addressed in subsequent chapters.

1.3 Radiation Therapy Delivery

1.3.1 Treatment Planning

As technology advances and computers become faster and more powerful, radiation therapy planning becomes a more computationally manageable task. When a patient must undergo radiation therapy, a Computed Tomography (CT) scan of the region to be treated is performed. CT scanning was developed in the 1970’s and uses radiographic projections obtained at different angles in cross-sections (a few mm thick) around the patient to digitally reconstruct a 3D distribution of the physical density inside the patient [17]. The location of anatomical structures are defined and the tumour, target contours and organs-at-risk (OAR) are drawn on each CT slice by a radiation oncologist. Adequate target information is not always provided by CT images, yet these
images are required as input by the treatment planning system. Magnetic resonance imaging (MRI) and/or positron emission tomography (PET) scans can be performed in addition to CT to better localize the target volume. MRI, for example, is better than CT for highlighting soft-tissue contrast, therefore small lesions may be more defined using the MR technique. The CT and MR/PET images are fused together digitally [18].

Three dimensional treatment planning systems require CT images with the contoured volumes as the starting point. The treatment dose, number of fractions given and dose constraints to the target and OAR are specified by the oncologist. Multiple beams are arranged and weighted so that the final dose distribution will conform to the oncologist's prescription. Dose calculation methods, as described in Section 1.2.4, are used by the treatment planning system to calculate the dose distributions resulting from the given beam arrangement.

1.3.2 Conformal Radiation Therapy

The goal of radiation therapy, as stated previously, is to deliver a uniform dose to the target while sparing surrounding critical structures as much as possible. Several delivery approaches are used to attain this goal, with multiple fields and field shaping being two common techniques.

Multiple Fields

When a beam of radiation is incident on a patient, a region near the surface exists where electronic equilibrium has not yet been achieved and the dose is primarily deposited downstream of where interaction is taking place. This region is known as the build-up region. At a specific depth the maximum dose deposited is achieved and is followed by a slow decrease in dose due to attenuation and increasing distance from the source [5, 7]. For a 6 MV photon beam, the location of maximum dose in tissue is at 1.5 cm. This distance (location) is dependent on the energy of the beam; beams of higher energy are more penetrating and the location of maximum dose will be deeper in the tissue. Figure 1.6 shows a plot of dose as a function of depth for a 5×5 cm² 6 MV photon beam. Dose decreases with depth, making it difficult to obtain a uniform dose in the tumour volume with a single field.

To achieve a uniform dose in the target volume, multiple beams at different gantry angles can be oriented with the centre of gantry-rotation located at one point within the tumour volume [7]. The majority of dose deposited will be focused uniformly
Chapter 1. Introduction

Figure 1.6: A percent depth-dose curve for a 6 MV photon beam. The maximum dose deposited is at a depth of 1.5 cm in tissue.

in the target area because the dose within the tumour is the cumulative sum of the contributions from all beams. The dose in surrounding tissue is only a contribution from a single field or small portion of the fields. An example of the dose profiles resulting from one set of parallel-opposed field pairs (POPs) is shown in Figure 1.7(a) and from two orthogonal POPs in Figure 1.7(b). Increasing the number of fields, especially at gantry angles not limited to orthogonal pairs, decreases the dose to surrounding tissue.

Field Shaping

As the electron beam passes through the components of the linac to the patient, the beam is filtered, shaped and modified. The beam passes through the primary collimator, a stationary device defining the maximum allowable field size, then through a secondary collimator. The secondary collimator consists of two pairs of attenuating metal blocks, or jaws, that move in and out of the radiation field to define arbitrary, rectangular fields required for treating the target volume, up to a maximum of $40 \times 40 \text{ cm}^2$.

While the secondary jaws sufficiently define rectangular field shapes, they are not practical in conforming to irregular-shaped tumours. Various tools have been devised to shape the beam as it enters the patient in order to improve dose conformity in the target and reduce dose to surrounding healthy tissue. One method of accomplishing this is to use metal-alloy blocks created specifically to match the tumour shape [7]. Often, low temperature alloy materials, such as Cerrobend, are used. The disadvantage of using custom-produced blocks is that a different set of blocks is required for each field because the beams’-eye-view (BEV) of the tumour will be different for each field. Also,
Chapter 1. Introduction

The blocks, which are mounted below the secondary jaws, need to be changed manually for each field. This requires a radiation therapist to enter the linac vault between the delivery of each field, a cumbersome and time-consuming process.

A relatively new device for 2D field shaping has been introduced in recent years. 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. The MLC is usually mounted (permanently) below the secondary collimator. Each leaf is controlled by its own motor, allowing the leaves to move independently of each other [19] to form arbitrary field shapes tightly conforming to the target volume. The MLC is a computer-controlled device, therefore for each field the MLC shape changes by selecting a new MLC file, making this device much less time- and labour-intensive than using and producing custom-designed blocks.

1.3.3 Intensity Modulated Radiation Therapy

The previous section introduced techniques used to conform uniform photon beams to the target area. However, in some situations, due to irregularities in the patient surface for example, it is desirable to also modify the intensity of the beam across its 2D extent [5, 7]. An example of this is to place a wedge-shaped attenuator in the field to create a dose gradient across the field.

In the past, the 2D modification of photon beam intensities has been limited to sim-
Chapter 1. Introduction

Figure 1.8: A photograph of an MLC assembly. The leaves move in and out of the radiation field via independent motors to form arbitrary shapes.

Intensity Modulated Radiation Therapy (IMRT) is an advanced method of 3D Conformal Radiation Therapy (3DCRT) that uses sophisticated computer-controlled beam delivery to tailor the conformality of the dose distributions to the shape of the tumour [20]. With IMRT, beam intensities are varied across each field using a multileaf collimator, whereas with 3DCRT the beam intensities are uniform across the field. Both IMRT and 3DCRT require multiple fields delivered at different gantry angles, however, in IMRT each field is divided into smaller sub-fields uniquely defined by the MLC. The cumulative summation of each sub-field produces a complex fluence map of varying intensity. The fluence from each sub-field is relatively constant but the size and shape of each sub-field can be quite diverse. The total fluence for each field is the sum of fluences from each sub-field and so at locations where multiple sub-fields overlap the total fluence will be the sum of each overlapping sub-field contribution and where no sub-fields overlap the total fluence at that location is zero [21].

IMRT, including the planning, optimization methods and delivery techniques are described in more detail in Chapter 2.
Chapter 1. Introduction

1.4 Dosimeters Used for IMRT

Dosimeters used for IMRT verification must be acceptable for both static and dynamic fields and must be characterized for linearity, energy dependence and angular response. Dosimeters are categorized into three groups: point, 2D and 3D dose detectors.

1.4.1 Point Dose Detectors

Ion Chambers

Ionization chamber dosimeters are used primarily for precision measurements required in radiotherapy clinics. Various types of ion chambers exist, including free-air chambers, thimble-type chambers, flat-cavity or extrapolation chambers and transmission-monitor chambers. The description of these can be found in any standard textbook [5, 6] and only the basic thimble ion chamber will be described as it is most commonly used in the radiotherapy setting.

The standard thimble ion chamber, shown in Figure 1.9, is cylindrical or spherical in shape with a diameter of less than 6 mm and a length of approximately 3 cm. The chamber wall is typically composed of air or a water-equivalent material and the chamber, often unsealed, encompasses a volume of gas, usually air, of approximately 0.1 – 3 cm$^3$. Incoming radiation interacts with and ionizes the surrounding gas and the ions are collected by the anode of the central electrode. The current is measured by an electrometer.

The ion chamber can be placed in a phantom to make point dose measurements and the relative dose measurements are related back to an absolute absorbed dose through a cross-calibration method.

Figure 1.9: A typical thimble ionization chamber.
Chapter 1. Introduction

Figure 1.10: A typical silicon p-n junction diode.

Diodes

Diode detectors made of semiconductor materials can be used for clinical relative dosimetry. A typical diode detector used in radiotherapy is the silicon p-n junction diode, which contains a wafer of silicon into which impurities are introduced to make p- and n-type silicon. The p-type junction is an electron acceptor and has an excess of positive ions, or holes. The n-type junction is an electron donor and has an excess of electrons [5, 6, 22]. When a reverse-bias potential of \( \approx 300 \) mV is applied across the p-n junction, the charge carriers are attracted toward the ends of the device, creating a depletion region (~ 2 \( \mu \)m thick) across the junction. When incident radiation passes through this region, the silicon becomes ionized and electron-hole pairs are produced. Due to the reverse bias, electrons and holes are swept to the n and p sides of the junction, respectively. The potential difference can be measured and related to relative dose.

Like ion chambers, the diode detector is essentially a point-dose detector but can be fashioned to measure dose in 2D and 3D by placing them in an array. These detectors can be made very small and have a high spatial resolution and low signal-to-noise ratio, making them appealing for phantom measurements and in-vivo dosimetry. However, diodes are susceptible to radiation damage over time, are not absolute dosimeters and their response is temperature and energy dependent.

Thermoluminescence Detectors

The phenomenon of thermoluminescence is the process by which light is emitted from a crystalline material that has been exposed to heat. In measuring the light emitted, the
Chapter 1. Introduction

Conduction Band

Valence Band

(a) (b)

Figure 1.11: (a) A diagram of the thermoluminescence cycle, where $M$ is the metastable state and $L$ is the luminescence state. Recombination of an electron and a hole releases an amount of light having an intensity proportional to the rate of escape of the electrons from the valence band. (b) The Glow Curve indicates the rate of electron escape, thus the intensity, as a function of temperature applied to the TLD as it is heated.

amount of radiation dose received by the material can be determined.

Perfect, uniform crystals are not ideal for thermoluminescent detectors (TLDs), as vacancy and impurity defects are required to trap electrons until the crystals are heated [5, 6, 7]. A negative ion vacancy is known as an electron trap and a positive ion vacancy is called a hole trap. When an electron and a hole meet, light is emitted. The thermoluminescence process is diagrammed in Figure 1.11(a). An ionization event raises the electron into the conduction band (A) where it migrates (B) to the electron trap (C). In this scenario the hole remains in the valence band and will migrate towards a hole trap. At room temperature or the temperature before irradiation, these traps should be deep enough, in terms of potential energy, to prevent the electron or hole from escaping for long periods of time. When the TLD is heated, the electron or hole or both will be released from their trap (D). When the electron and hole recombine (E), light ($h\nu$) is released. The reverse would hold true if the hole was raised into the conduction band and the electron remained in the valence band.

If a single trap is considered, the probability of release, $P$, is given by the Boltzmann distribution

$$P = \frac{1}{\tau} = se^{-\frac{E}{kT}},$$

(1.7)

where $\tau$ is the mean lifetime of the electron in the trap, $s$ is the frequency factor and is crystal dependent, $E$ is the energy gap between the trap and the conduction band, $T$
is the temperature of the crystal measured in Kelvin and \( k \) is the Boltzman constant. Assuming that the values \( k, E \) and \( s \) remain constant, Equation 1.7 implies that as the temperature \( T \) applied to the TLD increases, the rate at which the electrons escape from their traps will increase and eventually reach a maximum at temperature \( T_m \). After this point the supply of trapped electrons is exhausted, thereby reducing the number of escaping electrons. It is assumed that the intensity of light emitted is proportional to the rate of escape of electrons, therefore at \( T_m \) a corresponding peak in thermoluminescence brightness occurs. The accumulation of glow peaks of different trap energies produces a glow curve, as illustrated in Figure 1.11(b).

TLDs can be manufactured into very small compartments which makes them useful in radiotherapy settings. They can be placed on the body of a patient or into a solid phantom to make point dose measurements in 3D. Also, because the excited electrons can be held in their traps for extended time periods (days or months), TLDs work well as radiation dosimeter badges worn by personnel working with radiation. The badges can be tested every few months for the amount of radiation they have received and, in this way, can be used to monitor the radiation dose received by the person wearing the badge. However, TLDs must be characterized, which is time consuming, and the accuracy is limited to 2% to 3%. For IMRT field verification a large number of TLDs are required (60 or more) and so an automatic TLD reader is preferable.

MOSFET Detectors

Metal oxide silicon field effect transistors (MOSFETs), sometimes used for IMRT dosimetry, have excellent spatial resolution due to their small size, have automatic and immediate readout capabilities and can be reused. These attributes are advantageous compared to TLDs and film. MOSFETs have a response of ±3% reproducibility, ±2.5% angular dependence and a linear response for doses less than 30 cGy. Measured depth doses agree with ion chamber measurements within ±3% [6]. The main disadvantage is the decrease in linearity for doses greater than 30 cGy, which limits MOSFETs to lower dose applications [6].
1.4.2 Two Dimensional Dose Detectors

Radiographic Film

Radiographic film is one of the most common materials used to obtain and verify 2D dose distributions. This relative dosimeter has a high degree of spatial resolution, limited only by the resolution capabilities of the scanner or digitizer used. Film over­responds to low-energy photons because of its composition [6, 23, 24, 25] and is also sensitive to processor and digitizer quality assurance (QA) as well as its storage conditions. Differences in film batches require that separate dose-response curves be measured for each experiment in order to ensure accurate results.

Radiographic Film Structure and Image Formation

The sensitive photographic emulsion consists of microscopic grains (~ 3 μm) of silver bromide (AgBr) embedded in a gelatin layer. If the film is single emulsion the gelatin layer is bonded only to one side of the polyester-base substrate. With double-emulsion film, bonding is to both sides of the base, doubling the sensitivity of the film. A hardened gelatin supercoat protects the emulsion and prevents it from damage during processing [6, 26]. A schematic diagram of double-emulsion film is shown in Figure 1.12.

The grains are composed mainly of AgBr but may also contain small amounts of AgI. The Ag⁺, Br⁻ and I⁻ ions form a crystal lattice within a single emulsion grain. The small amounts of I⁻ ions form imperfections in the crystal, known as sensitivity specks, which increase the sensitivity of the crystal lattice to radiation. Upon exposure to radiation, energy is transferred to the emulsion and an electron is removed from the bromine atom. The electron migrates through the crystal until it is trapped at the...
sensitivity speck. The speck becomes negative, attracting interstitial Ag\(^+\) ions, then the ion combines with an electron to form an uncharged silver atom. Any uncharged bromine atoms are taken up by the emulsion. This cycle is repeated as more photons are absorbed by the emulsion and a certain number of silver ions migrate to the sensitive speck [27].

When developing a film, the developer acts as an electron donor and diffuses into the emulsion to be absorbed by the latent image centre. The chemical process follows:

\[
2\text{AgBr} + \text{H}_2\text{O} + \text{NaSO}_3 \rightarrow 2\text{Ag} + \text{HBr} + \text{H}_2\text{SO}_3\text{Na} + \text{NaBr}
\]  \hspace{1cm} (1.8)

where H\(_2\)O, hydroquinone, is the developer, NaSO\(_3\), sodium sulfite, is a preservative and clarifying agent, HBr is hydrobromic acid, HQSO\(_3\)Na is hydroquinonemonsulfonate and NaBr is sodium bromide [26]. This process reduces the Ag\(^+\) ions in the grains into metallic silver until eventually the grains are rendered entirely as metallic silver. Developer molecules will reduce Ag\(^+\) ions to Ag atoms in all grains eventually, whether ionized or not, however, those grains having a latent image are reduced much more quickly and the developer process can be terminated. Within a single grain, development is an all-or-none process and developed grains appear black [6].

A critical amount of time is required to develop the entire film, after which the emulsion is immersed in a dilute acetic acid stop bath to quickly terminate any further development. A sodium thiosulfate (hypo) solution is used to dissolve out any remain-
ing grains of AgBr that did not contain a latent image. The film is then washed in pure water and dried. The number of grains per cm\(^2\) indicate how much light will get through the film and is proportional to the electron fluence.

**Radiographic Film Dosimetry**

Film used in radiotherapy dosimetry applications is often placed into a phantom made of water or water-equivalent plastic. The film must not be exposed directly to light or water, so it is usually sandwiched between slabs of tissue-equivalent plastic and sealed tight in a darkroom. Some film comes in ready packs in which individual films are sealed in their own light-tight package. The advantage of the ready-pack film is that phantom loading does not need to take place in the darkroom, speeding the dosimetry process. The disadvantage of using ready packs is that a layer of paper sits between the film and the phantom material. There may also be an air gap between the ready pack and the film itself which may affect the recorded optical density if not taken into account. Once the exposed film is developed it is usually scanned with an optical scanner or scanning densitometer to measure the transmission of light through the film. The quantity measured is known as the optical density, \(OD\), and is given by

\[
OD = \log_{10} \left( \frac{I_0}{I} \right),
\]

where \(I_0\) is the incident light intensity measured in the absence of film and \(I\) is the intensity transmitted through the film. The optical density is then related to dose using a calibration graph known as a sensitometric curve. The sensitometric curve is created by measuring the optical densities of a number of known doses. The curve is specific to the batch of film used, imposing the criterion that a different set of calibration films need to be produced for every separate set of measurements taken. The relationship between \(OD\) and dose may or may not be linear depending on the type of film used as well as the range of doses delivered [28]. All types of film have a saturation point at some dose level where all grains produce a latent image.

Most of the measurements carried out for this work used Kodak EDR2 (Extended Dose Range) film as it has a responsive-dose range of 25 to 400 cGy and does not saturate until \(~700\) cGy [29]. This feature makes it a good choice for IMRT applications due to the higher doses required for these treatments. Kodak X-Omat V film was used for a small number of quality assurance measurements in which the doses delivered were substantially lower than 100 cGy. The responsive-dose range of the X-Omat V
Chapter 1. Introduction

film is 5 to 100 cGy and saturates around 200 cGy [29]. The AVID IMRT verification phantom (MDX Medical Systems, Vancouver, BC, Can) was used for all measurements and is described in detail in Chapter 6.

Radiochromic Film

Radiochromic film is a relatively new type of film in radiotherapy dosimetry. The most commonly used is a GafChromic™ film, which is colourless. This film contains a dye that polymerizes upon exposure to radiation [6, 30, 31], turning it a blue colour. The polymer absorbs light and a densitometer can measure the transmission of light through the film. Compared to radiographic film, radiochromic film is not as sensitive to low-energy photons and does not require a darkroom, film cassette or processor, making it easier to use than radiographic film. The response of the film to radiation, however, is non-uniform and is time and temperature dependent [32, 33]. Also, it is very expensive to purchase.

Electronic Portal Imaging Devices

Electronic Portal Imaging Devices (EPIDs) are used to verify patient positioning prior to treatment in order to reduce set-up errors. Treatment verification usually involves comparing a portal image acquired during a treatment fraction with a reference image generated prior to the treatment. The portal image is formed by the megavoltage beam used to treat the patient, but the reference image can be kilovoltage (simulation film), megavoltage, or a digitally reconstructed radiograph (DRR). Portal imaging may lead to various strategies for improving patient positioning, adjusting margins in combination with dose escalation and incorporating set-up uncertainty in treatment planning, all important considerations for IMRT treatments [18, 34].

Different varieties of EPIDs available. One is a camera-based system consisting of a metal plate and phosphor screen. Upon irradiation, high-energy electrons generated in the plate and screen are converted into light in the phosphor screen. With the aid of an angled mirror, the light from the phosphor screen is viewed by a sensitive video camera and converted into a video signal. The video signal from the camera can be digitized and viewed on a monitor located in the control area of the accelerator. The requirement of the mirror in this apparatus makes this type of EPID bulky [34].

A second type of EPID is the matrix ionization chamber device which consists of two sets of electrodes oriented perpendicularly to each other and separated by a small
fluid-filled gap. The fluid becomes ionized when the device is irradiated. One set of electrodes is connected to 256 electrometers and the other set of electrodes to a high voltage supply. The matrix ion chamber array is read out by applying a high voltage to each of the electrodes in succession and measuring the generated signal. This device is compact and of similar dimensions as a film cassette [35].

Another type of EPID is the amorphous silicon flat panel system which uses photodiode arrays in its image formation [36, 37]. This new generation of EPID is capable of providing IMRT verification.

1.4.3 Three dimensional Dose Detectors

Gel Dosimetry

Gel systems are inherently 3D dosimeters, useful for relative dose measurements. Gel is nearly tissue equivalent and can take the shape of any container, making it useful as a phantom that can measure the absorbed dose distribution in a full 3D geometry. The most common types of gel dosimeters include the Fricke and polyacrylamide polymer gels [6, 38]. The Fricke gel embeds ferrous sulfate in a gelatin matrix and when radiation interacts with the ferrous sulfate, ferrous ions are reduced to ferric ions. The gelatin matrix prevents the ferric ions from drifting, thus preserving the 3D dose distribution. With the polyacrylamide gel (PAG), acrylamide and \(N,N'\) methylene-bis-acrylamide polymerize into long cross-linked macromolecules upon irradiation. The concentration of macromolecules is proportional to the amount of dose given [6]. Again, the gelatin matrix inhibits any polymer molecule migration, ensuring the 3D spatial dose information remains intact. Imaging modalities including MRI [39], x-ray CT [40] and optical CT [41, 42] can image the gels to extract dose information. Gels are relative dosimeters and calibration curves are required to relate signal intensity to absorbed dose.

1.5 Thesis Objectives

In the last decade much research has gone into developing treatment techniques that will improve target conformity and minimize dose to organs at risk. IMRT is an advanced method of 3D conformal radiation therapy which uses the MLC to modulate the fluence across a treatment field. Recently a novel method of generating IMRT dose distributions has been developed whereby the MLC is rotated during the delivery of an
Chapter 1. Introduction

IMRT treatment [43]. With this method of IMRT delivery the total fluence is the cumulative summation of rotated sub-fields. Each sub-field is produced by an MLC-defined aperture, but instead of keeping the MLC at a fixed angle throughout the delivery of each field, the collimator is rotated by a specific angle between each sub-field. The initial introduction of this novel delivery technique showed that IMRT treatment plans could be delivered using a rotating collimator but did not include a rigorous testing of the delivery of the rotated fluence distributions.

The purpose of this thesis is two-fold. One objective is to evaluate whether IMRT plans created by incorporating collimator rotation can be delivered on a standard radiotherapy linac. This involves testing the accuracy and reproducibility of the collimator to rotate as well as testing the capability of a custom algorithm to model the rotated fluence maps. The second objective of this thesis is to evaluate whether dose distributions produced with rotating MLC IMRT are comparable to conventionally-delivered (non-rotated) IMRT plans.

The remainder of this thesis is organized as follows. Chapter 2 provides an overview of conventional IMRT (no collimator rotation). Rotating MLC IMRT is described in Chapter 3. This discussion includes the derivation of the rotating leaf motions and the optimization procedure. Chapter 4 discusses the commissioning and quality assurance testing performed for the Dynamic Beam Delivery (DBD) toolbox, the interface required to control collimator rotation. Chapter 5 presents results for modeling and dosimetrically verifying non-rotated and rotated fields. Chapter 6 describes the procedure used to produce and deliver rotating IMRT treatment plans. Dosimetric verification results are presented for plans delivered with and without MLC rotation. Finally, Chapter 7 gives a summary and recommendations for future work.
2.1 Conventional IMRT Overview

Intensity modulated radiation therapy (IMRT) is a treatment delivery technique that uses optimized, intensity-modulated radiation beams incident on a patient. Most IMRT treatment plans are generated using inverse planning whereby computer optimization techniques are used to determine the distribution of intensities across the treatment beams. The complexity of IMRT is greater than other traditional forms of radiation therapy and large improvements in target coverage and dose conformity as well as an increased normal-tissue sparing potentially are achievable. These advantages are evident especially for target volumes and/or critical structures having complex shapes and/or concave regions [20, 44].

Beam modifiers, including wedges and compensators, have been used for many years to modulate the beam intensity across the field and, although intensity modulated radiotherapy could be extended to incorporate these techniques, modern IMRT refers to those treatment plans generated using inverse planning to optimize the shape of the dose distribution. In virtually all cases a multileaf collimator (MLC) replaces the wedge and compensators as the beam intensity modulator. Many of the treatment geometries are complex, making it very important that the IMRT planning software can produce an optimal beam fluence map deliverable by the linac. The goal of IMRT is to derive the physically deliverable modulated-fluence profiles required to produce a dose
Chapter 2. Intensity Modulated Radiation Therapy

distribution closely matching the desired distribution.

Various IMRT delivery techniques are available for commercial use, including tomo-therapy IMRT [45, 46], conventional MLC IMRT [47, 48, 49, 50], intensity modulated arc therapy (IMAT) [51, 52, 53] and physical modulator (compensating filter) IMRT [54, 55, 56, 57, 58, 59]. The remainder of this thesis uses MLC-based IMRT and so only this delivery technique will be described in any detail.

A schematic diagram of the IMRT process is shown in Figure 2.1. Before planning begins the patient undergoes a treatment simulation involving a CT scan and, if necessary, MRI and/or PET scans in the treatment position. Custom-designed immobilization molds are created and used during the simulation to facilitate accurate patient repositioning throughout the treatment duration. The 3D images produced from the CT simulation are transferred to the treatment planning system (TPS) where the target and organs-at-risk (OAR) are contoured on each CT slice by the radiation oncologist. The beam directions to be used and the desired doses to be delivered to the tumour and normal tissue are defined as well. Inverse treatment planning is used for IMRT and the beam shapes and intensities are optimized (adjusted) to best meet the dose criteria. The optimization procedure is an iterative process performed by the computer and will be discussed further in the next section.

Once the plan has been optimized the MLC leaf sequences are derived. The dose distribution is calculated by the TPS and the treatment plan is evaluated. If the plan is not acceptable it is re-optimized using different dosimetric parameters until a suitable plan is achieved. At this point the plan is dosimetrically verified with film, ion chamber and EPID measurements prior to patient treatment. An IMRT plan can be delivered in either dynamic (sliding window) or static (step-and-shoot) mode. These delivery techniques are explained in more detail in Sections 2.3.1 and 2.3.2.

2.2 Treatment Plan Optimization

2.2.1 Forward Treatment Planning

Forward treatment planning is used for 3DCRT plans whereby the beam geometry (beam orientation, beam weights, shape, modifiers, etc.) is defined by the planner and the 3D dose distribution is calculated by the treatment planning system. The plan is reviewed and the initial geometry is modified, if necessary, to improve the final dose distribution by way of improving target coverage and/or decreasing the dose to critical
structures. The beam geometry is routinely modified by the planner until a satisfactory plan has been obtained.

2.2.2 Inverse Treatment Planning

Instead of specifying the initial beam geometry the planner outlines the desired outcome of the plan by specifying dose limits for the PTV and OARs. Through optimization methods the TPS iteratively adjusts the dose intensity pattern within each structure until the goals of the treatment have been met. The inverse planning process can be
Chapter 2. Intensity Modulated Radiation Therapy

separated into two components: (1) the cost function, or objective function and (2) the optimization criteria, or planning objectives and constraints.

The cost function is a mathematical definition of the quality of the treatment plan. Every iteration in the optimization process alters the intensity pattern within a defined structure in order to minimize the cost function. Objective functions can be based on biological criteria or dose criteria. The latter is most often used because a universally accepted biological model that predicts treatment outcome has yet to be developed [20]. The dose-based objective function should incorporate the clinical criteria used in routine planning, including the prescription dose to the target, target dose-homogeneity, critical organ maximum dose and critical organ dose-volume constraints. The criteria and constraints are specified on a cumulative dose-volume histogram (DVH). In an ideal situation 100% of the target volume will receive 100% of the prescribed dose and 0% of the OAR volume will receive 0% of the prescribed dose. This is never possible, therefore tolerance windows are defined on the DVH for different structures. A schematic diagram depicting a typical ideal and actual DVH are shown in Figures 2.2(a) and 2.2(b), respectively.

Each constraint has a mathematical cost function associated with it, the most common form being quadratic. For the target, the cost function, $F_{target}$, is given by

$$F_{target} = \frac{1}{N_t} \left[ \sum_{i=1}^{N_t} (D_i - D_{presc})^2 + w_{t,min} \cdot \sum_{i=1}^{N_t} (D_i - D_{min})^2 \cdot \Theta(D_{min} - D_i) ight]^{1/2},$$

(2.1)

where $N_t$ is the number of target points, $D_i$ is the dose to point $i$ and $D_{presc}$ is the prescription dose. The terms on the second and third lines include the target homogeneity criterion, where $D_{min}$ and $D_{max}$ are the minimum and maximum tolerance doses and $w_{t,min}$ and $w_{t,max}$ are penalties associated with under- and overdosing, respectively [20]. The Heaviside function, $\Theta(x)$, is defined as

$$\Theta(x) = \begin{cases} 1 & x \geq 0 \\ 0 & x < 0 \end{cases}.$$  

(2.2)
Chapter 2. Intensity Modulated Radiation Therapy

Figure 2.2: (a) The ideal DVH for a target and OAR. In this situation the target would receive 100% of the prescription dose and the OAR would receive 0% of the prescription dose. (b) The actual DVH for a target and OAR. The majority of the target volume is receiving 100% of the prescription dose and, while the OAR is still receiving some dose, the percentage of the volume receiving that dose is small. Acceptance criteria provide guidelines for the dose constraints for a given volume.

The objective function for the critical structures, $F_{OAR}$, is given by

$$F_{OAR} = \frac{1}{N_{OAR}} \left[ w_{OAR,max} \sum_{i=1}^{N_{OAR}} (D_i - D_{max})^2 \cdot \Theta(D_i - D_{max}) + w_{OAR,dv} \sum_{i=1}^{N_{dv}} (D_i - D_{dv})^2 \cdot \Theta(D_i - D_{dv}) \right], \quad (2.3)$$

where the first term inside the brackets is for the maximum dose constraint on the OAR and the second term is a dose-volume constraint. The relative penalty-weight for these constraints are $w_{OAR,max}$ and $w_{OAR,dv}$, respectively. $N_{OAR}$ is the number of points in the OAR and $N_{dv}$ is the number of points whose dose must be below the dose-volume constraint $D_{dv}$ [20].

Optimization Algorithms

Two basic types of algorithms used for IMRT optimization are gradient-based and stochastic methods [60, 61, 62, 63]. In general, a typical optimization algorithm works in the following way. A patient’s 3D image is divided into small 3D voxels (volume elements) and beamlets are traced from the radiation source through the patient. For an
initial set of beamlet weights the dose in each voxel is calculated and the dose distribution is used to calculate the value of the objective function. Each new iteration changes the beamlet weighting. If the change in beamlet weight minimizes the cost function and brings the calculated dose distribution closer to the desired distribution, the change is accepted; if not the change is rejected. This process is repeated for all beamlets. At the end of each iteration a new pattern of beamlet intensities is used to calculate a new dose distribution and cost function value. When no further improvement to the dose distribution is obtained from changes in beamlet weights, the optimization is stopped and the optimum plan is assumed to have been found.

Gradient-based methods are, by far, the fastest computationally [60, 64, 65, 66, 67], however, this technique assumes that there is a single global minimum. When multiple global minima exist, using dose-volume-based objectives for example [68], the gradient search method will converge to the nearest minima. This scenario, in theory, may lead to an unsatisfactory treatment plan that is far from the best solution, although the existence of multiple minima has not been found to adversely affect the final results.

If multiple minima cause problems in the creating acceptable plans, stochastic-based methods often are adopted [69, 70, 71, 72, 73]. Stochastic methods, such as simulated annealing techniques, allow the optimization process to escape from local minima to find the global minimum. Random processes are used whereby the values of the arguments jump from one local minima to another. The objective function is still used to evaluate the progress of the optimization but more stringent acceptance criteria are used to determine the argument values for the next iteration. The degree of randomness or jumping is usually adjusted as the optimization comes closer to finding the optimal solution. With this method, a large number of configurations are tested and there is no guarantee that the optimal solution will be found, only that the best solution among those considered will be found.

2.3 Leaf Sequencing

At the end of the optimization an optimal fluence map has been determined but limitations of the MLC used to produce these fluences have not been taken into account. The final process of deriving the MLC apertures needed to produce the actual fluence maps is referred to as the leaf-sequencing step. Many leaf-sequencing techniques have been proposed in the literature [74, 75, 76, 49, 77, 78, 79, 80]. In general, a one-dimensional approach is taken and the fluence profile between each leaf and its opposing leaf pair is
Chapter 2. Intensity Modulated Radiation Therapy

2.3. Distance (cm) Leaf Position (cm)

Figure 2.3: (a) The fluence for a dynamic MLC delivery is composed of a large number of levels. (b) Each leaf pair moves continuously during a sliding window treatment to create a MLC aperture that slides from one side of the field to the other.

considered separately. The position and size of gap between MLC leaves for each sub-field can be modified to produce almost any fluence profile. At the end of this procedure a set of actual, deliverable fluence maps will be produced that match the optimal fluence as closely as possible. Either a dynamic (sliding window) or static (step-and-shoot) leaf sequencing technique can be used.

2.3.1 Dynamic (Sliding Window) IMRT

With dynamic IMRT the radiation beam is on for the entire delivery and the MLC leaf pairs slide across the field to form the desired apertures. The computer-controlled leaves move continuously across the field at varying speeds to obtain the desired intensity profile [49, 77, 78].

As an example, consider the fluence distribution of a pyramid pattern: the intensity is greatest at the centre and decreases outward in steps. Figure 2.3 shows a schematic diagram of the fluence profile of an arbitrary leaf pair through this pyramid fluence distribution. The fluence profile in Figure 2.3(a) is composed of a large number of discrete levels. As the leaves continuously move across the field, the fluence is built up. The position of each leaf is a function of the fluence index, the relative quantity of radiation produced by the linac. 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, making the fluence profile in Figure 2.3(a) equal to the area between the two
Chapter 2. Intensity Modulated Radiation Therapy

Figure 2.4: (a) The build-up of fluence for a static MLC delivery. (b) The motion of the leaf banks as they move across the field.

leaf trajectories in Figure 2.3(b). As the fluence index increases, in Figure 2.3(b), the position and distance between the two leaves is translated across the desired area in a smooth, sliding-window, movement.

2.3.2 Static (Step-and-Shoot) IMRT

In this approach a series of multiple segments (sub-fields) make up the entire field with each sub-field being composed of a different MLC-defined aperture. When the beam is off, the computer-controlled leaves step to their pre-defined position. Once the leaves stop at that position, the beam is turned on to deliver, or shoot, a set amount of radiation. The beam is stopped while the leaves form the next aperture. This step-and-shoot pattern is continued until the leaf sequence for each sub-field has been delivered for that field [21, 75, 81]. This form of IMRT was used for all plans delivered for this thesis.

The delivery of a static IMRT plan is similar to the dynamic case except that the fluence is composed of larger discrete fluence levels resulting from the step-and-shoot nature of the delivery. Again, consider a pyramid-design fluence distribution. Figure 2.4(a) describes how the fluence is built up. Each leaf must move along a specific trajectory to produce the desired fluence profile. As shown in Figure 2.4(b), the leaf positions for each leaf are derived as a function of the fluence index.
2.4 Delivery

The leaf positions for each sub-field are transferred from the treatment planning system to the MLC control computer prior to treatment. The gantry angle, collimator jaw positions and MLC leaf positions are set for the first field segment. The radiation beam is turned on and the computer moves the leaves to the required position. Once the sequence has been delivered for this field, the gantry is moved to another angle for the delivery of the next field. When all fields have been delivered the treatment is complete.
To improve upon existing MLC-based IMRT techniques, collimator rotation can be incorporated into IMRT delivery. Conventional IMRT has certain physical limitations that can be remedied by rotating the collimator during beam delivery. Limitations include interleaf leakage and tongue-and-groove effects, smaller deliverable field sizes due to leaf-motion limits, the large number of monitor units required for a treatment and limited spatial resolution due to leaf width.

The concept of incorporating collimator rotation into IMRT treatment delivery was first introduced by Otto et al. [43], whose work provided information regarding the leaf motion derivations, optimization algorithm and generation of rotational fluence maps. This chapter includes a further discussion and summary of the advantages of rotating multileaf collimator (RMLC) IMRT as well as a description of the leaf motion derivation and generation of rotated fluence maps.

3.1 Advantages of Collimator Rotation

The leaves of the Varian Millennium 120-leaf MLC are 6 cm deep, made of tungsten alloy and are designed to block more than 95% of incoming photons [63, 82]. Each leaf is operated by a separate motor allowing independent leaf movement, with a small gap between adjacent leaves ensuring smooth interleaf movement. A tongue-and-groove design is introduced to minimize the increased photon transmission due to the gap. With conventionally-delivered IMRT fields, the MLC is restricted to movement in a single plane, in and out of the radiation field. The leaves move across the field but
Chapter 3. Rotating Multileaf Collimator IMRT

Figure 3.1: (a) An intermediate step of fluence exist when the MLC tongue is open in the field. (b) A similar situation occurs when the MLC groove is exposed. (c) An under-dosing occurs at the tongue-groove interface, an effect known as the tongue-and-groove effect.

the relative position of each leaf pair never changes during the delivery of the treatment. The immediate effect of this design is that radiation is constantly transmitted through the MLC at the same location, building up interleaf effects and causing a non-uniformity in the delivered dose distributions [83, 84].

The tongue-and-groove leaf design also increases the likelihood of under-dosing effects appearing along leaf edges [84]. This effect often arises during an IMRT treatment and is most prominent at locations where a leaf is open in one step and closed in the next, as illustrated in Figure 3.1.

When treatment fields are delivered using conventional IMRT the location of the leaf edge is fixed so that dosimetric errors resulting from interleaf leakage and tongue-and-groove effects are concentrated along the leaf edges, as shown in Figure 3.2(a). Rotating the MLC between each sub-field minimizes these effects because the leaves continually move to a different location. Any effects due to transmission and the tongue-and-groove design are effectively spread out over the 2D field area (Figure 3.2(b)). Consequently, dosimetric errors appearing throughout a RMLC IMRT treatment will only be a fraction of those appearing in a conventional treatment.

The spatial resolution of a fluence map can be increased by incorporating collimator rotation. This can be visualized by considering the size of a fluence beamlet, the
Chapter 3. Rotating Multileaf Collimator IMRT

Figure 3.2: (a) In conventional IMRT the location of the MLC leaf edge is fixed throughout the delivery thereby accumulating transmission and interleaf leakage dose along the same lines. (b) When the collimator is rotated between each sub-field the location of the leaf edge constantly changes thereby reducing interleaf effects.

minimum beamlet size that can be obtained from the leaf width. With conventional IMRT the MLC leaves move in a single plane, and although the MLC is capable of moving in less than 1 mm increments in the direction of leaf motion, the leaf widths do not change. The minimum beamlet that can be defined is thus a rectangle, as shown in Figure 3.3(a). With RMLC IMRT the direction of leaf motion is not limited to a single plane and changes with each sub-field rotation, as shown in Figure 3.3(b). This allows the minimum beamlet size achievable to be much smaller than with conventional delivery, consequently increasing the inherent spatial resolution of the fluence beamlet.

3.2 Rotating Leaf Motion Derivation

Although MLC rotation provides an additional degree of freedom and flexibility when delivering a desired fluence map, deriving the necessary leaf motions increases the complexity of this technique. The inherently 1D approach associated with the conventional IMRT leaf motion calculation is no longer applicable when MLC rotation is introduced. The leaves move to a different location as the collimator rotates, with each leaf affecting different points in the desired fluence map. The leaf pairs capable of modulating the fluence at a specific point change, an effect that is not straight-forward to calculate.
Chapter 3. Rotating Multileaf Collimator IMRT

Figures 3.3: (a) The minimum beamlet size achievable for conventional IMRT. (b) The minimum beamlet size achievable for rotational IMRT. The inherent spatial resolution of the fluence beamlet increases when collimator rotation is incorporated.

3.2.1 Rotating Leaf Motion Optimization

A custom-designed optimization algorithm, stochastic in nature, has been created that randomly varies leaf positions at pre-specified collimator angles to bring dose distributions of actual and desired fluence maps closer together. After each variation in leaf position the fluence map, \( \Phi_{\text{optimal}} \), is updated and compared to the desired fluence map, \( \Phi_{\text{actual}} \), by calculating the objective function, \( \text{Obj} \),

\[
\text{Obj} = \sum_{x,y} | \Phi_{\text{optimal}}(x,y) - \Phi_{\text{actual}}(x,y) |. \quad (3.1)
\]

If the objective function decreases and the calculated fluence is brought closer to the desired fluence, the change is accepted and the objective function is updated. If the change is not accepted, the individual fluence beamlets affected by the change are examined. If those beamlets lie within a dynamically-controlled range of acceptable error, the change is also accepted. If neither of these criteria are met the leaf position change is discarded and another set of randomly-selected leaf positions is tried.

The process of updating fluences after each leaf position change is computationally-intensive but is made more efficient by creating a reference matrix incorporating each rotated position of the MLC leaves with respect to beamlets in the fluence map for all possible leaf positions at each rotated segment. Each time the optimization is run, the location of any changes to the calculated fluence from varying leaf positions is obtained directly from the reference matrix.
Chapter 3. Rotating Multileaf Collimator IMRT

3.2.2 Rotated Leaf Motion Parameters

Before the optimization begins various parameters need to be defined. The rotating leaf motion calculator (RLMC) is specified (stochastic algorithm) along with the type of MLC used for the delivery (Varian Millennium 120-leaf MLC (Varian Medical Systems, Palo Alto, CA, USA)) and the delivery mode required (static or dynamic). Only multiple static segment delivery is considered for this work. The initial number of segments, rotation range and radiation efficiency are also defined; for the cases shown in this thesis, a radiation efficiency of 95% and rotation range of 180° are used. The initial number of segments for each plan is chosen such that the same plan can be delivered conventionally or with rotation using the same number of segments for a fair comparison. For static delivery, the MLC leaf position at each segment is pre-defined with the closing point of each leaf pair placed at the location of maximum fluence along the trajectory of those leaves. Therefore, when the optimization begins the leaves will open at the location where the MLC is open for the largest portion of the treatment. Finally, the optimal fluence maps need to be imported to the custom algorithm. As mentioned previously, the algorithm uses an optimal fluence map, generated initially by the CadPlan TPS, as its starting basis to find the actual fluence distribution that is as similar to the optimal fluence as possible.

3.3 IMRT Delivery With a Rotating MLC

The Varian CL21EX linac (Varian Medical Systems, Palo Alto, CA, USA) was used for all measurements performed in this work. In order to control collimator rotation accurately and reliably, a specialized interface, the Dynamic Beam Delivery (DBD) toolbox, was installed on the linac. The interface is capable of controlling all dynamic motion of the linac, although only collimator rotation control was explored. Chapter 4 describes the quality assurance and commissioning tests performed.

As with any standard IMRT delivery the fluence maps for each field need to be loaded into the MLC control computer and the correct information regarding beam energy, field sizes, treatment time and dose must be entered into the linac control console. The only difference between RMLC and conventional IMRT delivery is that a DBD toolbox file must be imported to the linac control computer for RMLC IMRT. A separate file for each field delivered is required by the linac and these files are generated by the RLMC algorithm.
Testing of the Dynamic Beam Delivery Interface

Rotating MLC aperture delivery is controlled by the Dynamic Beam Delivery (DBD) toolbox, a special interface required for the Varian CL21EX linear accelerator (Varian Medical Systems, Palo Alto, CA, USA). The DBD toolbox can control a variety of motions dynamically, including collimator and gantry rotation, couch and all jaw motions; however, for this thesis MLC rotation only is controlled. Commissioning and quality assurance testing must be performed before the DBD toolbox can be used clinically and tests were carried out to evaluate the accuracy of the toolbox to rotate the collimator to a given angle and the stability of the monitor unit output. All tests were delivered in a step-and-shoot fashion whereby the collimator rotates and the MLC leaves move to the next sub-field position while the radiation beam is off.

4.1 Collimator Angle Accuracy

The DBD toolbox must accurately and reproducibly control collimator rotation. As a test, a pattern was produced whereby the MLC leaf bank was opened completely and a long narrow aperture (0.5×8.0 cm²) was defined by the secondary jaws of the linac. The collimator, under the control of the DBD toolbox, was programmed to rotate the aperture in 10° increments over a 180° span.

Kodak X-Omat V film was placed on the central plane of the AVID IMRT verification phantom, perpendicular to the incoming beam, at a source-to-axis distance (SAD)
Chapter 4. Testing of the Dynamic Beam Delivery Interface

Figure 4.1: (a) The 2D test pattern used to determine the accuracy of the collimator angle positions. (b) The angle between each adjacent segment is measured.

of 100.0 cm and a depth of 12.8 cm in the phantom. Measurements were taken with the gantry at 0° and 270° to evaluate whether gantry sag had any effect on the capability of the collimator to rotate to precisely-defined angles. When the gantry is rotated away from the 0° or 180° position, gravity effects due to the mass of the gantry increase and causes the machine to sag to an angle slightly different than the intended angle. The effect is most prominent when the gantry is rotated to either 90° or 270°. Gantry sagging is an important consideration on how accurately the DBD toolbox is able to control collimator rotation because each field in a clinical treatment plan is delivered at a different gantry angle and so it is essential to be aware of any effects the gantry angle may have.

Figure 4.1(a) shows the beams'-eye-view distribution generated from the test pattern. The angle between each rotated aperture would be exactly 10° if the DBD toolbox was capable of rotating the collimator precisely. To validate this, the centre of each segment was located and the angle between the central lines of two adjacent segments was measured, as shown in Figure 4.1(b).

Presented in Figure 4.2 is a plot comparing the measured and expected collimator angular positions. The values agree within ± 0.5° for measurements taken at 0° and 270°, a variation within the specifications given by the manufacturers for the DBD toolbox. These results also make apparent that the change in gantry angle does not adversely affect collimator rotation thus gantry sag is not considered to be an issue in further measurements.

Once it was confirmed that the DBD toolbox could reliably control MLC rotation, reproducibility of the collimator had to be validated. Evaluations were performed
Chapter 4. Testing of the Dynamic Beam Delivery Interface

Figure 4.2: The measured collimator angle is plotted against the expected collimator angle. Results are shown for gantry angles 0° and 270°.

monthly\(^1\), beginning from the installation and commissioning of the linac, over a period of 16 months. An established quality assurance procedure was used to verify the collimator angular position. The gantry was rotated to 90° to ensure the axis of collimator rotation was oriented perpendicular to the direction of the force of gravity. Using a spirit level placed on the upper edge of the collimator housing, collimator angles of 0°, 90° and 270° were evaluated by selecting them from the control computer and allowing the collimator to freely rotate to the desired angle. Figure 4.3 shows measured angles at 0°, 90° and 270°. Apart from one measurement at 270°, the collimator rotation angle is accurate to within 0.5°. The anomalous result occurred near the beginning of linac use, after which the measurements have remained relatively constant. The results are adequately reproducible over the 16-month time period to conclude that rotation of the collimator is not a limiting factor.

4.2 Monitor Unit Accuracy

The pattern used in Section 4.1 was also used to test the stability of the monitor unit output with Kodak X-Omat V film placed in the same orientation. The purpose of this test is to determine, first, whether MU variation throughout a delivery is constant and

\(^1\)Measurements and results completed and collected by the Physics Assistants at the BC Cancer Agency in Vancouver, BC.
Chapter 4. Testing of the Dynamic Beam Delivery Interface

Figure 4.3: (a) Collimator-angle reproducibility results at 0°. (b) Collimator-angle reproducibility results at 90°. (c) Collimator-angle reproducibility results at 270°. The collimator angle was verified once per month over a period of 16 months, starting from the installation of linac Unit 6.
second, whether dose values can be linearly scaled, an important consideration with single field delivery when the doses are often modified to lie within the responsive range of the film used. In one set of measurements 760 MU was delivered to the film (40 MU per segment) and in a second set 76 MU was delivered (4 MU per segment). Both measurements were performed with the gantry at 0° and 270° and, after scaling the second measurement by a factor of ten, the dose per segment delivered to each film was expected to agree within 3% (the inherent error associated with radiographic film measurements [23, 24, 25]).

A three dose-level calibration technique, described fully in Chapter 6, was performed for both the collimator angle and MU accuracy tests. The calibration films were scanned with a VIDAR DosimetryPRO film scanner (VIDAR Systems Corporation, Herndon, VA, USA) and custom software was used to construct the required sensitometric curve to convert the optical density into dose values.

The histograms in Figures 4.4(a) and 4.4(b) compare the total dose values when the gantry is at 0° and 270° and indicate that the relative doses scaled by a factor of ten are consistently lower at almost all collimator angles. The average dose for both measurements, however, agrees to within 3%. One explanation is that a dose of 4 MU per segment is too low to be delivered accurately, although previous work [85] indicated that 4 MU was the minimum dose-threshold deliverable before obtaining a non-linear response. This is attributed to the results obtained.

Figures 4.4(c) and 4.4(d) compare results from the two gantry angles for a total film dose of 760 MU and 76 MU. For a given total film dose, the same dose is delivered at both gantry angles, therefore large dose discrepancies between the two gantry angles should not exist. The average relative dose and standard deviation of the segments in each scenario is calculated and the results are summarized in Table 4.1. The average dose values are consistent, within error, for both gantry angles and total film dose and the standard deviations agree within 3% of the average calculated dose, signalling acceptable agreement between the dose and gantry parameters considered. The standard deviation calculated for the film receiving 10×76 MU at gantry 270° is substantially higher than the other three values, indicating the spread of dose values about the mean is larger and may result from a combination of delivering both single aperture doses that are too low and small gantry sag effects (the standard deviation values are higher when the gantry is at 270°, as shown in Table 4.1).
Figure 4.4: (a) MU output comparisons at gantry 0°. (b) MU output comparisons at gantry 270°. (c) Gantry angle comparisons for 760 MU. (d) Gantry angle comparisons for 76 MU. The bars labeled 10×76 MU correspond to the measurements where dose values are scaled by a factor of ten.
Chapter 4. Testing of the Dynamic Beam Delivery Interface

<table>
<thead>
<tr>
<th>MU Accuracy Output</th>
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<tbody>
<tr>
<td>Total Dose (MU)</td>
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<tr>
<td>760</td>
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<tr>
<td></td>
</tr>
<tr>
<td>10×76</td>
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Table 4.1: Summary table of statistical results for the MU accuracy test. The average doses of the bars at each collimator angle (from 90° to 270°) are given for each of the four cases and their standard deviation is calculated.

4.3 Conclusion

In conclusion, the accuracy and reproducibility of the DBD toolbox was validated during commissioning and quality assurance testing. The DBD toolbox was found to accurately rotate the MLC to within 0.5° of its intended collimator angle, a result within the limitations specified for the interface. Testing was performed at gantry angles 0° and 270° and results indicated no significant discrepancies between the two gantry angles, concluding that gantry sag was not adversely affecting the ability of the DBD toolbox to rotate the collimator. The accuracy of delivered MU was also considered and, within 3% (the inherent error in the dose response of radiographic film), the relative dose of segments at each collimator angle was accurate and reproducible, although extremely low doses should be avoided.
5.1 Fluence and Dose Calculation

Measurements were taken with the Varian CadPlan Helios inverse treatment planning system v.6.27 (Varian Medical Systems, Palo Alto, CA, USA) which uses a modified pencil-beam model for the dose calculation algorithm. Each pencil beam defined in the radiation field is associated with a dose kernel describing the physical distribution of absorbed dose about a small, but finite, beam of photons incident on matter. The total dose to a point in the medium is a superposition of the dose contribution from all pencil beams in the field and can be represented by a convolution of the dose kernel with the primary fluence [10, 12, 86]. The shape of the DSK depends on the photon energy spectrum at the depth of interest as well as the production and distribution of secondary electrons in the material, thus the DSK shape has an effect on how the deposited dose is calculated.

Dose kernels are not modeled accurately by most treatment planning systems and are often too wide [10, 15]. Figure 5.1 depicts two different-sized dose kernels used to generate a bar pattern distribution. When a wide DSK is used to approximate a thin pencil beam, as in Figure 5.1(a), the lower region of the kernels overlap and the calculated dose is a poor representation of the measured dose. Usage of a tighter and thinner DSK, as in Figure 5.1(b), results in less dose blurring. The DSKs used in the CadPlan TPS dose calculation algorithm are not tightly modeled and, while a tighter DSK does
Chapter 5. Fluence Modeling and Dose Calculation

Figure 5.1: (a) A schematic diagram representing the dose resolution of a wide dose spread kernel (DSK) for a bar pattern. (b) A schematic diagram representing the dose resolution of a narrow DSK for a bar pattern. (c) Sample dose profiles comparing measured and calculated distributions using the wide DSK. (d) Sample dose profiles comparing measured and calculated distributions using the narrow DSK.

not necessarily translate into better agreement between measured and calculated dose distributions, it does represent more accurately the thin pencil beams used in the dose model. Figure 5.1(c) is a hypothetical dose profile comparing measured and calculated (TPS) profiles of a dose distribution when a wide, poorly-modeled DSK is used. Often the dose calculation algorithm under-estimates dose in the peaks and over-estimates dose in the valleys, as shown. Figure 5.1(d) illustrates the same hypothetical dose profiles but with a thinner DSK used in the dose calculation. The effects represented in these diagrams may not exist in every dosimetric comparison and depend on the level of fluence modulation, however, when discrepancies arise in peak and valley regions, the shape of the dose kernel used in the dose calculation will often be an important contribution to the uncertainty. The custom optimization algorithm uses a modified dose kernel [10] to calculate dose distributions, however, the CadPlan TPS used to calculate the final dose distribution for all clinical plans shown does not use the modified kernel.

Another limitation of most TPS dose calculation algorithms is that extra-focal ra-
Chapter 5. Fluence Modeling and Dose Calculation

diation, comprised of photons and electrons scattered in the head of the linac, is not modeled explicitly. As a result, treatment planning systems, CadPlan included, will under-estimate the dose outside the field.

5.2 Modeling of Non-Rotating Fields

The rotation MLC segmentation algorithm, described in Chapter 3, is designed to produce optimal fluence distributions incorporating collimator rotation. The goal of this thesis is to evaluate whether IMRT plans can be delivered accurately using collimator rotation, however, it is important to first test that the RMLC algorithm is modeling fluence accurately in general. Two simple test patterns not involving MLC rotation were generated with the algorithm and the validity of the fluence modeling was verified by examining the dose distributions. Kodak EDR2 film was placed on the isocentre-plane of the AVID IMRT phantom and a 6 MV photon beam was used to deliver test patterns in a step-and-shoot fashion. For the remainder of the thesis the film results will be denoted as the measured distribution and the modeled distribution (MatLab or CadPlan) as the calculated distribution.

5.2.1 Square Aperture (Open Field)

A 10.5 x 10.5 cm² aperture, the simplest fluence pattern that can be delivered, was modeled using the RMLC segmentation algorithm. A total of 200 cGy was delivered to a phantom in five non-rotated segments; the measured and calculated dose distributions are shown in Figures 5.2(a) and 5.2(b), respectively. Orthogonal profiles taken at the isocentre in the x- and y-planes are shown in Figures 5.2(c) and 5.2(d). The agreement between the measured and calculated distributions is excellent, confirming that the fluence map of a simple, non-rotated field can be modeled properly by the RMLC algorithm. This test pattern is not, however, representative of highly complex fields seen in IMRT plans and so accuracy limitations of the dose calculation will not be apparent in the results of this test.

5.2.2 Bar Pattern

A five-segment bar pattern was produced with the RMLC algorithm such that the MLC leaves moved across the field from left to right, defining a new segment (bar) for each
Chapter 5. Fluence Modeling and Dose Calculation

Figure 5.2: (a) The measured dose distribution of an open field pattern. (b) The calculated dose distribution of the same pattern. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. The profiles taken across the measured and calculated distributions are in the solid and dashed lines, respectively.

sub-field. The pattern was delivered with the MLC rotated to both 90° and 180° because interleaf leakage and transmission effects become more prevalent in one case over another.

A total of 1000 MU was delivered to Kodak EDR2 film in the AVID IMRT phantom with each segment receiving 200 MU. The measured and calculated dose distributions are shown in Figures 5.3(a) and 5.3(b), respectively, and a schematic diagram depicting the location of the MLC leaves with respect to the apertures formed is presented in Figure 5.3(c). An orthogonal profile taken along the x-plane, Figure 5.3(d), compares the measured and calculated dose distributions. The rounded leaf-tips of the Varian Millennium 120-leaf MLC are not modeled adequately by the RMLC algorithm so leaf-end transmission and interleaf leakage effects are not accounted for. This results
in a higher measured film-dose in regions between each aperture (valley) of the profile plot. Transmission and interleaf leakage effects are maximal when the MLC is oriented in this position, increasing the dose in those regions. The RMLC algorithm does not account for interleaf effects and calculates a very small dose, mainly due to photons scattered from within the linac head. Although differences between the measured and calculated dose distributions are seen, these discrepancies are not unexpected due to modeling limitations.

Figure 5.4 shows the results from rotating the MLC 180°. Compared to the previous case, interleaf leakage and transmission effects are not as prevalent because the apertures are now formed using the length of the leaves instead of the leaf tips, as in Figure 5.4(c). Interleaf leakage and transmission effects are reduced and the level of
dose agreement is increased in the peaks and valleys, as expected. An increased dose along the superior portion of the measured dose distribution is an effect of the closed leaf pairs not being completely blocked by the secondary jaws of the treatment head.

The results of the square aperture and bar tests verify that the RMLC algorithm can model simple, standard aperture shapes and can also calculate the dose to a level that is accurate, within the current modeling limitations of the MLC and interleaf effects.
Chapter 5. Fluence Modeling and Dose Calculation

Figure 5.5: (a) The dimensions of a conventional beamlet used for IMRT optimization. (b) The sub-sampling of a beamlet used for rotational IMRT optimization.

5.3 Modeling of Rotated Fields

5.3.1 Sub-sampling

One important difference between plans optimized with conventional and rotational IMRT is the minimum beamlet size used. In conventional IMRT the beamlet size is limited by the width of the MLC leaf: for a 5.0 mm MLC leaf and a minimum step size of 2.5 mm, the minimum beamlet size is 5.0x2.5 mm$^2$. The MLC leaves are also restricted to movement in one direction, in and out of the field. In one leaf step the beamlet will either be covered completely or uncovered completely, meaning that only one beamlet can ever be affected by moving a leaf one position. A schematic of this is illustrated in Figure 5.5(a). For this reason, conventional IMRT is effectively a one-dimensional problem.

When MLC leaf segmentation is incorporated into an IMRT treatment, the beamlet size is no longer limited by the leaf width. The entire MLC is rotated between each segment and more than one beamlet can be blocked or partially blocked by one leaf. To account for this, the concept of sub-sampling is introduced and the minimum size used for each beamlet is reduced to 2.5x2.5 mm$^2$, the smallest size allowed by CadPlan. Consider Figure 5.5(b) in which the MLC has been rotated with respect to the beamlet grid. The leaf in question is affecting more than one beamlet and the covered portion of each beamlet is different, so the beamlet intensity is weighted or scaled by the area uncovered by the MLC. The rotated positions of the MLC leaves with re-
Chapter 5. Fluence Modeling and Dose Calculation

5.3.2 Rotating Square

The fluence-modeling capability of the RMLC algorithm was further evaluated by producing a pattern in which two identical square apertures were created at a distance shifted off the isocentre and diagonal from each other, as shown in Figure 5.6. The apertures were rotated in five segments at equidistant angles, creating a pattern with regions of high-dose gradients as well as a central region of very low dose. Clinical IMRT plans may exhibit dose gradients and regions of very high and low doses similar to those found in this test pattern, therefore it is important for the rotating aperture algorithm to model accurately patterns such as this.

The rotating square pattern was delivered to Kodak EDR2 film in the AVID IMRT phantom. A total of 200 cGy was delivered in five segments; Figure 5.7 shows the dose distribution acquired from this rotating square pattern. The measured and calculated dose distributions are compared using a custom-designed program and good agreement exists between the two distributions, even in high-dose gradient regions. Dose agreement in the central region is not as good, with the measured dose being approximately 25% higher than the calculated. Accuracy limitations of the dose calculation and inadequate inclusion of extra-focal radiation may be the cause, as similar results from the bar pattern test indicated that the calculated dose was under-estimated in the low-dose regions.
Chapter 5. Fluence Modeling and Dose Calculation

Figure 5.7: The rotating square pattern. (a) The measured dose distribution. (b) The calculated distribution. (c) Orthogonal profiles taken along the x-plane. (d) Orthogonal profiles taken along the y-plane. The profiles taken across the measured and calculated distributions are in the solid and dashed lines, respectively.

(valley) regions. Another possibility for the discrepancy may arise from scattering effects not modeled extensively by the algorithm; electrons scattered from the edge of the MLC aperture will increase the dose to the central region.

Results from the rotating square test indicate that the RMLC algorithm can accurately produce fluence maps of rotating apertures within the limits of the accuracy of the dose calculation algorithm.

5.3.3 Rotating IMRT Results of Single Fields

A final measure of the fluence modeling and dose calculation capabilities of the RMLC algorithm is achieved by verifying an IMRT plan created with the RMLC algorithm. Only geometric shapes have been produced and verified and it is crucial for the al-
algorithm to work equally well for clinical treatment plans. A full description of the process required to create a RMLC IMRT plan is outlined in Chapter 6. For now it is sufficient to understand that an IMRT plan is initially created in CadPlan and the RMLC algorithm uses CadPlan-produced fluence maps to create RMLC fluence maps. These fluence maps can be delivered by the linac and the dose distributions can be verified.

Single-field RMLC IMRT measurements of a prostate carcinoma treatment are shown in Figure 5.8. Fluence maps for the five-field plan were created with the RMLC algorithm and delivered to Kodak EDR2 film in the AVID IMRT verification phantom. Each single-field fluence map was delivered to a separate film with the gantry at 0° and the dose distributions were compared to the distributions calculated by both the RMLC algorithm and the CadPlan TPS. In Figure 5.8 the measured and calculated dose distributions are shown for all five fields. The doses are absolute in that there has been no normalization and the dose delivered to the film is the dose per fraction calculated by the CadPlan TPS. The RMLC algorithm is not a treatment planning system and does not calculate dose so the CadPlan TPS is relied upon for the final dose calculation. Comparing absolute dose values is preferable because no scaling is required and subtle differences in dose distributions are enhanced.

Qualitatively, the measured dose values tend to be higher than calculated values, particularly in regions of very high dose. One explanation for the large discrepancies is because the rounded MLC leaf-tips are not modeled properly in the RMLC algorithm, leading to an over-dosing. Single-field measurements provide a more sensitive dose comparison, therefore, while this problem will likely be evident in a composite distribution, the magnitude of the errors may not be as pronounced for composite verifications.

In conclusion, discrepancies between measured and calculated dose distributions are apparent for single-field results delivered with MLC rotation, particularly in the high-dose regions. The next section discusses the leaf gap concept and provides a solution to problems introduced by inadequate modeling of the rounded leaf ends.

5.4 Dosimetric Leaf Gap

The design of multileaf collimators can be either double or single focused. The leaves of a double focused MLC have a flat leaf-tip and move along an arc about the x-ray target to mimic beam divergence off-axis. In this design, fluence transmission is unity on the open side of the leaf and immediately reduces to the full leaf-thickness transmission
Figure 5.8: The measured and calculated distributions and a comparison profile for each of the five fields of a prostate carcinoma plan.
Chapter 5. Fluence Modeling and Dose Calculation

Figure 5.9: Small penumbral effects are caused by the rounded leaf-tip design of the Varian Millennium 120-leaf MLC.

below the leaf. The leaves of a single-focused MLC, like the Varian 120-leaf Millennium MLC used in this thesis, move perpendicular to the beam central axis. In this design the MLC leaves do not follow the divergence of the beam as they move across the field so a rounded leaf-tip design is used to maintain a relatively constant penumbra for different leaf displacement values [87, 88, 89]. Fluence transmission through the rounded leaf-tip is different than through the thickest part of the leaf, leading to a larger overall penumbra (see Figure 5.9). As a result, the single-focused MLC design can lead to hot spots between neighbouring beam segments during IMRT dose delivery if the dose calculation algorithm does not properly account for the additional fluence through the rounded leaf-tip. This problem is exaggerated in dynamic IMRT but is still a concern in step-and-shoot IMRT when the leaf-pair separation is small (< 5.0 mm).

The additional fluence through the rounded leaf-tips can be regarded as being equivalent to an effective increase in the space between opposing leaves. The concept of a leaf gap or leaf offset was devised in order to compensate for this problem [63, 90, 91, 92, 93]. This measured quantity indicates the distance in or out of the radiation field by which the leaves must be shifted in order to minimize the effects of additional fluence and transmission. The leaf gap is a modification used to bring the measured dose distribution closer to the calculated distribution when MLC leaf-tips are not modeled adequately. The leaf gap value should be the same, over time, for one linac and its MLC, but may vary between different linacs and MLCs.

Different methods of determining the dosimetric leaf gap are reported in the literature [90, 91, 92, 93] and the following terminology has been adopted: the leaf offset
Chapter 5. Fluence Modeling and Dose Calculation

Figure 5.10: A schematic diagram of the relative MLC positions used to determine the optimal leaf gap, based on work done by Tangboonduangjit et al. [92]. The dark colour of MLC represents the MLC bank for a leaf offset of 0.0 cm and a leaf gap of 2.0 cm. The lighter colour of MLC indicates the MLC bank for 0.5 cm per leaf offset value and a leaf gap of 1.90 cm.

The match-line method [92] was chosen to determine the leaf gap and uses five 2.0 x 10.0 cm$^2$ bars delivered to radiographic film to produce a 10.0 cm$^2$ square at the end of the delivery. Each time this pattern is delivered, a different leaf offset value is used. The leaf offset value that produces a final dose distribution that most closely resembles an open field (as in Section 5.2.1) is taken as the desired leaf offset for the linac and MLC used. In another commonly-used method [90], 10.0 x 10.0 cm$^2$ fields are delivered to film with the leaf-bank separation adjusted from 0.5 to 10.0 cm for separate irradiations. The leaf offset is determined by plotting the measured versus set leaf-bank separations and calculating the y-intercept.

5.4.1 Determining the Leaf Gap

To find the optimal leaf offset value, MLC files required to control the leaf bank motion were created using the RMLC program. A 2.0 x 10.0 cm$^2$ bar defined by the MLC was programmed to move across the field in five segments moving in 2.0 cm increments. In an ideal scenario the dose distribution produced would be an open field, however, because of interleaf leakage and added transmission through the rounded leaf-tips, re-
Figure 5.11: Leaf gap test results. The dose values are normalized to 100% and the ideal distribution, which has been a profile taken through an open square aperture of the same dimensions as the size of the final leaf gap test.

regions of under- and over-dosing exist. The optimal leaf offset is achieved when the dose distribution closest to the ideal (open field) situation is obtained.

In the original set-up the leaf offset was 0.0 cm with a leaf bank separation of 2.0 cm. Five subsequent MLC files were produced in which the leaf offset was increased, leading to a decreased separation window between opposing leaf pairs. The leaf offset values used were 0.05, 0.07, 0.10, 0.11 and 0.12 cm, resulting in leaf bank separations of 1.90, 1.86, 1.80, 1.78 and 1.76 cm, respectively. A schematic diagram of the relative MLC positions is shown in Figure 5.10.

Kodak EDR2 film was used for all measurements, with each film placed in the AVID IMRT verification phantom at a SAD of 100.0 cm and depth of 12.8 cm in phantom. A 6 MV photon beam was used to deliver a total dose of 1000 MU to each film (200 MU/bar). The films were scanned with the Vidar DosimetryPRO Film Digitizer and the images imported to the custom dosimetry system for analysis.

For each film a profile across the x-plane is taken and normalized to a relative dose of 100%, as seen in Figure 5.11. The flat profile (thick black line) is taken across a 10.0×10.0 cm² open-field measurement. The graph explicitly indicates that a com-
Chapter 5. Fluence Modeling and Dose Calculation

<table>
<thead>
<tr>
<th>Leaf Gap (mm)</th>
<th>P/V</th>
<th>Average Relative Dose (cGy)</th>
<th>Standard Deviation (cGy)</th>
</tr>
</thead>
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<tr>
<td>0.0</td>
<td>1.19</td>
<td>105.12</td>
<td>0.32</td>
</tr>
<tr>
<td>1.0</td>
<td>1.02</td>
<td>101.76</td>
<td>0.53</td>
</tr>
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<td>1.4</td>
<td>0.94</td>
<td>100.31</td>
<td>0.45</td>
</tr>
<tr>
<td>2.0</td>
<td>0.85</td>
<td>98.36</td>
<td>0.53</td>
</tr>
<tr>
<td>2.2</td>
<td>0.81</td>
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</tr>
<tr>
<td>2.4</td>
<td>0.79</td>
<td>97.26</td>
<td>2.30</td>
</tr>
<tr>
<td>Ideal</td>
<td>1.00</td>
<td>99.54</td>
<td>0.30</td>
</tr>
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Table 5.1: *Summary of the average peak (P) to valley (V) dose ratios and the average relative doses for each profile as well as the standard deviation at various leaf gap values.*

...completely flat profile is never achievable, although certain leaf offset values prove more effective than others. To determine the optimum leaf offset value, the method used by Tangboonduangjit *et al.* [92] was followed, in which the ratio of peak-to-valley dose for each profile, the average relative dose and the corresponding standard deviation were evaluated.

For the spikes along each profile in Figure 5.11, the average dose of every peak (P) and valley (V) is calculated to obtain the P/V ratio, and the average relative dose and standard deviation for each profile is calculated from the relative doses. The results, compiled in Table 5.1, show that the peak-dose point decreases with increasing offset value. A leaf gap of 1.4 mm is chosen as the most appropriate value, as the average dose along the corresponding beam profile is closest to the ideal average dose and the low standard deviation reflects less variation in dose values relative to the other leaf gap results.

Single-field measurements were carried out using different leaf gap values to examine how the different gap values affected the dose distribution and if an optimum leaf gap value of 1.4 mm worked best in clinical situations. The prostate carcinoma presented in Figure 5.8 was used, focussing on field 3 (Figures 5.8(g)–5.8(i)). This field was delivered to three separate films in the AVID phantom, using leaf gaps of 0.0, 1.4 and 2.0 mm. The measured and calculated dose distributions for different leaf gaps are shown in the following figures: Figure 5.12 for the 0.0 mm gap, Figure 5.13 for the 1.4 mm gap and Figure 5.14 for the 2.0 mm gap. Profiles were taken at the same location for each leaf gap to provide a meaningful dosimetric evaluation.

As the leaf gap increases the agreement between the measured and calculated dose...
Chapter 5. Fluence Modeling and Dose Calculation

<table>
<thead>
<tr>
<th>Dose Region</th>
<th>Leaf Gap (mm)</th>
<th>Measured Dose (cGy)</th>
<th>Calculated Dose (cGy)</th>
<th>Absolute Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low</td>
<td>0.0</td>
<td>91.86</td>
<td>97.19</td>
<td>5.48</td>
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<tr>
<td></td>
<td>1.4</td>
<td>89.98</td>
<td>97.19</td>
<td>7.42</td>
</tr>
<tr>
<td></td>
<td>2.0</td>
<td>88.67</td>
<td>97.19</td>
<td>8.77</td>
</tr>
<tr>
<td>High</td>
<td>0.0</td>
<td>160.81</td>
<td>144.69</td>
<td>11.14</td>
</tr>
<tr>
<td></td>
<td>1.4</td>
<td>157.51</td>
<td>144.69</td>
<td>8.86</td>
</tr>
<tr>
<td></td>
<td>2.0</td>
<td>150.12</td>
<td>144.69</td>
<td>3.75</td>
</tr>
</tbody>
</table>

Table 5.2: Point dose measurements taken within a prostate carcinoma plan. In the low-dose region there is an over-compensation of dose and in the high-dose region there is an under-compensation.

distributions improves in the high-dose regions but worsens in the low-dose regions. Table 5.2 summarizes the dose values taken at the profile intersection in these regions. The dose values are from a single point only, however, results from using the 1.4 mm leaf gap yield the greatest dosimetric agreement between the high- and low-dose areas. The average dose to the target volume for both the measured and calculated distributions in Figures 5.12–5.14 are calculated and summarized in Table 5.3 along with the absolute percent difference values. Distributions using the 1.4 mm leaf gap are, again, found to have the best dosimetric agreement and have the lowest percent difference value. The average measured and calculated doses are 118.73 cGy and 117.24 cGy, respectively, yielding an absolute percent difference of 1.27%. The absolute percent difference between the measured and calculated distribution is 5.52% using a 0.0 mm leaf gap and 3.71% using a 2.0 mm leaf gap. This verifies that the 1.4 mm leaf gap is the optimum gap value to use for this particular linac and MLC thus for all further measurements taken, a leaf gap of 1.4 mm is incorporated.
Chapter 5. Fluence Modeling and Dose Calculation

Figure 5.12: Prostate carcinoma, field 3, RMLC IMRT. (a) The measured dose distribution using a 0.0 mm leaf gap, optimized with a 5.0 mm MLC. (b) The calculated distribution using a 0.0 mm leaf gap, optimized with a 5.0 mm width beamlet. (c) X-plane orthogonal profiles taken in a high-dose region (∗). (d) Y-plane orthogonal profiles taken in a high-dose region (∗). (e) X-plane orthogonal profiles taken in a low-dose region (+). (f) Y-plane orthogonal profiles taken in a low-dose region (+).
Figure 5.13: Prostate carcinoma, field 3, RMLC IMRT. (a) The measured dose distribution using a 1.4 mm leaf gap, optimized with a 5.0 mm MLC. (b) The calculated distribution using a 1.4 mm leaf gap, optimized with a 5.0 mm width beamlet. (c) X-plane orthogonal profiles taken in a high-dose region (†). (d) Y-plane orthogonal profiles taken in a high-dose region (†). (e) X-plane orthogonal profiles taken in a low-dose region (+). (f) Y-plane orthogonal profiles taken in a low-dose region (+).
Figure 5.14: Prostate carcinoma, field 3, RMLC IMRT. (a) The measured dose distribution using a 2.0 mm leaf gap, optimized with a 5.0 mm MLC. (b) The calculated distribution using a 2.0 mm leaf gap, optimized with a 5.0 mm beamlet. (c) X-plane orthogonal profiles taken in a high-dose region (†). (d) Y-plane orthogonal profiles taken in a high-dose region (†). (e) X-plane orthogonal profiles taken in a low-dose region (+). (f) Y-plane orthogonal profiles taken in a low-dose region (+).
Chapter 5. Fluence Modeling and Dose Calculation

## Table 5.3

<table>
<thead>
<tr>
<th>Leaf Gap (mm)</th>
<th>Measured Dose (cGy)</th>
<th>Calculated Dose (cGy)</th>
<th>Absolute Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.0</td>
<td>123.96</td>
<td>117.48</td>
<td>5.52</td>
</tr>
<tr>
<td>1.4</td>
<td>118.73</td>
<td>117.24</td>
<td>1.27</td>
</tr>
<tr>
<td>2.0</td>
<td>112.69</td>
<td>117.03</td>
<td>3.71</td>
</tr>
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</table>

Table 5.3: The average target dose in the prostate carcinoma for both the measured and calculated dose distributions for each leaf gap test and the absolute percent difference. The distribution using the 1.4 mm leaf gap is the most accurate.
Dosimetric Treatment Verification
Using RMLC IMRT

In previous chapters it was shown that the DBD toolbox could accurately and reliably rotate the MLC and that the custom algorithm was capable of generating rotated fluence maps deliverable on the linac. In this chapter the procedure used to generate an IMRT plan using collimator rotation will be described in detail as will the delivery and dosimetric verification procedures. At the end of the chapter, IMRT treatment plans generated for three target areas will be described and the dosimetric results of each case will be shown. The intention of this thesis is to evaluate whether IMRT plans can be delivered using RMLC and whether the dose distributions are comparable to conventional IMRT treatment plans.

6.1 Generating Deliverable IMRT Plans with RMLC

6.1.1 Treatment Plan Generation

An IMRT treatment plan is created initially with the CadPlan treatment planning system. A research (virtual) linac has been configured to use a MLC with 2.5 mm wide leaves instead of the commercially-available MLC with 5.0 mm wide leaves. The leaf width of this virtual MLC corresponds with the $2.5 \times 2.5$ mm$^2$ beamlets used in the RMLC algorithm. When the optimization process is started, constraints for the target and critical structures (volume, dose and priority) are defined on the dose-volume his-
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

togram (DVH). For the plans shown, the optimization uses a termination criteria of 0.1 and a maximum of 900 iterations. The termination criteria is set such that the optimization will stop before 900 iterations if, after each iteration, substantial improvements in the DVH are not seen; this avoids creating very small aperture sizes for each field. For example, if the optimization is allowed to stop after achieving an acceptable plan (the DVH criteria are met), this may require 600 iterations only. However, if the optimization must complete all 900 iterations, any adjustments made to the fluence map after 600 iterations will not be necessary. Often these changes modify the leaf positions such that the apertures become quite small thereby increasing the possibility of incurring dosimetric errors.

Once the optimization is complete and the optimal fluence maps are created, actual, deliverable fluence maps must be produced. This is done by using either the sliding window or step-and-shoot approach to generate the appropriate leaf sequences. When a conventional IMRT plan is generated, the leaf sequences required to created actual fluence maps are those leaf sequences used for the treatment delivery. This is not the same when generating an IMRT plan with collimator rotation. For RMLC IMRT the optimal (non-rotated) fluence maps are transferred from CadPlan to the custom RLMC algorithm, which runs under MatLab (The Mathworks, Natick, MA, USA). The algorithm specifies the number of segments per field to be used, the rotation range of the MLC and the radiation efficiency. The type of leaf-motion calculator and rotation (sub-sampling) reference matrix required for the plan are selected and the leaf motion optimization is initiated to calculate the RMLC sequences for each field. The generated fluence maps are the actual, deliverable, rotated fluence maps.

The RMLC algorithm determines the X and Y jaw positions and collimator start angles for each field, the rotation file required for the DBD toolbox and the MLC leaf sequencing file for the MLC controller. The gantry angles are preset by the CadPlan TPS and are not optimized. The actual fluence maps are transferred back to the CadPlan TPS where existing fluence maps are over-written with those containing collimator rotation information. The jaw settings generated by the RMLC algorithm are necessary when the 3D dose distribution is calculated by the CadPlan TPS and at this stage the MU for each field are calculated.

69
6.1.2 Treatment Plan Delivery

The treatment plan incorporating collimator rotation can be delivered on the linac with the aid of the DBD toolbox to control collimator rotation. The fields can be delivered statically with the collimator rotating and leaves moving only when the beam is stopped, or dynamically with both the collimator rotating and leaves moving continuously. The RMLC IMRT plans are delivered with the Varian CL21EX linac and Millennium 120-leaf MLC. The MLC model is composed of leaves 0.5 cm wide over the central 20.0 cm and leaves 1.0 cm wide outside.

The information required to deliver a RMLC IMRT plan on the linear accelerator include a file for the MLC-controller computer and a file for the DBD toolbox. The MLC file contains information specifying the position of the MLC leaves for each sub-field. The DBD toolbox requires information regarding the dynamic motion of the collimator, specifically the start and stop collimator angles for each field and the angle through which the collimator must rotate between each sub-field. The following parameters are required to deliver each field: the beam energy (6 MV for all cases presented), dose rate, number of MU, gantry angle, starting collimator angle, X and Y jaw positions and correct DBD toolbox and MLC file for that field.

6.2 AVID IMRT Verification Phantom

The treatment plans are delivered to radiographic film placed in a water-equivalent IMRT phantom and the measured dose distributions are compared to the dose distributions calculated by the TPS. This section discusses the phantom used for all thesis measurements as well as the process required to compute and extract the calculated dose distribution.

6.2.1 Phantom Measurements

The AVID IMRT verification phantom, used for all dosimetric verification measurements presented, consists of an inner cartridge and outer shell. The inner cartridge is composed of black, high-impact polystyrene and the shell is made of Rexolite, both water-equivalent materials. The inner cartridge, shown in Figure 6.1(a), houses removable polystyrene spacers, between which film can be placed. The spacers are of varying thickness and allow up to 21 planes of film to be irradiated in a single delivery. The inner cartridge can be placed in the shell in one of three orientations so that the film
can be irradiated in either the sagittal, coronal, or axial plane. Cross-hairs etched onto the top, left and right sides of the shell allow the phantom, and thus the film inside, to be accurately aligned using the isocentre lasers in the linac vault. The set-up of the phantom on the treatment couch is shown in Figure 6.1(b). For the plans shown in this chapter, Kodak EDR2 film was placed in the isocentre plane of the phantom at a SAD of 100.0 cm and a depth of 12.8 cm, and the phantom was oriented perpendicular to the incoming beam. Unless otherwise stated, all measurements were performed with the film in the coronal plane.

### 6.2.2 Dose to Phantom: TPS Calculation

The 3D dose distribution calculated by the treatment planning system determines the dose to a patient, but the treatment plan is delivered to a phantom. The cube-shaped phantom does not approximate the shape of an actual patient, and the measured and calculated dose distributions must be in the same medium for an acceptable comparison, thus a 3D dose distribution must be calculated in the phantom as well.

The AVID IMRT verification phantom has been imaged with a CT scanner and the images imported into the CadPlan system act as the phantom patient. The original IMRT plan, including the fluence maps, field sizes and gantry angles, is transferred to the phantom patient and the dose is calculated. This 3D dose distribution will be equivalent to the distribution delivered to the film in the AVID IMRT phantom.
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

Figure 6.2: (a) First percent depth dose (PDD) curve (low dose). (b) Second PDD (medium dose). (c) Third PDD (high dose). (d) A calibration curve converts pixel values into dose.

6.3 Film Calibration Technique

Film acts as a relative dosimeter and requires a calibration technique to convert the OD values into dose; in this work percent depth dose (PDD) curves are used. Film is placed in the axial plane with the incident beam parallel to film placement. A 6 MV photon beam with a $5.0 \times 5.0$ cm$^2$ field size was used to irradiate three films with different MU. The MU were chosen so that the dose values overlapped between all three films and the entire range of values in the delivered IMRT distribution were contained within the curves. Problems due to calibration error, film emulsion non-uniformity, or processing variations would be evident as a discontinuity between the three curves at the overlapping points.

The OD of an exposed, processed film is proportional to the amount of dose deposited in the film and the OD is determined at a point by measuring the transmission of a collimated light source through the film. The films were scanned with a Vidar DosimetryPRO film scanner using a resolution of 71 pixels/inch ($0.0358$ cm/pixel) and 16-bit depth settings. Scanning the film produced a 2D matrix of pixel values di-
6.4 Dosimetric Verification Methods

For RMLC IMRT to become a clinically viable procedure, treatment plans must be delivered reliably and reproducibly and the dose distributions must yield equivalent (or better) results to what a conventionally-delivered IMRT treatment plan would yield.

Dosimetric verification techniques are used to determine the acceptability of a plan. This section describes the techniques used to quantify how closely the measured and calculated distributions compare.

6.4.1 Orthogonal Profiles

A simple but effective way to visually quantify the agreement between measured and calculated distributions is to generate orthogonal profiles through various points in the distribution. A dosimetric analysis program is used to plot the $x$ and $y$ profiles for the measured and calculated distributions at a given location. This technique is useful for locating geometric and dosimetric shifts between two distributions.

6.4.2 Gamma Factor Analysis

The gamma ($\gamma$) factor analysis is another technique commonly used to quantitatively compare two distributions whereby the dose distributions are subdivided into regions of high and low dose-gradients and agreement is considered in those areas. Van Dyk *et al.* [94] describes the quality assurance procedure of treatment planning systems and states the acceptance criterion in high and low dose-gradient regions. In low dose-gradient regions, doses can be compared directly by subtracting the calculated from the measured dose distribution and the remaining dose-difference distribution will highlight dosimetric discrepancies between the two. In high dose-gradient regions a small spatial error can result in large dose differences, especially along field edges. Poor agreement in these regions does not necessarily translate to a poor overall agreement between the two distributions, therefore the concept of distance-to-agreement (DTA) is
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

devised to determine the acceptability of the distributions in these high dose-gradient regions. The DTA is the distance between a measured data-point and the nearest point in the calculated dose distribution that exhibits the same dose.

The $\gamma$ factor composite analysis uses a pass-fail criteria combining both the dose difference and DTA. Each measured point in the distribution is evaluated to determine if both the dose difference and DTA exceed the specified tolerance. Points that fail both criteria are identified on a composite distribution. One limitation of this technique is that a 2D dose distribution is simplified into a binary outcome.

Two graphical representations of the $\gamma$ factor criteria are shown in Figure 6.3. For this discussion the calculated dose distribution is taken as the reference distribution and the measured dose distribution as the evaluated distribution. The mathematical formalism describing the dose difference, DTA and $\gamma$ distribution has been derived by Low et al. [95, 96, 97] but will be summarized here for clarity. The dose difference criterion is referred to as $\Delta D_M$ and the DTA criterion is referred to as $\Delta d_M$: in most clinical situations the standard passing criteria is $\Delta D_M = 3\%$ and $\Delta d_M = 3$ mm.

A single point, $r_m$, is considered to be lying at the origin. The $x$- and $y$-axes define the spatial location of the reference point, $r_c$ and the third axis, $\delta$-axis, represents the difference between the measured, $D_m(r_m)$, and calculated, $D_c(r_c)$, doses. The dose difference criterion, $\Delta D_M$, is represented by the vertical lines in the $\delta$ direction. If the calculated distribution surface crosses the line

$$|D_c(r_m) - D_m(r_m)| \leq \Delta D_M,$$

the calculated distribution at that measurement point will pass the dose difference test. The DTA criteria, $\Delta d_M$, is represented by the disk in the $r_m - r_c$ plane with a radius equal to $\Delta d_M$. If the calculated distribution surface, $D_c(r_c)$ intersects the disk, the DTA is within the accepted criterion and the calculated distribution at that point passes the DTA test.

Figure 6.3(b) considers both the dose difference and DTA criteria simultaneously for the $\gamma$ evaluation. An ellipsoid is selected as the surface representing the acceptance criterion and is defined as

$$1 = \sqrt{\frac{r^2(r_m, r)}{\Delta d_M^2}} + \frac{\delta^2(r_m, r)}{\Delta D_M^2}$$  \hspace{1cm} (6.1)
where

$$r(r_m, r) = |r - r_m|$$  \hspace{1cm} (6.2)$$

and the dose difference at the position $r_m$ is

$$\delta(r_m, r) = D(r) - D_m(r_m).$$  \hspace{1cm} (6.3)$$

If any portion of the surface $D_c(r_c)$ intersects the ellipsoid defined by Equation 6.1 then the calculation passes at that point, $r_m$.

The quantity on the right-hand-side of Equation 6.1 is used to quantify the $\gamma$ index.
at each point in the evaluation plane \( r_c \rightarrow r_m \) for the measurement point \( r_m \), therefore

\[
\gamma(r_m) = \min \{ \Gamma(r_m, r_c) \} \forall \{ r_c \}
\]

(6.4)

where

\[
\Gamma(r_m, r_c) = \sqrt{\frac{r^2(r_m, r)}{\Delta d_M^2} + \frac{\delta^2(r_m, r)}{\Delta D_M^2}}
\]

(6.5)

The pass-fail criteria for the composite \( \gamma \) distribution is as follows:

\[
\begin{align*}
\gamma(r_m) & \leq 1 \quad \text{test passes} \\
\gamma(r_m) & > 1 \quad \text{test fails.}
\end{align*}
\]

It is known only if the dose distributions agree or disagree at a point and no information regarding the severity of the failed points is provided. To obtain more information, a scale is incorporated on the \( \gamma \) composition to distinguish points that have a \( \gamma \) between 1 and 2 or between 2 and 3 and so on. If points fail the \( \gamma \) criteria a scaling will indicate how severely the points have failed. If many \( \gamma \) index values fail but lie between 1 to 2 these regions may be analyzed on individual dose distributions to determine their significance. This scaling technique is used commonly for quantitative IMRT dose verification [97, 98, 99, 100, 101].

### 6.5 Treatment Plans and Results

The prostate, nasopharynx and a complex geometric c-shape target (representative of a treatment site where the target is wrapped around a critical structure) are the treatment areas considered. Five plans were created and delivered to Kodak EDR2 film placed on the isocentre plane of the AVID IMRT phantom. The fields were centered on the target volume in all cases and optimized such that a uniform dose was given to the target volume with minimal dose to surrounding critical structures. Conventional IMRT plans, identical to the rotational cases apart from the collimator rotation, were also created for the prostate and nasopharynx target volumes and delivered to the phantom for additional verification.
6.5.1 Prostate Carcinoma

A prostate carcinoma was planned using five equidistant beams at gantry angles 144°, 72°, 0°, 288° and 216°. The rectum and rectal wall were the dose-limiting structures and a total of 74 Gy in 37 fractions was prescribed. Two separate plans were produced from the same CT data-set. A different target shape and location was used for each plan, thus optimizing the targets yielded two different treatment plans. Dose distributions calculated in the patient and in the phantom by the CadPlan TPS are shown in Figures 6.4(a) and 6.4(b), respectively, for one of the target contours.

RMLC IMRT (2.5 mm Width Beamlet): PTV 1

To begin, a prostate carcinoma case was optimized to the target volume PTV 1 and the treatment plan was generated with collimator rotation using the research linac, a fictitious linac having a 2.5 mm leaf width MLC corresponding to the 2.5 mm width beamlets used during the optimization. A composite treatment, with each field at their respective gantry angle delivered to the same film, produces a distribution similar to what one would expect within the patient. The measured (film) and calculated (CadPlan) dose distributions for this target are shown in Figures 6.5(a) and 6.5(b), respectively. The two distributions have consistent dose patterns, however, along superior and inferior regions of the target, the measured distribution displays higher doses not predicted by the CadPlan TPS. The magnitude of disagreement is evident from the orthogonal profiles and $\gamma$ factor distributions. Orthogonal profiles were taken along the
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

Figure 6.5: (a) The measured dose distribution for the prostate carcinoma plan delivered with RMLC IMRT (2.5 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The $\gamma$ factor analysis using a dose difference criteria of 3% and a distance-to-agreement criteria of 3 mm. The maximum $\gamma$ value is 13.1 and 83.9% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.

78
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

$x$- and $y$-planes, as depicted in Figures 6.5(c) and 6.5(d), with the profile along the $x$-plane running from left (-40 mm) to right (+40 mm) of the target and the profile along the $y$-plane running from the superior (-40 mm) to the inferior (+40 mm) portion of the target. Along the $x$-plane there is a well-defined dose disagreement at the centre of the profile, from approximately -10 mm to +10 mm. Along the $y$ plane good agreement exists within the central portion of the dose distributions but discrepancies arise at approximately -25 mm and +25 mm. These locations correspond to the superior and inferior portions of the target area, respectively. Orthogonal profiles are shown only at this one location but the results are representative of the level of agreement between the measured and calculated distributions.

Aside from the discrepancies along the superior and inferior bands of the target region, good dose agreement generally exists between the two distributions, as validated with the results of the $y$ distribution (Figure 6.5(e)). On the plot shown, any point in the distributions satisfying the $y$ criteria (a dose-difference of 3% and DTA of 3 mm for all plans shown) will have a $y$ less than or equal to one and will appear black. Any points not satisfying the criteria are plotted according to the gradient scale. For this target, the agreement within the central portion generally is acceptable and the discrepancies along the superior and inferior region evident with orthogonal profiles are also quantified with the $y$ analysis.

**Conventional IMRT (5.0 mm Width Beamlet): PTV 1**

The prostate plan was also created and delivered with conventional IMRT. The plan was optimized using the same target (PTV 1) and dose volume constraints but an actual linac, with a 5.0 mm leaf width MLC, was selected during the optimization. The measured and calculated dose distributions resulting from the conventional treatment plan are shown in Figures 6.6(a) and 6.6(b), respectively.

The measured and calculated distributions were compared using orthogonal profiles and $y$ analysis. The profiles were taken at the same location used for the RMLC plan. The profile along the $x$-plane, in Figure 6.6(c), runs from the left to right of the target and the profile along the $y$-plane, in Figure 6.6(d), runs from the superior to inferior part of the target. The conventional plan shows improved dosimetric agreement between the two distributions compared to the rotational plan and while dose discrepancies continue to exist along the inferior region, the disagreement is not as significant. It could be argued that, while the agreement between the two distributions is not ideal,
Figure 6.6: (a) The measured dose distribution for the prostate carcinoma plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is approximately 3.75 and 86.5% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
the conventionally-delivered plans show better agreement in the superior region than the RMLC plan. Agreement in the central area, outlined in the $\gamma$ profile, is still acceptable.

The $\gamma$ map in Figure 6.6(e) shows improved agreement between the measured and calculated distributions over the rotated IMRT plan. The $\gamma$ maps for the RMLC and conventional IMRT plans are displayed using the same gradient scale and, while dosimetric discrepancies are still apparent along the inferior portion of the target volume, the level of dissimilarity is reduced greatly compared to the rotational plan. The $\gamma$ value is approximately 3.75 in this region, compared to 13.1 for the rotational case in the same region.

Delivering an IMRT treatment with collimator rotation should yield superior dose distributions compared to delivering the same treatment conventionally (without MLC rotation). The dosimetric advantages of applying collimator rotation, including reduced interleaf leakage and transmission, reduced number of MU required for a delivery and inherently increased spatial accuracy, were outlined in Chapter 3. Yet for the prostate carcinoma plan delivered with RMLC and conventional IMRT, improved dosimetric accuracy in the rotated case is not apparent. One hypothesis for the lack in superior dosimetric agreement has been mentioned previously and involves the accuracy of the dose calculation algorithm. The CadPlan TPS used for all dose calculations has inherent limitations due to the shape of the dose kernel used. Improvements to the kernel have been made by Bergman et al. [10], however, these improvements have not been implemented into the clinically-used treatment planning systems. Therefore, while the RMLC plan is superior theoretically, the advantages of producing higher-resolution fluence maps are lost if a poor-accuracy dose calculation algorithm is used for the final dose calculation. Until significant improvements are made in the dose calculation algorithms used clinically, it is unlikely that RMLC IMRT will show substantial improvement over the conventional case.

In most treatment planning systems a fluence smoothing is applied along the $x$-direction of a fluence map to minimize large dose-gradients and fluctuations along the direction of leaf travel. In the CadPlan TPS no fluence smoothing is applied along the $y$-direction, perpendicular to leaf travel, therefore the probability of incurring large dose-gradients and spikes of dose in the dose distribution along the $y$-plane is increased. The diagrams in Figure 6.7 outline the dosimetric implications of using 5.0 mm and 2.5 mm width beamlets. In Figure 6.7(a) a step pattern corresponding to a 5.0 mm width beamlet is shown and Figure 6.7(b) indicates the expected measured and calculated...
Figure 6.7: (a) Schematic diagram showing a bar pattern produced with a 5.0 mm width beamlet. (b) Typical measured (film) and calculated (CadPlan) dose profiles for the 5.0 mm width beamlet. (c) Schematic diagram showing a bar pattern produced with a 2.5 mm width beamlet. (d) Typical measured and calculated dose profiles for the 2.5 mm width beamlet. In most cases the discrepancies in the peaks and valleys are more pronounced with the smaller beamlets used. The 2.5 mm width beamlet is used for the RMLC IMRT optimization.

(CadPlan) dose profiles across the beamlets. Additionally, Figures 6.7(c) and 6.7(d) show a step pattern corresponding to a 2.5 mm width beamlet and the expected measured and calculated dose profiles across the leaves, respectively. When wider beamlets are used to generate fluence distributions, the calculated dose profile will generally match the measured dose profile more closely than for the narrower, 2.5 mm width beamlets. This graphical representation does not always hold true, however these effects are evident in Figures 6.5(d) and 6.6(d) for the RMLC (2.5 mm width beamlet) and conventional (5.0 mm width beamlet) profiles, respectively. Therefore, if both the RMLC and conventional IMRT plans are optimized using the 5.0 mm width beamlets, a more appropriate comparison of dosimetric accuracy between the two delivery techniques can be made.
Figure 6.8: (a) Conventional IMRT uses 5.0×2.5 mm² width beamlets. (b) RMLC IMRT uses 2.5×2.5 mm² width beamlets. These diagrams depict how the beamlet size used for the optimization affects how the dose is deposited.

Another explanation concerning the discrepancies observed between the rotated and conventional IMRT plans, especially along superior and inferior target regions, is an extension to the beamlet-size problem. During an optimization, the objective is to meet the constraints of the plan such that the target receives a maximum dose and surrounding critical structures a minimal dose. A balance must be maintained between these two constraints. The optimization algorithm works such that each beamlet in the treatment area is considered individually, in terms of its weighting and how the dose deposited in each beamlet affects surrounding beamlets. When critical structures are located close to the target, which is often the case, beamlets in these areas cannot be weighted heavily and given a large dose because this will increase the dose to the nearby critical structure. If the dose to the target is not high enough, beamlets around the target that are not close to any critical structures will be weighted heavily to boost the dose to the target by way of primary and scattered radiation.

In cases where there are no critical structures directly above or below the target, as with the prostate carcinoma case shown, beamlets directly above (superior to) and below (inferior to) the target will receive a very large dose to boost the target dose. There are no critical structures nearby therefore there are no consequences for the optimization objective, however, when the dose distribution is calculated by the TPS, the effects of having a large dose directly outside the target are apparent. This is diagrammed in Figure 6.8(b) when 2.5×2.5 mm² beamlets are used during the optimization. The beamlets directly above and below the target may be receiving a very large dose in an attempt to boost the target dose up to its prescription value. No smoothing exists in the
y-direction, making large dose-spikes superior and inferior of the target more noticeable. When the 5.0×2.5 mm² beamlets are used during an optimization, these effects may not be as pronounced.

Consider the schematic in Figure 6.8(a) where 5.0 mm width beamlets are used. The beamlets are twice as wide and a margin exists around the superior and inferior regions of the target. The optimization algorithm only considers what a beamlet contains (target, critical structure, neither) and assumes the beamlet to be completely filled with that structure. Therefore, in this scenario, if the row of beamlets directly above and below the target are given a large dose to boost the target dose, the contribution of scatter and primary radiation will not affect the target dose as severely as in Figure 6.8(b) because of the margin area surrounding the target.

For the reasons described above it was decided to optimize a RMLC IMRT plan using the 5.0 mm width beamlets in an attempt to improve the agreement between measured and calculated dose distributions.

RMLC IMRT (5.0 mm Width Beamlet): PTV 1

To validate the hypothesis that the beamlet size used in the optimization affects the delivered dose distribution, fluence maps of the prostate target PTV 1 were modified to use the 5.0 mm width beamlet during the RMLC optimization instead of the 2.5 mm width beamlet.

The composite distribution was delivered to Kodak EDR2 film in the AVID phantom and the measured and calculated dose distributions are shown in Figures 6.9(a) and 6.9(b), respectively. Orthogonal profiles and the γ factor analysis were performed to analyze the dosimetric conformity of the two distributions. The dashed lines on Figures 6.9(a) and 6.9(b) define the locations of the orthogonal profiles and are at the same location as used for the previous two cases (RMLC plan with 2.5 mm width beamlet and conventional plan with 5.0 mm width beamlet). There is a significant improvement in dosimetric agreement between the two distributions along both the x- and y-planes, as shown in Figures 6.9(c) and 6.9(d), respectively, especially in the superior/inferior regions. The calculated dose in the central region is higher than the measured dose, but only by 4.2%, at most. The level of dosimetric agreement throughout the target area is representative of the results shown in these two orthogonal profiles.

In the resulting distribution, shown in Figure 6.9(e), a maximum γ value of 4.0 is located in the central portion of the target region. Agreement in the previously prob-
Figure 6.9: (a) The measured dose distribution for a prostate carcinoma plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 4.0 and 77.4% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
lematic inferior region has been attained by using the standard 5.0 mm width beamlet when optimizing the target for RMLC delivery. Dosimetric agreement in the central target region is not as good as for the previous two cases (83.9% and 86.5%) as 77.4% of the pixels in the region of interest pass the $\gamma < 1$ criteria. The same $\gamma$ gradient scale is used in all three cases (2.5 mm RMLC, 5.0 mm conventional and 5.0 mm RMLC) to compare the $\gamma$ analysis results easily.

Considering the orthogonal profile and $\gamma$ distribution results, it is clear that optimizing a plan with 5.0 mm width beamlets for a RMLC delivery is superior to the 2.5 mm width beamlets. Until improvements are made in the accuracy of the dose calculation, the benefits of using a smaller beamlet size during the optimization process are lost. The results for the maximum $\gamma$ value and the percentage of pixels in the target area passing the $\gamma \leq 1$ criteria are summarized in Table 6.1.

**RMLC IMRT (5.0 mm Width Beamlet): PTV 2**

A second target volume, PTV 2, was contoured on the prostate carcinoma CT images and the plan was optimized using the standard linac with the 5.0 mm width beamlets for both the RMLC and conventional IMRT plans. The target volumes were optimized with the 5.0 mm width beamlet instead of the 2.5 mm width beamlet for the RMLC case because significant improvements in the dose distributions have been identified when optimizing with the wider beamlet. These improvements have been noted in Figure 6.9 as well as with the nasopharynx carcinoma case, which will be presented in Section 6.5.2.

The measured and calculated dose distributions for PTV 2 are presented in Figures 6.10(a) and 6.10(b), respectively, and orthogonal profiles taken along the $x$ and $y$ planes are shown in Figures 6.10(c) and 6.10(d), respectively. Along the $x$-plane, the profile is taken from the left (-40 mm) to the right (+40 mm) of the target. Small discrepancies exist at approximately -10 mm and +10 mm with a magnitude of approximately 4.4% and 5.1%, respectively. Along the $y$-plane, the profile runs from the superior (-40 mm) to the inferior (+40 mm) part of the target. In the central portion of the target the resolution of the dose calculation algorithm is not sensitive enough to distinguish the subtleties of small dose fluctuations evident in the measured distribution. This result is not an effect of the RMLC algorithm or delivery but of the TPS dose calculation.

A $\gamma$ factor analysis, shown in Figure 6.10(e), has a maximum $\gamma$ value of 4.7, located
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

Figure 6.10: (a) The measured dose distribution for the prostate carcinoma plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 2. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 4.7 and 79.1% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

along the superior ridge of the target, with 79.1% of the pixels in the region of interest passing the $\gamma \leq 1$ criteria. The majority of discrepancies are found in the central target area and, as stated previously, this is due mainly to the limitation of the dose calculation algorithm being unable to distinctly calculate doses.

**Conventional IMRT (5.0 mm Width Beamlet): PTV 2**

The treatment plan for the prostate carcinoma target PTV 2 was also created and delivered with conventional IMRT with the measured and calculated dose distributions shown in Figures 6.11(a) and 6.11(b), respectively. The horizontally-defined bands across the measured distribution are a direct result of MLC transmission and inter-leaf leakage, an effect not as noticeable in Figure 6.10 because the rotating collimator smears out these effects, for reasons described in Section 3.1.

Orthogonal profiles taken across the x- and y-planes are shown in Figures 6.11(c) and 6.11(d), respectively. At this location the agreement between the two distributions is acceptable, although the TPS does not model the step-like transmission and leakage effects seen in the measured distribution.

A $\gamma$ factor analysis, shown in Figure 6.11(e), has a maximum $\gamma$ value of 13.4 located along the inferior part of the target, however, the superior ridge of the target also shows disagreement. The $\gamma$ analysis results combined with the orthogonal profiles verifies the that the location of the superior/inferior discrepancies are non-negligible. While 85.3% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria, the maximum $\gamma$ value and values of disagreement in the centre of the target are higher than for the RMLC case (Figure 6.10(e)).

**RAO IMRT (2.5 mm Width Beamlet): PTV 1**

The final measurement for the prostate carcinoma considers the original target volume, PTV 1, optimized with rotating aperture optimization (RAO). RAO is a technique that combines direct aperture optimization (DAO) and collimator rotation. For the treatment cases shown previously the fluence maps were optimized using the rotating leaf motion calculator (RLMC) followed by a leaf sequencing step to generate deliverable fluence maps. MLC constraints are not accounted for during the dose optimization, resulting in leaf sequences that are generally inefficient and require a large number of segments and MU. An alternative approach to fluence-based optimization is DAO whereby the leaf positions and weights of the segmented fields are optimized directly. With this
Figure 6.11: (a) The measured dose distribution for the prostate carcinoma plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 2. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y plane orthogonal profiles. (e) The maximum $\gamma$ value is 13.4 and 85.3% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

technique, the leaf sequencing step is eliminated and the MLC constraints are included in the optimization. At the end of the optimization an actual, deliverable fluence map is produced. Many DAO algorithms can be found in the literature [62, 102, 103], but the method used in this thesis utilizes a simulated-annealing technique to optimize the segment shapes and weights. Current research has combined MLC rotation with DAO to exploit the advantages of both techniques. This approach is referred to as RAO. Although the optimization techniques are different, the set-up, delivery of the plan to the phantom and the verification methods used remain the same for all cases shown. The 2.5 mm width beamlets were used in the RAO optimization.

The measured and calculated dose distributions for the RAO prostate carcinoma plan are shown in Figures 6.12(a) and 6.12(b), respectively. Compared to the results for PTV 1 (both RMLC and conventional IMRT) the dose distributions for the RAO plan are more homogeneous and do not have large dose-gradient areas. This is because the apertures making up each subfield are much larger than those created from both the RMLC and conventional IMRT optimization algorithms. A minimum aperture size is specified in the optimization algorithm so that very small apertures can never be produced, therefore dosimetric error due to small aperture sizes will be reduced as will the penumbral effects associated with MLC leaf-tips (see Section 5.4).

Orthogonal profiles taken along the x- and y-planes are shown in Figures 6.12(c) and 6.12(d), respectively, using the same measurement location as in the RMLC and conventional IMRT plans (Figures 6.5–6.9). Dosimetric agreement along both planes is very good and has improved with respect to the results obtained from the orthogonal profiles taken for the RMLC (2.5 and 5.0 mm leaf width MLC) and conventional plans.

The areas of disagreement in the y map, Figure 6.12(e), are localized to the inferior portion of the target. This region of disagreement is just noticeable on the y-plane profile at a location of approximately +30 mm.

Prostate Carcinoma Summary

For each target and optimization method discussed, the maximum γ value and percentage of pixels passing the γ criteria are summarized in Table 6.1. Ionization chamber measurements were performed for each of the plans using the Wellhöfer Dosimetric IC 15 ion chamber (Wellhöfer, Schwarzenbruck, Germany) with the Victoreen Model 530 Precision Electrometer/Dosimeter (Elimpex Medizintechnik, Austria). The ion chamber, with an inner diameter of 0.6 cm and an active detection volume of 0.13 cm³, was
Figure 6.12: (a) The measured dose distribution for the prostate carcinoma plan delivered with RAO IMRT (2.5 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 6.7 and 89.0% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

### Table 6.1: Summary of the measurements taken for the prostate carcinoma. The percentage of pixels in the target region passing the $\gamma \leq 1$ criteria and the maximum $\gamma$ value are summarized for the two planning target volumes, PTV 1 and PTV 2.

<table>
<thead>
<tr>
<th>Target Volume</th>
<th>Delivery Method</th>
<th>Beamlet Width</th>
<th>Maximum Gamma</th>
<th>% Pixels in Target Passing $\gamma \leq 1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>PTV 1</td>
<td>Rotation (RLMC)</td>
<td>2.5 mm</td>
<td>13.1</td>
<td>83.9</td>
</tr>
<tr>
<td></td>
<td>Rotation (RLMC)</td>
<td>5.0 mm</td>
<td>4.0</td>
<td>77.4</td>
</tr>
<tr>
<td></td>
<td>Conventional</td>
<td>5.0 mm</td>
<td>3.75</td>
<td>86.5</td>
</tr>
<tr>
<td></td>
<td>Rotation (RAO)</td>
<td>2.5 mm</td>
<td>6.7</td>
<td>89.0</td>
</tr>
<tr>
<td>PTV 2</td>
<td>Rotation (RLMC)</td>
<td>5.0 mm</td>
<td>4.7</td>
<td>79.1</td>
</tr>
<tr>
<td></td>
<td>Conventional</td>
<td>5.0 mm</td>
<td>13.4</td>
<td>85.3</td>
</tr>
</tbody>
</table>

placed into the centre of the phantom to measure dose at the isocentre. The ion chamber dose values were compared with the CadPlan values at the isocentre and the results are summarized in Table 6.2. The RAO plan shows the most consistent agreement between the ion chamber and CadPlan dose points. From all of the results presented thus far, it can be concluded that the RMLC IMRT plans optimized with the 5.0 mm width beamlets yield results that are equivalent to the conventional IMRT plans. It could be argued that the RAO IMRT plan generates the best results because the maximum $\gamma$ value is similar to the RMLC and conventional cases but a larger portion of pixels in the gamma map pass the given criteria, all accomplished using the 2.5 mm width beamlets. Also, the ion chamber measurements agreed very well for this case and the number of apertures required for each field and the total number of MU needed is reduced dramatically from both the RMLC and conventional IMRT plans.

#### 6.5.2 Nasopharynx Carcinoma

A recurring nasopharynx carcinoma was planned using seven beams at gantry angles $114^\circ$, $76^\circ$, $38^\circ$, $0^\circ$, $322^\circ$, $284^\circ$ and $246^\circ$. The temporal lobes and brainstem were the dose-limiting structures and a total of 60 Gy in 30 fractions was prescribed. Two plans were produced from the CT data set by optimizing different target areas. Dose distributions calculated for a patient and phantom by the CadPlan TPS are shown for one target contour in Figures 6.13(a) and 6.13(b), respectively. In the following discussion the plans will be denoted as PTV 1 and PTV 2.
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

<table>
<thead>
<tr>
<th>Target Volume</th>
<th>Delivery Method</th>
<th>Ion Chamber Dose (cGy)</th>
<th>Calculated Dose, CadPlan (cGy)</th>
<th>Absolute Difference (%)</th>
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</thead>
<tbody>
<tr>
<td>PTV 1</td>
<td>RMLC (2.5mm)</td>
<td>161.9</td>
<td>171.6</td>
<td>5.6</td>
</tr>
<tr>
<td></td>
<td>RMLC (5mm)</td>
<td>170.5</td>
<td>174.4</td>
<td>2.2</td>
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<tr>
<td></td>
<td>Conventional (5mm)</td>
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<td>175.6</td>
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<td></td>
<td>RAO (2.5mm)</td>
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<td>185.2</td>
<td>0.8</td>
</tr>
<tr>
<td>PTV 2</td>
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<td>177.0</td>
<td>8.4</td>
</tr>
<tr>
<td></td>
<td>Conventional (5mm)</td>
<td>175.2</td>
<td>190.8</td>
<td>8.2</td>
</tr>
</tbody>
</table>

Table 6.2: Summary of the ion chamber dose measurements for the prostate carcinoma along with the isocentre doses of the calculated (CadPlan) distribution and the absolute differences in percent.

(a) (b)

Figure 6.13: (a) The dose distribution calculated in a patient for a nasopharynx carcinoma. (b) The distribution calculated in the phantom for the same carcinoma.

RMLC IMRT (2.5 mm Width Beamlet): PTV 1

The nasopharynx carcinoma plan optimized to the target volume PTV 1 was created and delivered with RMLC IMRT using 2.5 mm width beamlets during the optimization. The measured and calculated composite dose distributions are shown in Figures 6.14(a) and 6.14(b), respectively, and as with the prostate carcinoma plan optimized with the smaller beamlets, dose discrepancies exist primarily along the superior and inferior regions of the target volume. Orthogonal profiles and the $\gamma$ distribution highlight the magnitude of these differences.

Orthogonal profiles taken across the x- and y-planes are shown in Figures 6.14(c) and 6.14(d), respectively. Good dosimetric agreement exists between the measured...
Figure 6.14: (a) The measured dose distribution for the nasopharynx carcinoma plan delivered with RMLC IMRT (2.5 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 19.2 and 80.0% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
and calculated distributions along the x-plane profile until approximately -7 mm. At this point a discrepancy of approximately 9.2% exists between the measured and calculated distributions, with the calculated distribution over-estimating the dose. The profile along the y-plane shows acceptable dose agreement until approximately +25 mm, where the dose of the measured distribution is under-estimated by approximately 10.8%. The location, corresponding to the inferior target region, is a problematic region reminiscent of the prostate carcinoma plan shown in Figure 6.5. Minor fluctuations throughout the central portion of the target are most likely due to the poor accuracy of the dose calculation algorithm.

The $\gamma$ map, Figure 6.14(e), graphically presents the calculated agreement between the two distributions. The maximum $\gamma$ value, 19.2, is located along the inferior portion of the map with the majority of values in that region being 11.0 to 13.0. While these values are still high, a large fraction (80%) of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.

**Conventional IMRT (5.0 mm Width Beamlet): PTV 1**

The nasopharynx carcinoma plan for PTV 1 was also created and delivered using conventional IMRT with the 5mm width beamlets used during the optimization. The measured and calculated dose distributions from the treatment plan are shown in Figures 6.15(a) and 6.15(b), respectively. The effects of interleaf leakage and transmission for the conventional IMRT delivery are well-defined on the measured distribution, compared to the RMLC case in Figure 6.14(a). The MLC was not rotating during the treatment delivery therefore radiation leakage through the leaves was consistently at the same location. For this treatment plan, visible improvements arising from collimator rotation are evident.

The two distributions are dosimetrically compared using orthogonal profiles taken along the x- and y-planes, as shown in Figures 6.15(c) and 6.15(d), respectively. The dosimetric trend along the x-plane is virtually identical to the results obtained for the RMLC case (Figure 6.14(c)), with the dose disagreement at -7 mm still apparent. However, the measured dose is under-estimated by approximately 6.4% only, as opposed to 9.2% at the same location in the RMLC case. Similar results exist along the y-profile with the measured dose in the inferior region of the distribution, at a location of +20 mm, being greater than the calculated dose by approximately 6.2%. The dosimetric agreement has improved, compared to the RMLC case which showed 10.8% differ-
Figure 6.15: (a) The measured dose distribution for the nasopharynx carcinoma plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 10.0 and 82.9% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
The effects of interleaf leakage and transmission are noticeable in Figure 6.15(d) where step-like ridges along the y-profile correspond to the width of the MLC leaves. The y-plane profile in the RMLC case (Figure 6.14(d)) is noticeably smoother as these effects have been spread out from MLC rotation.

The y distribution, shown in Figure 6.15(e), has a maximum γ value of 10.0, located along the inferior portion of the target, and 82.9% of the pixels in the region of interest pass the γ ≤ 1 criteria. The same gradient scale is used for both the RMLC and conventional plans so a direct comparison can be made. The RMLC and conventional plans both yield similar γ values in the centre of the target but along the inferior region in the conventional plan, the maximum γ value is lower than the RMLC plan by almost half. A reduced γ value corresponds to improved dosimetric agreement between the calculated and measured dose distributions. The percentage of pixels in the region of interest passing the γ ≤ 1 criteria differs only by 2.5% between the RMLC and conventional plans, which is insignificant considering that the estimated uncertainty in film dosimetry with Kodak EDR2 film is on the order of 3% [23, 24, 25]. However, the reduction in the maximum γ value between the RMLC and conventional plans is significant. This effect, as stated previously, is due most likely to the size of the beamlets used for the optimization (5.0 mm for the conventional and 2.5 mm for the rotational). To validate this hypothesis the nasopharynx carcinoma for this target volume was re-optimized for IMRT delivery with a rotating collimator using the 5.0 mm width beamlets.

**RMLC IMRT (5.0 mm Width Beamlet): PTV 1**

A second plan was created for the nasopharynx target PTV 1 using the 5 mm width beamlets during the RMLC optimization. The reason for this is based on the discussion introduced in Section 6.5.1 regarding beamlet sizes used in the optimization procedure. The measured and calculated dose distributions delivered to Kodak EDR2 film in the AVID phantom are shown in Figures 6.16(a) and 6.16(b), respectively with the dashed lines on the two distributions indicating the location across which the orthogonal profiles are taken. The profiles across the x- and y-planes are shown in Figures 6.16(c) and 6.16(d), respectively, with the x-plane profile taken from left to right of the target and the y-plane profile taken from the superior to inferior part of the target. While good dosimetric agreement exists along the x-plane, large dose discrepancies are apparent on the y-profile at approximately -20 mm and +10 mm. At these locations the measured doses are 14.1% and 7.4% higher than the calculated doses, respectively.
Figure 6.16: (a) The measured dose distribution for the nasopharynx carcinoma plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum \( \gamma \) value is 10.2 and 72.7% of the pixels in the region of interest pass the \( \gamma \leq 1 \) criteria.
Chapter 6. Dosimetric Treatment Verification Using RMLC IMRT

The $\gamma$ factor analysis shown in Figure 6.16(e) displays a maximum $\gamma$ value of 10.2 with 72.7\% of the pixels in the region of interest passing the $\gamma \leq 1$ criteria. The dosimetric agreement in the central area has decreased compared with the 2.5 mm RMLC plan as fewer pixels are passing the $\gamma$ criteria, however, the maximum $\gamma$ value has been reduced by almost half, which is important. Of the pixels that do not pass the $\gamma$ criteria, the majority range from 2.0 to 4.0, which is consistent with previous results. Hence, it appears that optimizing RMLC plans with the 5.0 mm width beamlets instead of the 2.5 mm width beamlets yield more acceptable results.

RMLC IMRT (5.0 mm Width Beamlet): PTV 2

A second target volume, PTV 2, was contoured onto the existing nasopharynx carcinoma CT slices and the target was optimized using the 5.0 mm width beamlets for the RMLC IMRT plans. The measured and calculated distributions are shown in Figures 6.17(a) and 6.17(b), respectively. Orthogonal profiles, taken across the x- and y-planes at the locations indicated by the dashed lines on the composite distributions, are found in Figures 6.17(c) and 6.17(d), respectively. The doses in the calculated distribution are generally under-estimated in the peaks along both planes and over-estimated in the valleys, an effect due mainly to the limitations of the dose calculation algorithm.

The $\gamma$ factor analysis, Figure 6.17(e), shows a maximum $\gamma$ value of 14.4 with 83.4\% of the pixels in the region of interest passing the $\gamma \leq 1$ criteria. The dosimetric agreement in both the orthogonal profiles and $\gamma$ analysis is similar to the other treatment cases shown, for both the prostate and nasopharynx carcinoma. The $\gamma$-gradient-scale for the RMLC and conventional $\gamma$ maps is the same for comparison purposes. In the centre of the target, the $\gamma$ values are approximately 2.0 to 5.0 and are consistent with the other $\gamma$ maps shown.

Conventional IMRT (5.0 mm Width Beamlet): PTV 2

The RMLC plan for PTV 2 can be compared to the conventional IMRT plan for the same target. The conventional plan was delivered to film in the AVID phantom and the measured and calculated dose distributions are shown in Figures 6.18(a) and 6.18(b), respectively. The effects of transmission and interleaf leakage through the MLC leaves are apparent in the measured distribution. Orthogonal profiles along the x- and y-planes, shown in Figures 6.18(c) and 6.18(d), respectively, were taken at the same location as for the RMLC plan to enable a direct comparison. Dosimetric discrepancies along the
Figure 6.17: (a) The measured dose distribution for the nasopharynx carcinoma plan delivered with RMLC IMRT (5.0 mm width beamlet) for target PTV 2. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 14.4 and 83.4% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
Figure 6.18: (a) The measured dose distribution for the nasopharynx carcinoma plan delivered with conventional IMRT (5.0 mm width beamlet) for target PTV 2. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 21.2 and 67.4% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
profiles exist at the same location for both the conventional and RMLC plans, however, the dose variations are larger for the conventional IMRT plan. When the dose difference is maximal, the measured dose is under-estimated by almost 20% with the majority of discrepancies along these profiles on the order of 10 – 15%. It is clear that the measured doses of the conventional plan are higher than for the RMLC plan. RMLC plans have two advantages: they usually require a lower number of MU and also elicit reduced interleaf leakage and transmission effects.

The discrepancies are also depicted on the $\gamma$ map in Figure 6.18(e). Compared to the RMLC case for this target, the maximum $\gamma$ value is much greater (21.2 versus 14.4) and the percentage of pixels passing the $\gamma$ criteria is much lower (67.4% versus 83.4%). In the central target region the $\gamma$ values are approximately 3.0 to 7.0. For target PTV 2, the RMLC IMRT appears to have produced a better plan than the conventional IMRT.

**RAO IMRT (2.5 mm Width Beamlet): PTV 1**

The nasopharynx carcinoma plan was optimized also to PTV1 using RAO. The plan was delivered to Kodak EDR2 film in the AVID phantom with the measured and calculated distributions shown in Figures 6.19(a) and 6.19(b), respectively. Orthogonal profiles taken along the $x$- and $y$-planes are shown in Figures 6.19(c) and 6.19(d), respectively. Good agreement exists throughout both profiles with minor discrepancies appearing where the dose calculation algorithm is not sensitive enough to properly model small dose fluctuations.

The $\gamma$ factor analysis in Figure 6.19(e) displays a maximum $\gamma$ value of 11.8 with 70.9% of the pixels in the target area passing the $\gamma \leq 1$ criteria. Hot-spot regions localized on the $\gamma$ map are evident on the composite distributions where, in certain regions, the measured dose is higher than the calculated dose. The dose throughout the distributions is more homogeneous than for the RMLC and conventional distributions because the apertures used to generate the RAO distributions are much larger.

**Nasopharynx Carcinoma Summary**

For each target and optimization method discussed for the nasopharynx carcinoma, the maximum $\gamma$ value and percentage of pixels passing the $\gamma$ criteria are summarized in Table 6.3. Ionization chamber measurements for each plan were obtained and the results summarized in Table 6.4. The same ion chamber and electrometer used for the prostate carcinoma plans were used for these measurements with the ion chamber
Figure 6.19: (a) The measured dose distribution for the nasopharynx carcinoma plan delivered with RAO IMRT (2.5 mm width beamlet) for target PTV 1. (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum $\gamma$ value is 11.8 and 70.9% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
Table 6.3: Summary of the measurements taken for the nasopharynx carcinoma. The percentage of pixels in the target region passing the $\gamma \leq 1$ criteria and the maximum $\gamma$ values are summarized for the planning target volumes, PTV 1 and PTV 2.

Table 6.4: Summary of the ion chamber dose measurements for the nasopharynx carcinoma along with the isocentre doses of the calculated (CadPlan) distribution and the absolute difference (%) between the two values.
6.5.3 Geometric C-shape

The complex geometric c-shape is a challenging plan to deliver and requires IMRT because of the location of the critical structure with respect to the target. The plan used seven equidistant beams placed at gantry angles 155°, 103°, 52°, 0°, 309°, 258° and 206°. A total dose of 60 Gy in 30 fractions was prescribed.

This plan, optimized using RAO, was slightly different than the prostate and nasopharynx carcinoma cases in that the contouring of the target and critical structures was done directly on the CT images of the phantom, rather than on actual patient anatomy. Kodak EDR2 film was placed in the axial plane of the AVID IMRT verification phantom and the measured and calculated dose distributions are shown in Figures 6.21(a) and 6.21(b), respectively. The outline of the c-shape is visible in both distributions. Orthogonal profiles taken along the $x$ and $y$ planes are shown in Figures 6.21(c) and 6.21(d), respectively. There are minor dose discrepancies between the two distributions along the $x$-plane which are most likely due to effects of the inadequate modeling and poor accuracy of the dose calculation algorithm. The profile along the $y$-plane indicates minor discrepancies with the largest error occurring at approximately +40 mm. This correlates to the location just inferior to the critical structure region.

The $\gamma$ analysis, shown in Figure 6.21(e), graphically presents the level of agreement between the measured and calculated dose distributions. The two distributions agree to a high level, as 90% of the pixels in the region of interest pass the $\gamma \leq 1$ criterion and...
Table 6.5: Summary of the measurements taken for the complex geometric c-shape. The percentage of pixels in the target region passing the $\gamma \leq 1$ criteria is very high.

Of all the targets considered (prostate, nasopharynx and c-shape) and optimization modalities used (RMLC, conventional and RAO), the geometric c-shape generated with RAO yields the best agreement between measured and calculated distributions and the best overall quality in planning results.
Figure 6.21: (a) The measured dose distribution for the geometric c-shape plan delivered with RAO IMRT (2.5 mm width beamlet). (b) The calculated distribution for the same plan. (c) X-plane orthogonal profiles. (d) Y-plane orthogonal profiles. (e) The maximum dose value is approximately 2.5 and 90.0% of the pixels in the region of interest pass the $\gamma \leq 1$ criteria.
7.1 Summary and Conclusion

The two aims of this thesis were to evaluate whether IMRT plans incorporating collimator rotation could be delivered on a standard radiotherapy linac and whether the dose distributions produced with RMLC IMRT were of comparable accuracy to IMRT plans delivered conventionally.

The first part of this thesis involved testing the capability of the Dynamic Beam Delivery toolbox to control collimator rotation independently. The DBD toolbox was programmed to rotate the collimator to precisely-defined angles and results showed this could be achieved to within ±0.5°, a value within the accuracy specifications of the manufacturer. It was also shown that gantry sag did not affect collimator rotation, with similar dosimetric results obtained with the gantry at 0° and 270°. The second part of this thesis focused on using the RMLC algorithm to model fluence and calculate dose of rotated and non-rotated apertures. Dosimetric analysis of non-rotated fields provided information regarding the limitations of MLC modeling and dose resolution, and results from the rotated-square pattern confirmed that the RMLC algorithm could generate deliverable rotated fluence maps accurately. It was also shown that rotated fluence maps could be produced for clinically-relevant treatment sites and, after incorporating a 1.4 mm leaf gap requirement, that a good agreement could be obtained between the measured and calculated dose distributions for single field deliveries.

Composite dose distributions were analyzed to evaluate whether treatment plans optimized for RMLC IMRT delivery were comparable to plans optimized for conven-
Chapter 7. Conclusion

Traditional IMRT delivery. The prostate, nasopharynx and c-shape structures (typical sites treated with IMRT) were considered. RMLC IMRT plans were optimized using the 2.5 mm and 5.0 mm width beamlets, the conventional plans with the 5.0 mm width beamlets only and the RAO plans with the 2.5 mm width beamlets only. For the prostate and nasopharynx carcinomas, dosimetric agreement between the measured and calculated dose distributions obtained for the conventional IMRT plans was better than for the RMLC IMRT plans optimized with 2.5 mm width beamlets. However, when the same target volumes were optimized for RMLC IMRT with 5.0 mm width beamlets, dosimetric agreement was comparable between the composite distributions generated with RMLC and conventional IMRT.

In determining whether IMRT plans delivered with collimator rotation were equivalent to plans delivered conventionally, the following information was considered: orthogonal profiles taken throughout the dose distributions, a gamma factor analysis for each plan, the maximum gamma value, the percentage of pixels in each target passing the $\gamma \leq 1$ criteria, results from ion chamber measurements and transmission and interleaf leakage effects. For the prostate carcinoma plans, the results for PTV 1 exhibited similar maximum gamma values for the conventional and RMLC IMRT plans optimized with 5.0 mm width beamlets, however, the conventional plan had a greater number of pixels in the target area passing the gamma criteria. Orthogonal profile results for both treatment deliveries were similar. For prostate target PTV 2, the RMLC plan had fewer pixels agreeing with the gamma criteria than the conventional plan (by 7.3%) but the maximum gamma value for the conventional plan was considerably higher (~65%). Similar results were shown for the two nasopharynx targets. For PTV 1, the maximum gamma value was similar in both the RMLC and conventional plans but fewer pixels in the target area of the RMLC plan passed the gamma criteria. For PTV 2, the RMLC results showed a decreased maximum gamma value (~32%) as well as a higher percentage of pixels (~24%) in the target area passing the gamma criteria. Interleaf leakage and transmission effects were visible in the conventional IMRT plans and were reduced and smoothed out substantially when collimator rotation was incorporated, signalling a dosimetric improvement due to collimator rotation.

The prostate and nasopharynx carcinomas and the geometric c-shape target volumes were also optimized using RAO and, in all three cases, the agreement between the measured and calculated distributions was as good as or better than both the RMLC and conventionally-delivered IMRT plans. The benefit of using RAO is that less segments are required for each field, translating into fewer MU required for each treat-
Chapter 7. Conclusion

ment. Clinically, these advantages are very important. Also, a minimum aperture size is defined in the optimization algorithm so apertures of resulting sub-fields will, very often, be larger than those generated with either RMLC or conventional IMRT plans. This results directly in smaller dosimetric errors due to the rounded MLC leaf tips. The CadPlan TPS was used to calculate the final dose distributions of all IMRT plans, hence it is important to recognize that the advantages and benefits of RAO are apparent despite the limitations imposed by the dose calculation algorithm.

There is a degree of subjectiveness in determining the acceptability of an IMRT plan and the results of this thesis indicate that the RMLC IMRT plans optimized with a 5.0 mm width beamlet yield distributions that are dosimetrically similar to conventional IMRT plans. For the plans considered, it is apparent that target volumes optimized with RAO are dosimetrically superior to those optimized with either RMLC or conventional IMRT. Therefore it has been shown that treatment plans generated with collimator rotation yield dose distributions that are as good or better than conventional IMRT plans, depending on the optimization method used.

The main limitation hindering better results is the CadPlan TPS dose calculation algorithm used to calculate the final dose distributions. Until fundamental changes in the dose calculation algorithm are made, it is unlikely that RMLC IMRT will show substantial improvement over the conventional case. Modifications would involve adapting the dose spread kernel, as it is has already been shown that sharper dose kernels improve the accuracy of the dose calculation. Also, the MLC needs to be modeled more accurately. For MLCs having rounded leaf tips implementing a leaf gap shift approximates additional transmission through the tips, although, the shift does not model properly the dose effects resulting from the rounded tips. As well, dose contributions due to extrafocal scattered radiation may need to be modeled more explicitly in the dose calculation to include scattered radiation from the sides of the linac head.

7.2 Future Work

Observations regarding the limitations of conventional IMRT highlighted the effects of insufficient accuracy of the dose calculation algorithm and insufficient modeling of leaf transmission, rounded MLC leaf tips and beam-scatter effects. Regardless of whether collimator rotation is incorporated into IMRT delivery or not, the limitations associated with conventional IMRT must also be addressed.

For collimator rotation to be incorporated into clinical IMRT treatment deliveries,
Chapter 7. Conclusion

The problems and limitations of RMLC IMRT need to be considered in further detail. One of the primary restrictions of this technique is the dose calculation algorithm used. The accuracy of the algorithm is not sensitive enough to identify the benefits of incorporating collimator rotation into an IMRT treatment. Research is currently in progress to develop a new dose calculation platform to further improve inaccuracies in calculations.

Advancements are being made in RAO IMRT. Multiple plans are being generated and the delivery and subsequent analysis of the dose distributions produced will increase the statistical significance of the results and provide further insight into the potential benefits associated with collimator rotation.

Finally, substantial improvements are being made to reduce the optimization time required to complete a full Monte Carlo-calculated IMRT plan. Using Monte Carlo simulations, RMLC and conventional IMRT plans will eliminate the need for the dose calculation algorithm used by standard treatment planning systems and will represent more accurately the full potential of collimator rotation. A MLC with the 2.5 mm width beamlet could be modeled explicitly using Monte Carlo to provide meaningful calculated dose distributions to be compared against the measured, film, distributions. Dosimetric accuracy problems apparent throughout this thesis would no longer be the main concern.

Preliminary results have shown that if the rotating MLC technique is to be applied clinically then using the RAO optimization platform will be most effective dosimetrically, and provide the greatest benefit in treatment delivery.


Bibliography


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