QUEENSLAND UNIVERSITY OF TECHNOLOGY

MONTE CARLO SIMULATIONS OF DYNAMIC RADIOTHERAPY TREATMENTS

Muhammad Basim Kakakhel

Submitted to the Faculty of Science and Technology, Queensland University of Technology in fulfilment of the requirements of the degree of Doctor of Philosophy

February, 2012
Keywords

Enhanced dynamic wedges

Gel dosimetry

Interplay effects

X-ray computed tomography

Radiation therapy

Monte Carlo simulations

BEAMnrc

DOSXYZnrc
Abstract

The effects of tumour motion during radiation therapy delivery have been widely investigated. Motion effects have become increasingly important with the introduction of dynamic radiotherapy delivery modalities such as enhanced dynamic wedges (EDWs) and intensity modulated radiation therapy (IMRT) where a dynamically collimated radiation beam is delivered to the moving target, resulting in dose blurring and interplay effects which are a consequence of the combined tumor and beam motion. Prior to this work, reported studies on the EDW based interplay effects have been restricted to the use of experimental methods for assessing single-field non-fractionated treatments. In this work, the interplay effects have been investigated for EDW treatments. Single and multiple field treatments have been studied using experimental and Monte Carlo (MC) methods.

Initially this work experimentally studies interplay effects for single-field non-fractionated EDW treatments, using radiation dosimetry systems placed on a sinusoidaly moving platform. A number of wedge angles (60°, 45° and 15°), field sizes (20 × 20, 10 × 10 and 5 × 5 cm²), amplitudes (10-40 mm in step of 10 mm) and periods (2 s, 3 s, 4.5 s and 6 s) of tumor motion are analysed (using gamma analysis) for parallel and perpendicular motions (where the tumor and jaw motions are either parallel or perpendicular to each other). For parallel motion it was found that both the amplitude and period of tumor motion affect the interplay, this becomes more prominent where the collimator tumor speeds become identical. For perpendicular motion the amplitude of tumor motion is the dominant factor where as varying the period of tumor motion has no observable effect on the dose distribution. The wedge angle results suggest that the use of a large wedge angle generates greater dose variation for both parallel and perpendicular motions. The use of small field size with a large tumor motion results in the loss of wedged dose distribution for both parallel and perpendicular motion. From these single
field measurements a motion amplitude and period have been identified which show the poorest agreement between the target motion and dynamic delivery and these are used as the ‘worst case motion parameters’.

The experimental work is then extended to multiple-field fractionated treatments. Here a number of pre-existing, multiple-field, wedged lung plans are delivered to the radiation dosimetry systems, employing the worst case motion parameters. Moreover a four field EDW lung plan (using a 4D CT data set) is delivered to the IMRT quality control phantom with dummy tumor insert over four fractions using the worst case parameters i.e. 40 mm amplitude and 6 s period values. The analysis of the film doses using gamma analysis at 3%-3mm indicate the non averaging of the interplay effects for this particular study with a gamma pass rate of 49%.

To enable Monte Carlo modelling of the problem, the DYNJAWS component module (CM) of the BEAMnrc user code is validated and automated. DYNJAWS has been recently introduced to model the dynamic wedges. DYNJAWS is therefore commissioned for 6 MV and 10 MV photon energies. It is shown that this CM can accurately model the EDWs for a number of wedge angles and field sizes. The dynamic and step and shoot modes of the CM are compared for their accuracy in modelling the EDW. It is shown that dynamic mode is more accurate. An automation of the DYNJAWS specific input file has been carried out. This file specifies the probability of selection of a subfield and the respective jaw coordinates. This automation simplifies the generation of the BEAMnrc input files for DYNJAWS.

The DYNJAWS commissioned model is then used to study multiple field EDW treatments using MC methods. The 4D CT data of an IMRT phantom with the dummy tumor is used to produce a set of Monte Carlo simulation phantoms, onto which the delivery of single field and
multiple field EDW treatments is simulated. A number of static and motion multiple field EDW plans have been simulated. The comparison of dose volume histograms (DVHs) and gamma volume histograms (GVHs) for four field EDW treatments (where the collimator and patient motion is in the same direction) using small (15°) and large wedge angles (60°) indicates a greater mismatch between the static and motion cases for the large wedge angle.

Finally, to use gel dosimetry as a validation tool, a new technique called the ‘zero-scan method’ is developed for reading the gel dosimeters with x-ray computed tomography (CT). It has been shown that multiple scans of a gel dosimeter (in this case 360 scans) can be used to reconstruct a zero scan image. This zero scan image has a similar precision to an image obtained by averaging the CT images, without the additional dose delivered by the CT scans.

In this investigation the interplay effects have been studied for single and multiple field fractionated EDW treatments using experimental and Monte Carlo methods. For using the Monte Carlo methods the DYNIAWS component module of the BEAMnrc code has been validated and automated and further used to study the interplay for multiple field EDW treatments. Zero-scan method, a new gel dosimetry readout technique has been developed for reading the gel images using x-ray CT without losing the precision and accuracy.
# Table of Contents

Keywords ............................................................................................................................................ ii  
Abstract ............................................................................................................................................... iii  
Table of Contents ............................................................................................................................. vi  
List of Figures ...................................................................................................................................... xi  
List of Tables ....................................................................................................................................... xv  
List of Abbreviations .......................................................................................................................... xvi  
List of Publications ............................................................................................................................ xix  
Acknowledgements ............................................................................................................................ xxi  

Chapter 1 : Introduction ..................................................................................................................... 1  

Chapter 2 Background ......................................................................................................................... 4  

2.1 Cancer and Radiotherapy .............................................................................................................. 4  

2.2 Linear accelerators and EBRT techniques .................................................................................. 6  

2.3 Wedge filters ............................................................................................................................... 10  

2.5 Monte Carlo modelling .................................................................................................................. 12  

2.5.2 Monte Carlo modelling of dynamic radiotherapy treatments ................................................. 14  

2.5.3 Modelling the Interplay effects in dynamic radiotherapy delivery ......................................... 16  

2.6 Treatment verification ................................................................................................................... 20  

2.7 Investigative aims ......................................................................................................................... 24
Chapter 3: Interplay for single field enhanced dynamic wedge deliveries ........................................ 25

3.1 Introduction ................................................................................................................................... 25

3.2 Methods and Materials .................................................................................................................. 25

3.2.1 Equipment .................................................................................................................................. 25

3.2.2 Effect of amplitude and period variation .................................................................................. 27

3.2.3 Effect of wedge angle and field size ....................................................................................... 29

3.3 Results .......................................................................................................................................... 29

3.3.1 Effect of amplitude and period for parallel motion ................................................................. 29

3.3.2 Amplitude and period variation for perpendicular motion .................................................. 31

3.3.3 The effect of wedge angle and field size for parallel motion ................................................. 33

3.3.4 The effect on a multiple fraction single field delivery ............................................................ 33

3.3.5 The effect of wedge angle and field size for perpendicular motion ....................................... 35

3.4 Discussion and summary ............................................................................................................. 36

Chapter 4 Interplay for fractionated multiple-field enhanced dynamic wedge deliveries ............ 39

4.1 Introduction ................................................................................................................................... 39

4.2 Methods and Materials .................................................................................................................. 39

4.2.1 Overview .................................................................................................................................. 39

4.2.2 Existing patient plans delivered to MapCHECK2 and EBT2 films ...................................... 42

4.2.3 A multi-fractionated 4D CT planned patient delivery to EBT2 films ..................................... 43

4.3 Results .......................................................................................................................................... 47

4.3.1 Patient plans delivered to MapCHECK2 and EBT2 .............................................................. 47
Chapter 5: Validation of the Monte Carlo model using DYNJAWS component module

5.1 Introduction
5.1.1 Background
5.1.2 Input file for DYNJAWS

5.2 Methods and Materials
5.2.1 Experimental measurements
5.2.2 Probability calculation
5.2.3 Automation of the input file generation
5.2.4 MC simulations
5.2.5 Back scatter correction

5.3 Results
5.3.1 Input file Generation
5.3.2 Validation

Chapter 6: Modelling Enhanced Dynamic Wedge Interplay using Monte Carlo methods

6.1 Introduction
6.2 Methods and Materials
6.2.1 Monte Carlo simulation of the treatment head
6.2.2 Phantom dose calculation
6.2.3 Simulating individual respiratory phases using DYNJAWS
6.2.4 Single field EDW deliveries ................................................................. 75
6.2.5 Comparison of four field EDW treatment for 15 and 60 wedge angles......... 76
6.2.6 Comparison of large and small wedge angles ........................................ 77
6.2.7 Analysis ............................................................................................... 77
6.3 Results .................................................................................................. 78
6.3.1 Individual respiratory phases ................................................................ 78
6.3.2 Single field deliveries ........................................................................ 78
6.3.3 Four field deliveries ........................................................................... 82
6.3.4 Comparison of large and small wedge angles ....................................... 83
6.4 Discussion and Conclusions .................................................................. 87

Chapter 7 : Improving the gel image readout using X-ray computed tomography (CT) .... 89
7.1 Introduction .......................................................................................... 89
7.2 Methods and Materials ........................................................................ 89
7.2.1 Gel preparation and irradiation ......................................................... 90
7.2.2 X-ray CT imaging ........................................................................... 91
7.2.3 Image analysis ................................................................................ 92
7.3 Results ................................................................................................ 94
7.3.1 Qualitative .................................................................................... 94
7.3.2 Quantitative ................................................................................ 96
7.2.3 Optimum number of scans for reconstruction .................................. 102
7.4 Zero-scan method extended to 3D image ............................................. 103
7.5 Discussion and summary ................................................................. 108

Chapter 8 Summary and discussion .................................................................. 111

8.1 Summary .......................................................................................... 111

8.2 Overall discussion and conclusions ......................................................... 115

8.3 Future Recommendations ........................................................................ 120

References .................................................................................................. 121
List of Figures

Figure 2.1: A Theratron Cobalt 60 machine ................................................................. 6

Figure 2.2: a) A Varian linear accelerator b) Block diagram of a linear accelerator head. ....... 7

Figure 2.3: Conventional radiation delivery showing a single field technique. .................... 8

Figure 2.4: a) Multileaf collimator b) More tightly conformed tumor dose as result of MLC use. ......................................................................................................................... 8

Figure 2.5: A comparison of conventional conformal and IMRT tumor coverage, a) Conventional radiotherapy b) Conformal radiotherapy c) IMRT delivery depicting the non uniform dose delivery to the tumor. .................................................................................. 9

Figure 2.6: Wedge dose distribution .................................................................................... 11

Figure 2.7: BEAMnrc GUI .......................................................................................... 13

Figure 2.8: DOSXYZnrc GUI. ......................................................................................... 14

Figure 2.9: An irradiated gel sample, the square field in the middle is visible. .................... 22

Figure 3.1: MapCHECK 2 diode array ............................................................................. 26

Figure 3.2: Respiratory motion platform .......................................................................... 26

Figure 3.3: Detector platform geometry ............................................................................ 28

Figure 3.4: Effect of varying the amplitude and period of motion for a 20 × 20 cm² field size, 45° EDW for parallel motion .......................................................................................... 30

Figure 3.5: Effect of varying the amplitude and period of motion for a 20 × 20 cm² field size, 45° EDW delivery for perpendicular motion ................................................................. 32

Figure 3.6: Field size and wedge angle comparison, with motion parameters of 40 mm and 6.0 s for parallel motion .............................................................................................. 35

Figure 3.7: Field size and wedge angle comparison with motion parameters of 40 mm and 6.0 s for perpendicular motion ...................................................................................... 36
Figure 4.1: Epson Perfection V700 flatbed scanner. ................................................................. 40
Figure 4.2: IMRT quality assurance phantom. ........................................................................ 41
Figure 4.3: a) Lung insert and tumour in the IMRT phantom b) Tumour film geometry. ...... 44
Figure 4.4: 4D CT acquisition setup..................................................................................... 44
Figure 4.5: Treatment planning screen shot. ......................................................................... 45
Figure 4.6: Treatment delivery setup..................................................................................... 47
Figure 4.7: Two lung and a breast plans delivered to MapCHECK 2 and EBT2. Motion parameters were 40 mm and 6.0 s ................................................................. 48
Figure 4.8: Planned lung patient (dose in Gy indicated by the color palette)....................... 50
Figure 5.1: Coordinate specification in DYNAJWS CM....................................................... 55
Figure 5.2: Experimental setup showing the MatriXX diode array with solid water placed on top. ...................................................................................................................... 56
Figure 5.3: Probability calculation flow chart....................................................................... 58
Figure 5.4: AUTODJAWS GUI .............................................................................................. 60
Figure 5.5: Input file generated by the AUTODJAWS code.................................................. 63
Figure 5.6: Comparison of measured and simulated wedged profiles for $20 \times 20 \text{ cm}^2$ and $10 \times 10 \text{ cm}^2$ field size at 100 cm SSD for four wedge angles at $d_{\text{max}}$ and 10 cm depth. ........... 66
Figure 5.7: Comparison of measured and simulated open field and wedged profiles at $d_{\text{max}}$. Both the simulated and measured doses are normalized to 100% at the centre of the field. .... 67
Figure 5.8: Comparison of dynamic and step-and-shoot mode for 6 MV at $d_{\text{max}}$. Both the simulated and measured doses are normalized to 100% at the centre of the field. .......... 67
Figure 6.1: 4D CT acquisition, respiratory trace of the moving phantom. ............................. 72
Figure 6.2: Motion of the phantom as a function of slice offset............................................. 73
Figure 6.3: Treatment planning snapshot for individual phase simulation.......................... 74
Figure 6.4: Treatment planning snapshot for 15° parallel motion. ........................................ 75
Figure 6.5: Treatment planning snapshot for 60º parallel motion. ..................................................... 76

Figure 6.6: Individual phase dose volume histogram................................................................. 79

Figure 6.7: Comparison of single field 15º and 60º treatments a) Dose volume histograms for 15º and 60º EDWs for parallel motion b) Dose volume histograms for 15º and 60º EDWs for perpendicular motion................................................................. 80

Figure 6.8: Gamma dose images 15º and 60º EDWs for parallel collimator motion. .......... 81

Figure 6.9: Gamma dose images 15º and 60º EDWs for perpendicular collimator motion. ... 81

Figure 6.10: Gamma dose volume histograms for 15º and 60º EDWs for parallel and perpendicular collimator motion................................................................. 82

Figure 6.11: Dose volume histograms for four field 15º and four field 60º EDWs comparing the static and motion images........................................................................ 83

Figure 6.12: Gamma volume histograms for four field 15º and four field 60º EDWs. ......... 84

Figure 6.13: Gamma dose images for four field 15º and four field 60º EDWs. ................. 85

Figure 6.14: Dose volume histograms for four fields large and four field small EDWs comparing the static and motion images. ......................................................... 85

Figure 6.15: Gamma volume histograms for four field large and four field small EDWs..... 86

Figure 6.16: Gamma dose images for four field 15º and four field 60º EDWs................. 86

Figure 7.1: Gel irradiation on the linear accelerator................................................................. 90

Figure 7.2: Irradiated gel showing the different ROIs as indicated by the arrows. .......... 91

Figure 7.3: CT scan setup for the irradiated gel................................................................. 92

Figure 7.4: Zero-scan reconstructed images........................................................................ 95

Figure 7.5: (a) HU value of a single pixel inside the irradiated ROI (686 cGy), over the 360 images. (b) HU value of a single pixel outside the irradiated ROIs, over the 360 gel images c) Averaged HU values within irradiated and non irradiated ROIs (121 pixels) with linear and exponential fits. ................................................................................. 98
Figure 7.6: Plot of mean HU versus dose in cGy for all types of fit. ................................. 101

Figure 7.7: Plot of average % error (ROI’s compared to the linear reconstructed image) versus the number of images used in image reconstruction.......................................................... 103

Figure 7.8: Irradiated gel container. .................................................................................. 105

Figure 7.9: Reconstructed images using the zero-scan method........................................... 106

Figure 7.10: Coronal ((a) and (b)), transverse ((c) and (d)) and sagittal ((e) and (f)) views, through the Zero-scan 3D image of the gel. ................................................................. 107
List of Tables

Table 3.1: Gamma analysis pass rates with a 3%-3 mm acceptance criteria for parallel and perpendicular deliveries........................................................................................................... 31

Table 4.1: Breast and lung patient treatment parameters................................................................. 43

Table 4.2: Lung patient # 3, treatment plan parameters. ................................................................. 46

Table 5.1: Comparison of gamma values for dynamic and step-and-shoot modes for 6 MV photon beams.......................................................................................................................... 68

Table 7.1: Analysis of linear and exponential fits. ........................................................................... 99

Table 7.2: Plan parameters for zero-scan extension treatment plan. ............................................ 104
**List of Abbreviations**

1D: One dimensional

2D: Two dimensional

3D: Three dimensional

3DCRT: Three dimensional conformal radiation therapy

4D: Four dimensional

AAA: Analytic anisotropic algorithm

CM: Component module

CT: Computed tomography

CTV: Clinical target volume

DAH: Dose area histogram

DIMRT: Dynamic intensity modulated radiation therapy

DVH: Dose volume histogram

EBRT: External beam radiation therapy

ECUT: Electron cut off

EDW: Enhanced dynamic wedge

EGS: Electron gamma shower

FWHM: Full width half maximum
GUI: Graphical user interface

GVH: Gamma volume histogram

IMAT: Intensity modulated arc therapy

IMRT: Intensity modulated radiation therapy

MC: Monte Carlo

MCDTK: Monte Carlo DICOM Tool Kit

MIP: Maximum intensity projection.

MLC: Multi leaf collimator

MRI: Magnetic resonance imaging

OCT: Optical computed tomography

OF: Output factors

PAGAT: Polyacrylamide gel dosimeter (with THPC as antioxidant)

PCUT: Photon cut off

PDD: Percentage depth dose

PET: Polyethylene terephthalate

PPS: Position probability sampling

PTV: Planning target volume

ROI: Region of interest
RPM: Real time position monitoring

SCS: Static component simulation

STT: Segmented treatment table
List of Publications


STATEMENT OF ORIGINAL AUTHORSHIP

The work contained in this thesis has not been previously submitted to meet requirements for an award at this or any other higher education institution. To the best of my knowledge and belief, the thesis contains no material previously published or written by another person except where due reference is made.

Signature: _______________________

Date: _______________________

xx
Acknowledgements

First of all I would like to thank Allah almighty the creator for giving me the ability to complete this project. Then my parents whose support and prayers have constantly encouraged me throughout this period to stay focused and dedicated. I would like to sincerely thank my wife Sumaira who has accompanied and helped me during these three very long years. How can I forget my little son; Hamza who was always there smiling and cheering me up after the long and never ending days at the university.

The first person in my university life to be thanked and acknowledged is my principal supervisor Dr. Jamie Trapp. Without indulging into any sort of flattery I have rarely met a person such as him who is always smiling, supportive and helping his students. As an international student usually adjustment at the beginning of the project is difficult especially if you have arrived with a family carrying a two months old baby. He has been consistently helping me out with experimental investigations thesis writing and overall project support. I really appreciate his support. One thing that I have learnt from Jamie is the quick response to emails. Usually his email response time is less than 30 minutes which is fantastic considering that besides supervising the PhD students he has a lot of other more important and interesting things to do.

I would also like to thank my associate supervisor Dr. Andrew Fielding. His clinical insight and expertise in Monte Carlo simulations has been really helpful. Andrew along with Jamie has been consistent in meeting me weekly. Andrew has always provided very useful clinically relevant feedback on all my write ups.

Dr. Tanya Kairn who is also my associate supervisor has been extremely helpful and supportive throughout the duration of the project. She has been very helpful, cheering and
encouraging. Always going out of the way and helping me out. To start with she has been assisting with the MC simulations. Later on she has been present at all experiments carried out in Premion Wesley hospital and simultaneously providing valuable feedback on my write ups. I am really grateful for all the help that she has extended over the years.

I would also like to specially thank Dr. Scott Crowe for his extensive support in Monte Carlo simulations and data analysis. Scott has always been able and willing to help. Scott has helped with BEAMnrc and DOSXYZnrc simulations by providing some of the input files and dose map analysis using his MCDTK software suite.

How can I forget my fellow colleagues who have encouraged and helped a lot throughout this time. Rodney Hewins, Michael Hirning, Sean Paul, Nora Tischler, Mathew Shortell, Martin Kurth, Eugene Tan, Juna Sathian and the list goes on. I would like to specially mention Shadi Khoei and Mohammad Al Roumi and Chris Poole. Shadi has helped me on a number of occasions in preparing the gels, besides that her usual morning chats have been inspirational. Mohammad Al Roumi has been very supportive in discussing various ideas related to Monte Carlo simulations.

Professor Christian Langton’s weekly tea gathering, during the initial years of my PhD was a very nice and memorable experience, where I got to know a lot of new students and academics and acquired ideas pertaining to my research. Similarly Jacqueline Broad has provided numerous supports in terms of administration matters etc.

The Physics departments at the Princess Alexandra, Premion Wesley and Mater hospitals are acknowledged for their support in terms of various experimental investigations. John Kenny has been very helpful sparing a lot of time for various experiments. Emmanuel Baveas, Katrina Seet, Paul Charles all have helped in the hospital.
There are number of people across QUT that deserve special mention. John Barrett at the Radiological lab, Simon Belton at the BEE workshops.

Computational resources and services used in this work were provided by the HPC and Research Support Group, Queensland University of Technology, Brisbane, Australia. Ashley Wright and David Warne from HPC have especially helped out with software installations and programming help.

Blake Walters and Dr. Frédéric Tessier at NRCC are acknowledged for their help with providing DYNJAWS component module before its final release in the BEAMnrc code.

Finally I would like to thank the Australian government for providing me with the prestigious IPRS scholarship and QUT for providing the living allowance.
Chapter 1: Introduction

Radiotherapy is a frequently used cancer treatment modality where ionizing radiation is employed to kill the cancer cells. The aim of radiotherapy is to deliver maximum dose to the tumor while minimizing the dose to the surrounding normal tissues.

External beam radiotherapy treatments can be delivered either statically, using conventional or three dimensional conformal radiotherapy (3D-CRT) techniques, or dynamically, using techniques such as enhanced dynamic wedges (EDWs) and dynamic intensity modulated radiotherapy (IMRT). In dynamic treatments one or more components of the radiation beam collimation system (orthogonal jaws or multi-leaf collimators) is in motion while the radiation is delivered.

In conventional and conformal radiotherapy the beam is delivered with constant fluence, while the tumor or target is considered to be static i.e. it is assumed that there is no motion in the patient. However, in reality there is tumor motion during the course of radiation delivery (breathing, internal organ motion etc.) which causes several issues including a blurring of the radiation field relative to the patient’s anatomy.

Breathing and organ motion results in further issues for treatments where the collimators dynamically modulates the beam across a potentially moving tumor volume, resulting in additional dose discrepancy. This additional effect is termed as interplay which is a combination of tumor and beam motion. For example EDW interplay effects were highlighted by Pemler et al., where they investigated the significance of interplay effects for single field non fractionated treatments. They studied the effect of wedge angle, dose rate, tumor-collimator motion, amplitude and frequency of tumor motion. In this study a single
direction of motion between the tumor and collimator was considered and the dose distributions not experimentally measured; rather these were calculated in the treatment planning system. A second study by Sidhu et al.\textsuperscript{15} was an experimental investigation comparing the physical wedge, EDW and IMRT treatments for a 2 field non-fractionated breast plan, considering a single tumor-collimator direction of motion. More thorough studies are required to specifically examine EDW based interplay effects where multiple field treatments combining different wedge angles are delivered over several fractions.

The purpose of this research is to study the interplay effects for EDW based treatments using experimental and Monte Carlo (MC) methods. A series of experimental measurements have been made starting with single field wedged deliveries studying the dependence of tumor speed and frequency, wedge angle, field size and the direction of collimator-tumor motion. These have been extended to multiple field patient plans. The experimental work is backed by MC modelling which required that the BEAMnrc Component Module (CM)\textsuperscript{16} be validated and automated.\textsuperscript{17}

Chapter 2 provides a background to the research and introduces the current research problem in the context of the already published literature by presenting a detailed literature survey. The first series of experimental interplay investigations for single field EDW deliveries is presented in Chapter 3. A number of parameters have been studied for their potential contribution to the interplay effects including the period and amplitude of tumor motion, wedge angle and field size.

The experimental work of Chapter 3 is extended further to patient specific treatment plans using fractionated deliveries in Chapter 4. A four field EDW treatment has been delivered employing 4D CT planned data to the IMRT phantom with a dummy tumor insert. MC commissioning of the linac model using the DYNJAWS CM is presented in Chapter 5.
BEAMnrc and DOSXYznrc codes have been used to simulate the treatment head simulations and the phantom dose calculation respectively. To aid this the DYNJAWS CM has been validated and the input file generation has been automated and tested.\textsuperscript{17} The commissioned MC model is further used to study interplay effects using single and multiple field treatment plans in chapter 6. A number of wedge angle and combinations have been investigated using the DYNJAWS CM.

Gel dosimetry is a potentially powerful tool for verification of wedge dosimetry. For this technique to be used here, further optimization of a gel readout technique was required. In Chapter 7 the gel readout using x-ray computed tomography (CT) has been improved by introducing a zero-scan methodology.\textsuperscript{18} This method enables the generation of a reconstructed low noise gel image by taking subsequent scans of the irradiated gel without adding further dose to the gel as a result of repeated x-ray CT scans. The zero-scan methodology has been extended to scan the entire gel image produced from a multiple field EDW delivery.

Chapter 8 is the concluding chapter of the document where the result summary and overall discussion and future recommendations are presented.
Chapter 2 Background

2.1 Cancer and Radiotherapy

Cancer is one of the lethal diseases faced by humanity today. Each year more than 10.9 million people are diagnosed with cancer worldwide.\textsuperscript{19} According to the World Health Organization (WHO) in 2008 cancer was responsible for 7.6 million deaths i.e. around 13\% of all deaths worldwide and this number is projected to rise to over 11 million in 2030.\textsuperscript{20} As defined by WHO “Cancer is a generic term for a large group of diseases that can affect any part of the body. Other terms used are malignant tumours and neoplasms. One defining feature of cancer is the rapid creation of abnormal cells that grow beyond their usual boundaries, which can then invade adjoining parts of the body and spread to other organs. This process is referred to as metastasis. Metastases are the major cause of death from cancer”.\textsuperscript{21}

A number of cancer treatments are currently available; amongst them the most widely used ones are surgery, radiation therapy and chemotherapy or a combination of these. Usually the first step is to remove the malignant tissue after which a combination of radiation and chemotherapy is used depending on the tumor site, grading and extent of the tumor spread.\textsuperscript{2} Radiotherapy is a cancer treatment modality where radiation (photons, electrons, neutrons or ion beams) is delivered to the tumour with the purpose of killing the cancer cells and stopping their further spread or metastasis.

There are two processes by which cells are damaged by radiation. The first involves direct damage of the cells by a radiation hit to the DNA,\textsuperscript{22} and the second more prominent process involves indirect damage through the creation of free radicals in the intracellular space.\textsuperscript{2, 22, 23} These free radicals can react with DNA and other molecules in the intracellular matrix and can result in damage. Hall has suggested that more than 75\% of the DNA damage in mammalian
cells is caused by the hydroxyl radical. The amount of cellular damage induced by radiation depends on a number of factors, including the type of radiation, rate of dose delivery, radiosensitivity and oxygenation of the targeted tissue and the cell cycle phase of the individual cells. Because radiation damages both healthy and cancerous cells, the main goal of radiotherapy is to deliver the maximum possible radiation dose to the tumour while simultaneously minimizing the dose to the surrounding normal tissues and the organs at risk (OARs). Absorbed dose, which describes the energy deposited in the tissue by the radiation beam, is measured in units of gray (Gy) where one Gy corresponds to the absorption of one joule of energy per kilogram of mass.

Delivery of radiotherapy is generally categorized into two techniques i) external beam radiation therapy (EBRT), which is the focus of this study, where the radiation source is outside the patient body at some distance and the tumor within patient is treated with external radiation beam and ii) Brachytherapy, where the radiation source is inserted into the patient’s body directly into the tumor. The first step in an EBRT treatment is the localization and identification of the extent of the tumor through patient imaging. In the second step a treatment plan is prepared based on the patient’s anatomical data. Here the radiation beams are defined and dose calculation engines are used to predict the dose in tumor and surrounding healthy tissues. Once the treatment plan is finalized the treatment delivery is carried out. Dosimetric measurements verifying the accuracy of the radiation dose delivery can be made either before treatment, using phantom studies, or during and after treatment, using in vivo patient dose measurements.
2.2 Linear accelerators and EBRT techniques

Modern day EBRT is delivered with the use of linear accelerators. Before the introduction of linear accelerators, the Cobalt-60 unit (shown in figure 2.1) was the workhorse of radiation therapy departments.

![A Theratron Cobalt 60 machine](image)

**Figure 2.1**: A Theratron Cobalt 60 machine. Manufactured by MDS Nordion, Ottawa, Canada. [Reprinted with permission].

A linear accelerator (also called a linac as shown in figure 2.2a) accelerates a beam of electrons using a wave guide. This electron beam can be directed into a metallic target (shown in figure 2.2b) to generate bremsstrahlung photons. The resulting high-energy photon beam can be collimated using the jaws and further tailored with the help of multileaf collimators.
Early linac-based radiotherapy treatments were planned using two dimensional (2D) anatomical images and 2D dose calculations. Boxed or parallel opposed radiation beams were common field orientations, where jaws alone were used for field collimation. These conventional radiotherapy techniques were successfully applied for the treatment of various types of cancers, although significant irradiation of healthy tissues was unavoidable. This is demonstrated in figure 2.3, where the tumor is adequately covered by the radiation field (shown in red) but at the same time a large segment of the normal tissue is also being irradiated.
Figure 2.3: Conventional radiation delivery showing a single field technique. Tumor and healthy tissues are represented by red and green colours respectively. [Reprinted with permission].

To achieve improved field conformity multi leaf collimators (figure 2.4a) have been introduced where individual leafs can be moved and any arbitrary tumor shape can be treated.

Figure 2.4: a) Multileaf collimator [Varian Oncology system (Palo Alto, CA: Varian)] b) More tightly conformed tumor dose as result of MLC use. [Reprinted with permission].
As demonstrated in figure 2.4b the dose margins around the tumor are far more restricted than the corresponding conventional radiotherapy case shown in figure 2.3. The delivery of multiple fields, all collimated to the shape of the tumour, is known as three dimensional conformal radiotherapy (3DCRT).

A major, recent development in cancer treatment is the introduction of intensity modulated radiation therapy (IMRT) where a non uniform fluence is delivered to the tumor. In step and shoot, the beam is delivered by sequential or discrete movement of the collimator leaves with the radiation being turned off between the segments. A second type of delivery is called dynamic delivery which involves continuous delivery of the radiation as the MLC leaves are swept across the field. In figure 2.5 a comparison of conventional (2.5a), conformal (2.5b) and IMRT treatments (2.5c) has been presented. It can be seen that IMRT conforms not only more tightly to the tumor shape but also the dose within the target can be varied.

**FIGURE REMOVED DUE TO COPY RIGHT**

**Figure 2.5:** A comparison of conventional conformal and IMRT tumor coverage, a) Conventional radiotherapy b) Conformal radiotherapy c) IMRT delivery depicting the non uniform dose delivery to the tumor.
2.3 Wedge filters

Historically in conventional radiotherapy physical or fixed wedges have been routinely implemented for dose shaping i.e. to alter the photon beam intensity and to provide oblique isodose distribution in patients by the introduction of a dose gradient determined by the wedge angle.\textsuperscript{27, 28} Figure 2.6 shows a wedged dose distribution where the dose is decreased at the thick end of the wedge.

Most external radiotherapy machines (Co-60 and linear accelerator) supports four wedge angles (i.e. 15°, 30°, 45° and 60°). Despite the immense utility of the physical wedges nevertheless there are associated problems, including the beam hardening, dose discrepancy due to occasional misalignment and the potential hazard to the patient due to the weight of the wedge.\textsuperscript{29}

FIGURE REMOVED DUE TO COPY RIGHT
With the introduction of computer control, it has now become possible to deliver wedged fields by moving the collimator jaws during the irradiation; such a computer controlled wedge is normally termed as virtual or dynamic wedge. The implementation of the dynamic wedge was attempted for the first time by Kijewski et al., on a Siemens Mevatron XII accelerator.

Varian (Varian Medical Systems, Palo Alto, CA) was the pioneer in the clinical introduction of the dynamic wedge. This feature was available in the Varian 2100C linear accelerator in 1991 based on the work of Leavitt et al. Although the dynamic wedge was a novel addition there was a desire for further improvement in its capabilities. This improvement was incorporated by Varian in their linear accelerators with the implementation of enhanced dynamic wedges (EDWs). Additional functionalities included the introduction of symmetric and non symmetric fields, availability of more wedge angles; larger field size (up to 30 cm) and fewer numbers of segmented treatment tables (STTs).

Siemens also introduced the virtual wedge in mid 1990’s (Siemens Medical Systems, Inc., concord, CA 94520). This model was validated by Desobery et al. Their conclusion was that the virtual wedge device was performing according to the specifications and was able to accurately model any wedge angle. In the same year the dosimetric evaluation of the Siemens dynamic wedge was carried out by Santvoort, it was reported that the dynamic wedge is a useful and accurate clinical tool. It provides extra flexibility and convenience in terms of clinical application.

Dynamic wedges have several advantages over physical wedges.

- Use of dynamic wedges can result in reduced treatment times, because the manual replacement of the wedge by the operator is not required.
- Dynamic wedges are not fixed to certain wedge angles as any arbitrary wedge angle can be generated.
From a dosimetric point of view the beam hardening associated with the physical wedge does not affect beams produced by dynamic wedges.\textsuperscript{36}

Also the application of dynamic wedges has resulted in a decrease in the peripheral dose.\textsuperscript{37}

One disadvantage of using the dynamic wedges is the additional quality control associated with the computer controlled collimator motion. Dynamic wedges are routinely used in a large number of non IMRT treatments involving anatomic sites like breast, lung, prostate and brain.

2.5 Monte Carlo modelling

Monte Carlo (MC) simulations are based on numerical techniques employing random numbers. These simulations have been widely used in medical physics.\textsuperscript{38, 39} There are various MC-based codes available for calculating the radiation dose. The Electron Gamma Shower system (EGS, updated to EGSnrc) is one of the most widely used MC codes for simulating coupled electron-photon transport in medical physics applications.\textsuperscript{40} This code is useful in simulating electrons and photons with energies in excess of a few keV up to several hundred GeV, and has been widely benchmarked.\textsuperscript{41} The code is based on MORTRAN language which is very similar to the FORTRAN language.

BEAMnrc is an EGSnrc-based MC user code used for simulating the linear accelerator head.\textsuperscript{42} The BEAMnrc code can model the therapy source with the Z-axis taken as the central axis. The code comprises of various component modules (CMs) perpendicular to the incident beam, each of these CMs can be fully specified by the user and can operate independently of each other in a non overlapping manner. The user can interact with the software through a graphical user interface (GUI) as shown in figure 2.7. The code allows the placement of scoring planes in the linac model and the phase space files are generated accordingly.
A phase space file contains the data related to each particle such as particle position, direction, charge total energy etc. In addition to phase space files fluence results, energy deposition and a processing time summary can be output. Currently there are 24 CMs available in the BEAMnrc code.\textsuperscript{43}

\textbf{Figure 2.7:} BEAMnrc GUI.\textsuperscript{43}

DOSXYZnrc\textsuperscript{44} is a general purpose EGSnrc user code for calculating absorbed dose in a 3D geometry as shown in figure 2.8. This code simulates the transport of the electrons and photons in a Cartesian volume and scores the energy deposited in the designated voxels. DOSXYZnrc has a number of important and unique features such as dose component calculations, a wide variety of source configurations and beam reconstruction techniques, CT to phantom conversion (via ctcreate), restart capabilities, phase-space redistribution, \textit{etc.}
ctcreate is a standalone program which converts CT data sets into the geometry needed for DOSXYZnrc to perform a simulation. The phase space files generated previously in the linac head simulations by BEAMnrc code are often used as input to the DOSXYZnrc code.

Figure 2.8: DOSXYZnrc GUI. 44

2.5.2 Monte Carlo modelling of dynamic radiotherapy treatments

One of the first reported studies was by Verhaegen et al., who performed the Monte Carlo modelling of the virtual wedge. 29 It was worth noticing that Monte Carlo was applied to model the time dependent geometries. For Monte Carlo (MC) simulations the linac head was modelled with the help of EGS/BEAM 42 for 6 MV and 10 MV photon beams. Siemens MD2 accelerator was used for dose delivery. The simulations were performed step wise by taking a step of 1 cm (the movement of jaws). Therefore 20 phsp (phase space) files were generated,
and then combined together into a single phase space. This procedure was repeated for four wedge angles (15°, 30°, 45° and 60°) using DOSXYZ code.\textsuperscript{45} For 10 MV photons, the agreement between measurements and MC simulations for all wedged fields was very good at most off-axis positions and all depths. Only in the penumbra region of the wedge toe the agreement was somewhat worse. For 6 MV photons, the overall agreement was slightly worse in the toe region compared with 10 MV. It was suggested that an alternate approach could be used if a probability distribution function for the position of the moving jaw is established from which the geometry could be sampled before transporting each particle. This would eliminate the need for generating phase space files at each discrete step.

Later Verhaegen \textit{et al.} \textsuperscript{46} proposed two techniques were proposed for modelling the EDWs in MC. These included the position probability sampling (PPS) method and the static-component simulation (SCS) method. In the PPS method the jaw position was treated as a random variable during the simulation, i.e. the jaw position was sampled from a cumulative probability distribution function when each particle was initiated to start its transport in the simulation. SCS involved the modelling of a dynamic field by simulating individual static component fields separately, each with different field sizes, and finally integrating the static components. The measured and calculated electron energies were compared and the best match electron energies were found out to be 6.25 MeV and 11.0 MeV for 6 MV and 10 MV photon beams respectively. It was also observed that for the PPS method the primary output of the machine (excluding the monitor chamber backscatter effect) was proportional to the number of simulation histories. It was proposed that the PPS method is more accurate in simulating the dynamic collimator motion, and can be used towards modelling the dynamic treatments, for instance IMRT delivery using MLCs.

EDWs have been modelled using BEAM code as early as 2001, however until recently there was no dedicated Component Module (CM) for this purpose. In 2010 a new CM named
DYNJAWS has been incorporated in the BEAMnrc code for explicitly modelling dynamic wedges. This code is capable of simulating a dynamic wedge using the step and shoot and dynamic delivery techniques. For a step and shoot and dynamic delivery the user is required to provide an input file which specifies the probability of selecting a sub field and the respective jaw coordinates. The DYNJAWS CM is an important addition to the BEAMnrc code and will enable the MC verification of patient treatments involving the use of dynamic wedges.

2.5.3 Modelling the Interplay effects in dynamic radiotherapy delivery

Treatment planners have compensated for organ motion during treatments by enlarging the clinical target volume (CTV) by a certain margin to form the internal target volume (ITV). The assumption is that the tumour is moving within the boundaries of the ITV, therefore if the ITV is fully irradiated that ultimately ensures the proper dose delivery to the tumour. However in IMRT treatments where dose escalation is used to treat the tumour the use of such large margins around the tumour may have undesirable dosimetric effects on the surrounding normal tissues. Therefore it has become imperative to incorporate the tumour/organ motion into the treatment planning process.

Tumour and internal organ motion has been classified into two categories namely inter-fractional (between treatment deliveries) and intra-fractional (during the treatment delivery) movements. Intra-fraction movements can be classified into various groups according to their origin. The major reported sources of intra-fractional movement can be traced to respiratory, cardiac, gastro intestinal and muscular systems. However the research has largely focused on respiratory motion.

Intra-fractional organ motion can result in two types of effects. The first one is called the dose blurring effect. Blurring effect results in the over/under dosage of the tumour in radiation
beam. The tumor receives a lower dose and an adjacent organ at risk more dose than in static case. This type of motion effect is independent of the treatment delivery technique and not the subject of this literature review.\textsuperscript{8}

The second type of motion effect which is termed as the interplay effect is only a problem in case of dynamic delivery of IMRT or dynamic treatments with EDWs.\textsuperscript{12} This effect is the result of interplay between the moving tumour and the motion of the radiation beam as defined by the EDWs or MLCs.\textsuperscript{9,14}

The seminal work studying the interplay effects was carried out by Yu \textit{et al}.\textsuperscript{56} They investigated the interplay effects over a span of 30 fractions with the aid of numerical techniques based on a simple analytical model where the variation of the primary photon fluence was quantified without considering the scatter contribution. Particularly the impact of collimator speed, beam width and dose fractionation was studied.

The moving target was simulated as a rigid line of fixed length (30 cm) and a slit radiation beam of fixed width was scanned across the moving target. The velocity of the moving beam was kept constant and the target motion was assumed to be sinusoidal. The interplay effect was quantified by calculating the primary photon fluence to the target. The beam width and velocities were varied during the course of the simulation. It was found that dose variation due to the interplay effects was related to the speed of the slit relative to that of the target motion and it was suggested to keep the collimator motion as slow as possible. Similarly, decreasing the slit width resulted in an increase in the dose error. Over a span of 30 fractions dose fractionation resulted in the reduction of the dose to clinically non significant values.

Respiratory induced organ motion has been more extensively studied than any other type of motion.\textsuperscript{55} The respiratory based interplay effects for EDWs have been studied by Pemler \textit{et al}.\textsuperscript{14} Numerical methods were employed to ascertain the impact of wedge angle, amplitude of organ motion, beam energy, dose rate and the phase of respiratory cycle with Varian Clinac
600C type linear accelerators. Dose distributions were calculated both with and without respiratory induced organ motion. CadPlan V 6.0.8 treatment planning system was employed as well for the dose calculations. The results were compared by using the percentage difference analysis. The error in dose delivery was quantified for 3 cm amplitude and 6 s period and was found to be less than 16 % for varying wedge angles. In addition the discrepancies were highest if the collimator speed matched the speed of the organ. The asymmetry of the respiratory cycle was found to be responsible for a mean over dosage of the organs. Another important conclusion of their study was that the results are not dependent on the phase shift i.e., the organ position at the time when the collimator starts moving, and the beam energy.

Sidhu et al. 15 studied the interplay effects using a two field tangential breast case study and compared physical wedges, dynamic wedges and IMRT plans for three different collimator speed scenarios where the motion of the tumour was in the anterior posterior direction. They studied four wedge angles (15°, 30°, 45° and 60°) for three speeds where the collimator motion is slower, equal or more compared to the tumor motion. For gated beam delivery Varian’s Real-Time Position Management (RPM) Respiratory Gating System 1.6 (Varian Medical Systems, Inc., Las Vegas, NV) was used. The dose comparison between static and motion cases were compared using dose area histograms (DAH) and normalized agreement test (NAT). They found out that the deviations from the static case are highest if the speed of the collimator and tumour motion are of the same magnitude. Moreover they also found out that gated beam delivery will improve the dose discrepancy between the static and motion cases.

However the studies by Pemler and Sidhu et al. have not considered some aspects as described further. The study by Pemler et al. was a treatment planning study for a single radiation beam in a non fractionated case where actual measurements were not carried out. Moreover the dose was calculated to a rectangular water phantom which is not an accurate representation of the
patient anatomy. Similarly they also assumed that organ does not pass the radiation field during the treatment delivery. The direction of the wedge motion was always from inferior to superior. The study by Sidhu et al. also considered the motion in one direction for a non fractionated treatment for 2 cm peak to peak tumor amplitude. However in some cases for deep inspiration a 5 cm peak to peak motion has been reported. Both these studies either involved single field or two wedged fields. However in actual treatment delivery usually more than 2 wedged fields are employed. Another important aspect is the EDW plan complexity which has not been examined in these investigations. For instance questions remain, such as whether the interplay will be more dominant for a two field wedged treatment than a 4 field wedged treatment, or alternatively which combination of wedge angles will result in worst agreement between the motion and static cases etc.

In many clinical situations for IMRT treatments the case is more complex, and the tumour moves in a cranial- caudal direction while the MLCs are moving from left to right. It was also highlighted that the mechanism of interplay effect is entirely different when the tumour and MLC motion are parallel and perpendicular. The Massachusetts General Hospital group published two publications in quick succession dealing with the interplay effects. The first paper by Bortfeld et al. performed a statistical analysis investigating the effects of intra-fraction motion on IMRT and calculated the expected dose values over a fractionated treatment regime using a linear quadratic model for considering the motion effects. Intensity maps were generated with Varian Helios inverse treatment planning system (Varian Medical Systems, Inc., Palo Alto, CA, USA) for a lung case where as the IMRT motion effects were simulated with the help of Matlab software (The MathWorks, Natick, MA, USA.). It was noted that expected dose value is independent of the dose delivery technique and the standard deviation over a 30 fraction treatment was generally within 1% of the
expected dose concluding that averaging of the dose occurs during the organ motion over a number of fractions.

In the second paper Jiang et al. experimentally investigated the intra-fractional organ motion effects in lung IMRT delivered by MLC on a Varian 2100C linear accelerator. The tumour motion was simulated by using a moving phantom, with amplitude of 1 cm and a period of 4 s. The MLC leaves were moving along left to right direction while the tumour motion was modelled from a cranial-caudal direction. A five field IMRT plan was constructed for two dose rates (300 MU/min and 500 MU/min) on the Helios inverse planning system V2.7 (Varian Medical Systems, Inc., Palo Alto, CA, USA) for three MLC delivery modes (dynamic, step and shoot with 10 and 20 intensity levels) and eight equally spaced starting phases of the tumour motion. The dose calculations were performed for a single and 30 fraction treatment respectively. A 0.6 cc farmer chamber was used for measurements in the moving phantom. It was determined that the average difference between a static and moving tumour was less than 3 %. Moreover it was also suggested to use lower dose rate as it reduced the discrepancy in the motion effects. The authors also raised the need for further investigation of treatments with the use of low number of monitor units and shorter delivery times where the interplay effects can be more significant.

2.6 Treatment verification

There are various tools for treatment verification including ion chamber measurements, portal films, radiochromic films, diode arrays and electronic portal imaging devices (EPIDS). Radiochromic films and gel offer significant advantage in terms of dose verification. Ion chambers are very useful for measuring point doses in regions of dose homogeneity, but ion chambers are susceptible to volume averaging effects, which compromise measurements in regions of varying dose (such as in a wedged field).
Point dose measurements provide only localised indications of the degree to which beam transmission is attenuated by the wedge and do not provide information on the overall effect of the wedge on the beam profile, given that the wedge thickness (and therefore the effect of the wedge) varies across the field, as shown in section 2.3.

Arrays of diodes and (very small) ion chambers can produce 2D dose images and dose profiles \(^6^0\), which are more useful for examining the effects of wedges, but currently available arrays suffer from low spatial resolution (detector spacing of around 5-7 mm) as well as questionable detector response (due to volume averaging\(^5^8\), for ion chamber arrays, and spectral variations, for diode arrays\(^6^1, 6^2\)). The angular dependence of current detector arrays\(^6^3, 6^4\) also means that fields need to be delivered from a zero gantry angle, so the combined effects of multiple fields from different angles cannot be evaluated.

Radiochromic film can be used to produce 2D dose images and profiles, with high spatial resolution and negligible angular dependence or energy dependence.\(^6^6\) Gel dosimeters can be used to produce 3D dose distributions, which can be compared with 3D dose distributions from the treatment planning systems or Monte Carlo calculations, as well as 2D images and profiles, with high spatial resolution, good dose-response linearity\(^7^0, 7^1\) and no angular dependence.

Consequently radiochromic film and gel are expected to provide more accurate and detailed dose measurements in wedged fields, compared to the other dosimetry systems. This is especially the case for wedged fields applied to moving phantoms, where high resolution is imperative and point dose measurements are uninformative.

Gel dosimetry has evolved as a promising technique for accurate three dimensional dose distribution predictions. Gel dosimeters, consisting of a radiation sensitive material infused in a 3D gel matrix, are increasingly being investigated for radiotherapy dose verification and quality assurance.\(^7^2, 7^3\) When a volume of gel is irradiated, the radiation sensitive material
undergoes a measurable change in the magnetic relaxation, density and optical density which is directly related to the radiation dose received, potentially providing a high-resolution three-dimensional measurement of the dose absorbed by the gel.\textsuperscript{74} An irradiated gel dosimeter is shown in figure 2.9.

![An irradiated gel dosimeter](image)

**Figure 2.9:** An irradiated gel sample, the square field in the middle is visible.

An important consideration for any dosimeter prior to clinical use is the spatial resolution and the accuracy in the measurement of the absorbed dose and many authors have investigated gel dosimetry as a solution.\textsuperscript{75-80}

One of the challenges in gel dosimetry is the extraction of the dose information once the gel has been irradiated. Various techniques have been employed for gel dose readout including magnetic resonance imaging (MRI)\textsuperscript{81-83}, optical CT scanning (OCT)\textsuperscript{84, 85}, x-ray computed tomography (CT)\textsuperscript{86, 87}, and ultrasound\textsuperscript{88}. In MRI imaging of polymer gel dosimeters the spin-spin relaxation rate (R\textsubscript{2}) is used to determine the radiation induced polymerization corresponding to the absorbed dose.\textsuperscript{81-83} One issue with MRI imaging is that artefacts are
significant issues affecting the accuracy of the gel dosimeters, which requires careful selection of scanning parameters to ensure accuracy.\textsuperscript{74} OCT has been demonstrated as viable readout technique for polymer gel dosimeters due to a post-irradiation change in optical density;\textsuperscript{84, 85} however this technique is susceptible to artefacts due to refraction of light \textsuperscript{89, 90}. Ultrasound imaging \textsuperscript{88} utilizes changes in acoustic speed of propagation, absorption and attenuation which vary with radiation induced polymerization.

X-ray CT has been employed to exploit post irradiation changes in linear attenuation coefficient in polymer gel dosimeters.\textsuperscript{73, 86, 87, 91-94} The availability of CT scanners in radiotherapy centres makes this imaging technique attractive as a routine technique for imaging of gel dosimeters. However, the small CT signal and relatively large noise level arising from radiation exposure means that this technique suffers from a low signal to noise ratio (SNR). Attempts to reduce stochastic noise by averaging several CT images result in an additional dosing of the gel.\textsuperscript{86, 87} An alternative approach for the reduction of noise in CT imaging of polymer gel dosimeters has been the application of image processing techniques.\textsuperscript{92, 95, 96}

It would be important to briefly mention the limitations of the film dosimetry include non uniformity, scanning artefacts and temperature depended on the polymerization \textsuperscript{97} and where as for gel dosimeters these include the handling of toxic chemicals.

Data analysis tools used in the current work include dose volume histograms (DVHs) and gamma analysis. DVHs are a standard clinical tool that helps oncologists and treatment planners to examine and interpret the dose levels delivered to specific volumes of specific organs.\textsuperscript{25} DVHs can summarize the 3D dose information in a large volume and reduce them to a single plot indicating the dose received by a percentage of volume for a tumor, organ at risk or any other anatomical site of interest. There are two types of DVHs; differential and integral.
In this study integral DVHs have been used (presented in Chapter 6) which display how much volume receives a certain amount of dose. The 2D counterparts of DVHs are the dose area histograms (DAHs).

For dose map comparisons the gamma analysis technique has been used. The use of enhanced dynamic wedges renders the dose difference technique less useful as gradients are involved. The gamma analysis technique was developed by Low et al. Here two criteria are included for comparison of the dose. First is the percentage dose difference and second is the distance to agreement (DTA). The DTA is the distance between a measured data point and the nearest data point with same dose in the overall dose map. For each gamma calculation a value is calculated indicated by $\gamma$. A gamma value of greater than 1 indicates that the region has not passed the dose-distance criteria. For the application of gamma analysis the dose values between 3% to 5% and distance criterion between 2 to 5 mm has been reported. In this study the gamma analysis has been performed using 3%-3mm values as suggested and used in clinical studies.

2.7 Investigative aims

This study aims to examine the interplay effects for Enhanced Dynamic Wedges (EDWs). Single field non fractionated EDW treatments (Chapter 3) and multiple field fractionated treatments will be investigated using experimental (Chapter 4) methods. Monte Carlo (MC) methods will be used to study the EDW interplay effects first by commission the MC model by validating the DYNJAWS component module (Chapter 5). The commissioned MC model will be used to further study the Interplay effects (chapter 6). For the experimental investigation of EDW treatments using gel dosimeters a new x-ray CT based readout technique called zero-scan method was developed (Chapter 7)
Chapter 3: Interplay for single field enhanced dynamic wedge deliveries

3.1 Introduction

The previous chapter has outlined the literature survey for the current study and contextualized the problem, emphasising the importance of understanding the interplay effects for enhanced dynamic wedge (EDW) treatments. This chapter examines the interplay effects for single field EDW deliveries. A series of experimental investigations have been carried out to assess the role of the amplitude of tumor motion, period of tumour motion, treatment field size, and wedge angle. The effects of the interplay have been studied for two types of motion between the jaw that produces the dynamic wedge effect and the motion of the irradiated phantom, which can be either parallel or perpendicular to each other.

3.2 Methods and Materials

3.2.1 Equipment

EDW lateral dose profiles were obtained for a static and moving phantom using a MapCHECK 2 (Sun Nuclear Corp., Melbourne, USA) diode array. The MapCHECK 2 is a two-dimensional diode array consisting of 1527 SunPoint diode detectors arranged in $32 \times 26$ cm matrix (shown in figure 3.1). The detector spacing is 7.7 mm with an active detector area of $0.64 \text{mm}^2$. The device has a buildup and backscatter to the active detectors of $2.0 \pm 0.1$ and $2.75 \pm 0.1 \text{g/cm}^3$ respectively.
Sinusoidal tumour motion was simulated by placing the MapCHECK 2 detector on a respiratory gating motion platform (Standard Imaging, Middleton, USA); with a single (1D) motion axis (as shown in figure 3.2). This platform can simulate longitudinal movement between 5 mm to 40 mm in 5 mm increments. The motion period can be adjusted from 2 to 6 seconds with increments of 0.5 seconds.

Figure 3.1: MapCHECK 2 diode array [Reprinted with permission].

Figure 3.2: Respiratory motion platform moving along the couch’s main axis.
For all EDW measurements 200 monitor units (MU) were delivered at 600 MU/min with the Y1-IN wedge orientation and 0° gantry angle with 6 MV photon beam and 100 cm Source to Surface Distance (SSD) to the top of the detector. The collimator was rotated to provide jaw motion parallel and perpendicular to the platform motion.

The results were compared qualitatively (visual comparison of the profiles) and quantitatively using 2D gamma analysis with 3%-3 mm criteria. For image manipulation and registration Matlab (version 7.8.0.342, The MathWorks, Natick, USA) and the open source utility ImageJ (Version 1.45d, Rasband, W.S., ImageJ, U. S. National Institutes of Health, Bethesda, USA,) were used.

**3.2.2 Effect of amplitude and period variation**

The first investigation was designed to assess the role of amplitude and period of tumour motion. Two geometries were investigated as shown in figure 3.3, one where the platform-jaw motion was parallel and the one where platform-jaw motion was perpendicular. The platform was always moving in the longitudinal couch direction. All deliveries were made using a 45° EDW. This wedge angle was chosen because of its frequent application in clinical treatments.

Initially the phantom was irradiated with a 20 × 20 cm² field, while stationary, with the MapCHECK2 placed on the respiratory platform under 10 cm of water equivalent plastic. There was no additional backscatter material placed beneath the diode array.

The platform was set in motion and further profiles were obtained, varying the amplitude (10-40 mm, in steps of 10 mm) and period values (3.0 s, 4.5 s and 6.0 s). These period values represented either a faster (3.0 s, 4.5 s) or similar (6.0 s) tumour speed relative to the collimator. The collimator speed was found to be around 10 mm/sec for a 45° wedge angle.
Figure 3.3: Detector platform geometry. The red arrows describe the direction of radiation beam sweep or jaw motion for parallel and perpendicular cases, platform motion direction is described by the white arrows.

From these 12 amplitude-period measurements (each measurement was delivered once) an amplitude-period pair was identified, defined as the worst case motion parameters, where the greatest difference between the static-phantom and the moving-phantom profiles was observed. This set of motion parameters was used as described further in section 3.2.3.
3.2.3 Effect of wedge angle and field size

In the second investigation the effect of wedge angle and field size was assessed by employing the previously identified worst case motion parameters for parallel and perpendicular phantom-collimator motions. Large and small wedge angles (60°, 15°) and three field sizes (20 × 20 cm², 10 × 10 cm² and 5 × 5 cm²) were examined.

Furthermore a fast moving tumour was tested (2.0 s period) for 45° delivery with 20 × 20 cm² field size and 40 mm amplitude. Finally to assess the impact of averaging in a multi-fractionated case a single field (45°, 20 × 20 cm², worst case motion parameters and parallel phantom-collimator motion) profile was delivered over 5 fractions and qualitatively compared with a single fraction delivery.

3.3 Results

3.3.1 Effect of amplitude and period for parallel motion

In figure 3.4(a) to (l) a comparison of 20 × 20 cm² static and motion profiles is presented for a 45° delivery with multiple amplitudes (10-40 mm) and periods (3.0, 4.5 and 6.0 s). Figure 3.4 show that both the amplitude and period of the phantom motion affect the shape of the profiles. The static and dynamic profiles show increasing disagreement as the period approaches 6.0 s where the platform and collimator speeds are approximately similar.

Two types of effects can be observed from the profiles in figure 3.4, firstly penumbral blurring, and secondly the appearance of a step function within the field. The blurring of the penumbra increases with increasing the amplitude of motion (figure 3.4a, d, g and j) and also causes a drop-off in the peak delivered dose. For any given amplitude, with increasing period the step function becomes more pronounced.
Figure 3.4: Effect of varying the amplitude and period of motion for a $20 \times 20$ cm$^2$ field size, 45° EDW for parallel motion. Continuous line represents the static and symbols (x) represent the motion cases respectively (a) 10 mm-3.0 s (b) 10 mm-4.5 s (c) 10 mm-6.0 s (d) 20 mm-3.0 s (e) 20 mm-4.5 s (f) 20 mm-6.0 s (g) 30 mm-3.0 s (h) 30 mm-4.5 s (i) 30 mm-6.0 s (j) 40 mm-3.0 s (k) 40 mm-4.5 s (l) 40 mm-6.0 s
Table 3.1 shows the results of a gamma analysis on the profiles where it can be observed that first for any given amplitude the 6.0 s period results in the lowest pass rate, and secondly that for all periods the gamma pass rate decreases with increasing amplitude.

**Table 3.1**: Gamma analysis pass rates with a 3%-3 mm acceptance criteria for parallel and perpendicular deliveries.

<table>
<thead>
<tr>
<th>Period (s)</th>
<th>Amplitude (mm)</th>
<th>3.0</th>
<th>4.5</th>
<th>6.0</th>
</tr>
</thead>
<tbody>
<tr>
<td>Parallel</td>
<td>Parallel</td>
<td>95</td>
<td>94</td>
<td>92</td>
</tr>
<tr>
<td>Perpendicular</td>
<td>Parallel</td>
<td>96</td>
<td>95</td>
<td>95</td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>20</td>
<td>90</td>
<td>91</td>
<td>87</td>
<td>87</td>
</tr>
<tr>
<td>30</td>
<td>87</td>
<td>87</td>
<td>81</td>
<td>79</td>
</tr>
<tr>
<td>40</td>
<td>81</td>
<td>82</td>
<td>81</td>
<td>80</td>
</tr>
</tbody>
</table>

Amplitude and period of phantom motion both are responsible for dose degradation in case of parallel phantom-collimator motion.

### 3.3.2 Amplitude and period variation for perpendicular motion

The effect of phantom motion that is perpendicular to the wedge motion appears as a broadening of the penumbra only, with no major dependence on the period as illustrated in figure 3.5.
Figure 3.5: Effect of varying the amplitude and period of motion for a 20 × 20 cm² field size, 45° EDW delivery for perpendicular motion. (a) Static (b) 10 mm-3.0 s (c) 10 mm-4.5 s (d) 10 mm-6.0 s (e) 20 mm-3.0 s (f) 20 mm-4.5 s (g) 20 mm-6.0 s (h) 30 mm-3.0 s (i) 30 mm-4.5 s (j) 30 mm-6.0 s (k) 40 mm-3.0 s (l) 40 mm-4.5 s (m) 40 mm-6.0 s

It can be seen that with increasing amplitude the penumbral cut off starts increasing down the column in figure 3.5, however across the row the dose distribution looks more or less the same. This is also quantitatively confirmed from data shown in table 3.1 which indicates that the gamma pass rates for perpendicular motion are almost constant at any amplitude irrespective of the period value. By contrast, for any period, the gamma pass rates decrease
with increasing amplitude. Evidently, the effects of perpendicular phantom-collimator motion depend on the amplitude rather than the period of phantom motion.

3.3.3 The effect of wedge angle and field size for parallel motion

In figure 3.6, 60º and 15º EDW deliveries are compared for three field sizes (20 × 20 cm², 10 × 10 cm² and 5 × 5 cm², employing worst case motion parameters i.e. 40 mm amplitude and 6s period). The selection of the worst case scenario was based on the results from previous section 3.3.1 where for single field cases the worst disagreement between the static and motion parameters was obtained. The result of changing the wedge angle from 60º to 15º, keeping the same amplitude-period parameters, (figure 3.6a versus 3.6b and 3.6c versus 3.6d) showed an increase in the difference in the dose profiles for a static and moving detector. For the smaller 5 × 5 cm² field size the wedged distribution has been entirely lost for both wedge angles (figures 3.6e-f). The penumbral blurring is evident for all field size and wedge angles, with a more pronounced step function for 60º wedge angle (figure-3.6a). In figure 3.6 (g-i) profiles for a higher tumour speed (2.0 s period) are shown depicting penumbra blurring without the appearance of the step function.

3.3.4 The effect on a multiple fraction single field delivery

Figure 3.6j-n shows the result of a multi-fractioned 45º delivery for 20 × 20 cm² field using worst case parameters with parallel motion where the first five fractions are shown. It can be seen that after 5 fractions the step function has smoothed out as visible in figure 3.6o, however the loss of penumbra still remains there.
Figure 3.6: Field size and wedge angle comparison (a-f), with motion parameters of 40 mm and 6.0 s for parallel motion, continuous line represents the static case and symbols (x) represent the motion case. (a) 20 × 20 cm²- 60° (b) 20 × 20 cm²-15° (c) 10 × 10 cm²-60° (d) 10 × 10 cm²-15° (e) 5 × 5 cm²-60° (f) 5 × 5 cm²-15°. A fast moving tumor simulated in g-i with motion parameters of 40 mm, 2.0 s 60° 20 × 20 cm² (g) 20 × 20 cm² (h) 10 × 10 cm² (i) 5 × 5 cm². Comparison of a single field 5 fraction delivery for 20 × 20 cm² 45° (j) fraction 1 (k) fraction 2 (l) fraction 3 (m) fraction 4 (n) fraction 5 (k) the combined image after 5 fractions.

3.3.5 The effect of wedge angle and field size for perpendicular motion

For two wedge angles (60° and 15°) and three field sizes (20 × 20 cm², 10 × 10 cm² and 5 × 5 cm², using worst case motion parameters) the perpendicular motion has been compared in figure 3.7. A penumbral broadening occurs for both wedge angles, however for the same field size this effect is more pronounced for the large wedge angle (figure 3.7b versus 3.7d). With reducing the field size to 10 × 10 cm² the penumbral cut off has increased further (figure 3.7h & f). Finally further reducing the field size to 5 × 5 cm² the wedged dose distribution has been completely lost for both wedge angles (figure 3.7 j & l).
Figure 3.7: Field size and wedge angle comparison with motion parameters of 40 mm and 6.0 s for perpendicular motion. (a) 20 × 20 cm² - 60° static (b) 20 × 20 cm²-60° motion (c) 20 × 20 cm²-15° static (d) 20 × 20 cm²-15° motion (e) 10 × 10 cm²-60° static (f) 10 × 10 cm²-60° motion (g) 10 × 10 cm²-15° static (h) 10 × 10 cm²-15° motion (i) 5 × 5 cm²-60° static (j) 5 × 5 cm²-60° motion (k) 5 × 5 cm²-15° static (l) 5 × 5 cm²-15° motion

3.4 Discussion and summary

For single field parallel delivery, the dose distribution is dependent both on the amplitude and the period of tumor motion, as shown in figure 3.4 with the appearance of a step function along with penumbral cut off. The disagreement between motion and static profiles increases when the collimator-tumor speeds are comparable (6 s case).
This observation is consistent with earlier findings by Pemler and Sidhu et al. The appearance of a prominent step function as the period approaches the 6 s value is similar to an interplay-induced step function illustrated by Pemler et al. The 45º EDW was selected as it is one of the commonly used wedge angles. For perpendicular phantom-collimator motion there is an increasing penumbral cut off as the motion amplitude is increased, however the period seems to have little effect as indicated by the constant gamma values across all rows in table 1. The 4 cm peak to peak amplitude of tumour motion has been reported for cases of deep inspiration.

Dose differences observed for parallel motion are greater for a larger wedge angle (60º) (Figure 4a) due to the steeper dose gradient involved (Figure 4b). With a reduced field size (5 x 5 cm²) the wedged dose distribution is lost for both large and small angles (60º and 15º); this is probably caused by the large amplitude of the tumor motion relative to the sizes of the smaller fields. For perpendicular motion for the same field size, there is a more pronounced penumbral broadening for large wedge angle (60º) compared to the smaller wedge angle (15º). Similar to the parallel case for small field size the wedged distribution is completely lost for both wedge angles.

In this chapter the interplay effects have been examined for single field EDW deliveries. A set of amplitudes, periods of motion, field sizes and wedge angles has been studied for both parallel and perpendicular motions (between the collimator and the tumour). It was observed that for parallel motion both amplitude and period affect the interplay resulting in two main effects. The first is penumbral cut off directly related to the magnitude of the amplitude. The second one is the appearance a step function especially for period values where the collimator tumour speed is similar resulting in higher discrepancy between static and motion profiles. The step function smooths out if the measurement is repeated a number of times. In case of higher tumour speed compared to the collimator, the step function is not visible even for a
single fraction. For perpendicular motion the amplitude is the only dominating factor resulting in a penumbral cut off directly related to the magnitude of the amplitude.

For the wedge angle and field size investigation the dosimetric discrepancy is higher for a large wedge angle (60°) as compared to a smaller wedge angle (15°). With reduction of the field size the large amplitude value causes the complete loss of the wedged profiles for both parallel and perpendicular motion.

This chapter has summarised the impact of various parameters for single field deliveries on the interplay effects. The next chapter will extend the current study to the assessment of the interplay effects for multi field fractionated cases employing more realistic treatment planning parameters.
Chapter 4 Interplay for fractionated multiple-field enhanced dynamic wedge deliveries

4.1 Introduction

In the previous chapter the interplay effects for single field enhanced dynamic wedge (EDW) deliveries have been examined. A number of parameters have been investigated for their potential contribution in the interplay between the collimator and the moving tumour for single field treatments. However in most clinical scenarios single field or single fraction deliveries are seldom employed for patient treatment. The treatment field usually combines a set of open and wedged fields delivered over several fractions as opposed to a single fraction. Therefore there is a need to study pertinent clinical scenarios with respect to their impact on the interplay effects. This chapter aims to further study the interplay effects for multi-field deliveries carried out over several fractions.

This chapter has been divided into three sections. The methods and materials are presented in section 4.2, with section 4.2.1 presenting the overview of the materials and methodology; the specific studies are further described in sections 4.2.2 and 4.2.3. This section is followed by the results in section 4.3 and finally the chapter is concluded with a brief discussion and chapter summary in section 4.4.

4.2 Methods and Materials

4.2.1 Overview

This study was divided into two parts. In the first part, existing patient plans were delivered, and in the second part a 4D CT treatment was planned and delivered over 4 fractions.
For the first part a MapCHECK 2 (Sun Nuclear Corp., Melbourne, USA) diode array and EBT2 gafchromic films (International Specialty Products, Wayne, USA) were used for dose image and profile extraction. The EBT2 films were scanned using an Epson Perfection V700 Photo flatbed scanner (Seiko Epson Corp., Nagano, Japan) operating in transmission mode as shown in figure 4.1.

![Epson Perfection V700 flatbed scanner](image)

**Figure 4.1:** Epson Perfection V700 flatbed scanner.

Each sheet of film was scanned before and after irradiation, elevated above the glass surface of the scanner, and analysed in the red-channel, without applying a blue-channel correction, as previously recommended. A set of small (5 × 5 cm²) pieces of film were irradiated to known doses between 0 and 450 cGy, and the resulting optical density data were plotted against delivered dose and used to generate a sensitometric curve for the film, which was fitted using a cubic polynomial. The net optical densities determined in each film image obtained during the interplay study were then converted into measurements of dose using the
sensitometric curve, producing two-dimensional dose maps of each radiation treatment examined.

For second part of the study, EBT2 films were placed in a lung IMRT dose verification phantom (Standard Imaging, Middleton, USA) as shown in figure 4.2 with lung inserts replaced with custom made foam inserts and a place holder for a dummy tumor. The lung inserts were made up of Styrofoam while a circular place holder was created for the placement of the tumor. The IMRT dose verification phantom models human anatomy and is helpful in patient dose verification. It also incorporates inhomogeneity structures with bone and lung equivalents. Radiation was delivered using a Varian iX linear accelerator, producing a 6 MV photon beam.

![Image](image.png)

**Figure 4.2:** IMRT quality assurance phantom.

The results were compared both qualitatively (visual comparison of the profiles) and qualitatively using 2D gamma analysis with 3 %–3 mm pass rate. For image manipulation and registration Matlab (Version 7.8.0.342, The MathWorks, Natick, USA) and the open source utility ImageJ (Version 1.45d, Rasband, W.S., ImageJ, U. S. National Institutes of Health, Bethesda, USA,) were used.
4.2.2 Existing patient plans delivered to MapCHECK2 and EBT2 films

Clinical patient plans were delivered to the MapCHECK 2 diode array and the EBT2 films. The number of treatment fields, gantry angle, collimator angle and wedge orientations were not restricted as they were in the previous chapter which described only single field delivery scenarios.

Three existing patient plans (a single breast and two lung plans, described in table 4.1) were delivered to MapCHECK 2 and the EBT2 film and dose images obtained with the static and moving phantom were compared using gamma analysis. The individual fractions were delivered to the static phantom and to the phantom moving with the worst case motion parameters identified in chapter 3. For all patient plans the tumor motion was simulated in the superior-inferior direction.

The first plan was a breast treatment. The MapCHECK 2 was placed on the couch (without any additional backscatter) for a single fraction, and the 2 tangential EDW-wedged fields (a combination of 25° and 20°) were delivered from their planned gantry angles (For both fields, the wedge direction was perpendicular to the phantom motion.

The second plan (Lung #1) was a four field lung EDW treatment, using a combination of 45° and 20° wedges. This plan was delivered to MapCHECK 2 over a single fraction. For three out of the four fields, the wedge direction was perpendicular to the phantom motion, and for the other field, the wedge direction was parallel to the phantom motion.

The third plan was a more complex lung case (lung #2), consisting of five coplanar wedged fields (combination of wedge angles: 45°, 25° and 20°) and an additional wedged field with a 320° couch rotation. This plan was delivered to the lung phantom over two fractions with the EBT2 film placed isocentrically between the two lung slabs in the coronal plane (EBT2 resolution was 0.35 mm × 0.35 mm²) to see the effect of a multiple wedged field delivery for a
certain number of fractions. The lung phantom was placed along the couch identical to the setup in figure 4.3.

Table 4.1: Breast and lung patient treatment parameters

<table>
<thead>
<tr>
<th></th>
<th>Lung # 1</th>
<th>Breast Plan</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field #</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>MU</td>
<td>55</td>
<td>89</td>
</tr>
<tr>
<td>Gantry (deg)</td>
<td>180</td>
<td>155</td>
</tr>
<tr>
<td>Collimator (deg)</td>
<td>90</td>
<td>90</td>
</tr>
<tr>
<td>Field Size (cm²)</td>
<td>15.7 × 11.7</td>
<td>15.4 × 10.1</td>
</tr>
<tr>
<td>Wedge angle</td>
<td>45-OUT</td>
<td>25-IN</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>Lung # 2</th>
<th>Breast Plan</th>
</tr>
</thead>
<tbody>
<tr>
<td>Field #</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>MU</td>
<td>59</td>
<td>48</td>
</tr>
<tr>
<td>Gantry (deg)</td>
<td>120</td>
<td>300.1</td>
</tr>
<tr>
<td>Collimator (deg)</td>
<td>90.0</td>
<td>35.0</td>
</tr>
<tr>
<td>Field Size (cm²)</td>
<td>6.2 × 6.5</td>
<td>6.2 × 6.5</td>
</tr>
<tr>
<td>Wedge angle</td>
<td>25-OUT</td>
<td>20-IN</td>
</tr>
</tbody>
</table>

The film dose data were read out using the methodology as outlined in the section 4.2.1 above and compared with the respective static cases using gamma analysis.

4.2.3 A multi-fractionated 4D CT planned patient delivery to EBT2 films

The lung inserts in the IMRT phantom were replaced with custom made foam inserts with a place holder for a dummy tumour (a cylindrical tumour made up of a polyethylene terephthalate (PET) container with a diameter of 7 cm and height of 2.5 cm-filled with gel as shown in figure 4.3a).
EBT2 film was placed between the two layers of the tumor in the coronal plane, as shown in figure 4.3b. The phantom was scanned on Toshiba 4D CT scanner (shown in figure 4.4) using Varian’s real time position monitoring (RPM) system and the 4D CT images were binned into ten respiratory phases. The scan parameters were 100 mAs, 120 kVp with 2 mm slice thickness.
The scan data was imported into the iPlanNET treatment planning system (Ver 2.5, BrainLab AG, Feldkirchen Germany) for sorting and a Planning Target Volume (PTV) was outlined based on a maximum intensity projection (MIP) reconstruction of the 10 respiratory phase correlated CT scans. MIP outlines the PTV considering the maximum data value for each pixel in the image data set.

Figure 4.5: Treatment planning screen shot.

The MIP CT was exported to Eclipse-External Beam Planning 8.6.17 (Varian Medical Systems, Palo Alto, USA) for dose calculation using the anisotropic analytical algorithm (AAA).
A four field EDW (60º and 30º) treatment was planned as shown in figure 4.5 and table 4.2, with a prescribed dose of 2.0 Gy/fraction. This plan was delivered over 4 fractions using worst case motion parameters identified in chapter 3 i.e. 40 mm tumor amplitude and 6 s period to assess the impact of averaging over initial few fractions. The treatment delivery setup is shown in figure 4.6. The jaw-phantom motion was not phase matched (the phantom motion and treatment delivery were not started simultaneously) because phase matching is not performed in standard clinical practise.

Table 4.2: Lung patient # 3, treatment plan parameters.

<table>
<thead>
<tr>
<th>Field #</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
</tr>
</thead>
<tbody>
<tr>
<td>MU</td>
<td>41</td>
<td>43</td>
<td>50</td>
<td>60</td>
</tr>
<tr>
<td>Gantry (deg)</td>
<td>320.1</td>
<td>39.0</td>
<td>90</td>
<td>0.0</td>
</tr>
<tr>
<td>Collimator (deg)</td>
<td>90</td>
<td>90</td>
<td>90.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Field Size (cm²)</td>
<td>13.2 × 9.4</td>
<td>9.5 × 11.6</td>
<td>11.6 × 6.6</td>
<td>8.3 × 11.6</td>
</tr>
<tr>
<td>Wedge angle</td>
<td>30-OUT</td>
<td>30-IN</td>
<td>60-OUT</td>
<td>60-OUT</td>
</tr>
</tbody>
</table>

The films were scanned according to the method described in section 4.2.1 and the measured dose distributions were averaged to produce a combined image after delivering four fractions. Motion and static images were compared using gamma analysis (3%-3mm) and dose area histogram (DAH).
Figure 4.6: Treatment delivery setup, the IMRT phantom was placed on top of the respiratory platform which was moving along the couch direction.

4.3 Results

4.3.1 Patient plans delivered to MapCHECK2 and EBT2

The results of delivering the first lung plan (lung #1) to the static and moving MapCHECK2 are respectively shown in figures 4.7(a) and 4.7(b). When comparing these two dose images, only 79% of pixels passed a gamma criterion of 3 %-3 mm, with figure 4.7b demonstrating dose blurring and a change in the shape of the dose distribution.

The results of delivering the breast plan to the static and moving MapCHECK2 are respectively shown in figures 4.7c and 4.7d. Comparison of these two dose images shows that motion has resulted in under dosage, with 89% of pixels passing a gamma evaluation with 3 %-3 mm criteria.
Figure 4.7: Two lung and a breast plans delivered to MapCHECK 2 and EBT2. Motion parameters were 40 mm and 6.0 s, a) Lung patient #1-static b) Lung patient #1-motion c) Breast-static d) Breast-motion e) Lung patient #2-static f) Lung patient #2-motion, fraction 1 g) Lung patient #2-motion, fraction 2 h) Lung patient #2-motion, averaged over two fractions.

The results for the second lung plan (lung #2) are shown in figures 4.7(e-h). This was a complex 6 field plan delivered over two fractions. It can be observed that as compared to the static case (figure 4.7e), both fractions 1 and 2 for the moving case (figure 4.7f and g) and the average of the two fractions for the moving case (figure 4.7h) show visible dose blurring and under dosage. Comparison of figures 4.7e and 4.7h results in a gamma pass rate of only 62% pass rate, with 3%-3mm criteria. It should be noted that no additional backscatter material was placed under the MapCHECK 2 and only the inherent backscatter of the diode array was present. This can result in a missing scatter contribution mimicking the patient body, however as the setup for both static and motion cases was similar, therefore the information about the dose discrepancy resulting from the motion of the tumor would not be affected.
4.3.2 4D CT Patient planned delivery Results

The result for the four field treatment planned using the MIP from the 4D CT of the moving phantom is shown in figure 4.8.
**Figure 4.8:** Planned lung patient (dose in Gy indicated by the color palate), a) Static b) Motion-fraction 1 c) Motion-fraction 2 d) Motion-fraction 3 e) Motion-fraction 4 f) Motion-Averaged over fractions 1&2 g) Motion-Averaged over fractions 1,2 &3 h) Motion-Averaged over all 4 fractions i) TPS dose image, j) Gamma analysis, comparing the individual fractions with static case(solid line), comparison of averaged fractions with the static case (dashed line), k) DAH of the PTV on the film image, red represent the motion case and blue representing the static case.

For the static image (Figure 4.8a) there is no visible dose degradation. The subsequent 4 motion fractions (Figure 4.8b-e) display a substantial but unequal increase in dose to the irradiated area. In figure 4.8f-h the motion averaged images have been presented. The overall result of delivering 4 fractions (Figure 4.8h) is a dose image with reduced sharpness of the edges and increased blurring, suggesting an increased dose outside the planned target area.

The gamma analysis at 3 %-3 mm (Figure 4.8j) also suggests that in this case, dose averaging over multiple factions has not improved the outcome; the overall pass rate (49%) has not improved after 4 fractions. The cumulative dose-area histogram in figure 4.8k describes the extent of loss of PTV coverage as a result of motion. Figure 4.8k indicates that whereas for the static phantom, more than 90% of the area of the PTV was covered by a dose of 2 Gy (although the clinical coverage should be around 95%, however for this simple phantom setup 90 % coverage was acceptable), for the moving phantom, less than 70% of the area of the PTV was covered by a dose of 2 Gy.

**4.4 Discussion and summary**

In this chapter the experimental results for interplay studies have been presented for multiple field treatments over a few fractions. This represents a more realistic scenario in terms of the number of fields employed and the use of custom made lung inserts with a dummy tumour.
For the multiple fraction treatment planned specifically for the moving phantom, difference between dose maps (figure 4.8b-d) can be attributed to the random starting phase of the motion. It is clear from figure 4.8j that for the combined fractions the gamma pass rate (corresponding to images in figure 4.8f-h) shows no visible improvement indicating the absence of averaging for this particular study.

The number of fractions studied here does not represent an actual treatment delivery of 25-30 fractions, however the intent was to observe the interplay averaging over the first 4-5 deliveries, as it has been reported that for IMRT treatments the interplay effects start averaging out within the first five fractions of the treatment. Mohn et al. recently observed interplay averaging out for a 25 fraction IMRT delivery, however they also pointed out that this averaging may be expected to be patient and plan specific. For the EDW plan examined in this study, it was not possible to verify/observe the dose averaging. Rather, the results of averaging the first 4 fractions are similar to the results obtained for the first fraction alone.

There is scope for further study of these effects. The tumor motion here is assumed to be a perfectly sinusoidal 1D motion; however actual lung tumor motion can be irregular with hysteresis and further studies using patient respiratory traces would be beneficial. Similarly the effect of target deformation has not been considered, assuming the tumor motion to be rigid. The interplay can be affected by other factors such as the schematics of the 4D CT and the possibility of setup errors; however these factors have not been explicitly studied here.

The relationship between the treatment field size and the extent of breathing motion also affects the outcome. Of the three plans measured over one fraction, lung #2 used the smallest field sizes and produced the poorest agreement between the static and moving cases. The smallest field length used in the lung #2 plan was 5.5 cm, which is just 1.4 times the amplitude
of phantom motion. Obviously this treatment plan was designed for a patient with a much smaller range of respiratory motion than was used in this experiment. Nonetheless, this result provides an indication of the negative outcomes that might arise if respiratory motions are not taken into account in treatment planning.

Of all of the treatment plans examined in this chapter, lung #3 produced the greatest deviation between the static and moving phantom measurements. The field sizes and the number of beams used in lung #3 were average, compared to lung #1, lung #2 and the breast treatment. However, lung #3 used the largest mean wedge angle and was the only treatment to include 60 degree EDWs (it included two). Combined with chapter 3’s conclusion that larger wedge angles result in more substantial motion interplay effects, the result for lung #3 seems to suggest that the dosimetric effects of using unusually large wedges to irradiate moving targets are substantial enough to overwhelm the effects of using more-standard field sizes and numbers of beams. However, due to the complexity and differences between the treatment plans examined in this chapter, it is not possible to make clinical recommendations at this stage. However it does indicate that there is a possibility of large over and under dosage with in a first few fractions that might have biological significance which needs to be further evaluated. The effects of plan complexity on motion interplay are investigated in detail, via Monte Carlo simulation, in the following chapters.
Chapter 5: Validation of the Monte Carlo model using DYNJAWS component module

5.1 Introduction

The previous chapter has assessed the interplay effects of dynamic wedges utilizing experimental methods. In this chapter and the following chapter Monte Carlo (MC) methods will be used to investigate effects of plan complexity, including the number of beams and different wedge angles, on the interplay effects. Before commencing the simulation of the actual treatments, the first stage is the commissioning of the MC model.

This chapter has been divided into four major sections, after the introduction and background in section 5.1, the Methods and Materials are presented in section 5.2. The commissioning and validations results are the subject of section 5.3. Finally the chapter is concluded with a brief discussion and chapter summary in section 5.4.

5.1.1 Background

For the commissioning of the MC model the DYNJAWS component module (CM) of the BEAMnrc code has been used. A CM is separate entity in the linear accelerator model that operates independently of the other components. Each CM is employed to specify different a particular part of the linear accelerator. DYNJAWS is specifically designed to simulate dynamic wedges. DYNJAWS is based on the JAWS CM available in earlier releases of the BEAMnrc distribution. In the DYNJAWS CM both dynamic (mode 1) and step-and-shoot (mode 2) modelling of the dynamic wedge is possible. Moreover this CM functions as a JAWS CM in static mode (mode 0) where the jaws are stationary during the simulation, representing standard orthogonal jaws.
Changes in the lateral position (opening) and longitudinal position (height, or distance from the source) of the jaws can be varied during a simulation using the CM in mode 1 or mode 2, through the use of a specific, detailed input file. This file is required in addition to the main input file used by the BEAMnrc simulation. (Hereafter the term ‘input file’ is used to refer to this specific DYNJAWS file and should be not be confused with the main BEAMnrc ‘egsinp’ input file). In mode 0, where the jaws are stationary, no additional input file is required.

5.1.2 Input file for DYNJAWS

The input file for DYNJAWS first specifies the number of sets of jaw settings or subfields which will be used in the simulation. The probability of selecting each subfield as well as the resulting jaw positions (for both X and Y jaws) and the Z-distance (from source to the front and back of the jaws) are also specified. The jaw position specification includes the front positive (FP), back positive (BP), front negative (FN) and back negative (BN) coordinates (for both X and Y jaws) as shown in figure 5.1. For example, in the case of the Y jaw, the front positive coordinate is abbreviated as YFP. The Z distance includes Z-min and Z-max for each pair of jaws, where Z-min is the distance from the source to the front of the jaw and Z-max represents the distance to the back of the jaw.
In DYNJAWS the probability of selecting a particular subfield is represented by the index variable which must be specified for all subfields. In the step-and-shoot method, a random number, $m_i$, is generated from the interval [0, 1] at the beginning of each incident history. The segment, $i$ is used if

$$index(i - 1) < m_i \leq index(i)$$  \hspace{1cm} (1)

Segment $i = 1$ is used if

$$m_i \leq index(1)$$  \hspace{1cm} (2)

To simulate the jaw motion in the dynamic mode, a similar comparison is carried out with a random number (using equation 1 and 2); however the jaw settings are selected after
interpolation between segment $i$ and $i-1$. By including this extra interpolation between the discrete subfields that define each jaw motion, the dynamic mode (mode 1) provides a more realistic representation of the physical sweeping motion of the jaw than the step-and-shoot mode (mode 2).

5.2 Methods and Materials

5.2.1 Experimental measurements

Profiles produced using the Varian enhanced dynamic wedge (EDW) were measured for 6 MV and 10 MV photon beams using the MatriXX device (shown in figure 5.2) with OmniPro I’mRT software (IBA dosimetry, Schwarzenbruck, Germany) for three field sizes ($5 \times 5 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$, and $20 \times 20 \text{ cm}^2$) and four wedge angles ($15^\circ$, $30^\circ$, $45^\circ$ and $60^\circ$), at $d_{\text{max}}$ and at 10 cm depth. The MatriXX ion chamber array consists of 1020 vented parallel plate ionization chambers (0.5 cm height, 0.45 cm diameter and 0.08 cc sensitive volume) arranged in a $32 \times 32 \text{ cm}^2$ matrix.

Figure 5.2: Experimental setup showing the MatriXX diode array with solid water placed on top.
Taking account of the inherent build up of the MatriXX, solid water was placed on the top of the array to achieve the depths of interest. The SSD of 100 cm was set to the top of the solid water. The dose delivery was carried out on a Varian Clinac iX machine (with up to 2 Gy delivered for each profile) with Y1-IN wedge orientation. The ion chamber array was corrected for the background signal and the relative ion chamber sensitivity. The dynalog files stored on the linear accelerator were recovered and used for the MC simulations. Dynalog files provide a complete treatment snapshot. The information about the expected and actual jaw movement versus the monitor unit delivery is stored in these files (for more details about the dynalog file refer to the Varian EDW documentation 33).

5.2.2 Probability calculation

The probability of selecting a subfield (the index variable in DYNJAWS) was calculated as shown in figure 5.3 from the segmented treatment tables (STTs) supplied by the vendor for each wedge angle and energy 33. An STT describes the relationship between the delivered dose versus the jaw position. There are 20 control points in each STT irrespective of the selected field size. For each energy there is a golden STT for 60º EDW from which the STT for the rest of the wedge angles is derived.
The STT specifies the jaw position versus the dose delivery information at different points in an EDW delivery. The first step was to select the relevant STT based on the photon energy and the wedge angle. The selected STT was then truncated according to the delivered EDW field size. The delivered jaw positions were read from the dynalog files. Jaw positions from the dynalog files were compared to the jaw positions in the STT. If they were matching then the corresponding dose value entry in the STT was used as the required probability for selecting that jaw position or subfield. However, if the dynalog derived jaw position did not match the STT jaw position entry, then interpolation was carried out to determine the probability value from the STT. This procedure was repeated for all subfields. After calculating the probability for all subfields, the probability values were renormalized with the probability for the last subfield set to unity (by dividing all the subfield probability values by the probability of the last subfield).

Figure 5.3: Probability calculation flow chart.

The STT specifies the jaw position versus the dose delivery information at different points in an EDW delivery. The first step was to select the relevant STT based on the photon energy and the wedge angle. The selected STT was then truncated according to the delivered EDW field size. The delivered jaw positions were read from the dynalog files. Jaw positions from the dynalog files were compared to the jaw positions in the STT. If they were matching then the corresponding dose value entry in the STT was used as the required probability for selecting that jaw position or subfield. However, if the dynalog derived jaw position did not match the STT jaw position entry, then interpolation was carried out to determine the probability value from the STT. This procedure was repeated for all subfields. After calculating the probability for all subfields, the probability values were renormalized with the probability for the last subfield set to unity (by dividing all the subfield probability values by the probability of the last subfield).
The probability values and the jaw positions were then written to the input file, with each jaw position corrected for the trigonometric difference between the jaw position projected to the isocentre (listed in the dynalog files) and the physical distance between the jaw and the central axis (required by the BEAMnrc input file).

5.2.3 Automation of the input file generation

As part of this study, a script named AUTODJAWS was written in Matlab (version 7.8.0.342, The MathWorks, Natick, USA) to automate and generate the input file. The GUI of the AUTODJAWS software is shown in figure 5.4. When using AUTODJAWS, the user selects the relevant dynalog file, with the ‘Select Dynalog File’ button, and then selects the wedge angle, jaw orientation, beam energy and collimator angle. For a given field size the coordinates in the dynalog file are identical for all wedge angles, however the number of monitor units (MU) delivered at each coordinate position differ according to the wedge angle. Therefore for a given field size, the user only needs one dynalog file and can select the appropriate wedge angle to generate the required input files for any wedge angle. Similarly, for 6 and 10 MV the jaw coordinates are the same for a particular field size; however the proportion of the total number of MUs that are delivered at a particular position is different. The user may also specify the position of the X-jaws, which by default will be parked at 10 cm on each side of the central axis. Finally by selecting the ‘Generate Input File’ button the required file is created (this includes the probability calculation described in the previous section).
5.2.4 MC simulations

All MC simulations were carried out on a SGI Altix XE Cluster, with 200 cores. Simulations were performed to generate the EDW dose profiles at photon beam energies of 6 MV and 10 MV for three different field sizes ($5 \times 5 \text{ cm}^2$, $10 \times 10 \text{ cm}^2$ and $20 \times 20 \text{ cm}^2$) at two depths ($d_{\text{max}}$ and 10 cm) for four wedge angles ($15^\circ$, $30^\circ$, $45^\circ$ and $60^\circ$). The optimized electron beam FWHM and incident electron energy combination for the models were found to be 1 mm, 5.875 MeV and 1 mm, 9.8 MeV for the linear accelerator operating at 6 MV and 10 MV respectively during the MC commissioning process. During commissioning the measured lateral dose profiles and percentage depth doses (PDDs) were compared with their MC counterparts for a number of field sizes (starting from $40 \times 40 \text{ cm}^2$ and going down to $5 \times 5 \text{ cm}^2$) at various depths. The measured and simulated profiles were compared at 2%-2mm gamma criteria. A phase-space file, containing the positions, trajectories, energies and charges of all particles exiting the linear accelerator was scored (starting with 22 million initial electron histories, earlier while calculating incident electron energy it was found that 22...
million histories give a reasonable particle population in the phase space without increasing the file size and computational overhead too much and further the phantom dose calculation as described further error was well below 1% suggesting the sufficiency of the particles) at a distance of 55 cm from the photon source. This scored phase-space file was used as an input in the phantom dose calculations in the second stage.

The simulated phantom dimensions were 56 cm × 32 cm × 6 cm (length x width x height). The MatriXX array was modelled in 3 layers, with reference to the manufacturer’s specifications. The first (build-up) layer has a physical thickness of 3.3 cm, and an unknown composition, but is described in the manufacturer’s documentation as having a 0.33 cm water-equivalent thickness. This layer was therefore modelled as a 3.3 cm thickness of water with a density of 0.1 g/cm³. The second (active) layer consists of an array of 0.5 cm thick ionisation chambers embedded in a water-equivalent medium. This layer was modelled as a 0.5 cm thickness of water, according to the common procedure of modelling ion chambers as water equivalent in MC simulations. The final (backscatter) layer of the MatriXX consists of a 2.2 cm thickness of polystyrene (98%) and titanium dioxide (2%), with an unspecified structure, at a density of 1.045 g/cm³. Given that a material composed of 98% polystyrene and 2% titanium dioxide has an electron density 0.998 times the electron density of standard polystyrene (evaluated using the mass fractions, molecular masses and atomic numbers of the component elements), this backscatter layer was modelled, for simplicity, as a 2.2 cm thickness of polystyrene with a density of 1.045 g/cm³. An atomic number of 22 could lead to problems for scattered radiation however as the intent was to compare the relative doses therefore this simplification was justified. The additional solid water placed on the MatriXX surface to achieve the desired depths was also modelled in the phantom design. The voxel size used was 7.62 mm along the wedge direction, which is similar to the inherent resolution of the MatriXX array. For the dose calculations in this simulated phantom, 6 billion histories were run.
producing calculated doses with a statistical uncertainty was well below 1% for all the cases. The statistical uncertainty (further referred as MC error) is calculated on history by history bases in the BEAMnrc code and is dependent on the number of histories and the variance reduction technique. Scored quantities such as fluence, energy etc are grouped according to primary history and then the root mean square standard deviation for the mean of groupings is calculated.112

5.2.5 Back scatter correction

An important consideration for modelling an EDW is the effect of backscatter into the monitor chamber from the upper Y-jaws; this usually results in shorter delivery time for small field sizes. Various authors have quantified the backscatter into the monitor chamber.46, 109, 113, 114 In this study the empirical correction suggested by Ahmed et al. has been incorporated into the STTs to account for the back scatter.115 Equation 3 and 4 has been used for 6 and 10 MV photons respectively.

\[
F_{bc}(y) = 1.03 - 7.03 \times 10^{-4} y \quad (3)
\]

\[
F_{bc}(y) = 1.02 - 4.9 \times 10^{-4} y \quad (4)
\]

Whereas \( F_{bc} \) represents the back scatter correction function, and \( y \) represents the Y-jaw position in cm. This correction is reflected in the probability values calculated from the backscatter corrected STTs.
5.3 Results

5.3.1 Input file Generation

Input files were generated using the AUTODJAWS script. These input files specified the details for all the subfields including the probability of selection of a subfield, X and Y jaw coordinates and the respective Z distance to the front and back of the jaws from the target plane. Figure 5.5 illustrates a sample input file, where the entries for first 2 fields are only shown.

Figure 5.5: Input file generated by the AUTODJAWS code.

The accuracy of these input files was confirmed by independent manual calculation of the probabilities and the jaw positions.

5.3.2 Validation

5.3.2.1 Wedged profiles

Comparison of wedged profiles for the 6 and 10 MV beams at $d_{\text{max}}$ and 10 cm depth is shown in figure 5.6 for $20 \times 20$ cm$^2$ and $10 \times 10$ cm$^2$ field sizes. For a particular field size, energy and depth combination, all wedge angles simulated are plotted on the same graph. It can be seen in figure 5.6 that there is an excellent agreement between the measured and simulated data for 6 MV and 10 MV photons at both depths (within 3% or 3 mm for all data points). Similar
agreement was observed for $5 \times 5 \text{ cm}^2$. All the simulations were carried out using the dynamic mode of the DYNJAWS (mode 1).

5.3.2.2 Open field and off axis profiles

In figure 5.7(a-c) measured and simulated wedged profiles for 6 MV photons have been compared at $d_{\text{max}}$ using the static mode. Figure 5.7d compares measured and simulated profiles for a corner field (using a 60 degree dynamic wedge and a $10 \times 10 \text{ cm}^2$ off-axis field). For all data points the results agreed within 3% or 3 mm criteria. Similar results were found for 10 MV.
Figure 5.6: Comparison of measured and simulated wedged profiles for 20 × 20 cm² and 10 × 10 cm² field size at 100 cm SSD for four wedge angles (from top to bottom 60°, 45°, 30° and 15°) at d_{max}(1.5 cm for 6 MV and 2 cm for 10 MV) and 10 cm depth. Both the simulated and measured doses are normalized to 100% at the centre of the field. a) 20 × 20 cm² - 6 MV profiles at d_{max} b) 20 × 20 cm² - 6 MV profile at 10 cm depth c) 20 × 20 cm² - 10 MV profile at d_{max} d) 20 × 20 cm² - 10 MV profile at 10 cm depth. e) 10 × 10 cm² - 6 MV profiles at d_{max} f) 10 × 10 cm² - 6 MV profile at 10 cm depth g) 10 × 10 cm² - 10 MV profile at d_{max} and h) 10 × 10 cm² - 10 MV profile at 10 cm depth. The error in Monte Carlo dose calculation was less than 1%. The 10 × 10 cm measured profiles were smoothed using a spline interpolation.
Figure 5.7: Comparison of measured and simulated open field and wedged profiles at $d_{max}$. Both the simulated and measured doses are normalized to 100% at the centre of the field. a) $20 \times 20$ cm$^2$ open field for 6 MV b) $10 \times 10$ cm$^2$ open field for 6 MV c) $5 \times 5$ cm$^2$ open field for 6 MV d) 6 MV Corner field (60°) $10 \times 10$ cm$^2$ profile. The error in Monte Carlo dose calculation was less than 1%.

5.3.2.3 Dynamic versus step-and-shoot

Figure 5.8 illustrates the comparison of dynamic and step-and-shoot modes for 60 and 15 degree simulated wedged profiles with the measurements for two field sizes ($20 \times 20$ cm$^2$ and $10 \times 10$ cm$^2$) for 6 MV photons at $d_{max}$.

Figure 5.8: Comparison of dynamic and step-and-shoot mode for 6 MV at $d_{max}$. Both the simulated and measured doses are normalized to 100% at the centre of the field. a) $20 \times 20$ cm$^2$-6 MV-60° b) $10 \times 10$ cm$^2$-6 MV-60° c) $10 \times 10$ cm$^2$-6 MV-60° d) $5 \times 5$ cm$^2$-6 MV-60°.
A gamma comparison (at 3% or 3 mm) of the dynamic and step-and-shoot modes was carried out and the mean gamma values are reported in table 5.1. The results suggest that the dynamic mode is more accurate, especially for smaller wedge angles at large field sizes. For example, the dynamic mode was more accurate than the step-and-shoot mode when simulating a 15° dynamic wedge with a $20 \times 20$ cm$^2$ field.

**Table 5.1:** Comparison of gamma values for dynamic and step-and-shoot modes for 6 MV photon beams.

<table>
<thead>
<tr>
<th>Field size (cm$^2$)</th>
<th>Dynamic 60°</th>
<th>Dynamic 15°</th>
<th>Step-and-shoot 60°</th>
<th>Step-and-shoot 15°</th>
</tr>
</thead>
<tbody>
<tr>
<td>10 x 10</td>
<td>0.17</td>
<td>0.17</td>
<td>0.22</td>
<td>0.19</td>
</tr>
<tr>
<td>20 x 20</td>
<td>0.40</td>
<td>0.17</td>
<td>0.46</td>
<td>0.30</td>
</tr>
</tbody>
</table>

**5.4 Discussion and summary**

The results confirm the suitability of the new DYNJAWS CM for modelling the dynamic wedges. In addition to being a dedicated CM for modelling dynamic wedges, DYNJAWS eliminates the need for the intermediate phase space files that were required in some previously reported approaches.\(^{116}\) In the static component simulation (SCS) method a number of phase space files were created at different jaw positions and then summed.\(^{46}\) When using DYNJAWS, a single phase space file is created whether simulating in dynamic or step-and-shoot modes. The dynamic mode of the new CM is truly dynamic because it carries out an extra interpolation for selecting the jaw positions after a particular subfield is selected for the delivery. The step-and-shoot mode in the DYNJAWS CM is based on Verhaegen and Liu’s probability positioning sampling (PPS) approach\(^ {46}\), while the dynamic mode is a new
addition. The open field profiles in figure 5.7 demonstrate that the DYNJAWS CM can be used as a standard JAWS CM when applied in the static mode. Figure 5.7d also provides a useful example of the use of the DYNJAWS CM to simulate a non-symmetric wedged field.

Data shown in table 5.1 and figure 5.8 suggest that the use of the dynamic mode simulations is especially beneficial when wedge angles are small, field sizes are large or the region under examination is located away from the central axis. The overall mean gamma value results indicate that dynamic mode is superior to the step and shoot mode for simulating the EDWs. In these circumstances, the step-and-shoot mode’s lack of interpolation between the jaw positions at each subfield defined in the dynalog file leads to noticeable inaccuracy in the positioning of the simulated jaw. We have observed that the dynamic mode appears to require slightly more CPU time than the step-and-shoot mode, possibly due to the additional interpolation of the jaw values that is carried out in this mode.

The AUTODJAWS script accurately generates the required input files which simplifies the use of the DYNJAWS CM. These results validate the capability of the DYNJAWS CM for simulating the dynamic wedges. The use of this CM can be made further efficient by automating the process of the input file generation. The DYNJAWS CM is an important addition to the BEAMnrc code and will allow the simulation and Monte Carlo validation of complex patient treatments involving the use of dynamic wedges.

In this chapter the Monte Carlo model has been commissioned using the DYNJAWS CM. The next chapter will utilize the results to examine the interplay effects using Monte Carlo methods and exploring the plan complexity issue.
Chapter 6 : Modelling Enhanced Dynamic Wedge Interplay using Monte Carlo methods

6.1 Introduction

This chapter investigates the interplay effects further using Monte Carlo (MC) methods. Previously in chapter 3 and 4 experimental methods have been employed to study the interplay effects. However those studies have not investigated a few aspects. Questions such as the combination of different wedged fields and its potential impact on the interplay have not been examined. Furthermore the effect of direction of motion between the wedge and tumor have not been studied in case of a multiple field treatment. The 4D CT data set acquired in chapter 4 can be used to initiate MC studies, which are generates three dimensional dose maps which can be further analysed using three dimensional gamma analysis.

The previous chapter has validated a Monte Carlo enhanced dynamic wedge (EDW) model using the DYNJAWS component module (CM) of the BEAMnrc user code. In this chapter the interplay effects will be simulated using the BEAMnrc and the DOSXYZnrc MC codes. Two main sets of simulations have been carried out. In the first stage the interplay related dose discrepancy has been investigated using the individual respiratory phases from the 4D CT data set. For simulating the dynamic wedges the previously commissioned DYNJAWS model has been used.

This chapter has been divided into three main sections. The methods and materials are presented in section 6.3 where the treatment plan parameters and the MC specific details are presented. Section 6.4 presents the results along with the analysis. Finally the conclusions and a brief discussion are the subject of section 6.5.
6.2 Methods and Materials

6.2.1 Monte Carlo simulation of the treatment head

For the Monte Carlo simulation of the treatment head the BEAMnrc user code was used. The EDWs were simulated using the dynamic mode of the DYNJAWS component module (validated and described in the previous chapter). The coordinate input files for DYNJAWS were generated using the AUTODJAWS code (detailed in section 5.2.3). The FWHM and energy of the incident beam were 1 mm and 5.875 MeV respectively for 6 MV photon beam. Each treatment field was 10 ×10 cm². For variance reduction the directional bremsstrahlung splitting technique was used with 12 cm diameter. The photon and electron energy cut off values were 0.01 MeV and 0.7 MeV, respectively. For the treatment head simulation 22 million primary particle histories were utilized and the phase space was scored at 55 cm from the source for 10°, 15°, 20°, 25°, 30°, 45° and 60° EDWs for both Y1-IN and Y2-OUT jaw motions. These phase space files containing the positions, trajectories and charges of all particles were used for the DOSXYZnrc phantom dose calculation as described in the next sections.

6.2.2 Phantom dose calculation

The phantom dose calculation was carried out using the DOSXYZnrc user code. The 4D CT data acquired for the IMRT quality control phantom in section 4.2.3 was further used to create CT based phantoms to be used in DOSXYZnrc BEAMnrc’s inbuilt ctcreate code was not suited for creating the CT phantom because of its memory limitations and its incompatibility with the Toshiba Acquilion CT scanner with which the 4D CT scan was obtained. Therefore the Monte Carlo DICOM tool kit (MCDTK) developed by Crowe et al. ¹⁰³ was employed instead of ctcreate. MCDTK is a flexible Java-based MC suite which can generate CT phantoms and automate the generation of MC input files from a treatment plan. This toolkit is very useful for
the automation of the MC input file generation from a treatment plan and for the subsequent image analysis including the production of gamma maps and the dose volume histograms.

Ten MC phantoms were generated, corresponding to the ten respiratory phases obtained from the 4D CT. Figure 6.1 shows an illustration of the respiratory trace from the 4D CT acquisition and Figure 6.2 illustrates the longitudinal displacement of the moving phantom in each of the ten respiratory phases.

**Figure 6.1**: 4D CT acquisition, respiratory trace of the moving phantom.

The resolution of each phantom created for each respiratory phase was 200 cm × 125 cm × 160 cm. After the generation of the MC phantoms corresponding to each respiratory phase, the treatment fields were simulated in the DOSXYZnrc code using the phase space files generated in treatment head simulations. For phantom dose calculation 9 billion primary particle histories
were used. The phase 5 scan was used to model the static case because the phantom was in the middle of motion, as evident from figure 6.2.

![Graph: Motion of the phantom as a function of slice offset](image)

**Figure 6.2:** Motion of the phantom as a function of slice offset

### 6.2.3 Simulating individual respiratory phases using DYNJAWS

In the first investigation a four field wedged treatment was simulated to demonstrate the dose discrepancy between the various respiratory phases. A four field wedged treatment was planned as shown in figure 6.3, this plan was similar to the one reported in table 4.2, however here a symmetric field size was used (10 × 10 cm²). Two fields had a wedge angle of 30 degrees and the other two fields had a wedge angle of 60º. Initially both 30º EDWs were oriented so that the direction of jaw motion was parallel to the phantom motion, while one of the 60º EDWs was oriented parallel to the phantom motion and the other was perpendicular. This treatment was
also re-simulated with all collimator angles rotated by 90°, to investigate the effect of orienting three EDWs in parallel and one EDW perpendicular to the phantom motion.

For both the initial and the collimator-rotated treatment plan, nine DOSXYZnrc simulations were carried out using the DOSXYZnrc phantoms created from the individual respiratory phase CT datasets. For each treatment to each respiratory phase phantom, the four fields were summed together and a dose volume histogram (DVH) was generated for a region of interest representing the tumor. The DVHs for all phases were averaged in order to provide an approximate DVH for the moving tumour which was compared with the phase 5 (static phantom) DVH.

![Figure 6.3: Treatment planning snapshot for individual phase simulation.](image-url)
6.2.4 Single field EDW deliveries

Single field 15° (figure 6.4) and 60° (figure 6.5) EDWs were simulated for a fixed 0° gantry angle, while utilizing parallel (0°) and perpendicular (90°) collimator motions. The Phase 5 phantom was used in DOSXYZnrc for scoring the dose. For each treatment field, 11 phantom simulations were carried out shifting the isocentre coordinates in z plane by 2 mm to mimic the phantom motion.

Figure 6.4: Treatment planning snapshot for 15° parallel motion.
6.2.5 Comparison of four field EDW treatment for 15 and 60 wedge angles.

Two four field plans were compared for studying the effect of using a combination of large (60°) and small (15°) wedge angles in a multiple field delivery. The gantry angles were similar to those described in figure 6.4 and figure 6.5 while the collimator motion was parallel to the movement of the phantom. The Phase 5 phantom was used in DOSXYZnrc for scoring the dose. For each treatment field 11 simulations were carried out shifting the isocentre coordinates in z plane by 2 mm to mimic the phantom motion.
6.2.6 Comparison of large and small wedge angles

A further two sets of simulations from the predefined gantry angles were carried out, combining two different sets of wedge angle, the first combination was referred to as the large wedge angle plan, comprising of 60º, 45º, 30º and 25º EDWs. The second set of wedge angles was called the small wedge angle plan, consisting of 10º, 15º, 20º and 25º EDWs. For each treatment field 11 simulations were carried out shifting the isocentre in z plane by 2 mm to mimic the phantom motion. The collimator motion was parallel to the phantom motion, with a collimator angle of 0º.

6.2.7 Analysis

For the analysis of the Monte Carlo images, MCDTK was used. The phase 5 image was taken as the reference image compared with the motion images. All the dose maps from each respiratory phase were averaged together and compared with the Phase 5 image. A three dimensional gamma analysis was carried out at 3%-3mm. A region of interest (ROI) was selected from the MC dose images comprising of the tumor volume and for this ROI dose volume histograms (DVHs) and gamma volume histograms (GVH) were generated and compared. GVH quantitatively specify the gamma values versus volume of the region of interest information. They are used for assessing the failure/passage of the gamma criterion in the volume of interest. The treatment plans used in this part of the study were designed for analysing the effects of EDW angle, EDW orientations and number of EDWs only. Many simplifications have been made so that these effects could be analysed independently of other treatment planning parameters. Beam weights and gantry angles have been kept constant, despite the varying effects that the different EDWs have on the dose distribution. As a result, the dose to the planning target volume (PTV) shows, in many cases, much more heterogeneity than would be acceptable in a patient plan. In summary, these plans were designed to enable relatively
straightforward data analysis, rather than to provide a clinically valid representation of the dose that might be delivered to a PTV in actual treatments.

6.3 Results

6.3.1 Individual respiratory phases

The results of simulating the 4-field, 30 and 60 degree EDW treatment, with a combination of different wedge angles, with and without a 90 degree collimator rotation (as described in section 6.2.3) are shown in figure 6.6. This figure compares the DVHs obtained from simulations of the treatments to respiratory phase 5, the static case, with DVHs obtained by averaging the 9 DVHs obtained from simulating the each treatment to all 9 respiratory phase phantoms (the motion case). Although a slightly larger difference can be observed in figure 6.6(b), where most EDWs are moving perpendicularly to the tumour, the only small differences between the static and motion cases for both rotated and non-rotated collimator angles can be observed in the DVHs shown in Figures 6.6(a) and (b). This suggests that the interplay effects resulting from the different wedges and wedge orientations maybe cancelling out. For this reason, the effects of single and multiple wedged fields, with uniform wedge and collimator angles, are examined in the following sections.

6.3.2 Single field deliveries

The DVHs and GVHs are presented in figure 6.7 for static and motion cases for both 15º and 60º EDWs for parallel jaw-phantom motion. There are small differences between the DVHs obtained for the static and motion cases for both wedge angles when the collimator and phantom motion are parallel as indicated by the plots in figure 6.7a. For perpendicular jaw-phantom motion (figure 6.7b), there are larger differences between the static and motion cases, and the relative discrepancy between the 60º static and motion cases is larger than between the 15º static
and motion cases. This is evident from the large distance between the green and purple lines compared to red and blue lines in figure 6.7b.

Figure 6.6: Individual phase dose volume histogram, a) Individual phase without collimator rotation (two 30° EDWs and one 60° EDW moving parallel to the phantom and one 60° EDW moving perpendicularly to the phantom), b) Individual phase with collimator rotation (two 30° EDWs and one 60° EDW moving perpendicularly to the phantom and one 60° EDW moving parallel to the phantom).
Figure 6.7: Comparison of single field 15° and 60° treatments a) Dose volume histograms for 15° and 60° EDWs for parallel motion b) Dose volume histograms for 15° and 60° EDWs for perpendicular motion.
The gamma analysis results comparing the EDW static versus EDW dynamic at 3%-3mm for 15° and 60° EDW are presented in figures 6.8 and 6.9 for parallel and perpendicular motions, respectively.

**Figure 6.8:** Gamma dose images 15° (a-c, representing the x, y and z dimensions) and 60° (d-e, representing the x, y and z dimensions) EDWs for parallel collimator motion.

For parallel motion in all three planes (x, y and z) the disagreement between the static and motion cases is higher for the 60° EDW (figures 6.8d-f).

**Figure 6.9:** Gamma dose images 15° (a-c, representing the x, y and z dimensions) and 60° (d-e, representing the x, y and z dimensions) EDWs for perpendicular collimator motion.
Additionally, for perpendicular motion, apart from the x plane (figure 6.9d), the disagreement between the static and motion cases is higher for the 60º EDW (figures 6.9e-f).

The gamma volume histograms for both wedge angles and collimator angles for the selected ROI are shown in figure 6.10. These plots also suggest that gamma indices are higher for 60 degree wedges than for 15 degree wedges and higher for perpendicular motion than for parallel motion.

![Gamma dose volume histograms for 15º and 60º EDWs for parallel and perpendicular collimator motion.](image)

Figure 6.10: Gamma dose volume histograms for 15º and 60º EDWs for parallel and perpendicular collimator motion.

6.3.3 Four field deliveries

DVHs obtained from simulating the four-field, uniform-wedge-angle plans shown in figures 6.4 and 6.5 are presented in figure 6.11. Here all wedges were oriented parallel to the direction of phantom motion. It can be seen that the difference between static and motion cases is more
pronounced when all four EDWs have a 60° wedge angle than when all four EDWs have a 15 degree wedge angle.

![Four field-15 and 60 EDW-DVHs](image)

**Figure 6.11:** Dose volume histograms for four field 15° and four field 60° EDWs comparing the static and motion images.

This trend can also be further confirmed from the GVHs presented in figure 6.12. In figure 6.12 the gamma values for the 60° EDW are higher than the corresponding 15° case, indicating higher disagreement in the selected ROI between the static and motion images for the larger wedge angle.

### 6.3.4 Comparison of large and small wedge angles

The last set of results compare the four-field treatment simulations (described in section 6.2.6) performed for the large wedge angle plan (which used 60°, 45°, 30° and 25° EDWs) and the small wedge angle plan (which used 10°, 15°, 20° and 25° EDWs). Here all wedges were
oriented parallel to the direction of phantom motion. The DVHs for both sets of simulations are similar, as indicated by the plots in figure 6.14 and the gamma volume histograms shown in figure 6.15. However a close inspection reveals that the difference between the static and motion for the large wedge angle plan is slightly higher than for the small wedge angle plan.

**Figure 6.12:** Gamma volume histograms for four field 15º and four field 60º EDWs.

The gamma analysis results at 3%-3mm for four field 15º and 60º EDW are presented in figures 6.13.
Figure 6.13: Gamma dose images for four field 15° (a-c, representing the x, y and z dimensions) and four field 60° (d-e, representing the x, y and z dimensions) EDWs.

Figure 6.14: Dose volume histograms for four fields large and four field small EDWs comparing the static and motion images.
Figure 6.15: Gamma volume histograms for four field large and four field small EDWs.

The gamma images for the four field delivery comparing the 15° and 60° EDWs at 3% 3 mm are shown in figure 6.16.

Figure 6.16: Gamma dose images for four field 15° (a-c, representing the x, y and z dimensions) and four field 60° (d-e, representing the x, y and z dimensions) EDWs.
6.4 Discussion and Conclusions

Simulation of a four field EDW treatment with 2 wedge angles and 2 wedge orientations resulted in small differences between DVHs for the static and moving phantom when the simulations were performed using all respiratory phases, if most of the wedges were oriented perpendicularly to the phantom motion direction. When the same treatment plan was modified by rotating all collimator angles by 90°, so that most of the wedges were oriented parallel to the phantom motion direction, then the differences between the DVHs for the static and moving phantom were reduced. To elucidate the causes for this behaviour, some simpler treatment plans were simulated using the phase 5 respiratory phantom.

Comparisons of the results of simulations of single field treatments with 15° and 60° EDWs indicated that there were small differences between DVHs obtained for the static and moving phantom when the EDWs were moving parallel to the phantom motion and much more substantial differences between DVHs obtained for the static and moving phantom when the EDWs were moving perpendicularly to the phantom motion.

In chapter 3 it was observed that perpendicular jaw-phantom motion results in dose blurring, while parallel jaw-phantom motion results in a stepped dose profile. Evidently, the dose blurring, which produces large regions of under dosage in the tumour, has a more noticeable effect on the tumour DVH and leads to the perpendicular motion results showing greater deviation from the static simulation results.

Phantom motion has a larger effect on dose for large wedge angles than small wedge angles, and this effect is increased when the wedge is moving perpendicularly to the phantom. This has been observed in the DVH data for both one and four field plans and confirmed by the gamma analyses reported in this chapter. Again, this confirms the observation made in chapter 3, that dose differences caused by phantom motion are greater for larger wedge angles.
Examination of two four-field EDW plans, which each used four different wedge angles, indicated that motion caused a greater effect on the tumour DVH when the wedge angles were large than when the wedge angles were small.

Importantly, the dose differences caused by phantom motion were greater for the four field plan which used a uniform 60° wedge angle and a uniform parallel orientation, than for the four field plan which used four different wedge angles and a uniform parallel orientation. Additionally, the latter plan showed more substantial dose differences caused by phantom motion than were seen for the initial four field plan which used two different wedge orientations. Specifically, comparison of the results shown in figure 6.11, where all EDWs in each plan have the same orientation and the same wedge angle, with the results shown in figure 6.6, where the four EDWs have two different orientations and two different wedge angles, suggests that the effects of phantom motion on the dose to the tumour can be exacerbated by the selection of uniform wedge angles and orientations.

It is therefore possible to make the following clinical recommendations, for the planning of lung treatments that utilises EDWs: If the patient has a noticeably extensive breathing motion, or if they are prone to frequent coughing, or if their tumour is located in the inferior lobe, close to the diaphragm, then EDWs with small wedge angles should be used in preference to EDWs with large wedge angles, where possible. Additionally, where multiple beams are used, the gantry and table angles should be selected so that EDWs can be used with different orientations relative to the major axis of patient respiratory motion. If the patient breathes shallowly or if their tumour is located in the upper lobe of the lung or fixed to the mediastinum, then the above recommendations can be relaxed.
Chapter 7 : Improving the gel image readout using X-ray computed tomography (CT)

7.1 Introduction

The preceding chapters have used 2D dose measurements and MC simulations to investigate the dose interplay effects caused by moving targets in dynamically wedged beams. This chapter describes the development of a method to improve the 3D dose readout from dosimetry gels scanned with x-ray CT, which can be helpful in providing a 3D dose verification of the interplay effects. CT evaluation was chosen because of the local expertise and because motion is more likely to impact on regions with strong dose gradients where the high spatial resolution and fidelity of CT would be advantageous. The aim of the current work is to investigate the feasibility of a simple image analysis technique whereby data from multiple scans is used to provide a hypothetical ‘zero-scan’ image representing the irradiated gel prior to CT scanning. A simple example of a normoxic polymer gel irradiated to a range of doses is used to establish that this method is capable of appreciably improving CT image quality.

7.2 Methods and Materials

This investigation is divided into two main sections. In first section the zero-scan method is developed. After the establishment of the proof of the principle in the second section a wedged plan is delivered to the gel dosimeter and the zero scan method is used to reconstruct the gel image.
7.2.1 Gel preparation and irradiation

A PAGAT gel dosimeter was prepared as described by Venning et al.\textsuperscript{71} with 8 mM of Tetrakis (Hydroxymethyl) Phosphonium Chloride for improved stability.\textsuperscript{70} The gel dosimeter was prepared under normal atmospheric conditions and poured into a cylindrical Polyethylene terephthalate (PET) container of 10 cm height and 5 cm radius. It was then stored at 4°C for 24 hours before irradiation. The gel dosimeter was irradiated (as shown in figure 7.1 and 7.2) with three small (1.5 × 1.5 cm\textsuperscript{2}) fields of 118 cGy, 233 cGy and 384 cGy parallel to the central axis of the container with a Varian linear accelerator using a 6 MV photon beam at 600 MU/min.

![Image](image.jpg)

**Figure 7.1**: Gel irradiation on the linear accelerator

A further 686 cGy was delivered using a fourth test field, close to the centre of the container. The area of this high-dose field was reduced to 1.0 × 1.0 cm\textsuperscript{2} to minimise possible scatter into the other test regions of the gel.
Figure 7.2: Irradiated gel showing the different ROIs as indicated by the arrows.

7.2.2. X-ray CT imaging

One day after irradiation, the gel was imaged using a GE Lightspeed RT 4 CT scanner (see figure 7.3). The CT scans used an X-ray tube load of 300 mA with 1s rotation, beam energy of 120 kVp, 5 mm slice thickness, image size of 512 × 512 and 25 cm field of view. The CT dose from this protocol is estimated as 83.4 mGy per scan at the scan centre, based on the ImPACT CTDI_{100}(soft tissue) (ImPACT CT Patient Dosimetry Calculator V1.0, ImPACT, London, UK). The gel was scanned 360 times at a single slice location while placed in a cylindrical water tank as shown in figure 7.3 similar to that described by Trapp et al.\textsuperscript{87}. The choice of acquiring 360 CT scans was made to gather enough data for reconstruction as there was no previous guideline available for the number of scans. The gel was positioned in the tank such that the CT scanning plane was orthogonal to the radiation beam direction. All scans consisted of one slice only, with
no couch motion, providing a transverse image of the phantom showing all four irradiated regions at a depth of 4 cm from the surface onto which the radiation was incident.

**Figure 7.3**: CT scan setup for the irradiated gel.

An additional set of 60 scans of the tank were obtained, with the gel removed such that the tank only contained water.

### 7.2.3 Image analysis

The CT images, in DICOM format, were imported into Matlab (version 7.8.0.342, The MathWorks, Natick, USA). The 60 images of the water phantom were averaged to reduce random noise.\(^8^7\) This averaged water image was then subtracted from each of the gel images, to remove CT artefacts from the gel images, as described by Trapp *et al.*\(^8^7\).

Three composite data sets were then obtained from the processed CT images, using the following method. For each pixel, the Hounsfield unit (HU) was acquired sequentially from the series of 360 processed CT images to form a data set of 360 data points for each gel voxel. For
each dataset linear, quadratic and exponential fits were applied to the data. Thus, estimates of the relationship between scan number and HU for each voxel were obtained using:

\[ HU(i, j) = L_n(i, j)N + L_0(i, j) \]  
(5)

\[ HU(i, j) = Q_n(i, j)N^2 + Q_0(i, j)N + Q_0(i, j) \]  
(6)

\[ HU(i, j) = E_n(i, j) + (A(i, j) \cdot e^{-N/B(i, j)}) \]  
(7)

where \( N \) is the scan number and the arrays \( L_n(i, j), Q_n(i, j) \) and \( E_n(i, j), B(i, j) \) and \( A(i, j) \) are free parameters for the linear, quadratic and exponential fits respectively, evaluated for each pixel \( (i,j) \). A new image was then constructed, referred to here as the zero-scan image, whereby each pixel in the new image is the intercept of the fit for the corresponding pixel in the processed CT images. In other words, the zero-scan images are maps of \( L_0(i, j), Q_0(i, j) \) and \( E_0(i, j) \). The rationale behind this technique is that the intercept of the fitted data most closely matches the properties of the gel dosimeter immediately before CT scanning commences.

In each of these three new images, four regions of interest (ROIs) were selected; each consisting of 121 pixels centred at the location of the 118, 233 and 384 and 686 cGy fields. One ROI was also selected corresponding to an un-irradiated region in the gel container. To provide measurements of the signal and noise in the images, the mean HU value and the standard deviation were calculated for these selected ROIs.

An analysis was also carried out to find out the optimum number of CT scans required for reconstructing the zero-scan image. Starting from the first 50 images and subsequently adding images up to 300, the average percentage error was calculated for each group of CT images. As a first step the linear reconstructed image using all the 360 scans was considered as the standard image. For each of the four ROIs described above, the mean HU value was calculated and
compared with the standard image, and the percentage error was calculated for each ROI. In the final step all the individual percentage errors in the four ROIs were averaged out and a single averaged percentage error was calculated for each group of CT images.

7.3 Results

7.3.1 Qualitative

In figure 7.4 three sets of zero-scan images are shown, reconstructed from sets of all 360 images, the first 50 images, and the first 16 CT images respectively. The relatively poor signal to noise affecting un-processed CT scans of dosimetric gels is apparent in Figure 7.4 (a) top left and top middle panels, which show the first and last of the 360 CT images of the gel dosimeter series. Both images are windowed to the same pixel values and an overall increase in HU is evident, above noise, in the overall lighter appearance of 360th image. The source of this increase in HU is the gel density increase due to the radiation dose delivered during the CT imaging. Figure 7.4(a) top right panel shows an average of all of the CT images of the gel dosimeter, and a reduction of noise compared to the single images can clearly be seen. Lower left, middle and right panels show the zero-scan images created from exponential, linear and quadratic fits to the data as described in Section 7.2.3. A reduction of noise compared to the upper panels is clearly evident.

Figure 7.4(b) and 7.4(c) show results where only the first 50 and first 16 images of the dataset respectively were used to create the zero-scan image. As fewer images are used an increase in
Figure 7.4: Zero-scan reconstructed images. All images are windowed to the range 19-24 HU. a) Top left-First CT image, top center-360th CT image, Top-right-Averaged CT image. Lower left- zero-scan image from exponential fit, Lower middle- zero-scan image from linear fit, Lower right- zero-scan image from quadratic fit. b) Top left-First CT image, top center-50th CT image, Top-right-Averaged CT image. Lower left- zero-scan image from exponential fit, Lower middle- zero-scan image from linear fit, Lower left- zero-scan image from quadratic fit. c) Top left-First CT image, top center-16th CT image, Top-right-Averaged CT image. Lower left- zero-scan image from exponential fit, Lower middle- zero-scan image from linear fit, Lower left- zero-scan image from quadratic fit.
noise in the zero-scan image is clearly visible when a quadratic fit is used and less apparent when exponential and linear fits are used.

7.3.2 Quantitative

Figures 7.5(a) and 7.4(b) show examples of the variation in the HU values of two individual pixels throughout the acquisition of the 360 CT images. A linear fit to the data for the pixel that received 686 cGy (Figure 7.5(a) has a steeper gradient (0.007 ± 0.001)) than the linear fit to the data for the non-irradiated pixel (0.004 ± 0.001) as shown in figure 7.5(b). This suggests that the un-irradiated gel is less sensitive to the additional dose increments delivered during the scanning process, possibly due to inhibition of the low-dose response caused by the residual presence of oxygen within the non-irradiated gel, as described by DeDeene et al. 87, 117, 118 Moreover, variations in manufacturing conditions may lead to different concentrations of residual oxygen between batches leading to a variations in response to the CT dose utilized in this technique. Therefore, this technique remains suitable only for relative dosimetry unless an internal absolute calibration is undertaken.119-121
Figure 7.5: (a) HU value of a single pixel inside the irradiated ROI (686 cGy), over the 360 images. (b) HU value of a single pixel outside the irradiated ROIs, over the 360 gel images c) Averaged HU values within irradiated and non irradiated ROIs (121 pixels) with linear and exponential fits. The error bars represent the mean standard deviation within the ROI.

Figure 7.5(c) shows the mean pixel value of an ROI of 121 pixels calculated in the regions corresponding to irradiated and non irradiated portions of the gel, and further averaged over increasing numbers of images starting with the first raw CT/zero image. The data in figure 7.5(c) has been fitted with linear and exponential fits. Results from statistical analysis of the exponential and linear fits using 360 images are shown in table 7.1, which suggests that the linear and exponential fits yield similar results. Although only the data for a single pixel within the 686 cGy field is shown, similar results are noted for other pixels in this and other fields.
Table 7.1: Analysis of linear and exponential fits.

<table>
<thead>
<tr>
<th>Fit type</th>
<th>Equation</th>
<th>Coefficients(95 % confidence bounds)</th>
<th>SSE</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>Linear</td>
<td>$HU = L_1N + L_0$</td>
<td>$L_1 = 0.005804$ (0.00463, 0.006979)</td>
<td>0.04509</td>
<td>0.9792</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$L_0 = 21.91$ (21.63, 22.18)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Exponential</td>
<td>$HU = Aexp(-N/B) + E_0$</td>
<td>$A = -5632$ (-2.246e+007, 2.246e+007)</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$B = 9.701e+005$ (-3.867e+009, 3.867e+009)</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$E_0 = 5654$ (-2.244e+007, 2.244e+007)</td>
<td>0.04511</td>
<td>0.9792</td>
</tr>
</tbody>
</table>

Figure 7.6(a-c) shows the mean HU in each ROI in the zero-scan images of the gel, with error bars representing +/- one standard deviation from the mean for the three data sets (i.e. 360, 50 and 16 images). Since the first CT image in the series is the closest in value to the irradiated gel dosimeter prior to imaging, it is used for comparison with the other images.
Figure 7.6: Plot of mean HU versus dose in cGy for all types of fit. ‘CT # 1’ refers to the fist gel image. ‘CT Averaged’ refers to the average of all gel CT images. On the x-axis the data points were shifted for visual clarity a) using 360 images b) using 50 images c) using 16 images. Error bars represent one standard deviation of the pixel values and therefore 68% confidence interval.

Figure 7.6(a) shows that when all CT images are averaged together stochastic noise is reduced (as indicated by the smaller error bars on the data from the averaged scans), but the resulting mean HU values in the ROIs are consistently higher than the mean CT numbers from the corresponding ROIs in the first CT image. This suggests that although the averaging of CT images may produce more precise images, the accuracy suffers. In fact, for almost all ROIs, the mean HU value in the first CT image falls outside the error bars of the averaged image, indicating the severity of the inaccuracy of the data from the averaged image.

Figure 7.6(a) shows that when using 360 CT images to create the zero-scan image, all of the fitting functions produce mean HU values in each ROI which are of a close match to the values
from the first CT image. In fact, the mean ROI values from CT image 1 falls within the error bars of all corresponding zero-scan images.

In figures 7.6 (b) and 7.6 (c), where 50 and 16 CT images respectively are used to create the zero-scan image, there is an increase in noise compared to figure 7.4 (a) for the zero-scan images using linear, exponential and quadratic fits. The noise in the averaged image is less as compared to the fitted data however the mean value remains inaccurate indicating the deposition of the dose due to averaging. The minimum dose limit of the PAGAT gel was found to be 2.3 Gy for 0 Gy dose at 95% confidence interval using the approach reported by Trapp et al.\textsuperscript{94} with a linear fit to the data.

7.2.3 Optimum number of scans for reconstruction

In figure 7.7 the average % error (in number of scans used to reconstruct the image) is plotted against the number of CT images used. It clearly demonstrates that, in this case, the error remains within 1% if 60 or more scans are used and within 0.5% if 100 or more CT scans are used. The number of images required will naturally vary according to the uncertainty required by the user together with the CT imaging parameters (for example, see Baxter et al.\textsuperscript{122}), and can be calculated by using statistical methods.
Figure 7.7: Plot of average % error (ROI’s compared to the linear reconstructed image) versus the number of images used in image reconstruction.

7.4 Zero-scan method extended to 3D image

In the above sections the zero-scan method has been established and it has been shown capable of reconstructing single slice gel images. In this section the zero-scan method has been employed to reconstruct the entire 3D gel image with a clinically relevant EDW treatment plan.

The gel was prepared according the method described in section 7.2.1 and poured in to a PET container with 10 cm height and 7 cm diameter. A three field wedged treatment plan was generated on the Varian treatment planning system (External Beam Planning 8.6.17). The plan parameters are shown in table 7.2.
Table 7.2: Plan parameters for zero-scan extension treatment plan.

<table>
<thead>
<tr>
<th>Field #</th>
<th>1</th>
<th>2</th>
<th>3</th>
</tr>
</thead>
<tbody>
<tr>
<td>MU</td>
<td>90</td>
<td>87</td>
<td>137</td>
</tr>
<tr>
<td>Gantry (deg)</td>
<td>320.1</td>
<td>39.0</td>
<td>219</td>
</tr>
<tr>
<td>Collimator (deg)</td>
<td>0</td>
<td>0</td>
<td>0.0</td>
</tr>
<tr>
<td>Field Size (cm(^2))</td>
<td>10.0 × 10.0</td>
<td>10.0 × 10.0</td>
<td>10.0 × 10.0</td>
</tr>
<tr>
<td>Wedge angle</td>
<td>30-OUT</td>
<td>15-IN</td>
<td>45-OUT</td>
</tr>
</tbody>
</table>

A total of 6 Gy was delivered in fraction (2 Gy/fraction using 6 MV photon beam) on Varian linear accelerator with 600 MU/min dose rate. The gel container is shown in figure 7.8 where the irradiated portion is clearly visible. The x-ray CT of the gel was carried out six hours after irradiation. The gel was imaged using a GE Lightspeed RT 4 CT scanner with an X-ray tube load of 230 mA with 1s rotation, beam energy of 120 kVp, 10 mm slice thickness, image size of 512 × 512 and 25 cm field of view. A 12 cm length of the gel was scanned, yielding 12 image slices per scan.
Figure 7.8: Irradiated gel container.

Two sets of images were acquired. First the water only images where the gel was not placed in the phantom, two scans were acquired (providing a total of 24 image slices). Then the gel container was placed inside the phantom and 63 scans were acquired (providing a total of 756 image slices). The averaged water images were subtracted from the respective gel images and further processed using the Matlab code as discussed in section 7.2.3. The zero-scan image was obtained by applying a linear fit to each pixel in the 3D data set as it has been shown in section 7.3.2 and table 1 that linear fit can be used for reconstruction. The result of reconstruction is shown in figure 7.9 where it can be seen that for each image slice the zero-scan image shows a clearer signal with less noise than the unreconstructed image.
Figure 7.9: Reconstructed images using the zero-scan method.
Figure 7.10: Coronal ((a) and (b)), transverse ((c) and (d)) and sagittal ((e) and (f)) views, through the Zero-scan 3D image of the gel. White lines in (c) indicate the positions of the coronal (horizontal line) and sagittal (vertical line) planes shown in (a) and (e), respectively. Images (b), (d) and (f) are duplicates of (a), (c) and (e). Regions A, B and A + B are described in the text.

Figures 7.10(a) to (f) show slices through the Zero-scan 3D image of the gel, in the coronal, transverse and sagittal planes. The low resolution of the coronal and sagittal images arises from the 1.0 cm CT slice thickness used to obtain the data. Despite the low resolution of these images, it is nonetheless possible to observe the following irradiated regions in the gel in figures 7.10(a) to (f): Region A was irradiated with one beam at a gantry angle of 320 degrees, with a 30 degree EDW. Region B was irradiated with an opposed pair of beams, at gantry angles of 40 and 220 degrees, one of which had a 15 degree EDW and the other had a 45 degree EDW. Region A+B was irradiated with all three beams. It is evident from the greyscale values shown
in figures 7.10(a) to (f), where regions of low dose appear darker than regions of high dose, that region A received a lower dose than region B, which received a lower dose than region A+B. It is also apparent, although less obvious, from an examination of figures 7.10(a) to (f), that in the coronal and sagittal images, the EDW motion direction is up the page, with the thin edge of the wedge at the bottom of the page and the thick end of the wedge at the top of the page. In the transverse image, the wedge motion direction is into the page.

This proof-of-concept study indicates the potential value of using dosimetric gels, read out using x-ray CT, to study the effects of EDWs in three dimensions. Higher-resolution images of dosimetric gels could be potentially be obtained by using a smaller slice thickness and the noise in these images could be further reduced by further performing additional CT scans and obtaining the Zero-scan images from a fit to a larger data set. Such detailed three dimensional information could be used to verify EDW treatment dose calculations made using conventional treatment planning systems or Monte Carlo simulations, and could be used in further studies of the effects of the interplay between patient and beam motion.

7.5 Discussion and summary

Reducing noise in CT imaging of gel dosimeters by averaging 360 images results in an overall increase in CT number of the final image due to the gradual increase in gel density caused by the radiation dose delivered during CT scanning. This is illustrated by the comparison between figure 7.4(a), where a visible difference is seen between images from the first and last scans, and by examination of the data in figure 7.6(a).

Comparison of the response of the non-irradiated and irradiated regions of the gel, exemplified by the different gradients of the linear fits to the individual pixel response data in Figures 7.5(a) and 7.5(b) indicates that this PAGAT gel responds differently to the incremental absorption of small radiation doses (from repeated CT scanning) depending on its degree of pre-irradiation.
This is additional confirmation of behaviour observed by several authors. Previous authors have shown that some polymer gel dosimeters undergo post-irradiation changes at a rate which depends upon the absorbed dose, resulting in an edge enhancement effect. In the present work the maximum delivered dose of 686 cGy is below that shown to produce measurable changes in the 24 hour irradiation-to-imaging time; however if this technique is used with larger doses then earlier imaging may be required to ensure accuracy.

Fitting a function to the pixel data and using this to evaluate a zero-scan image reduces image noise, while providing accurate measurements (see figures 7.5 and 7.6). Fitting the exponential fit does not result in any additional advantage in analysing CT images according to the method described here as the level of extra dose delivered via CT scanning is low and the gel’s subsequent response shows no obvious non-linearity. When analysing CT data by producing a zero-scan image, data shown in figure 7.6 suggest that the use of a simple, linear fitting relationship is suitably accurate in situations where a user does not have access to exponential fitting software.

The technique presented here can be accompanied by further techniques for noise improvement including image filtering, however the application of a particular filtering strategy is dependent on the nature of the dose distribution and the noise present in the original CT data. Reduction of noise by averaging the CT images will result in inaccuracies as evident from figure 7.6.

The data presented here represents results from specific scanning parameters on a specific scanner and gel dosimeter. If parameters are varied the technique presented here will continue to work if the chosen gel dosimeter responds to radiation dose with a change in density and therefore a change in CT number. For example, using a smaller slice thickness will increase the stochastic noise in each acquisition, thus increasing the noise in figure 7.5, however if a large
enough sample size is acquired the fitted function (and therefore intercept) will not significantly alter. Similarly, using a more sensitive gel dosimeter\textsuperscript{123, 124} may reduce the number of images required for suitable results to be obtained using this technique.

Extending the zero-scan method from a single slice reconstruction to a multi slice reconstruction has been shown to be successful and is illustrated in figure 7.9. The zero-scan slices demonstrate visible reduction in the noise. Three dimensional dose information obtained from this gel clearly shows the effects of combining different beams with different wedge angles, despite the low resolution of the CT scan, in the longitudinal direction.

This technique is likely to be valuable for clinical investigation of gel images, due to the widespread availability of CT scanners in radiotherapy departments. Although the images in figure 7.9 were acquired for an EDW plan with no patient motion, this technique is potentially valuable for studying the interplay where the phantom is moving relative to the radiation beam.

A simple method has been proposed for improving the image quality of polymer gel dosimeters imaged with x-ray CT. It has been shown that a simple analysis of the increase in HU with repeated imaging can be used to produce an accurate, low-noise ‘zero-scan’ image of the gel. The zero-scan image prediction method described here has been shown to be capable of improving the precision while maintaining the accuracy of a two-dimensional single-slice CT image of a gel sample irradiated to a range of doses. Additionally, repeated helical scanning of a volume of gel irradiated with clinical EDW fields has allowed this method to be applied in three dimensions, indicating the potential value of CT-scanned dosimetric gels for the 3D analysis of wedged and modulated radiotherapy fields. The next chapter will conclude the document by presenting a summary of the main results.
Chapter 8 Summary and discussion

8.1 Summary

In chapter 1 the research topic has been briefly introduced emphasising the need to investigate the interplay effects for Enhanced Dynamic Wedges (EDWs) using experimental and Monte Carlo (MC) methods. Moreover the use of x-ray Computed Tomography (CT) as an imaging modality for polymer gel dosimeters is highlighted. It is suggested that the improvements in the x-ray based gel readout techniques will be helpful in the experimental verification of the EDW treatments. A comprehensive background to the current research and the aim of the projects is outlined in a detailed literature survey that contextually situates the current research interest.

The first experimental investigation of the interplay effects has been carried out for single field EDWs. A number of parameters have been studied for their potential contribution to the interplay including the amplitude and frequency of tumor motion, wedge angle and the treatment field size. All of these parameters have been studied for two types of motions termed as parallel and perpendicular (i.e. either the jaw-tumor motion is parallel or perpendicular). It is shown that for a similar jaw-tumor motion speeds (i.e. the 6 s period scenario) the dose profile mismatch increases in case of parallel motion with the appearance of a step function and penumbral cut-off suggesting the dependence of interplay on the period in addition to the amplitude of tumor motion. While for perpendicular motion only a penumbral broadening occurs as a result of changing period and increasing the amplitude without the appearance of the step function. This indicates that interplay for perpendicular motion is not dependent on the frequency of tumor motion. It is also shown that for small field size (5 × 5 cm²) with large amplitude of motion (i.e. 4 cm) the wedged dose distribution is completely lost for both parallel and perpendicular cases. An amplitude period combination resulting in worst agreement
between the motion and static profiles was identified and called as worst case parameters and has been further used.

Although this section has dealt with single field cases nevertheless it provides valuable information about the dependence of the interplay on period and amplitude which can be further considered during the treatment planning process such as for instance if the direction of tumor collimator motion is the same there will be additional dose discrepancy as opposed to perpendicular motion, or for a large wedge angle the relative dose discrepancy is more than a small wedge angle. Also for small field size the loss of the wedged dose distribution for small and large angle is important, because if using non MC methods (which underestimates the dose by several percent) for planning the lung treatment with a large tumor motion, the margins might need to be extended further.

The single field experimental work of has been extended further to patient specific treatment plans using fractionated deliveries. Here pre-existing patient plans have been delivered using the worst case, motion parameters for single and multiple fractions. Also a four field EDW plan (using 4D CT planned data) have been delivered to the IMRT phantom with a dummy tumor insert for worst case motion parameters over 4 fractions. The gamma analysis of the film images from the four field EDW plan suggest that after the delivery of four fractions the dose has not averaged out and the overall gamma pass rate stands at 49%. These results indicate that there are possible scenarios where the dose averaging might not occur over the first few fractions for a multiple field EDW treatment resulting in local hot and cold spots. Although beyond the scope of this study in such cases evaluating the biological significance of these daily dose variations would be of interest. One more contributing factor to the dose discrepancy is the use of large wedge angles (for this four field study two fields were composed of 60° EDWs).
After the conclusion of the experimental study of the interplay the next extension of the study was to conduct the Monte Carlo counterpart. Monte Carlo commissioning of the linac model using the DYNJAWS component module (CM) has been presented. The measured and simulated EDW profiles have been compared using 3%-3mm gamma analysis. The DYNJAWS CM has been tested for two photon energies three field sizes, four wedges angles. It has been demonstrated that this CM can be used to accurately model the Varian EDWs. The dynamic and step and shoot modes of DYNJAWS have been compared and it is shown that dynamic mode more accurately models the EDW delivery especially for small wedge angles at small field sizes. Therefore for dynamic wedge simulations it is recommend that the CM should be used in the dynamic mode, despite the fact that dynamic mode takes slightly more computational time because of the additional interpolation carried out.

The generation of the input file specifying the jaw coordinates and probability of selecting a particular subfield for using DYNJAWS has been automated as well. This will help in the more efficient use of the CM as the user will not have to worry about the calculation of the probability values from the segmented treatment tables or for instance incorporating the back scatter from the upper jaws. This automation is also valid for any other variant of DYNJAWS which is used to simulate the EDWs. For instance currently a new CM called SYNCJAWS 125 is being incorporated in the BEAMnrc code and the developed automation can also generate the input files for SYNCJAWS.

The DYNJAWS commissioned MC model has been further used to study interplay effects. The EDWs have been simulated using the dynamic mode of the DYNJAWS. For the phantom dose employing the DOSXYZnrc code the 4D CT data of the IMRT phantom has been used. For each respiratory phase a phantom has been created. The static and motion dose maps have been compared using 3D gamma analysis at 3%-3mm criterion. For a selected ROI encompassing the dummy tumor the DVHs and GVHs for static and, motion cases have been presented as well. It
has been shown that for single field deliveries for perpendicular motion the discrepancy for 60 degree EDW is more than the corresponding 15 degree case. Similarly the comparison of four field 60 and 15 degree plans indicate a greater mismatch between the motion and static cases for 60 degree EDW. The comparison of DVH and GVH for a scenario combining large (60°, 45°, 30° and 25°) and small wedge angles (10°, 15°, 20° and 25°) in a four field delivery suggests no major difference between the two cases. Therefore it is possible to suggest the following clinical recommendations, for the planning of lung treatments that utilises EDWs: If the patient has a noticeably extensive breathing motion, or if they are prone to frequent coughing, or if their tumour is located in the inferior lobe, close to the diaphragm, then EDWs with small wedge angles should be used in preference to EDWs with large wedge angles, where possible.

Additionally, where multiple beams are used, the gantry and floor angles should be selected so that EDWs can be used with different orientations relative to the major axis of patient respiratory motion. If the patient breathes shallowly or if their tumour is located in the upper lobe of the lung or fixed to the mediastinum, then the above recommendations can be relaxed.

After studying the interplay using experimental and MC methods the next step was to improve gel dosimeter imaging with x-ray Computed Tomography (CT) which is a valuable tool for the experimental verification of the EDW deliveries and can be potentially further used in other studies for the experimental verification of the interplay effects. This technique provides a simple method for improving precision of x-ray computed tomography (CT) scans of irradiated polymer gel dosimetry. The noise affecting CT scans of irradiated gels has been an impediment to the use of clinical CT scanners for gel dosimetry studies. It has been shown that multiple scans of a single PAGAT gel dosimeter can be used to extrapolate a “zero-scan” image which displays a similar level of precision to an image obtained by averaging multiple CT images, without the compromised dose measurement resulting from the exposure of the gel to radiation from the CT scanner. When extrapolating the zero-scan image, it is shown that exponential and
simple linear fits to the relationship between Hounsfield unit and scan number, for each pixel in the image, provide an accurate indication of gel density.

It is expected that this work will be utilized in the analysis of three-dimensional gel volumes irradiated using complex radiotherapy treatments. The zero-scan methodology originally developed for a single slice reconstruction has been extended to scan the entire gel image produced from a clinical enhanced dynamic wedge plan.

It has been shown that interplay affects the dose delivery for EDW deliveries. The wedge angle and tumor amplitude plays an important role in the Interplay. The use of large wedge angles in a multiple field delivery results in higher dose discrepancy. It has been shown that DYNJAWS CM accurately models the EDW simulation and its automation will help in further using this CM. Finally zero-scan method for x-ray CT based gel readout produces reconstructed low noise images without adding further dose to the gel. It has been shown to work for scanning the entire gel volume for an EDW treatment.

8.2 Overall discussion and conclusions

The aim of this study was to investigate the interplay effects arising from the delivery of enhanced dynamic wedge (EDW) treatments to the moving tumor. The present investigation consisted of three major parts where experiments and MC simulations have been conducted for the assessment of the interplay between the dynamic wedge and the moving tumor.

In the first part a series of two dimensional experimental analysis of the interaction of dynamic wedge delivery and target was carried out as explained in chapter 3 and 4. The single field studies as described in chapter 3 provided a sizeable data set for identifying the potential contribution of the various parameters affecting the interplay such as field size, amplitude and period of tumor motion, direction of tumor-collimator motion etc. These measurements
confirmed and reproduced some of the earlier treatment planning and experimental results.\textsuperscript{14, 15} Besides reconfirming the earlier findings further insight towards understanding the interplay effects was achieved. As the direction of tumor-collimator motion was not restricted to the parallel case only, this provided an indication of the dependence of the interplay on the amplitude and period of tumor motion in the case of perpendicular delivery. As it has been suggested by Jiang \textit{et al.} that the mechanism of interplay is different for parallel and perpendicular motion.\textsuperscript{8} The observation that for perpendicular motion the dose discrepancy is only dependent on the tumor motion amplitude and not the period of motion was important because it suggests that if the tumor collimator direction is parallel the dose discrepancy will be more severe than its perpendicular counterpart and this information is useful for the treatment planner while deciding the orientation of collimator motion in various treatment beams.

One important factor affecting the interplay is the beam dose rate which was kept constant for the measurements in the first two stages. Pemler \textit{et al.}\textsuperscript{14} have suggested that using low dose rate the organ motion becomes faster compared to the collimator case and the dose deviation approaches the static case while using a 60º EDW. However the results by Sidhu \textit{et al.} suggested otherwise and no dose smoothing was observed in the case of a faster organ motion. The possible explanation for that was the assumption by Pemler \textit{et al.}\textsuperscript{14} that organ does not pass the boundaries of the radiation beam. Another contributing factor was that for the chosen wedge angle of 60º the dose gradient is higher and the collimator speed is lowered compared to the other wedge angles even for a similar dose rate. Therefore in this study the dose rate was kept constant as the tumor movement was not restricted within the radiation field size.

The tumor motion was considered to be sinusoidal for all the cases in the first two stages. Many previous studies have assumed the respiratory motion as sinusoidal.\textsuperscript{126, 127} The amplitude of lung tumor motion was between10 mm to 40 mm. There is a need to put this range of motion in to perspective with the clinical observed values. A comprehensive collection of lung tumor motion
data has been provided in the AAPM task report 76.\textsuperscript{55} In Table I of the report the lung tumor motion in superior inferior (SI) direction of up to 50 mm has been reported. The amplitude of tumor motion is also depended on the breathing mode. For instance in the same report in Table II for deep breathing in the case of diaphragm up to 100 mm of movement in the SI direction has been documented.

The experimental measurements in chapter 3 were carried a single time except for the measurements in figure 3.6. Therefore there is a possibility of phase dependence between the collimator and tumor motion. However as it can be seen from figure 3.6 that phase dependence is minimum and this has been confirmed earlier.\textsuperscript{14} Moreover in actual treatment delivery usually the phase matching is not performed.

The 2D gamma analysis in the first two stages was carried out with DTA of 3%-3mm. In the case of diode arrays where the detector spatial resolution was poor (5-7 mm distance between dose points), the gamma evaluation software carried out interpolation to ensure that the step size is less than the acceptance criteria which results in smoothed dose map which are reasonable for further analysis.\textsuperscript{103}

Although single field EDW treatments are rarely used in a clinical treatment perspective, however still these simple set of measurements provide a picture of the potential contribution of the period in interplay effects when the motion is parallel where amplitude and period both are contributing to the interplay. An amplitude-period pair termed as worst case was identified where the static and motion EDW profiles had the maximum deviation. Secondly the effect of reduction of field size and its negative consequences on the dosimetric outcomes were experimentally shown as often these effects are ignored in clinical practice while performing 3D conformal radiotherapy treatments.
The single field studies presented in chapter 3 are indicative of the potential contribution of the various parameters towards the interplay. However in routine clinical application of the EDWs multiple field treatments are used over several fractions as reported in chapter 4. The important conclusion drawn from these sets of measurements was that there can be potential scenarios where in the initial few fractions for the particular EDW treatment plan that was studied. The clinical implications of these results can be further augmented by performing extra experiments or simulations where the biological significance of these variations can be further examined.

In the second stage of the study the complexity of the interplay was the next aim after the acquisition of base line single field and multiple fields experimental data. The intent was to consider the combination of different wedge angles with different collimator rotations so as to assess the combined effect of wedge angles where EDW tumor motion is parallel or perpendicular. One obvious choice was to carry out experimental studies using films in the IMRT phantom setup as described in chapter 4. However this does require a considerable machine time for treatment delivery and film calibration. Film scanning and data processing time are the additional overheads. Finally the results will yield only 2D information as opposed to a 3D dose map which is a more realistic and desired outcome. One second alternative was to carry out EDW deliveries using MC simulations. MC user codes such as BEAMnrc have been used as a gold standard for dose calculation. EDW delivery can be simulated on the 4D CT data set of the IMRT phantom setup discussed in chapter 4. This will generate 3D dose maps for the assessment of the complexity on the combination of wedged fields and different collimator angles. The decision to choose MC simulations over measurements was further aided by a number of factors. First the availability of a high powered local computational cluster resulted in the reduction of the treatment simulation times drastically, and the availability of several in house data processing tools for MC simulations which were capable of generating phantoms for MC simulations form the 4D CT data set further processing of the results. Finally the
availability of dedicated component module (CM) named as DYNJAWS for the modelling the
dynamic wedges in BEAMnrc made the EDW simulations even more convenient where in a
single BEAMnrc simulation the entire treatment head was simulated.\textsuperscript{16} This CM was not
previously available in the BEAMnrc user code.

For carrying out the MC simulations with aim to study the interplay complexity the first step
was to commission the model. This is the subject of Chapter 5 where the new CM has been
validated and automated.\textsuperscript{17} This validation of the CM involved the comparison of a series of
systematic measurements with the simulations. The automation of the CM was important
because it provided an efficient way of the generation of the input file which specified the
probability and jaw coordinates. The validation of the component also demonstrated the
suitability of the dynamic mode to model the EDWs as compared to the step and shoot mode
especially in the case of large field sizes where small wedge angles were used.

The DYNJAWS CM was further used to simulate a series of multiple field EDW treatments to
IMRT MC phantoms as reported in chapter 6. The results from these investigations are of
particular interest with respect to the combination of different wedge angles and their potential
impact on the interplay. It has been demonstrated that the combination of different wedge angles
along with the direction of collimator motion can affect the interplay. In particular if the
collimator angles are aligned along the direction of motion of the tumor the interplay effects are
further exacerbated. Despite the simplified phantom design where organ deformation has not
been taken into account these results do suggest that for the reduction of the interplay a
combination of lower angle wedged fields where collimator tumor motion is not parallel will be
advisable. Therefore a scenario where excessive tumor motion is present these
recommendations can be incorporated at the treatment planning stage or alternatively a gated
treatment delivery can be performed. However even with the availability of the gated treatments
at times it is not possible to treat every patient with gated deliveries.\textsuperscript{55} In that case these
recommendations will be helpful in minimizing the interplay. The results in this stage also
confirm the suitability of the developed MC framework using the DYNJAWS CM, which be
used to further study the interplay effects where different collimator rotations, beam weights can
be used.

The third stage of the study a new imaging technique was developed which can be further used
for verification of the interplay treatments or any other treatment with the help of the gel
dosimeters. Here the evaluation of gel dosimeters using x-ray CT data was proposed because of
the local expertise and because motion is more likely to impact on regions with strong dose
gradients where high spatial resolution and fidelity of x-ray CT would be advantageous. The
technique was called a zero scan method and explained in chapter 7. This technique can be
useful in clinic for reading out gel dosimeters without the need for specialized equipment like
Optical CT scanner or MRI machines. Such imaging techniques are can be employed for the
study of gel dosimeters irradiated

8.3 Future Recommendations

The use of a more realistic tumor/phantom will be desirable as described by Court et al.\textsuperscript{105} Also
for simulating the tumor motion the use of patient respiratory traces will be a better choice
because the tumor motion is not always perfectly sinusoidal\textsuperscript{107} and can vary from patient to
patient. In this study organ deformation has not been incorporated into dose calculations. A new
DOSXYZnrc based user code called defDOSXYZnrc can be used for studying the organ
deformation.\textsuperscript{108} Finally the setup errors can be incorporated which have not been studied here.
Moreover for EDW interplay a complexity score can be developed predicting the interplay
based on the planning parameters such as the field size, wedge angle, amplitude and direction of
motion etc.
References

64. Z. Han, S. K. Ng, M. S. Bhagwat, Y. Lyatskaya and P. Zygmanski, Evaluation of MatriXX for IMRT and VMAT dose verifications in peripheral dose regions Med. Phys. 37 (7), 3704-3714 (2010).
70. S. Kohei, J. B. Moorrees, C. M. Langtom and J. V. Trapp, in IC3DDose, 6th International Conference on 3D Radiation Dosimetry, edited by M. Oldham (Duke University Medical Center, Hilton Head Island, SC, USA, 2010).


123. A. Jirasek, Alternative imaging modality for polymer gel dosimetry, in IC3DDose, 6th International Conference on 3D Radiation Dosimetry, edited by M. Oldham (Duke University Medical Center, Hilton Head Island, SC, USA, 2010).


127