DEVELOPMENT OF EPID-BASED
REAL-TIME DOSIMETRY AND
GEOMETRY IN RADIATION THERAPY

Todsaporn FUANGROD

BEng

A thesis submitted for the degree of
Master of Philosophy (Medical physics) from the
Faculty of Science and Information Technology,
University of Newcastle

AUGUST 2012
DECLARATION

This thesis contains no material which has been accepted for the award of any other degree or diploma in any university or other tertiary institution and, to the best of my knowledge and belief, contains no material previously published or written by another person, except where due reference has been made in the text. I give consent to this copy of my thesis, when deposited in the University Library, being made available for loan and photocopying subject to the provisions of the Copyright Act 1968.

Todsaporn FUANGROD
ACKNOWLEDGEMENT OF AUTHORSHIP

I hereby certify that the work embodied in this thesis has been done in collaboration with other researchers. I have included as part of the thesis a statement clearly outlining the extent of collaboration, with whom and under what auspices.

Todsaporn FUANGROD
ACKNOWLEDGEMENTS

First, I wish to express my sincerest gratitude to my supervisor, Conjoint Associate Professor Peter Greer, who has supported me throughout my studies. When I made the jump from an engineering background into the medical physics world, no one trusted me except for Peter: thank you for choosing my application. Peter attempts to understand me and gives me the room to do research work in my own style. Peter, you are the best supervisor I have ever had (and I have had a few). I would also like to say a big thank you to my co-supervisor Professor John O’Connor, for his great supervision and valuable advice. Your suggestions always bring my focus back on track, even though we do not have many opportunities to meet in person. I found that Prof. John never says “no” when help is needed, and that’s impressive.

Second, I would like to thank the University of Newcastle for providing me with my scholarships. This changed my life and fulfilled my dream. I had been fighting for a long while to get a scholarship...

I personally thank Dr. Pejman Rowshanfarzad and Dr. Mahsheed Sabet, you taught me that I should know in medical physics. My questions always met your answers. Thanks Pejman for believing in me, you are an amazing person. Thanks Dr. Brian King for helping me in Matlab and mathematical problems. Thanks Jidi Sun for your kind considerations and help with the mobile phone contract.

Next, I would like to say thank you to Dr. Henry Woodruff, who is the “new” postdoctoral researcher in our group. Thanks for making the prediction model work, thanks for your time, enjoying having coffee with me, and thanks for your friendship.

I have had the pleasure of working with the Winnipeg group from CancerCare Manitoba, Canada: Professor Boyd McCurdy and Dr. Eric van Uytten. I would like to thank them all for your advice and sharing knowledge.

On my personal note, I would like to thank my family, my parents for their support throughout 28 years. Finally Benjawan, my wife, thank you for your love and for spending long hours with me in the office. Thank you for making my life in Australia meaningful, I am very lucky to have you at this moment.
PUBLICATION AND PRESENTATION ASSOCIATED WITH THIS RESEARCH

Publications


Conference presentations

- **T. Fuangrod, D.J. O’Connor, and P.B. Greer**, “Synchronisation of EPID and Predicted Images Used in Real-time Treatment Verification System for Radiation Therapy”, The 12th International Conference on Electronic Patient Imaging, Sydney, Australia, 2012 (Received Travel Award)
- **T. Fuangrod, D.J. O’Connor, and P.B. Greer**, “Development of EPID-based real-time treatment verification in radiation therapy: system design and simulation”, The 6th Student Research Symposium of the ACT/NSW Branch of the Australasian College of Physical Scientists and Engineering in Medicine (MedPhys 11), Australia, 2011 (Received Best Presentation Award)

Other presentation

# TABLE OF CONTENTS

ABSTRACT..........................................................................................................................1

CHAPTER 1: INTRODUCTION.................................................................................................2

1.1 MOTIVATION..................................................................................................................2

1.2 BACKGROUND.............................................................................................................6

1.2.1 Process of a radiation therapy.................................................................................6

1.2.1.1 Error in the delivery of radiation therapy .........................................................7

1.2.1.2 Linear accelerator, IMRT and arc-IMRT .........................................................9

1.2.1.3 Imaging devices: a-Si Electronic Portal Imaging Device (EPID)..................11

1.2.1.4 Failure Mode and Effect Analysis (FMEA)....................................................13

1.3 THESIS OUTLINE .......................................................................................................17

CHAPTER 2: LITERATURE REVIEW......................................................................................18

2.1 A FEASIBILITY STUDY FOR REAL-TIME DOSIMETRY USING EPID..............18

2.2 EPID DOSIMETRY.......................................................................................................21

2.3 EPID GEOMETRY .......................................................................................................26

2.4 DOSE DISTRIBUTION COMPARISON TOOLS.......................................................29

CHAPTER 3: MATERIALS AND METHODS.......................................................................34

3.1 LINAC AND EPID.......................................................................................................34

3.2 SYSTEM DEVELOPMENT TOOLS...........................................................................35

3.3 PREDICTION MODEL.................................................................................................35

3.4 IMAGE SYNCHRONISATION.......................................................................................36

3.5 AN AUTOMATIC MLC LEAF POSITIONING ALGORITHM...................................38

3.5.1 Leaf-end edge detection .......................................................................................39

3.5.2 MLC leaf template construction.................................................................40

3.5.3 MLC position matching .....................................................................................41

3.6 REAL-TIME DOSIMETRY USING EPID.................................................................43

3.7 REAL-TIME GEOMETRY USING EPID.................................................................45

3.7.1 Automatic jaw positioning..................................................................................45

3.7.1.1 Investigation of area shielded by jaws using fuzzy c-means

Clustering.........................................................................................................................47
CHAPTER 4: RESULTS……………………………………………………………………..55

4.1 PHYSICS-BASED PREDICTION MODEL ..............................................55
4.2 IMAGE SYNCHRONISATION.................................................................57
4.3 DOSE COMPARISON ...........................................................................60
4.4 REAL-TIME ERROR DETECTION USING REAL-TIME DOSIMETRY........61
  4.3.1 Case study I : Dose error ...............................................................61
  4.3.2 Case study II : MLC leaf error .......................................................62
4.5 AUTOMATIC MLC POSITIONING .........................................................63
4.6 AUTOMATIC JAW POSITIONING .........................................................64
4.7 ACCURACY OF AUTOMATIC COLLIMATOR ANGLE DETECTION.......66
4.8 CALCULATION TIME ...........................................................................67

CHAPTER 5: DISCUSSION ............................................................................68

CHAPTER 6: CONCLUSION AND FUTURE WORK .......................................71

BIBLIOGRAPHY ................................................................................................73
ABSTRACT

A real-time EPID-based dosimetric and geometric verification system is proposed. The system enables detection of gross treatment delivery errors prior to delivery of substantial radiation to the patient in IMRT treatment. A sophisticated physics-based model is utilized to generate an image stream of predicted cine EPID images as a reference dataset, and it then compares these to measured EPID images acquired during treatment. A new method that combines both geometrical and dosimetric comparison is used for synchronisation. In addition, jaw position and collimator angle are detected and verified in the beginning of treatment (approximately first 5 seconds). The system was simulated using MATLAB/SIMULINK. The synchronisation is shown to agree within 2 control points (~1% of the total dose delivered). In real-time dosimetry, two case studies were simulated: 1) Gross dose delivery error and 2) MLC leaf failure; the real-time dosimetry system was able to detect both of these errors in real-time (within 0.1 sec). In real-time geometry, jaw positioning and collimator angle were tested, and an average 0.5 mm error for jaw positioning and less than 0.5 degrees for collimator angle are shown in the tests. MLC leaf verification uses the advantageous result of synchronisation to monitor the MLC leaf motion behaviour during the treatment. A real-time verification system can prevent many of the major mistreatments that have recently occurred in radiation therapy. The system design is fast and can be easily applied in a clinical environment.
Chapter 1
Introduction

1.1 Motivation

Nowadays, most cancers are treated by radiotherapy, chemotherapy (drug treatment), or surgery. The treatment options are determined by the type and stage of the cancer. Radiotherapy is one of the most effective treatments, so approximately half of all cancer patients will receive radiotherapy (Delaney, Jacob et al. 2005), either as the only treatment or as an important part of their total treatment. Moreover, radiotherapy is used to treat solid tumours, such as cancers of the skin, brain, breast, prostate, etc. There are three principal modalities for the administration of radiotherapy: external beam radiotherapy, internal brachytherapy, and unsealed source therapy. In this research, the focus has been external beam radiotherapy.

High-energy beams of ionizing radiation of photons, electrons, and ions are used to destroy the DNA of the tumour cells inside the human body. Cancerous cells are unable to repair this damage as quickly, so their growth is curtailed, and the tumour volume shrinks accordingly. The primary aim of radiotherapy is to deliver the maximum possible radiation dose to the target tumour and the minimum dose to the healthy surrounding tissues (Washington and Leaver 2010). Although some healthy cells are affected by radiation, most of them appear to recover more fully from the effects of radiation than do tumour cells. However, there is a risk to patient safety if there is an overdose to healthy tissue or underdose to the target tumour. Even though the normal tissue can repair itself, overdose of irradiation are able to damage the tissue beyond self-recovery, potentially causing organ shutdown and death of the patient. An underdose of irradiation can lead to insufficient treatment and the inability to control the cancer growth (Washington and Leaver 2010).

In the last few decades, the complexity of dose delivery methods used in radiotherapy has been rapidly increasing with the advent of delivery techniques such as dynamic intensity modulated radiation therapy (IMRT) and intensity modulated arc therapy (VMAT). The high dose gradients typically associated with these treatments require rigorous verification of the radiation delivery, which is generally performed using a pre-treatment verification test (Chen, d’Errico et al. 2006). Although, considerable improvements in modern verification methods have been achieved, and there is a low rate of reported treatment errors and injuries using either...
IMRT or IMAT techniques (Advisory 2009), an interest in patient safety has been a recent focus of the radiation therapy community. In 2010, a news headline of the New York Times drew public attention to the potential dangers of radiation treatment delivery errors (Bogdanich 2010). The author pointed to the advancements of radiation technology as the catalyst for the radiation errors: “While this new technology allows doctors to more accurately attack tumours and reduce certain mistakes, its complexity has created new avenues for error through software flaws, faulty programming, poor safety procedures or inadequate staffing and training” (Bogdanich 2010). Unexpected errors still occur in clinical settings, which are very difficult to detect even with the current high level of technology used for clinical radiation treatment. An example is multi-leaf collimator (MLC) leaves being retracted or having misaligned positions during irradiation.

To ultimately catch and prevent these treatment errors as a quality improvement measure, it is necessary to monitor and verify the radiation dose delivered to the patient in real-time. As shown in Figure 1.1, an independent system that monitors and verifies dosimetry during treatment delivery would be an added level of quality assurance (QA) for complex delivery techniques, helping to prevent gross treatment errors. The error could be identified and the treatment could be interrupted before a significant dose has been delivered to the patient. One approach to real-time monitoring is the use of point dosimeters, e.g. polymethylmethacrylate (PMMA) optical fibres, which are set up either on the skin or inside the patient body, depending on the location of the tissue to be treated (S.O'Keeffe 2009). However, the main drawbacks of this approach are the lack of two-dimensional data and the difficulty of the setup, which is why these methods are not frequently used in clinical situations.
Figure 1.1 Radiation process of care diagram\textsuperscript{29} and relationship to the real-time treatment verification system using EPID—including prediction, verification, and treatment report—operating independently from the regular radiotherapy treatment process to detect and prevent treatment errors.

In the past decade, the idea of using electronic portal imaging devices (EPIDs) as dosimeters has been extensively studied due to their high resolution, large sensitive area, digital format, widespread availability, robustness, and easy setup. In terms of dosimetric QA, EPID dosimetry is currently used clinically for patient-specific IMRT verification (McDermott, Wendling et al. 2007; van Elmpt, McDermott et al. 2008; Mans, Remeijer et al. 2010), and IMAT dose verification using EPIDs has also been studied and recommended as a QA tool (Bakhtiar, Kumaraswamy et al. 2011). To perform time-resolved EPID dosimetry for IMRT and IMAT requires cine-mode imaging. McCurdy et al. demonstrated the feasibility of using cine-mode by comparing real-time EPID response and real-time ion-chamber data for selected points in the deliveries and finding high agreement (McCurdy and Greer 2009). Based on the potential of those investigations, the first system for real-time dose verification in IMRT using cine-EPID images was developed in this thesis. The system is intended to operate in a clinical environment as a testing mechanism to ensure that overdoses do not occur and underdoses are detected as soon as is practically possible (Figure 1.2). It should be noted that
the system implemented in the thesis does not cover the systematic physical detector offset issue, such as the detector panel is not positioned accurately. However, this work has the potential to make a significant impact on the current dose verification practice in radiation therapy.

Figure 1.2 System concept for EPID-based real-time patient dose delivery verification: (a) system monitor displays the result of dose verification in real-time during treatment (Adapted from http://www.ormc.org/hospital_services/snow_igrt.aspx), and (b) when the system detects an error, the preventative action must be performed, such as switching the beam off.
1.2 Background

1.2.1 Process of a radiation therapy

A radiotherapy treatment is a complex procedure that can be divided into four main steps, including treatment prescription, treatment preparation, treatment delivery, and treatment follow-up (see Figure 1.1). The treatment is prescribed in terms of a certain dose to a particular volume in a number of fractions with a specified technique.

Figure 1.3 Radiotherapy treatment procedure.

Firstly, the treatment prescription (beam shape, directions, energy etc) is selected based on the volumetric patient imaging. This is typically performed by using a CT-scan, which uses a rotating X-ray beam and corresponding rotating detector to provide a two dimensional “slice” of the patient. Several slices of CT-image are reconstructed into a 3D model of the tumour and its surroundings as a “virtual tumour volume.” The target volume is delineated based on clinical examination, diagnostic images from the CT-scan, and knowledge of the disease. Other 3D displays of anatomical information, like Magnetic Resonance Imaging (MRI), may give better tissue differentiation for more accurate target volume delineation. Positron Emission Tomography is now an important tool for diagnosis or (re)staging of cancer in some cases.

In the treatment preparation step, the beam angles that conform optimally to the target tumour volume are selected. Multi-leaf collimator (MLC) shaping and dose calculation are performed using the treatment planning system (TPS). Dose is expressed in Joules per kg (J/kg) but is more commonly referred to as Gray (Gy). Next, the treatment delivery is performed and is repeated in each fraction over a number of weeks based on the TPS plan. Finally, the results of the treatment are evaluated in the treatment follow-up step. The follow-up step is very important for early assessment of treatment effectiveness and local regional recurrence. The acutely responding tissues manifest their damage immediately during the overall therapy, which is routinely accessed after the end of the treatment fraction. In addition, the follow-up investigation is to correlate the treatment outcome with a certain treatment delivery technique, and the results are essential to evaluate the quality of treatment.
1.2.1.1 Error in the delivery of radiation therapy

The rapid development of new technology has changed the method of radiation therapy. This has included three-dimensional computed tomography (CT)-based planning, multi-leaf collimator beam shaping, improved immobilization, more complex planning software, and highly conformal treatments. The treatment equipment and software systems have become more complex. This rapid adoption of new technologies might introduce new sources of error, and the potential for “tragic errors” to occur has increased even though these technologies aim to reduce the uncertainty and error of treatment. Implementation of a quality assurance (QA) programme is necessary to guarantee the entire plan and treatment delivered to the patient is accurate and satisfies the given requirements for quality.

G. Huang et al. (2004) investigated the error in delivery over the period 1997-2002 and found that errors were rare and generally thought to be of no or minor clinical significance. The use of beam-modifying devices increased the risk of error and could introduce new and unanticipated ways for errors to occur (Huang, Medlam et al. 2005). In addition, patients with multiphase plans (e.g. head-and-neck) or tumour types with diverse anatomic locations (e.g., sarcoma and lymphoma) were found to be associated with a greater risk of error, suggesting that the more complicated treatment increased the opportunity for error to occur.

As published by the New York Times in their examination series on patient safety in radiation therapy (Bogdanich 2010), a tragic error occurred with Scott Jerome-Parks who died...
due to an overdose of radiation in 2007. Two year earlier, Mr. Jerome-Parks was diagnosed with a form of tongue cancer and was offered treatment with IMRT, which is an advanced form of radiation treatment. After his fourth treatment session, Mr. Jerome-Parks’ radiation oncologist decided to revise the current treatment plan to minimize the radiation damage to Mr. Jerome-Parks’ teeth. The medical physicist, tasked to create the new treatment plan, encountered a software error while entering the plan into the software system that controls the radiation therapy machine. The error message asked the medical physicist if they would like to save the changes prior to the application aborting. The response to the message was “yes,” the application restarted, and the physicist resumed finalizing the plan. Once complete, the radiation oncologist approved the plan.

Undetected by the medical physicist, radiation oncologist, and the radiation therapists who were responsible for delivering the treatment, Mr. Jerome-Parks’ new treatment plan contained a massive error. The MLC leaves were programmed to be wide open, which would not occur during a typical prescribed treatment. Unfortunately, this was not noticed until Mr. Jerome-Parks had received seven times the prescribed delivery. As described in the New York Times article, the radiation overdose would eventually leave Mr. Jerome-Parks “deaf, struggling to see, unable to swallow, burned, with his teeth falling out, with ulcers in his mouth and throat, nauseated, in severe pain and finally unable to breathe” (Bogdanich 2010).

Figure 1.3 Effect of overdose irradiation to Mr. Jerome-Parks (Adapted from http://www.nytimes.com/imagepages/2010/01/24/us/24radiation_CA2.html, Bogdanich 2010).
1.2.1.2 Linear accelerator, IMRT, and arc-IMRT

The linear accelerator (LINAC) is a primary tool in external beam radiotherapy for patients with cancer. The LINAC uses high-frequency electromagnetic (EM) waves to accelerate electrons to high-speed through the LINAC waveguide. The LINAC is used to treat all body sites using conventional techniques, three-dimensional conformal radiotherapy (3D-CRT), Intensity-Modulated Radiation Therapy (IMRT), Image Guided Radiation Therapy (IGRT), Stereotactic Radiosurgery (SRS), and Stereotactic Body Radiation Therapy (SBRT).

There are two options for treatment: the electron beam can be used for superficial treatment or to generate a photon beam by sending electrons onto a target (usually tungsten material). In both cases, the energy of the particles can be varied over a wide range, clinically used between 4 and 23 MV (Johns, H. E. et al. 1983). With the recently technology, the LINAC is equipped with an MLC, which consists of small tungsten leaves automatically placed in the beam to allow the delivery of more complex fields that conform to the tumour shape; no shielding blocks have to be manufactured independently. The patient setup time can be decreased during treatment, and more treatment fields can be performed during one session. MLCs typically have 80 to 120 leaves arranged in opposing pairs, depending on the manufacturer’s design. (see example, Figure 1.4a)

(a) [Image of multi-leaf collimator]  (b) [Image of multi-leaf collimator]

Figure 1.4 (a) Varian's 120-leaf multi-leaf collimator, a device for shaping a radiation beam so that it conforms to the 3-dimensional shape of a targeted tumour. (b) The radiation beam passes through and is shaped by a device called a multi-leaf collimator so that it conforms to the shape of the tumour. (Varian Medical Systems)

The use of MLCs allows precise delivery of higher doses to the target volume with a steep fall-off of the dose to adjacent regions by the use of IMRT (Intensity Modulated Radiation Therapy), which is an advanced form of three-dimensional conformal radiotherapy (3D-CRT). However, regular 3D-CRT and IMRT differ in how the pattern and volume of radiation
delivered to the tumour is considered. In conventional 3D-CRT, the treatment is forward planned using shaped uniform intensity beams. In IMRT, the physician designates specific doses of radiation that the tumour and normal surrounding tissues should receive. The treatment planning system then derives the incident fluence distribution for each beam to achieve these doses, using an optimisation procedure. These non-uniform fluences are delivered by moving the MLC leaves during the treatment. Each pair of MLC leaves can produce a fluence profile depending on the gap between them and the speed of movement. Treatment with IMRT is slightly longer than with 3D-CRT, but generally produces fewer side effects due to the more conformal dose distributions.

The position of the leaves of the MLC can be varied in time with a fixed or moving gantry, which is able to rotate 360 degrees around the isocentre. The isocentre is set at a distance of 1 metre from the radiation source or target. The radiation beams can be delivered at any angle towards the patient in the plane of rotation. The patient lies on the couch, which can be adjusted in both the horizontal and vertical directions (see Figure 1.4b and 1.5a). MLCs can be applied with two strategies, step-and-shoot and dynamic MLC method. First, a single treatment field can be segmented into several static shapes or sub-fields, where the radiation beam is switched on when the leaves are stopped at each prescribed position. Secondly, dynamic MLC is automatic computer controlled movement of the leaves into beam segments during irradiation. Figure 1.4b demonstrates an IMRT treatment of the prostate by means of 5 intensity-modulated fields.

![Figure 1.5](a) LINAC rotates 360 degrees around the patient to deliver radiation beams from many different angles (Varian Medical Systems). (b)Intensity Modulated Radiotherapy Treatment (IMRT) of the prostate with 5 isocentric intensity modulated fields.
A recent development of the IMRT technique is called arc-IMRT (RapidArc from Varian and VMAT (volumetric arc therapy) from Elekta). In simple terms, arc-IMRT is a faster way to deliver conventional IMRT, which can take a long time because of the complexity of treatment. With arc-IMRT, the LINAC rotates around the patient to deliver radiation treatments from all angles. During the treatment, the radiation beam aperture is shaped and reshaped as it is delivered continuously from virtually every angle in a 360-degree revolution around the patient. (see Figure 1.6)

1.2.1.3 Imaging devices: a-Si Electronic Portal Imaging Device (EPID)

Several types of EPID have been developed over time, but only the video-based (VEPIDs), scanning liquid-filled ionization (SLIC), and amorphous silicon (a-Si) type EPIDs have evolved into commercially available systems. The EPIDs were originally designed for geometric verification of patient position setup during treatment, however, EPIDs are possible to use for dosimetric verification, such as in-vivo dosimetry and for pre and during treatment verification. A dosimetric calibration is necessary to convert a greyscale EPID image into a portal dose image (PDI). The use of the a-Si EPID for dosimetry has also been investigated by many research groups (Chapter 2). The a-Si EPID that is used in this study is the aSi1000 EPID (Varian Medical Systems, Palo Alto, CA). The Varian aSi1000 EPID is mounted on a robotic arm, which allows it to be positioned at a source to EPID distance (SDD) from 95 cm to 180 cm. It has an active imaging area of 40X30 cm² (at an SSD of 105 cm). The image matrix is created from an array of 1024X768 pixels. The maximum frame acquisition rate is 9.574 frames per second, the permitted dose range is 4-25 MV, and permitted dose rates are 50-600 MU/min.

One characteristic of a-Si material is that it exhibits high resistance to radiation damage. In an a-Si EPID, the a-Si array detectors are integrated with circuits, called an active-matrix array (see Figure 1.6). Active-matrix technology allows the deposition of semiconductors across large-area substrates in a well-controlled fashion such that the physical and electrical properties of the resulting structures can be modified and adapted for many different applications. The a-Si is deposited onto a thin substrate (typically 1 mm thick glass) using semiconductor fabrication techniques, such as plasma enhanced chemical vapour deposition, to form a two dimensional matrix of thin film transistors (TFTs) and photodiodes (see Figure 1.7). These integrated circuits form a thin, large-area light sensor because the arrays have direct contact with the metal plate/phosphor screens from the X-ray detector in the EPID. Each pixel in the a-Si array consists of a light sensitive photodiode connected to a TFT. The incident X-rays are converted by the scintillator screens to visible light, which generates electron-hole pairs in the photodiode. The photodiode acts like a capacitor because the received light is integrated and captured as an
electric charge. The charge carriers are trapped in the photodiode. The TFTs control the readout of the recorded signal; they are switched to conduct when the gate lines are enabled, and the charge held in the photodiode is then read out over the data line.

![Figure 1.6 a-Si electronic portal imager.](image)

![Figure 1.7 a-Si pixel generation and microscopic view of a-Si pixel.](image)

During irradiation, light that is generated in the X-ray detector results in charge buildup in the photodiode, which has had a bias voltage applied before the irradiation. The TFT is non-conducting during the irradiation. During readout, the TFT is controlled by applying a control voltage to its associated gate lines, and this allows charges to flow between the photodiodes of all columns in parallel to external amplifiers. The current recharges the photodiode to its original bias voltage, and a charge amplifier records the charge, which is proportional to the light reaching the photodiode during the irradiation. To acquire and image, the a-Si pixels are arranged in a matrix. The gate driver electronics enable the first row, i.e. all the TFTs of the entire pixel row are switched to conduct. The charges held in all photodiodes (capacitors) of this
particular row are conducted to the read-out electronics, which have a single charge amplifier per column. As soon as one row is read out, the system switches to the next row and so on until the whole image is generated. All signals of the columns are amplified in charge amplifiers and converted to digital format by analogue to digital converters (ADC’s).

Before each set of EPID images is acquired, it is advisable to first calibrate the detector. This can be accomplished by obtaining a dark field and delivering a flood field. The premise is that taking these images will allow for the elimination of background noise and provide a uniform response for imaging. Specifically, the dark field image provides information about background noise, and it is obtained by reading out each pixel in the absence of radiation.

![Image of a dark field](a)  ![Image of a flood field](b)

**Figure 1.8** (a) Image of a dark field taken with the aS1000 Varian and (b) image of a flood field using 6 MV photons, also taken with the aS1000 Varian.

Figure 1.8a is a series of narrow vertical stripes, which result from the photodiode leakage current and varying electrometer offsets. The flood field image (see Figure 1.8b), on the other hand, is taken with the entire matrix exposed to a uniform dose. This allows the image acquisition software to internally correct for individual pixel sensitivities.

### 1.2.1.4 Failure Mode and Effect Analysis (FMEA)

As complexity and sophistication of equipment and processes within the clinical imaging environment, unexpected errors or system failures are more likely to occur. Failure Mode and Effect Analysis (FMEA) is a tool that identifies the possible failures in complex processes and provides a basis for improvement (Thornton, Eavan et al. 2011). FMEA is to reduce, predict, or prevent errors. FMEA is a systematic approach to identify and understand
contribute factors, causes, and effects of potential failure. There are six sequential steps in the FMEA process (see Figure 1.9).

**Figure 1.9 Six sequential steps used in FMEA (Thornton, Eavan et al. 2011)**

**Step 1 Define the topic:** The topic should be selected on the basis of data gathered for QA and performance problems. The root cause from sentinel events should be considered during choosing the topic. For example, interventional procedures, such as scheduling, performance, follow-up, communication of results, patient throughput, specimen labelling, transport, and processing, equipment failures, room utilisation, and after-hours clinical coverage.

**Step 2 Assemble a committed team:** The team must agree for the improvement. Leadership buy-in and support are essential, and must understand that the process is worth. The resources should be supported the changes implemented by the team. Once FMEA is implemented, the team is responsible for identifying the process, systems, or procedures that can be improved. Appropriate team member should be chosen for particular tasks to achieve the goal.

**Step 3 Develop a process map:** A detailed chart of the process should be constructed by the team enabling complete understanding of the individual steps that are involved in the process, namely process map. The process map is different from the regular flow chart by showing inputs, outputs, and units of activity, as well as decision or action points. The process map contains cycle time and delays between stages, responsible persons, inventory, the value or cost added at each step, and wastage.
Step 4 Conduct a risk or hazard analysis for each subprocess: the subprocesses should be then identified as identification of the failure modes. In a radiology environment, something could go wrong because of any number of factors, including staffing, local environmental issues, policies and guidelines, poor communication, equipment problems, human error, and missing or misplaced medications. The probability of a failure mode occurring is determined and ranking (scale of 1-10) as shown in table 1.1 (McDermott et.al. 1996, Goodman et.al. 1996). In addition, the probability that a failure mode will be detected presented in table 1.2. The failure mode could potentially be detected by personnel working in the department or by the patient, or as an alert in a computer system.

**Table 1.1 Occurrence Rating Scale of Failure Modes** (McDermott et.al. 1996, Goodman et.al. 1996)

<table>
<thead>
<tr>
<th>Score</th>
<th>Probability of Occurrence</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Remote – no known occurrence</td>
</tr>
<tr>
<td>2</td>
<td>Low probability – rare failures (yearly)</td>
</tr>
<tr>
<td>3,4</td>
<td>Moderate probability – occasional failures (quarterly)</td>
</tr>
<tr>
<td>5,6</td>
<td>Moderate high probability – monthly</td>
</tr>
<tr>
<td>7,8</td>
<td>Very high probability – frequent (weekly)</td>
</tr>
<tr>
<td>9</td>
<td>Inevitable and predictable failure</td>
</tr>
<tr>
<td>10</td>
<td>Certain probability – daily, or every time</td>
</tr>
</tbody>
</table>

**Table 1.2 Detection Rating Scale of Failure Modes** (McDermott et.al. 1996, Goodman et.al. 1996)

<table>
<thead>
<tr>
<th>Score</th>
<th>Probability of Occurrence</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Certain – error will always be detected</td>
</tr>
<tr>
<td>2</td>
<td>Very high probability that error will be detected</td>
</tr>
<tr>
<td>3,4</td>
<td>High probability of detection</td>
</tr>
<tr>
<td>5</td>
<td>Moderate chance that error will be detected</td>
</tr>
<tr>
<td>6,7</td>
<td>Remote chance of detection only</td>
</tr>
<tr>
<td>8,9</td>
<td>Remote or low likelihood of detection</td>
</tr>
<tr>
<td>10</td>
<td>No chance that error will be detected; no mechanism exists</td>
</tr>
</tbody>
</table>
The risk priority number (RPN), or the criticality index, is a quantitative measurement, which is used to evaluate and assess the failure mode (Marwick et al. 2007). The RPN is derived from the product of the numeric ratings for severity, probability of occurrence, and detectability (see equation 1.1). The RPNs are ranked to allow prioritisation of the failure modes and to highlight the failure modes that exceed acceptable limits. An attention should focus on any domain where the severity ranking is high.

\[
RPN = \text{Severity} \times \text{Occurrence} \times \text{Detection} \tag{1.1}
\]

Step 5 Develop and implement an action plan to redesign the process: Once a failed process has been identified, strategies should be developed and implemented to prevent subsequent occurrence. For example, specific corrective action, coupled with defined outcome metrics and timelines, the person who responsible for implementing the plan, educating staff, and monitoring results.

Step 6 Monitor, sustain, share and re-evaluate the improvement: Since new advancing system is implemented, there will always be new high risk processes that can be analysed using FMEA. It is important that processes should be monitored, sustained, shared and re-evaluated to ensure this implementation meets the desired impact.
1.3 Thesis outline

This thesis consists of a detailed development of EPID-based real-time dosimetry for patient safety in radiation therapy and is divided into the following chapters:

Chapter 1: Motivation and Background

Chapter 2: Literature Review

Chapter 3: Materials and Methods

Chapter 4: Simulation Results

Chapter 5: Discussion

Chapter 6: Conclusion and Recommendation for Future Work

Chapter 2 provides detail of the current knowledge relevant to this project, IMRT verification, data analysis tools, EPIDdosimetry, time-resolved prediction models, and patient-specific QA.

Chapter 3 gives detail on the materials and methods used for this research and system development. This chapter is divided into two main sections, real-time dosimetry and real-time geometry verification.

Chapter 4 presents the results of system simulations for both real-time dosimetry and real-time geometry verification techniques.

Chapter 5 provides discussions that summarise the system potential and its limitations.

Finally, Chapter 6 concludes the thesis with recommendations for future work and clinical applications.
The use of EPIDs for verification is now routine in a growing number of clinics. EPID dosimetry plays a common role in verification procedures particularly since the widespread use of IMRT and IMAT commenced. Highly complex treatments demand an exceptionally high level of patient-specific verification. However, the complexity of IMRT and IMAT makes it more difficult to recognize and verify potential errors in treatment. Previously used generic QA approaches are no longer sufficient and more patient-specific verification procedures become necessary to detect possible errors for these complex treatments, especially real-time treatment verification.

2.1 A feasibility study for real-time dosimetry using EPID

One approach to real-time monitoring is the use of point dosimeters, e.g. polymethylmethacrylate (PMMA) optical fibres, which are set up either on the skin or inside the patient’s body, depending on the location of the tissue to be treated (S.O'Keeffe 2009). However, the main drawbacks of this approach are the lack of two-dimensional data and the difficulty of the setup, which is why these methods are not frequently used in clinical situations.

Since the amorphous silicon EPIDs were introduced, EPID dosimetry has been widely investigated for both research and clinical uses, particularly for IMRT and IMAT verification. An advantage of using EPIDs for IMRT dosimetry is the high resolution and digital format. EPID properties for dosimetry have been investigated, including linearity, reproducibility, MLC leaf speeds, and dose-rate fluctuations. The a-Si EPID has a linear dose response that makes it suitable for dosimetry (Greer and Popescu 2003). However, some studies have found that the a-Si EPID is non-linear at small monitor unit (Low, Harms et al.) settings but that this does not have a large effect on dose measurement when a typical treatment dose is integrated (McDermott, Nijsten et al. 2006). The EPID showed promise as a device for IMRT verification, but dead-time was identified to be a major limitation of the system (Greer and Popescu 2003).

Grein et al., found that the relationship between absorbed dose and pixel value was linear for all source-to-detector distance (SDDs) and field sizes, although the slope was field size dependent (Grein, Lee et al. 2002). Van Esch et al. described the dosimetric properties of the a-
Si EPID for an acquisition mode which reduced dead-time effects during delivery both in static
and dynamic IMRT. Detector saturation, linearity of detector response, reproducibility,
ghosting, and field size dependence were investigated (Van Esch, Depuydt et al. 2004). Saturation
effects could be reduced for all SDDs by restricting the LINAC dose rate setting up to 300
monitor units per minute (MU/min). The short- and long-term reproducibility was found
to be excellent, and the effect of ghosting was clinically negligible. Adding build-up to the
detector surface neither facilitated nor complicated the prediction of the EPID dose. The field
size dependency was consistent at different SDDs and could be modelled through a single
analytical function. Moreover, Kirkby et al. investigated the spectral response of the Varian
aS500 EPID and the implications for dosimetric calibration. They examined the specific case of
beam hardening by using compensator material and showed that the discrepancy between open
and attenuated beam calibration curves can be as high as 8% for 6 MV photon beams. To reduce
the maximum discrepancy to < 4%, about 0.7 cm copper was placed 15 cm above the EPID,
supported by Styrofoam (Kirkby and Sloboda 2005).

The a-Si EPID has shown good characteristics for dosimetric purposes, has a linear
response to dose and rate, and is stable with time. It is able to acquire images in real-time at a
high frame rate, and it was also demonstrated that the a-Si EPID signal can be calibrated in
terms of absolute absorbed dose (Grein, Lee et al. 2002). Furthermore, the idea of using cine
acquisition mode was sparked by Piermattei et al. They used a frame acquisition rate of one
image every 1.66 s and a low dose-rate (100MU/min) to test the signal stability, signal
reproducibility, and linearity. Over 3 months, the investigation showed very good conditions for
using cine mode (Piermattei, Fidanzio et al. 2009). McCurdy et al. studied the dosimetric
properties with a frame-rate of 10 frames per second and found good agreement between cine
mode and integrated mode beams. Real-time EPID measurement for IMRT and VMAT fields
were in high agreement (> 95%) compared with real-time ion-chamber data (McCurdy and
Greer 2009). Mans et al. modified a 3D back-projection algorithm to work with time-resolved
EPID measurements at 2.5 frames per second (Mans, Remeijer et al. 2010). In their experiment,
they found the image lag or image acquisition delay to be approximately 0.4 sec or 1 frame in
the gantry angle.

The continuous acquisition (cine) mode results in a series of images where EPID dose data
are captured as a function of time during the entire irradiation. Thus, instead of a single
integrated image, real-time images could be performed for real-time dose verification. McCurdy
et al. suggested that the continuous acquisition mode is suited for real-time dosimetry
applications, including dynamic and arc-IMRT, where time-dependent dosimetry data are
captured and stored. As a result, it is feasible to use EPID with the continuous acquisition mode
for real-time dose verification applications, including dose reconstruction using the EPID images combined with gantry angle information.

Bakhtiar et al. presented a QA method to compare the MLC shape generated by the TPS at each snapshot (control point) acquired at a specific gantry angle (Bakhtiar, Kumaraswamy et al. 2011).

![Figure 2.1](image)

Figure 2.1 (a)-(d) examples of real-time images for a head-and-neck intensity modulated radiation therapy plan. The modulated beam is formed by sweeping computer controlled collimation leaves across during the delivery; (e)-(h) Cumulative EPID dose images in greyscale units representing the cumulative dose that correspond to these time points in the delivery.

Based on the investigations performed in the above publications and the recently developed QA applications in real-time, EPID dosimetry shows the potential for time-resolved dosimetry with cine mode imaging. Consequently, EPID-based real-time dose verification should be able to detect errors during the radiation treatment.
2.2 EPID dosimetry

Van Elmpt et al. classified EPID dosimetry procedures according to EPID position, attenuating medium, and verification phase (see Figure 2.2 and Table 2.1) (van Elmpt, McDermott et al. 2008).

 Dosimetry methods can be categorised into EPID calibration methods: 1) the conversion of greyscale pixel values into a dose value and 2) the simulation of greyscale pixel values. (van Elmpt, McDermott et al. 2008). This first method is to calibrate greyscale pixel values to dose by inter-comparison with a calibrated dosimeter, such as dose-in-water. The second method uses analytical or Monte Carlo calculations to model the detector response or the EPID level.

Clinical EPID dosimetry can be performed pre-treatment or during treatment with transit measurements. **Pre-treatment verification** procedure is a verification of the whole or part of the treatment plan using measured EPID images before the treatment. This comparison can focus on various aspects of the treatment plan, such as comparing the predicted and measured leaf positions, dose delivered to a phantom, or incident energy fluence extracted from measurements. **During Treatment verification** procedures verifies the treatment plan with measured EPID images during the treatment. Real-time dosimetry is included in this type. These measurements can be used to monitor the dose delivered to the detector (non-transit measurement) or patient (transit measurement). Transmission dosimetry is a significantly more complex problem, due to scatter from the attenuating medium (either patient or phantom). Data analysis is performed usually with the gamma analysis method for dose comparison (Low, Harms et al. 1998) for treatment error detection.

Figure 2.2 Three classifications of EPID dosimetry method (van Elmpt, McDermott et al. 2008)
Table 2.1 A classification of EPID dosimetry based on van Elmpt et al. investigation (van Elmpt, McDermott et al. 2008).

<table>
<thead>
<tr>
<th>Type of EPID dosimetry</th>
<th>Methodology</th>
</tr>
</thead>
<tbody>
<tr>
<td>Non-transmission dosimetry</td>
<td>- 2D verification at the EPID level</td>
</tr>
<tr>
<td></td>
<td>- 2D verification at the patient level</td>
</tr>
<tr>
<td></td>
<td>- 3D dose reconstruction</td>
</tr>
<tr>
<td>Transmission dosimetry</td>
<td>- Point-dose verification</td>
</tr>
<tr>
<td></td>
<td>- 2D portal dose prediction at the EPID level</td>
</tr>
<tr>
<td></td>
<td>- 2D back-projection model at the patient level</td>
</tr>
<tr>
<td></td>
<td>- 3D dose reconstruction</td>
</tr>
</tbody>
</table>

*Non-transmission dosimetry* can be performed either pre-treatment or during treatment. The detector position must be set between the source and patient or phantom in during treatment non-transmission dosimetry and normally the detector is attached to the linac-head. Non-transit measurement data is very useful for performing QA related to the dosimetric and geometric characteristics of the linear accelerator. The purpose of non-transmission dosimetry is to make sure the linac delivers the correct treatment based on the TPS plan.

Non-transmission 2D verification is used to verify the delivery of the correct fluence by comparing with the fluence exported from the TPS or calculated from the TPS plan using the planned treatment parameters. Several approaches have been developed for the conversion of EPID image scale values into a dose plane within a water or solid water phantom. The importance and the consequence of the spectral response of a-Si EPIDs for dosimetric calibration was investigated (McCurdy 2001; Greer 2005; Kirkby and Sloboda 2005). It was found that EPIDs display an energy-dependent response due to the high atomic number of phosphor. Adding copper shielding could reduce the over-response of detectors to low-energy components. Prediction models of EPID response have been developed using different techniques, such as the Monte Carlo simulation model (Parent, Seco et al. 2006) and physical-based model (Chytyk and McCurdy 2006; Chytyk and McCurdy 2009), which is used in this project. A comparison between the planned and measured relative profiles of 25 prostate fields after applying correction methods to the EPID response was investigated (Chang and Ling 2003; Tateoka, Ouchi et al. 2006). Additionally, a method to verify the absolute point dose and the 2D geometrical parameters of the treatment fields at the level of the EPID was developed (Nijsten, Minken et al. 2004).
2D dose distribution reconstruction can be performed inside the patient data CT scan or a phantom at a specific depth. The method originally proposed was a deconvolution of the EPID images to a 2D distribution of primary fluence using Monte Carlo generated scatter kernels (Warkentin, Steciw et al. 2003). Alternatively a leaf sequence file extracted from EPID images can be used to reconstruct the dose (Lee, Mao et al.). The reconstructed dose distribution inside the phantom or patient from the EPID measurements can then be compared to the predicted dose reconstruction from the TPS. 3D dose distribution based on measured EPID images is also used for dosimetry. The 3D dose distribution is calculated by using a dose calculation algorithm. The EPID images can be acquired pre-treatment or during treatment. The reference dose distribution can be calculated using either the patient CT scan or the phantom CT scan.

Figure 2.3 shows a summary diagram of non-transmission dosimetry approaches that can be performed at either the patient level or detector level. The TPS can calculate the predicted dose distribution. EPID images can be converted to dose-in-water for comparison with the TPS calculation. In the case of moving the detector to be between the source and the patient or phantom, the 2D or 3D dose distribution at the patient level can be reconstructed. Monte Carlo models and physics-based models have then been used for calculating the predicted dose distribution.

![Diagram of non-transmission dosimetry approaches](image)

**Figure 2.3** Summary of non-transmission pre-treatment and during treatment dosimetry methodology.
Transmission EPID dosimetry is the procedure for patient-specific treatment verification that ensures the patient receives the correct dose during treatment. The verification can be performed at either the EPID level or the patient level (using the patient’s CT-scan). In Figure 2.4, the methodology of transmission EPID dosimetry can be divided into 4 parts: point-dose verification, 2D portal dose prediction, 2D back-projection model, and 3D dose reconstruction models. Point-dose verification based on transmission images was used to replace diode-based in-vivo dosimetry, (Leunens, Van Dam et al. 1990). Point-based in-vivo dosimetry is very useful for detecting large errors in the delivery of conformal treatment fields. Generally, in-vivo dosimetry is performed by using diodes positioned on the patient. Setting up the diodes on the patient is time consuming and so is maintaining an accurate calibration. As a result, diode in-vivo dosimetry is not used at many centres and on many patients. Point dose verification using an EPID has been investigated(Nijsten, Mijnheer et al. 2007; Piermattei, Fidanzio et al. 2007).

2D verification at the EPID level is the most popular method for transmission dosimetry that involves predicting the portal dose at the level of the imager behind a phantom or the patient. This method has been applied for this project by acquiring EPID images in real-time and comparing with the real-time predicted image at the EPID level with the phantom as the
attenuating medium. Pasma et al. first introduced 2D verification at the EPID level by comparing corrected EPID images to ionisation chamber measurements and found that there was agreement within 1% (Pasma, Heijmen et al. 1998). In addition, the comparison of EPID dose images and ionisation chamber measurements at the detector plane behind an absorber for various field shapes gave an average difference in absolute dose of 0.1±0.5% (1 SD) over the entire irradiation field, with no deviation larger than 2% (de Boer, Heijmen et al. 2000). Afterward, (Spezi and Lewis 2002) developed a planar calibration of the SLIC EPID response using full Monte Carlo simulation of the radiotherapy treatment machine, including the MLC and EPID. Computed and measured doses agreed within 2%. (Mohammadi, Bezak et al. 2006) introduced a method of calculating the transmitted dose map by using the TPS and reported that more than 90% agreement was achieved using dose-difference and distance-to-agreement criteria of 2%, 3mm or 3%, 2.5 mm.

Regarding verification at the patient or phantom level (in vivo), it is necessary to make a measurement at the same time that the dose is being delivered to the patient or phantom and use the measurement to determine the dose inside the patient or phantom. Besides the measurement, therefore, an algorithm to reconstruct the dose in the patient from the EPID image is required. Boellaard et al. introduced a back-projection midplane dose reconstruction model for the SLIC EPID (Boellaard, Essers et al. 1998). The calibration of the EPID for back-projection dosimetry consisted of two parts, including a dosimetric calibration—which established the dose response relationship to EPID pixel values and conversion to dose values at the position of the EPID imager—and the parameters for the back-projection algorithm, i.e. attenuation factors and scatter within the EPID and patient, and from patient to EPID. All parameters were determined for converting the dose at the EPID position to the dose inside the patient or phantom. An extension of this method to the a-Si EPID was investigated and reported close agreement between EPID and film measurements using gamma evaluation with 2%, 2mm dose-difference and distance-to-agreement criteria (Wendling, Louwe et al. 2006). Moreover, the back-projected 2D EPID dosimetry replaced verification with film, and subsequently in vivo verification replaced pre-treatment checks for prostate treatments in terms of clinical application (McDermott, Wendling et al. 2006).
Figure 2.5 Comparison of EPID and film dose distributions inside a phantom for pre-treatment verification of an IMRT field: (a) Two-dimensional dose distribution derived with the EPID, (b) EPID dose as solid line, film dose as dashed line, (c) Two-dimensional $\gamma$-distribution of EPID versus film dose. A dose-difference criterion of 2% of the maximum dose and a distance-to-agreement criterion of 2 mm were used (from Wendling et al., 2006).

Three-dimensional dose reconstruction based on transmission dosimetry was extended from the back-projection algorithm for 2D dose reconstruction. The method was to calculated parallel dose planes in the patient perpendicular to the beam direction; as a result, a 3D dose reconstruction was generated using the EPID images and CT data (McDermott, Wendling et al. 2008). McDermott reported that planned and measured in vivo 3D dose distributions showed excellent agreement based on 3D gamma evaluation and dose-volume histograms (DVHs).

2.3 EPID geometry

The main purpose of EPID geometry verification is to monitor mechanical performance. Figure 2.6 shows the geometrical linac characteristic including jaw positioning, MLC movement, and collimator rotation. Real-time geometry verification aims to ensure a high accuracy of jaw position, MLC leaf position, and collimator rotation. Gantry angle, jaw position and collimator rotation are stable during the beam delivery in IMRT. On the other hand, VMAT is more complex technology and the gantry moves during beam delivery.
Figure 2.6 The geometrical linac characteristics include jaw position, MLC movement, and collimator rotation.

Jaw positions and collimator rotation can be determined from interleaf leakage analysis. Gao et al. applied the Radon Transform and Cross-correlation algorithm to determine the position and orientation of interleaf leakage from the EPID (Gao, Szanto et al. 2007). This method showed an accuracy within 0.5 degrees for collimator angle and less than 1 mm for jaw positioning. However, this method used the integrated EPID image and real-time jaw and collimator rotation verification was not investigated.

MLC leaf position can be detected on film, EPID, or using the dynamic log file. The dynamic log file, or “DynaLog Files” in the case of Varian MLCs, contains the leaf position recorded by the control system every 0.055 seconds. Litzenberg et al. developed the analysis and assessment tools for DynaLog Files. Kinetic, dosimetric, and statistical properties of the treatment delivery is calculated (Litzenberg, Moran et al. 2002). However, the limitation of using a DynaLog File for QA is it cannot be used for a real-time system. Using EPID for MLC QA has become more popular because of the advantage of its real-time characteristics. The EPID detector has developed from CCD Camera-based to a-Si EPID. The new a-Si EPID improves the image quality and geometrical distortion and enhances the accuracy of MLC leaf positioning verification to within an error of approximately 0.1 mm.

Partridge et al. developed a method to detect the dynamic MLC leaf position using the EPID. The 50% intensity level on time-averaged portal images found from an edge detection algorithm can be used to define the MLC leaf position. The accuracy is good compared with the
manufacturer’s specification, with generally less than 2% errors (1 mm error). The mean deviation of average MLC leaf position and specified leaf positions at prescription control points was used for the geometric verification (Partridge, Evans et al. 1998). James et al. pioneered a real-time qualitative verification of leaf positioning by overlaying a template of expected leaf positions onto the EPID images for comparison (James, Atherton et al. 2000). Samant et al. tested the performance of edge detection algorithms, including Contour, Sobel, Prewitt, Roberts, Laplacian, Canny, and Morphological, in terms of the processing time and accuracy of leaf positioning. All of the edge detection algorithms performed well with errors less than 1.5 mm. The Canny edge detection showed the best testing results, but it is the most time consuming. Samant et al. recommended that the Contour and Morphological edge detection algorithms are suitable candidates for real-time image analysis (Samant, Wei et al. 2000). Moreover, they proposed a supervised Laplacian of Gaussian (Slog) as the edge detection algorithm for portal images (Samant, Wu et al. 2000). Besides edge detection algorithm development, a geometric distortion correction was necessary. Xu et al. and Partridge et al. provided a method for an automatic geometric distortion correction by analysing the grid pattern reconstructed from the EPID image of a calibration phantom (Partridge, Symonds-Tayler et al. 2000; Xu, Chuang et al. 2002). Xu et al. proposed an active template-matching algorithm, which finds the best-fitting rectangle that matches the radiation field boundary. Vieira et al. introduced a pre-processing step to improve the quality of the EPID image (Vieira, Dirkx et al. 2002). The accuracy was improved to within an error of approximately 0.2 mm.

The application of using CCD-camera based EPIDs for MLC leaf positioning is limited due to the drawbacks of resolution, image distortion, and signal-to-noise ratio of these images. Since the amorphous silicon EPIDs were introduced, there has been increasing use of EPID as a QA tool in clinical use. The a-Si EPID provides a robust, geometrically stable, non-distorted, and better resolution image compared with other types of EPIDs. Yang et al. developed a new technique defining the cumulative signal in a small region of interest (ROI) as an indicator of the leaf position instead of using the conventional edge detection, 50% intensity level (Yang and Xing 2004). Their results indicate that their technique can detect a leaf positional error of approximately 0.1 mm. Kang et al. developed an EPID image resolution calibration prior to edge detection, which uses an edge-differential algorithm (Kang, Deng et al. 2009). Baker et al. proposed MLC calibration using an a-Si EPID (Baker, Budgell et al. 2005). The feasibility of using cine EPID images for MLC leaf positioning was introduced by Lee et al. (Lee, Mao et al. 2008). The result of MLC leaf positioning determination was used to generate a leaf sequence file, which can be imported into the TPS for dose reconstruction. They also applied edge detection with 50% intensity level as the threshold.
2.4 Dose distribution comparison tools

EPID dosimetry is used to directly verify the measured dose distributions by comparing to the reference dose that is calculated from the TPS. This verification procedure requires a mathematical comparison using the dose-difference (DD) and distance to agreement (DTA) concept. Van Dyk et al. first introduced DD for dose distribution comparison (Van Dyk, Barnett et al. 1993). The DD method is the straightforward technique however it results in large differences in steep dose gradient regions. The relatively small misalignments which can produce these large differences are not taken into account. DTA was created to solve the DD limitation and can be applied for 2D dose distribution comparisons (Van Dyk, Barnett et al. 1993; Low, Harms et al. 1998). In addition, the gamma ($\gamma$) evaluation method (Low and Dempsey 2003) combines both DD and DTA techniques (that are overly sensitive in steep dose gradient regions and in shallow dose gradient regions respectively). This has become the most popular dose comparison tool that is applied both for research experiments and clinical use.

The gamma evaluation method can be applied in both 2D and 3D, requiring two dose distributions, one a reference and the second the evaluated dose distribution or measured dose distribution. The $\gamma$-index is calculated by comparing the reference and evaluated dose. The reference dose is queried point by point for calculating the $\gamma$-index so that the reference dose does not need to be the same type as the evaluated dose; for example, the reference dose is measured by an ionization chamber measurement point and compared with the evaluated dose measured from EPID. To compare using the gamma evaluation method, the criteria of DTA and DD must be defined, such as DTA is 3 mm and DD is 3%. The $\gamma$ index calculation searches the local evaluated dose distribution based on DTA criteria. The searching algorithm uses the Euclidean distance technique, which generalises the dose and distance space. The renormalized dose and distance space have been divided by the DTA and DD criteria.

$$
\Gamma(\vec{r}_e, \vec{r}_r) = \sqrt{\frac{|\vec{r}_e - \vec{r}_r|^2}{\Delta d^2} + \frac{[D_e(\vec{r}_e) - D_r(\vec{r}_r)]^2}{\Delta D^2}}
$$

(2.1)

Where $\vec{r}_e$ and $\vec{r}_r$ are the vector positions of the evaluated and reference points, respectively, $D_e(\vec{r}_e)$ and $D_r(\vec{r}_r)$ are the evaluated and reference doses, respectively, and $\Delta d$ and $\Delta D$ are the DTA and dose difference criteria, respectively. The generalized $\Gamma$ function can be computed for any pair $\vec{r}_e$ and $\vec{r}_r$, so for each reference point, there are as many values of $\Gamma$ as there are evaluated points (infinite number with interpolation). The minimum value of $\Gamma$ is the value of $\gamma$ as shown in equation 2.2.
\[
\chi(\vec{r}_r) = \min\{\Gamma(\vec{r}_e, \vec{r}_r)\} \forall (\vec{r}_e) \tag{2.2}
\]

Equation 2.2 states that \(\chi\) is simply the minimum value in all of the evaluated distribution search space of \(\Gamma\) (see figure 2.7). While equations 2.1 and 2.2 provide the factual definition of \(\Gamma\), they do not impart any intuition for what the \(\chi\) means and its utility. Finally, the output of the gamma evaluation method is the \(\chi\) map, which is the same spatial resolution as the reference distribution. The pass-fail criterion used is shown in equation 2.3.

\[
f(x) = \chi(\vec{r}_r) \begin{cases} f(x) \leq 1; & \text{pass} \\ f(x) > 1; & \text{fail} \end{cases} \tag{2.3}
\]

**Figure 2.7** Determination of the \(\chi\) value. The \(\Gamma\) is calculated for a specific point of the evaluated image. The same point of the reference image is compared with another point in the evaluated image. For all points, the \(\Gamma\) value is computed, and the minimum of these values is the \(\chi\) value, which corresponds to the reference point.
The spatial registration error or misalignment of the reference and evaluated dose distributions have been corrected by applying a user-defined distance tolerance (Moran, Radawski et al. 2005). Instead of implementing the distance tolerance, Stock et al. firstly defined the $\gamma$-angle for the 2D dose distribution that identifies the dose axis (Stock, Kroupa et al. 2005).

$$\varphi(\vec{r}_e, \vec{r}_r) = \tan^{-1} \left( \frac{r(\vec{r}_e, \vec{r}_r)}{\delta(\vec{r}_e, \vec{r}_r)} \frac{\Delta D}{\Delta d} \right)$$

(2.4)

Where $\varphi(\vec{r}_e, \vec{r}_r)$ is the $\gamma$-angle and $\delta(\vec{r}_e, \vec{r}_r)$ is the relative dose difference; the $\gamma$-angle indicates the parameter that most affects the gamma function. The angle at $0\pi$ is defined only for dose difference and $\pi/2$ is simply for distance difference.

$$\theta(\vec{r}_e) = \tan^{-1} \left( \frac{\Delta y}{\Delta x} \right)$$

(2.5)

Where the angle $\theta$ indicates the direction of the distance difference in the (x,y)-plane (angle between $0\pi$ and $2\pi$), where $0\pi$ is the displacement in the x direction and $\pi/2$ in the y direction (see 2.8 b). In addition, they introduced $\gamma$-histograms as a method for representing the complex distributions into a histogram plot. A histogram allows the user to more easily determine the results of the $\gamma$-index in terms of the pass-fail criteria (see 2.8 c).
Jiang et al. proposed a different method that converts the comparison in the spatial domain into a comparison in the dose domain, called the “equivalent dose tolerance” concept. A maximum allowed dose difference (MADD) was introduced as the predetermined dose tolerance. The MADD is used to set a condition for the dose comparison that the dose difference is smaller or equal to MADD, then the comparison passes in both the dose and spatial domains. As a result, only dose difference is compared with MADD, and dose difference in terms of spatial domain is not necessary. They also introduced the evaluation method by constructing the normalized dose difference (NDD) (Jiang, Sharp et al. 2006). This method was recommended for the problem of $\gamma$ artefacts caused by the limitation of pixel size spacing of measured EPID image or EPID resolution (Low 2010).

Another issue for dose comparison is the processing time. The original gamma evaluation method (Low and Dempsey 2003) requires a comprehensive search inside the evaluated dose distribution that causes a high level of processing time, especially for 2D or 3D dose comparison. This major limitation led a number of groups to develop methods for modifying or improving the dose comparison tool.

Depuydt et al. proposed a modification to the searching algorithm by applying a filter cascade of 3 levels that classifies accepted or rejected data points of the reference distribution. The algorithm focuses on the $\gamma$ index being greater or less than 1 (pass-fail criteria), instead of a fully evaluated $\Gamma$ function and $\gamma$ calculation. The filter cascade process determines which evaluated points are within or outside the pass-fail criteria. With this filter cascade condition, the processing time of dose calculation is improved (Depuydt, Van Esch et al. 2002). Bakai et al. modified the of $\gamma$-index calculation to use the gradient of the reference dose distribution. They presented the new evaluation factor $\chi$ as

$$\chi = \frac{D_e(\vec{r}) - D_r(\vec{r})}{\sqrt{\Delta D_{\text{max}}^2 + \Delta d_{\text{max}}^2} \|\vec{\nabla}D_r\|^2}$$

(2.6)

where $\Delta D_{\text{max}}$ and $\Delta d_{\text{max}}$ are DD and DTA criteria respectively, and $\|\vec{\nabla}D_r\|^2$ is the second derivative of the reference dose distribution. The result of the $\chi$ comparison result returns similar results to the $\gamma$ comparison result, so the pass/fail criteria can be used with the same condition as $|\chi| \leq 1$ for acceptance (Bakai, Alber et al. 2003). Using the Bakai et al. method results in computation times 120 times faster than the $\gamma$ comparison method. However, the $\chi$ comparison requires both evaluated and reference distribution in the same grid or resolution.
The comparison of 3D dose distribution reconstructed inside the patient data (CT-scan) is widely used and has been demonstrated in patient-specific QA, especially in VMAT (Teke, Bergman et al. 2010). In 3D the issue of the time-consuming gamma evaluation becomes of serious concern. Wendling et al. developed a fast method of 3D γ-index calculation (Wendling, Zijp et al. 2007). They changed the searching strategies from originally full interpolation of the evaluated distribution to using the search in a radial pattern, beginning with a reference point and storing the interpolation factors. This method shows a 75% reduction in calculation time, though resource consumption is potentially high. Ju et al. approached this problem from a geometrical perspective. They proposed a geometric interpretation of the γ-index evaluation technique (Ju, Simpson et al. 2008). This method is specially developed for 3D dose comparison and takes approximately one minute for γ calculation. The evaluated distribution is subdivided into simplexes: line (1D), segments (2D), and triangles (3D). The closest distance between a reference point and these simplexes is rapidly computed. This interpolation of γ distribution results in significantly faster calculation. Chen et al. applied the Euclidean distance transform (EDT) with the (k+1)-d spatial-dose space embedded in the reference distribution (Chen, Lu et al. 2009). Yuan et al. applied the k-d tree and nearest neighbour searching to improve the searching techniques in the gamma dose comparison (Yuan and Chen 2010).

Another method to reduce the speed of γ calculation is to apply the algorithm on a graphic processing unit (GPU). GPU has been introduced as a platform for calculating with massive parallel computation. In addition, the medical physics community has been using the GPU to accelerate the speed and sophistication of image reconstruction, dose calculation, treatment plan optimization, and image processing (Pratx and Xing 2011). Xuejun et al. applied a geometric interpretation technique (Ju, Simpson et al. 2008), operating with a pre-sorting technique(Wendling, Zijp et al. 2007) instead of using the original EDT method of searching implemented on GPU (Xuejun, Xun et al. 2011). The modified γ-index calculation on GPU gave the result 20 to 30 times faster than implementation on a CPU. Person et al. tested the original 3D γ calculation (Low and Dempsey 2003) on GPU and found it to be an average of 5% faster for the virtual phantom cases and an average of 30% faster for the patient cases (Persoon, Podesta et al. 2011).
3.1 Linac and EPID

All radiation fields were produced at 6 MV using a Clinac iX linear accelerator (Varian Medical Systems, Palo Alto, CA) equipped with a Millennium 120-leaf MLC. A megavoltage (MV) EPID (Varian aS1000 flat panel detector) was used to acquire the images. The detector has a size of 40\times 30 \text{ cm} with a matrix of 1024\times 768 pixels, each square pixel having a side length of 0.0392 \text{ cm}. The electrical signals are digitized by a 14-bit analogue-to-digital converter and processed into image data. All data was acquired in continuous (cine) acquisition mode controlled by the IAS3 (Image Acquisition System version 7.3.15) AM-Maintenance software module (Varian Medical Systems, Palo Alto, CA) and hence constitutes a time-lapse series of individual frames acquired during dose delivery. The MLC-defined IMRT fields used for the commissioning of the synchronization and verification systems were clinical fields, with 400 monitor units per minute (MU/min) and the EPID positioned at 150 \text{ cm} source-to-detector-distance (SDD). All aS1000 images were automatically dark-field and flood-field corrected by the IAS3.

3.2 System development tools

The development of a real-time system for IMRT consists of two separate parts, including real-time EPID dosimetry and real-time EPID geometry. Both dosimetry and geometry used the equivalent linac and EPID environment and were simulated using MATLAB® (Math Works, Natick, MA, USA) and Simulink® (Visual programming environment). MATLAB/Simulink was used as the main platform, due to its ubiquity and its capacity to provide a broad range of computational functionality. Adopting Simulink allows the ‘block’ algorithm components to be written in either MATLAB or C/C++, both commonly used computer languages for image processing and analysis. A virtual frame grabber environment was created to simulate real-time EPID data acquisition at the linac. The environment reads either previously acquired cine or predicted images from storage and feeds them to the
verification algorithms at the EPID imaging rate. This allows for a more accessible and controlled testing strategy.

3.3 Prediction model

Predicted EPID images were calculated representing segments of dose delivery centred at each control point using the physics-based model of McCurdy et al. (Chytyk and McCurdy 2006; Chytyk and McCurdy 2009). This models the energy fluence distribution from the head of the accelerator, the primary attenuation through the patient or phantom to the EPID, scattered radiation from the patient or phantom to the EPID, and the energy fluence absorption in the EPID incorporating EPID support arm backscatter (Rowshanfarzad, McCurdy et al. 2010). The model separates the energy-fluence into 10 energy bins to account for the energy-dependent response of the EPID. The predictions are in absolute greyscale units of the EPID with a calibration factor derived from the ratio of model prediction and measured EPID image at central axis for a 10x10 cm field at 150 cm SDD and 100 monitor units. The model was adapted to calculate snapshot EPID images or predicted frames during the delivery. Each snapshot represents sequential dose increments delivered between control points.

Figure 3.1 The physics-based prediction model concept that generates the image representing its control point and amount of dose collects between the middle of the previous control point through the middle of the next control points.
3.4 Image synchronisation

The basic concept of geometric-based synchronization is to find the closest match of the MLC position between the measured and the predicted EPID images. The leaf positions are automatically extracted from each image frame (predicted and measured). The cosine similarity (Fraass, Lash et al.) is used as a metric for the comparison, which indicates the similarity between two vectors by calculating the cosine of the angle between them. This technique is often used to measured cohesion within clusters in the field of data mining (Tan, Steinbach et al. 2006). The cosine similarity is defined as:
\[ CS_{k,n} = \frac{1}{2} \cdot \left( \frac{MP_A(k)-PP_A(n)}{||MP_A(k)||||PP_A(n)||} + \frac{MP_B(k)-PP_B(n)}{||MP_B(k)||||PP_B(n)||} \right) \]  

(3.1)

Where \( MP_A(k) \) and \( MP_B(k) \) are the results of the automatic MLC leaf positioning algorithm at phase (frame) \( k \) for leaf bank A and B respectively, and the CP number \( n \) is a monotonically rising function from 1 to number of control points (N) and \( PP_A(n) \) and \( PP_B(n) \) are the MLC leaf positions given by the MLC CP file, again for both leaf banks. The value of \( CS_{k,n} \) ranges from -1 to 1, where a value closer to unity indicates higher agreement and negative values indicate a reversal in leaf position.

[Diagram]

**Figure 3.3** Image synchronisation finds the closest match between measured and predicted EPID images based on the extracted MLC leaf positions.

Geometric synchronization cannot correctly identify the point in delivery of the image when multiple CPs (and hence predicted images) have identical MLC positions. Thus a dosimetry-based search is incorporated into the synchronization algorithm to make sure the synchronization matches the monotonically increasing delivered dose.

To reduce valuable computation time, the dose check points (DCPs) are calculated in advance from the predicted images by summing the predicted image greyscale values and saving them as a cumulative DCP function:

\[ DCP_n = \sum_{i=1}^{n} PD_n \]  

(3.2)
With $PD_n$ the sum of greyscale values from the $n^{th}$ predicted image. If the system encounters identical MLC positions in two or more CPs, it synchronises with the first of those control points along the DCP that has not been chosen before, hence making sure that the cumulative dose remains monotonically increasing.

In the first synchronization error analysis, seven sets of images were modelled using the same treatment plan (MLC file with 82 CPs). The only difference between these datasets was the number of frames interpolated from the original CPs, delivering snap shots at different intervals and frequencies, resulting in 9, 40, 60, 82, 100, 160, and 200 frames respectively. Each dataset was in turn interpreted by the system as either “predicted” or “measured” frames, until all combinations were analysed. The ratio of the number of frames (“measured”)/“predicted”) was set as the reference step function (stepped due to the discrete number of CPs).

For the second comparison strategy, direct comparisons between the gamma and CS methods were performed on five clinical IMRT fields imaged with a 20 cm water equivalent phantom (SolidWater 457 by Gammex RMI, Middleton, WI) in the beam. Reference images were modelled as snapshots at each CP, and the EPID images were acquired in cine-mode at 7.06 fps. The gamma criteria were set to 3%, 3 mm, and for each measured image the predicted image (corresponding to a CP) with the highest gamma pass fraction was found. The CP was then compared with the result delivered by the CS calculation.

### 3.5 An automatic MLC leaf positioning algorithm

The purpose of the MLC leaf position extraction is twofold: it is part of the synchronization process, and it forms the basis of the geometric verification. The algorithm used in this work can be divided into three components: edge detection, MLC template construction, and MLC position matching.

![Flowchart](image)

**Figure 3.4** Flowchart indicating the components and flow of information of the real-time automatic MLC positioning algorithm.
3.5.1 Leaf-end edge detection

There are several methods to detect stepped increases in intensity (‘edges’) in images, such as convolution with the Sobel and Laplace operators or the Canny edge detection method. Contour and morphological algorithms were the fastest, however, making them the most suitable candidates for real-time image analysis (Samant, Wei et al. 2000). Since MLC leaf motion is one-dimensional, we were able to construct a simple convolution mask for edge detection taking into consideration the minimization of processing time (see Figure 3.3a). The major drawback of this approach is that MLC positions can only be extracted reliably at a zero degree collimator angle. This is overcome by extracting the collimator angle from the DICOM image header and applying a rotational correction to the image or by applying the collimator angle detection algorithm (section 3.7.2). A global threshold of 38.5% of the maximum intensity is applied to reduce noise and to isolate the irradiated field. Figure 3.3b shows the image after convolution with the mask and the profile of the cross-plane, which shows the two peaks representing the left and the right edge of the MLC defined field.

![Convolution Mask 2X3](image)

(a) Convolution Mask 2X3

![Cross-plane direction](image)

(b) Cross-plane direction of image after leaf-end edge detection processing.

A 50% intensity level for global threshold was recommended by many publications (Bijhold, Gilhuijs et al. 1991; Partridge, Evans et al. 1998; James, Atherton et al. 2000; Woo, Lightstone et al. 2003; Lee, Mao et al. 2008). The 50% intensity level performed well either in the integrated image of a-Si EPID or real-time image from a CCD-camera based EPID. To develop a suitable of global threshold for cine-mode EPID images from the aS1000 EPID (Varian), an investigation of leaf edge detection was performed. The open-close leaf pattern, or Zebra pattern, was imported into the linac, using 6 MV and 100 MU/min for the beam delivery. We captured images using cine-mode and investigated the leaf-edge greyscale value compared with the maximum greyscale value of the image. Figures 3.6a and 3.6b show the captured cine-
mode images and profile at column 256. Over the entire experiment, we found that the edge of the leaf corresponds to an approximate 38.5% intensity level. Therefore, the 38.5% intensity level was used as the global threshold in the automatic MLC leaf positioning algorithm.

![Figure 3.6](image)

**Figure 3.6** Investigation of MLC leaf end edge greyscale value using a Zebra pattern (open-close leaf): (a) one of cine images captured by aS1000 EPID and (b) vertical profile at x= 256 pixels.

### 3.5.2 MLC leaf template construction

In order to reconstruct the MLC leaf template from acquired EPID data, the imaging geometry needs to be well defined. Figures 3.7a and 3.7b show the system setup and the definitions of the source-to-MLC distance (SMD), source-to-axis distance (SAD) and the source-to-imager distance (SID). Pixel spacing calibration is necessary to account for varying SDDs, hence a process using similar triangles is used to derive the pixel spacing at any plane perpendicular to the central axis (Lee, Mao et al. 2008; Kang, Deng et al. 2009).
Figure 3.7 Diagram showing the projection of MLC leafs on the imaging plane (a) and the geometrical parameters used for the calculation of pixel spacing. (b) Source-to-MLC distance(SMD), Source-to-axis distance(SAD), and Source-to-imager distance (SID).

Consider an X-ray beam originating from a point source passing consecutively through the MLC, isocenter, and EPID planes at an angle to the central axis. As can be seen in Figure 3.7a, the respective distances from the central axis are denoted a, b, and c, respectively. Assuming the planes are perfectly perpendicular to the central beam (Lee, Mao et al. 2008) and pixel spacing at a particular SID can be calculated from similar triangles:

\[ c' = \frac{SID}{SAD} \cdot b' = \frac{SID}{SMD} \cdot a' \]  

(3.3)

Assuming that c’ is the pixel spacing in the imaging plane (i.e. measured EPID pixel spacing), a’ and b’ become the pixel spacings in the MLC and isocentral planes, respectively. With knowledge of the spatial ratios, the MLC template can now be constructed and superimposed on the measured EPID (James, Atherton et al. 2000; Xu, Chuang et al. 2002).

3.5.3 MLC position matching

The final phase of the automatic MLC leaf positioning algorithm consists of finding the best fit of the leaf template to the leaf-end edge (active template matching algorithm) (Xu, Chuang et al. 2002). With knowledge of the leaf edge positions, each leaf in the MLC template is set to an initial distance from the central axis. Since the MLC is divided into two banks of leafs (right and left), the leaf edge detection algorithm should deliver two edges, which the
differential nature of the convolution mask delivers as a positive and a negative signal. For each leaf row, the left bank edge is then positioned to the corresponding location of the maximum positive value of the convolved function, while the minimum intensity represents the right bank leaf edge. The final MLC leaf positions can now be calculated using the geometry outlined.

**Figure 3.8** Matching a template to the best-fitting leaf edge.

For an automatic MLC positioning algorithm experiment, a diamond shaped MLC pattern were used (Xu et al. 2002). The images were acquired in cine-mode with 6 MV, 150 cm SDD, and 300 MU/min and 100 MU. The system acquired 74 images, which were compared the leaf position in the plan or MLC file. RMSE and average error ± 1SD were determined.

**Figure 3.9** Example of automatic MLC positioning: (a) original EPID image, (b) leaf positions for corresponding leaf numbers (blue: BANK A [left] and red: BANK B [right]).
3.6 Real-time dosimetry using EPID

Figure 3.10 System workflow of the real-time EPID dosimetry system, which is divided into four main steps: input, synchronisation, verification, and monitoring.

Figure 3.10 shows the system components and workflow of the real-time EPID dosimetry system. The images are acquired in cine-mode, and the current MLC leaf position is automatically extracted (see section 3.5). By calculating the cosine similarity (Fraass, Lash et al.; Steinbach, Karypis et al. 2000) of the MLC geometry between measured and predicted images, the measured image can be synchronised to the nearest predicted MLC position. Applying the same leaf positioning algorithm to both the measured and the predicted image results should cancel out any systematic errors. However, this approach may produce non-unique results since it is possible that different control points (CP) along the treatment delivery have identical MLC leaf positions. Thus, it is essential to incorporate a comparison with the monotonically increasing dose as part of the synchronization process. Once the measured image is synchronized to the predicted image stream, the dosimetric verification function performs a gamma comparison. In the final step, the system monitors the result of the dose comparison in terms of both cumulative and individual (frame-by-frame) gamma <1 pass rate results. Consequently, this system is able to detect dose errors that may occur during treatment in real-time, based on a user defined pass rate threshold. A possible user interface of the system is shown in Figure 3.11.
The performance of the real-time EPID dosimetry system was evaluated by comparing EPID image data with the predicted images using the gamma comparison. We examined 19 clinical IMRT fields (7 prostate, 8 head and neck, and 4 brain) at a source-to-detector distance (SDD) of 150 cm with a 20 cm water equivalent phantom (SolidWater 457 by Gammex RMI, Middleton, WI) in the beam by comparing individual measured frames acquired in cine mode with synchronized predicted images using the gamma comparison (3%, 3mm and 4%, 4mm). The verification process can be divided into frame-by-frame (individual) comparison and cumulative comparison. The individual verification process is more reactive to small differences between the reference and measured images and is mainly (but not solely) used for geometric verification. The cumulative dose comparison is slower to show any deviations from the treatment plan and is used mainly for dosimetric verification since it contains a history of the delivered dose. The influence of the ratio of the number of predicted to acquired images on the synchronisation and results was investigated for a wide range of cases. For additional robustness, each measured frame was also compared with the predicted images immediately adjacent to the synchronized predicted image, yielding three reference images for comparison.

**Figure 3.11** System user-interface consisting of predicted and measured image viewers, gamma-map viewer and pass-rate display (%) for both real-time individual and cumulative dose comparisons.
3.7 Real-time geometry using EPID

The real-time geometry system aims to automatically monitor and verify three geometrical aspects, including jaw position, collimator angle, and MLC position. In IMRT treatment, jaw position and collimator angle remain in a static position over the entire treatment, but MLC leaf is dynamic. The system was designed to verify the jaw and collimator angle in the first 3 seconds of the treatment, and a warning action will be performed if the system detects an error as judged by the threshold (2 mm for jaw, 1 degree error for collimator angle).

![Figure 3.13 System workflow for real-time geometry verification using EPID.](image)

Figure 3.13 System workflow for real-time geometry verification using EPID, detecting jaw position, collimator angle, and MLC position automatically. Jaw and collimator angle are verified based on TPS settings and the tolerance level, but MLC position verification is compared to predicted EPID images (which are in turn derived from planned MLC positions).
The automatic MLC positioning is described in section 3.5, however the result of collimator angle detection is used to rotate the EPID image to 0 degrees, due to the algorithm limitation. Automatic jaw positioning applies the fuzzy c-mean clustering technique for segmentation prior to detection (see Section 3.7.1). Collimator angle detection is identified by using the filled-by-mean (FBM) technique to suppress the unnecessary data in the image, and Fourier transform (FFT) is used to identify the collimator angle (see Section 3.7.2). The tolerance is defined by the user; for example, 2 mm for jaw position error, 1 degree for collimator angle error, and 0.9 maximum cosine similarity value for the MLC position measurement (described in Section 3.7.3).

3.7.1 Automatic jaw positioning

Finding the jaw positioning on a cine-EPID image in either IMRT or VMAT is difficult due to the low contrast of the jaw edge which is usually obscured by the MLC and noise from scatter. There are three zones of the cine-EPID image: 1) the irradiation field, 2) the shielded area by the MLC leaves, and 3) the shielded area by the solid jaws (see Figure 3.14).

Figure 3.14 A typical EPID image acquired in cine mode (example of 2nd frame of head and neck treatment case) and the three different zones showing the irradiation field, the region shielded by MLC leaves that is the region of interleaf leakage, and the shielded area by the solid jaws, divided into 4 segments.
The procedure of automatic jaw detection and positioning requires two main steps, segmentation and localisation. Fuzzy C-mean clustering (FCM) is an unsupervised technique focusing on feature analysis and classifier design for medical imaging and recognition. However, using FCM has the limitation of high calculation time. In this project, we tested 14 treatment cases (7 prostate cases, and 7 head and neck cases) by the FCM algorithm to find the appropriate percentage intensity level for a global threshold that localises the area shielded by the jaws.

3.7.1.1 Investigation of area shielded by jaws using fuzzy c-means clustering

Fuzzy c-means clustering was first introduced by Duda et al. and was modified by Bezdek et al. (Duda and Hart 1973; Bezdek, Ehrlich et al. 1984). The FCM algorithm uses fuzzy memberships to assign each category. In terms of EPID image (size 768X1024), the matrix is first reshaped into $X = (x_1, x_2, x_3, ..., x_N)$ where $x_i$ represents multispectral data. The algorithm is an iterative optimization that minimized the cost function, defined as $J$

$$J = \sum_{j=1}^{N} \sum_{i=1}^{c} u_{ij}^m \|x_j - v_i\|^2$$  \hspace{1cm} (3.4)

where $u_{ij}$ represents the membership of pixel $x_j$ in the i-th cluster, and the membership function is

$$u_{ij} = \frac{1}{\sum_{k=1}^{c} \left( \frac{\|x_j - v_i\|}{\|x_j - v_k\|} \right)^{2/(m-2)}}$$  \hspace{1cm} (3.5)

where $v_i$ is the i-th cluster center by the following:

$$v_i = \frac{\sum_{j=1}^{N} u_{ij}^m x_j}{\sum_{j=1}^{N} u_{ij}^m}$$  \hspace{1cm} (3.6)

Where $m$ is a constant and $m=2$ is used in this study.

When the pixels are close to the centroid of their cluster, the cost function ($J$) assigns high membership values. The probability of a pixel belonging to a specific cluster is presented in the form of the membership function ($u$). In the FCM algorithm, the probability depends on the distance between the pixel and each individual cluster center. In regards to the cine-EPID image, the cluster is defined into 5 clusters that are shown in Figure 3.15. The level 1 is used to determine the proper percentage intensity level for the global threshold. We investigated cine-EPID images with 14 treatment cases with 6 MV, SSD 150 cm and 20 cm phantom thickness water equivalent. Table 3.1 shows the resulting percentage intensity level by picking up level 1 from 5 clusters.
Figure 3.15 Testing the cine-EPID image with the FCM algorithm is defined in 5 clusters, shield by jaw, noise between shield by jaw and MLC leaves, shield by MLC leaves, noise between shield by MLC leaves and irradiation field, and irradiation field.

Table 3.1 The results of segmentation using the FCM algorithm defined 5 clusters and segments with level 1—14 treatment cases, including 7 prostate cases and 7 head and neck cases were tested.

<table>
<thead>
<tr>
<th>Field number</th>
<th>% intensity level</th>
<th>Field number</th>
<th>% intensity level</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (prostate)</td>
<td>1.69</td>
<td>8 (H&amp;N)</td>
<td>2.29</td>
</tr>
<tr>
<td>2 (prostate)</td>
<td>2.12</td>
<td>9 (H&amp;N)</td>
<td>2.29</td>
</tr>
<tr>
<td>3 (prostate)</td>
<td>2.37</td>
<td>10 (H&amp;N)</td>
<td>2.30</td>
</tr>
<tr>
<td>4 (prostate)</td>
<td>2.14</td>
<td>11 (H&amp;N)</td>
<td>2.48</td>
</tr>
<tr>
<td>5 (prostate)</td>
<td>2.26</td>
<td>12 (H&amp;N)</td>
<td>2.95</td>
</tr>
<tr>
<td>6 (prostate)</td>
<td>1.81</td>
<td>13 (H&amp;N)</td>
<td>3.44</td>
</tr>
<tr>
<td>7 (prostate)</td>
<td>2.37</td>
<td>14 (H&amp;N)</td>
<td>2.31</td>
</tr>
<tr>
<td>Average prostate cases</td>
<td>2.11</td>
<td>Average H&amp;N cases</td>
<td>2.58</td>
</tr>
</tbody>
</table>

As our investigation showed in Table 3.1, we found 2.34% intensity level is an appropriate global threshold to segment the cine-EPID image for the purpose of jaw detection. It should be note that 2.34% intensity not fit every images. The threshold level depends on filed shape monitor unit, type of EPID, and treatment types. However, we selected the average intensity level based on 14 IMRT cases. If the lowest intensity level was applied, it introduced more noise.
Figure 3.16 An example of using 2.34% intensity level threshold. (a and b) Raw image of head and neck and prostate, and (c and d) the result of thresholding respectively.

3.7.1.2 Jaw segmentation and localisation

Figure 3.17 The process of Jaw segmentation and localisation.

Figure 3.17 presents the process of jaw positioning determination, including segmentation and localisation steps. As we have investigated, the 2.34% intensity level is used for the global threshold to segment the area that is shielded by the jaws. Directly applying the
FCM algorithm for segmentation has the drawback of slower calculation speed. Time calculation becomes vital in a real-time geometry verification perspective. Moreover, using a connected component filter for noise reduction can improve the segmented image. After that, horizontal and vertical projection profiles are acquired (see Figure 3.18c and 3.18d). The horizontal projection is used to identify X1 and X2 position, and Y1 and Y2 is located from the vertical projection. The searching condition for individual Jaw is constructed which shows in table 3.2.

Table 3.2 The searching algorithm for jaw localisation. The EPID image size is MXN.

<table>
<thead>
<tr>
<th>JAW</th>
<th>Searching Condition</th>
</tr>
</thead>
</table>
| X1  | \( i\_count = 1; \)  
     | \( \text{WHILE} \ i\_count < M+1 \)  
     | \( \text{IF} \ ( \text{pixelValue}(i\_count) > 0.99*\text{maximum} ) \)  
     | \( \text{THEN} \ X1\_position = i\_count-1; \)  
     | \( \text{ENDIF} \ i\_count = i\_count + 1; \)  
     | \( \text{ENDWHILE} \)  |
| X2  | \( i\_count = M; \)  
     | \( \text{WHILE} \ i\_count > 0 \)  
     | \( \text{IF} \ ( \text{pixelValue}(i\_count) > 0.5*\text{maximum} ) \)  
     | \( \text{THEN} \ X2\_position = i\_count+1; \)  
     | \( \text{ENDIF} \ i\_count = i\_count - 1; \)  
     | \( \text{ENDWHILE} \)  |
| Y1  | \( i\_count = N; \)  
     | \( \text{WHILE} \ i\_count > 0 \)  
     | \( \text{IF} \ ( \text{pixelValue}(i\_count) > 0.1*\text{maximum} ) \)  
     | \( \text{THEN} \ Y1\_position = i\_count+1; \)  
     | \( \text{ENDIF} \ i\_count = i\_count - 1; \)  
     | \( \text{ENDWHILE} \)  |
| Y2  | \( i\_count = 1; \)  
     | \( \text{WHILE} \ i\_count > N+1 \)  
     | \( \text{IF} \ ( \text{pixelValue}(i\_count) > 0.1*\text{maximum} ) \)  
     | \( \text{THEN} \ Y2\_position = i\_count - 1; \)  
     | \( \text{ENDIF} \ i\_count = i\_count + 1; \)  
     | \( \text{ENDWHILE} \)  |

Based on observation of IMRT treatment, the irradiation field normally is located on the left side and is moving to the right side during the treatment. This automatic Jaw localisation method aims to apply the first few frames of cine EPID images. In figure 3.18a shows the result of image thresholding that contains high noise on the left side of the image. After horizontal projection (see figure 3.18c), X1 and X2 position can be evaluated by defining the edge threshold 99% and 50% of maximum value respectively. To identify the Y1 and Y2 position, the vertical projection is used. In this method, since the image centre (pixel number 512) though the pixel number 1024 is used for vertical projection (see figure 3.18d). Concerning half of
image can eliminate image noise caused by the image scatter. 10% threshold is applied to evaluate Y1 and Y2 position.

![Image](image1.png)

(a)                                                                 (b)

![Image](image2.png)

(c)

![Image](image3.png)

(d)

**Figure 3.18** An example of the output of each step: (a) threshold, (b) filter, (c) horizontal projection plot, and (d) vertical projection plot.

In terms of automatic Jaw positioning experiments, 7 prostate cases and 7 head and neck cases of IMRT treatment were test preforming through the 20 cm solid phantom. X1, X2, Y1 and Y2 position were investigated for each case and to compare against the plan.
3.7.2 Automatic collimator angle detection

The collimator angle detection procedure consists of three steps: pre-image processing, 2D DFT function, and angle measurement.

The automatic collimator angle detection contains three main steps: pre-image processing, Discrete Fourier Transform (DFT), and angle measurement. The pre-image processing improves the quality of the EPID image and eliminates non-essential information. A Gaussian filter is applied to reduce the noise in the image (see Figure 3.20b). After that, the image is cropped into a square image of size NXN. The DFT is normally used to decompose the EPID image into geometric components, such as sine and cosine. The 2D DFT is calculated by:

\[
F(k, l) = \sum_{i=0}^{N-1} \sum_{j=0}^{N-1} f(i, j) e^{-i2\pi \frac{(ki +lj)}{N}}
\]  

(3.7)

Where \(f(i, j)\) is the EPID image, and the exponential term is the basis function corresponding to each point \(F(k, l)\) in the frequency domain. An example output of the 2D DFT is shown in figure 3.20c which is shifted the zero-frequency component to the centre.

After the 2D DFT is applied, the circle mask with a radius of 37 pixels is implemented to subtract the image (see figure 3.20d). The maximum magnitude of image after masking is identified as vector u and v, shown in figure 3.20e, can be measured the angle which represents the collimator angle. The example of each step is shown in figure 3.20.
Figure 3.20 Example of the output of collimator angle detection: (a) the result after 10% intensity level threshold, (b) Gaussian filter, (c) 2D DFT functions, (d) thresholding with circle mask radius of 37 pixels, and (e) two magnitudes representing vector $u$ and $v$, which are used to calculate the collimator angle.
3.7.3 MLC position verification

Another advantage of the synchronisation method developed for the real-time dosimetry system is that it can be used for MLC position verification. The cosine similarity (Fraass, Lash et al.) is a tool used to measure the similarity between measured images and predicted images. In fact, the synchronisation between measured and predicted images is close to linear and goes “forwards.” These facts can be set as conditions as MLC position verification for assessing MLC leaf motion behaviour. In figure 3.21a shows the result of synchronisation between control point number and EPID image number without error in a prostate IMRT treatment case. However, in figure 3.21b is an example of MLC leaf position error since EPID image frame 101, the relationship of synchronisation does not follow the rule. The system is set to alarm when the relationship of synchronisation does not go linear and “forwards”.

![Synchronisation without geometric error](image)

(a)

![Synchronisation with introducing geometric error](image)

(b)

**Figure 3.21** An example of using synchronisation results to detect the geometric error; (a) Normal case without geometric error, and (b) introducing geometric error from control point number 47 by stopping the leaves from moving, displayed in red background.
4.1. Physics-based prediction model (Chytyk and McCurdy 2009)

The benchmarking of the physics-based prediction model used for reference image calculation is presented in this section. Integrated predicted dose images of prostate and head and neck are compared with the measured EPID images using gamma evaluation (2%, 2mm) and are given in Table 4.1. The results show good agreement between the integrated predicted image and the integrated EPID image, averaging 99.2% of points with gamma <1. A sample dose comparison between the predicted image and the EPID image of a prostate IMRT field with a 20 cm phantom in the beam is presented in Figure 4.1.

**Table 4.1** Gamma evaluation results comparing the relative dose images of the EPID and predicted image, generated from the physics-based prediction model for prostate and head and neck fields.

<table>
<thead>
<tr>
<th>Field</th>
<th>% points meeting $\gamma$ (2%,2mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (prostate)</td>
<td>99.4</td>
</tr>
<tr>
<td>2 (prostate)</td>
<td>99.3</td>
</tr>
<tr>
<td>3 (prostate)</td>
<td>99.4</td>
</tr>
<tr>
<td>4 (prostate)</td>
<td>98.6</td>
</tr>
<tr>
<td>5 (prostate)</td>
<td>99.6</td>
</tr>
<tr>
<td>6 (prostate)</td>
<td>97.7</td>
</tr>
<tr>
<td>7 (prostate)</td>
<td>99.3</td>
</tr>
<tr>
<td>8 (head and neck)</td>
<td>99.7</td>
</tr>
<tr>
<td>9 (head and neck)</td>
<td>98.6</td>
</tr>
<tr>
<td>10 (head and neck)</td>
<td>98.8</td>
</tr>
<tr>
<td>11 (head and neck)</td>
<td>99.9</td>
</tr>
<tr>
<td>12 (head and neck)</td>
<td>99.8</td>
</tr>
<tr>
<td>13 (head and neck)</td>
<td>98.9</td>
</tr>
<tr>
<td>14 (head and neck)</td>
<td>99.3</td>
</tr>
<tr>
<td>Mean</td>
<td>99.2</td>
</tr>
</tbody>
</table>
Figure 4.1 Comparison of the dose between the predicted image and the EPID image for a prostate IMRT field (field number 1) with 20 cm phantom and 6 MV beam: (a) Predicted dose image, (b) EPID dose image, (c) cross-plane profile, (d) in-plane profile, and (e) the gamma map (2%,2mm).


4.2 Image synchronisation

Table 4.2 shows the results of the testing performed with varying ratios of predicted to measured images. The table reports the root mean square error (RMSE) and the average synchronization error for the measured frames. The average RMSE over all scenarios is 0.769 ±1.497 control points; the average mean over all scenarios was 0.439±0.969 control points. The average dose error was calculated from the percentage dose error corresponding to the CP synchronization error and was 0.548±0.777%. The large number of measured images and predicted images results in very low CP synchronisation error.

<table>
<thead>
<tr>
<th>Number of “predicted” images</th>
<th>Number of “measured” images</th>
</tr>
</thead>
<tbody>
<tr>
<td>9</td>
<td>40</td>
</tr>
<tr>
<td>RMSE-error</td>
<td>0.000</td>
</tr>
<tr>
<td>Mean – error</td>
<td>0.000</td>
</tr>
<tr>
<td>Dose-error (%)</td>
<td>0.000</td>
</tr>
<tr>
<td></td>
<td>60</td>
</tr>
<tr>
<td>RMSE-error</td>
<td>1.247</td>
</tr>
<tr>
<td>Mean – error</td>
<td>0.889</td>
</tr>
<tr>
<td>Dose-error (%)</td>
<td>2.222</td>
</tr>
<tr>
<td></td>
<td>82</td>
</tr>
<tr>
<td>RMSE-error</td>
<td>2.211</td>
</tr>
<tr>
<td>Mean – error</td>
<td>1.333</td>
</tr>
<tr>
<td>Dose-error (%)</td>
<td>2.222</td>
</tr>
<tr>
<td></td>
<td>100</td>
</tr>
<tr>
<td>RMSE-error</td>
<td>2.944</td>
</tr>
<tr>
<td>Mean – error</td>
<td>1.778</td>
</tr>
<tr>
<td>Dose-error (%)</td>
<td>2.168</td>
</tr>
<tr>
<td></td>
<td>160</td>
</tr>
<tr>
<td>RMSE-error</td>
<td>6.289</td>
</tr>
<tr>
<td>Mean – error</td>
<td>4.000</td>
</tr>
<tr>
<td>Dose-error (%)</td>
<td>2.500</td>
</tr>
<tr>
<td></td>
<td>200</td>
</tr>
<tr>
<td>RMSE-error</td>
<td>7.812</td>
</tr>
<tr>
<td>Mean – error</td>
<td>5.000</td>
</tr>
<tr>
<td>Dose-error (%)</td>
<td>2.500</td>
</tr>
</tbody>
</table>
Figure 4.2 A sample result of synchronisation using the cosine similarity index compared with ideal synchronisation (linear curve). There are 190 EPID images synchronised with 106 control points.

For one of the prostate IMRT treatments, 106 control points images were predicted, and 190 EPID images were acquired during treatment in an environment using a 20 cm phantom and 6 MV. The result of synchronisation using the cosine similarity technique compared with ideal synchronisation, which is plotted as the linear curve, is shown in Figure 4.2. Results of the comparison of CS and gamma based synchronisation are given in Table 4.3, with the errors shown as fractional CPs in terms of root mean square error (RMSE) and mean errors ± 1SD.

Table 4.3 The control point errors of synchronization testing comparing CS and gamma-based synchronisation (3%, 3 mm criteria) in 5 different prostate IMRT sliding window fields.

<table>
<thead>
<tr>
<th>Field no.</th>
<th>#control point/ #EPID images</th>
<th>Control point errors</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>RMSE</td>
<td>Mean ± 1SD</td>
</tr>
<tr>
<td>1</td>
<td>107/150</td>
<td>2.016</td>
</tr>
<tr>
<td>2</td>
<td>83/134</td>
<td>1.943</td>
</tr>
<tr>
<td>3</td>
<td>82/110</td>
<td>1.358</td>
</tr>
<tr>
<td>4</td>
<td>99/114</td>
<td>1.295</td>
</tr>
<tr>
<td>5</td>
<td>82/117</td>
<td>1.850</td>
</tr>
<tr>
<td>Average</td>
<td></td>
<td>1.693</td>
</tr>
</tbody>
</table>
Figure 4.3 A sample result of synchronisation using the cosine similarity index compared with gamma-based synchronisation (2%, 2mm criteria). There are 190 EPID images synchronised with 106 control points.

Figure 4.3 presents the comparison of synchronisation using cosine similarity and gamma-based synchronisation. Gamma-based synchronisation compares every single EPID frame with the set of predicted images. The highest percent fraction less than 1 is chosen as the corresponded control point number to the EPID image.

Table 4.4 Calculation time comparison of cosine similarity and gamma-based synchronisation using an Intel Core i5-2400 3.10GHz CPU, 8 GB RAM, 64-bit operating system programmed by MATLAB, and 106 control points.

<table>
<thead>
<tr>
<th>Synchronisation</th>
<th>Time per EPID image in seconds</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cosine similarity</td>
<td>0.03±0.02</td>
</tr>
<tr>
<td>gamma-based</td>
<td>6.87±0.40</td>
</tr>
</tbody>
</table>

The comparison in synchronisation computation time between cosine similarity and Gamma-based synchronisation is presented in Table 4.4. Gamma-based synchronisation cannot be used in a real-time system environment due to time consumption of 6.87±0.40 seconds per EPID image. On the other hand, the proposed method, CS, computes 0.03±0.02 seconds per EPID image, which meets the real-time execution gap requirement (0.149 second per frame with 6.67 frame rate).
4.3 Dose comparison

Table 4.5 shows the average results of the gamma pass rate (percentage) for the patient IMRT deliveries performed through the solid water phantom. Note that each value represents the mean and standard deviation of the gamma comparison results for all the acquired frames during the delivery. Therefore, for case 1, the results are the mean of 190 gamma values.

<table>
<thead>
<tr>
<th>Case</th>
<th>#EPID image/ #CP</th>
<th>Individual comparison (%)</th>
<th>Cumulative comparison (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>3%, 3mm</td>
<td>4%, 4mm</td>
</tr>
<tr>
<td>1 (Prostate)</td>
<td>190/106</td>
<td>82.5±7.8</td>
<td>91.9±6.6</td>
</tr>
<tr>
<td>2 (Prostate)</td>
<td>169/82</td>
<td>81.2±11.0</td>
<td>90.2±10.8</td>
</tr>
<tr>
<td>3 (Prostate)</td>
<td>140/81</td>
<td>87.0±7.8</td>
<td>96.1±4.3</td>
</tr>
<tr>
<td>4 (Prostate)</td>
<td>157/89</td>
<td>81.9±10.7</td>
<td>91.0±10.7</td>
</tr>
<tr>
<td>5 (Prostate)</td>
<td>180/82</td>
<td>91.2±7.7</td>
<td>97.1±4.1</td>
</tr>
<tr>
<td>6 (Prostate)</td>
<td>143/98</td>
<td>73.1±8.1</td>
<td>86.4±7.4</td>
</tr>
<tr>
<td>7 (Prostate)</td>
<td>148/81</td>
<td>87.6±9.0</td>
<td>95.8±4.5</td>
</tr>
<tr>
<td>8 (H&amp;N)</td>
<td>113/98</td>
<td>88.3±6.1</td>
<td>96.5±4.3</td>
</tr>
<tr>
<td>9 (H&amp;N)</td>
<td>115/105</td>
<td>75.6±10.9</td>
<td>85.3±9.2</td>
</tr>
<tr>
<td>10 (H&amp;N)</td>
<td>142/153</td>
<td>88.9±8.0</td>
<td>95.2±6.7</td>
</tr>
<tr>
<td>11 (H&amp;N)</td>
<td>107/106</td>
<td>87.0±7.0</td>
<td>91.8±6.5</td>
</tr>
<tr>
<td>12 (H&amp;N)</td>
<td>113/98</td>
<td>80.1±16.8</td>
<td>87.3±17.1</td>
</tr>
<tr>
<td>13 (H&amp;N)</td>
<td>112/128</td>
<td>84.8±6.7</td>
<td>91.3±6.1</td>
</tr>
<tr>
<td>14 (H&amp;N)</td>
<td>96/97</td>
<td>89.2±5.1</td>
<td>96.7±3.3</td>
</tr>
<tr>
<td>15 (H&amp;N)</td>
<td>112/126</td>
<td>89.0±7.6</td>
<td>95.4±6.1</td>
</tr>
<tr>
<td>16 (Brain)</td>
<td>143/109</td>
<td>83.4±8.2</td>
<td>87.3±7.4</td>
</tr>
<tr>
<td>17 (Brain)</td>
<td>133/118</td>
<td>89.2±9.4</td>
<td>91.9±9.1</td>
</tr>
<tr>
<td>18 (Brain)</td>
<td>130/127</td>
<td>87.6±8.4</td>
<td>90.9±7.7</td>
</tr>
<tr>
<td>19 (Brain)</td>
<td>135/112</td>
<td>89.0±15.1</td>
<td>90.9±14.4</td>
</tr>
<tr>
<td>Average</td>
<td></td>
<td>85.0±9.0</td>
<td>92.0±7.7</td>
</tr>
</tbody>
</table>
4.4 Real-time error detection using real-time dosimetry

4.3.1 Case study I: Dose error

In this case study, the delivered dose was doubled just before the halfway point (at dose fraction = 0.4) of an IMRT prostate treatment. The individual frame dose comparison showed an immediate drop in the pass rate fraction, whereas the cumulative dose comparison is not as responsive and decreases slowly after the error is introduced, as shown in Figure 4.4.

Figure 4.4 Result of error detection in terms of introduced dose error during treatment delivery (a) maximum dose EPID response, (b) individual dose comparison, and (c) cumulative dose comparison.
4.3.1 Case study II: MLC leaf error

Computer malfunctions can cause the MLC leaves to behave erratically. This study simulates a case where a computer crash occurred while the medical physicist was saving a reformulated treatment plan. The instructions for the MLC were lost, and the leaves were fully opened. This was simulated by retracting the MLC leaves at dose fraction 0.26, where they remained until the end of treatment. Figure 4.5 shows the system quickly detecting this error. The jaw positions remained the same (9 cm x 6.6 cm field).

![Graph showing error detection and cumulative dose comparison]

Figure 4.5 Result of error detection for an introduced MLC leaf position error during treatment, which simulated the real situation.

The threshold criteria for a detected error can be set by the user, and in these cases was set to 60% and 50% pass-rate for cumulative and individual dose comparison respectively. In case study I, where the dose error was introduced at dose fraction 0.4 (EPID frame number 80), the system detected an error in the individual dose comparison at image number 81.
(approximately 0.1 seconds delay. However, the cumulative dose comparison lagged, and the dose error was detected at EPID frame number 108 (approximately 4.2 seconds delay). In case study II, where the MLC leaf position error was introduced at dose fraction 0.26 (EPID frame number 48), the system detected an error at frame 49 (approximately 0.1 seconds delay) using the individual frame comparison and frame 57 (approximately 1.3 seconds delay) using the cumulative frame comparison.

4.5 Automatic MLC positioning

The accuracy of the MLC positioning algorithm had been tested with the diamond shaped MLC pattern shown in Figure 4.6. The MLC file in Figure 4.6a was generated by the Shaper software (Varian). The images were acquired in cine-mode with 6MV, 150 cm SDD, and 300 MU/min and 100 MU. The system acquired 74 EPID images during this test. To compare the results between MLC leaf positions produced by shaper and MLC leaf position from the automatic MLC positioning algorithm, RMSE of bank A is 0.012 cm with RMSE of bank B at 0.015 cm. In addition, the average error of bank A and B are $0.049 \pm 0.018$ and $0.055 \pm 0.015$ cm respectively.

![Figure 4.6](image)

**Figure 4.6** Automatic MLC leaf positioning performance testing: (a) diamond shape generated by Shaper software, (b) EPID image in cine-mode, and (c) the result of MLC positioning.
4.6 Automatic jaw positioning

Table 4.6 shows the result of automatic jaw positioning for the patient IMRT deliveries performed through the 20 cm solid water phantom. 7 prostate cases and 7 head and neck cases of IMRT treatment were tested. The EPID images were acquired with cine-mode via the IAS3 software provided by Varian. The 2\textsuperscript{nd} frame acquired during treatment was used for detecting and positioning. In the experiment, the maximum error was 0.276 cm, minimum error was 0.0073 cm, and the average error was $0.0751 \pm 0.0603$ cm. In addition, the specific jaw deviation were investigated, including $X_1$, $X_2$, $Y_1$, $Y_2$ in different treatment cases. For prostate cases, the average deviation of $X_1$, $X_2$, $Y_1$, and $Y_2$ were $0.069 \pm 0.314$, $0.085 \pm 0.080$, $0.055 \pm 0.056$, and $0.079 \pm 0.065$ cm respectively. For head and neck cases, the average deviation of $X_1$, $X_2$, $Y_1$ and $Y_2$ were $0.085 \pm 0.055$, $0.047 \pm 0.026$, $0.074 \pm 0.032$, and $0.106 \pm 0.108$ cm respectively.
Table 4.6 The results of automatic jaw positioning (14 IMRT treatment cases).

<table>
<thead>
<tr>
<th>case</th>
<th>jaw</th>
<th>Plan (cm)</th>
<th>Measured (cm)</th>
<th>Error ((\Delta)) cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (prostate)</td>
<td>X1</td>
<td>4.5</td>
<td>4.4109</td>
<td>0.0891</td>
</tr>
<tr>
<td>1 (prostate)</td>
<td>X2</td>
<td>4.5</td>
<td>4.2543</td>
<td>0.2457</td>
</tr>
<tr>
<td>1 (prostate)</td>
<td>Y1</td>
<td>3.3</td>
<td>3.1320</td>
<td>0.1680</td>
</tr>
<tr>
<td>1 (prostate)</td>
<td>Y2</td>
<td>3.3</td>
<td>3.2886</td>
<td>0.0114</td>
</tr>
<tr>
<td>2 (prostate)</td>
<td>X1</td>
<td>4.5</td>
<td>4.4631</td>
<td>0.0369</td>
</tr>
<tr>
<td>2 (prostate)</td>
<td>X2</td>
<td>4.5</td>
<td>4.4631</td>
<td>0.0369</td>
</tr>
<tr>
<td>2 (prostate)</td>
<td>Y1</td>
<td>3.3</td>
<td>3.3147</td>
<td>0.0147</td>
</tr>
<tr>
<td>2 (prostate)</td>
<td>Y2</td>
<td>3.0</td>
<td>3.0798</td>
<td>0.0798</td>
</tr>
<tr>
<td>3 (prostate)</td>
<td>X1</td>
<td>4.5</td>
<td>4.6197</td>
<td>0.1197</td>
</tr>
<tr>
<td>3 (prostate)</td>
<td>X2</td>
<td>4.5</td>
<td>4.4631</td>
<td>0.0369</td>
</tr>
<tr>
<td>3 (prostate)</td>
<td>Y1</td>
<td>2.8</td>
<td>2.7927</td>
<td>0.0073</td>
</tr>
<tr>
<td>3 (prostate)</td>
<td>Y2</td>
<td>2.8</td>
<td>3.0015</td>
<td>0.2015</td>
</tr>
<tr>
<td>4 (prostate)</td>
<td>X1</td>
<td>4.5</td>
<td>4.5414</td>
<td>0.0414</td>
</tr>
<tr>
<td>4 (prostate)</td>
<td>X2</td>
<td>4.5</td>
<td>4.4370</td>
<td>0.063</td>
</tr>
<tr>
<td>4 (prostate)</td>
<td>Y1</td>
<td>3.3</td>
<td>3.2625</td>
<td>0.0375</td>
</tr>
<tr>
<td>4 (prostate)</td>
<td>Y2</td>
<td>3.0</td>
<td>3.0537</td>
<td>0.0537</td>
</tr>
<tr>
<td>5 (prostate)</td>
<td>X1</td>
<td>4.5</td>
<td>4.5414</td>
<td>0.0414</td>
</tr>
<tr>
<td>5 (prostate)</td>
<td>X2</td>
<td>4.5</td>
<td>4.4631</td>
<td>0.0369</td>
</tr>
<tr>
<td>5 (prostate)</td>
<td>Y1</td>
<td>3.0</td>
<td>3.0537</td>
<td>0.0537</td>
</tr>
<tr>
<td>5 (prostate)</td>
<td>Y2</td>
<td>3.3</td>
<td>3.3147</td>
<td>0.0147</td>
</tr>
<tr>
<td>6 (prostate)</td>
<td>X1</td>
<td>4.5</td>
<td>4.5675</td>
<td>0.0675</td>
</tr>
<tr>
<td>6 (prostate)</td>
<td>X2</td>
<td>4.5</td>
<td>4.4631</td>
<td>0.0369</td>
</tr>
<tr>
<td>6 (prostate)</td>
<td>Y1</td>
<td>2.8</td>
<td>2.8188</td>
<td>0.0188</td>
</tr>
<tr>
<td>6 (prostate)</td>
<td>Y2</td>
<td>3.0</td>
<td>3.1059</td>
<td>0.1059</td>
</tr>
<tr>
<td>1 (head and neck)</td>
<td>X1</td>
<td>6.8</td>
<td>6.7860</td>
<td>0.014</td>
</tr>
<tr>
<td>1 (head and neck)</td>
<td>X2</td>
<td>7.0</td>
<td>6.9426</td>
<td>0.0574</td>
</tr>
<tr>
<td>1 (head and neck)</td>
<td>Y1</td>
<td>5.3</td>
<td>5.2200</td>
<td>0.0800</td>
</tr>
<tr>
<td>1 (head and neck)</td>
<td>Y2</td>
<td>5.3</td>
<td>5.2200</td>
<td>0.0800</td>
</tr>
<tr>
<td>2 (head and neck)</td>
<td>X1</td>
<td>6.8</td>
<td>6.7338</td>
<td>0.0662</td>
</tr>
<tr>
<td>2 (head and neck)</td>
<td>X2</td>
<td>7.3</td>
<td>7.2297</td>
<td>0.0703</td>
</tr>
<tr>
<td>2 (head and neck)</td>
<td>Y1</td>
<td>6.8</td>
<td>6.6816</td>
<td>0.1184</td>
</tr>
<tr>
<td>2 (head and neck)</td>
<td>Y2</td>
<td>5.5</td>
<td>5.4810</td>
<td>0.0190</td>
</tr>
<tr>
<td>3 (head and neck)</td>
<td>X1</td>
<td>6.8</td>
<td>6.6555</td>
<td>0.1445</td>
</tr>
<tr>
<td>3 (head and neck)</td>
<td>X2</td>
<td>7.3</td>
<td>7.2297</td>
<td>0.0703</td>
</tr>
<tr>
<td>3 (head and neck)</td>
<td>Y1</td>
<td>6.3</td>
<td>6.2118</td>
<td>0.0882</td>
</tr>
<tr>
<td>3 (head and neck)</td>
<td>Y2</td>
<td>6.0</td>
<td>6.0291</td>
<td>0.0291</td>
</tr>
<tr>
<td>4 (head and neck)</td>
<td>X1</td>
<td>7.0</td>
<td>6.9687</td>
<td>0.0313</td>
</tr>
<tr>
<td>4 (head and neck)</td>
<td>X2</td>
<td>7.5</td>
<td>7.4907</td>
<td>0.0093</td>
</tr>
<tr>
<td>4 (head and neck)</td>
<td>Y1</td>
<td>6.5</td>
<td>6.5511</td>
<td>0.0511</td>
</tr>
<tr>
<td>4 (head and neck)</td>
<td>Y2</td>
<td>5.8</td>
<td>5.7681</td>
<td>0.0319</td>
</tr>
<tr>
<td>5 (head and neck)</td>
<td>X1</td>
<td>7.3</td>
<td>7.1514</td>
<td>0.1486</td>
</tr>
<tr>
<td>5 (head and neck)</td>
<td>X2</td>
<td>7.5</td>
<td>7.4646</td>
<td>0.0354</td>
</tr>
<tr>
<td>5 (head and neck)</td>
<td>Y1</td>
<td>6.0</td>
<td>6.0291</td>
<td>0.0291</td>
</tr>
<tr>
<td>5 (head and neck)</td>
<td>Y2</td>
<td>5.0</td>
<td>4.7241</td>
<td>0.2759</td>
</tr>
<tr>
<td>6 (head and neck)</td>
<td>X1</td>
<td>7.0</td>
<td>7.1253</td>
<td>0.1253</td>
</tr>
<tr>
<td>6 (head and neck)</td>
<td>X2</td>
<td>7.5</td>
<td>7.5168</td>
<td>0.0168</td>
</tr>
<tr>
<td>6 (head and neck)</td>
<td>Y1</td>
<td>5.8</td>
<td>5.8986</td>
<td>0.0986</td>
</tr>
<tr>
<td>6 (head and neck)</td>
<td>Y2</td>
<td>6.8</td>
<td>6.8643</td>
<td>0.0643</td>
</tr>
<tr>
<td>7 (head and neck)</td>
<td>X1</td>
<td>6.8</td>
<td>6.7338</td>
<td>0.0662</td>
</tr>
<tr>
<td>7 (head and neck)</td>
<td>X2</td>
<td>7.3</td>
<td>7.2297</td>
<td>0.0703</td>
</tr>
<tr>
<td>7 (head and neck)</td>
<td>Y1</td>
<td>6.0</td>
<td>5.9508</td>
<td>0.0492</td>
</tr>
<tr>
<td>7 (head and neck)</td>
<td>Y2</td>
<td>6.3</td>
<td>6.0552</td>
<td>0.2448</td>
</tr>
</tbody>
</table>
4.7 Accuracy of automatic collimator angle detection

Figure 4.7 shows the collimator angle detection accuracy in a prostate IMRT case. The experiment was tested with the same parameters except the collimator angle, which was set to 60, 45, and 30 degrees. In the case of 60 degree collimator angle, the average angle error is 0.26 ± 0.1 degrees, and 0.15 ± 0.04 degrees is the average angle error with 45 degree collimator angle. In addition, the average angle error with collimator angle 30 degrees is 0.19 ± 0.13 degrees. Based on our experiment, this algorithm is able to detect the collimator angle within 0.5 degrees in IMRT treatment by analysing the cine-EPID images.

![Performance testing of automatic collimator angle detection on cine-EPID images in prostate IMRT case](image)

**Figure 4.7** Performance testing of automatic collimator angle detection on cine-EPID images in prostate treatment IMRT case with different angles, including 60, 45, and 30 degrees.
4.8 Calculation time

Time of execution of the algorithm is a critical issue for real-time system implementation. Table 4.7 shows the average calculation time of each function for real-time dosimetry and geometry verification processing a single frame, including loading the EPID image, synchronisation, dose comparison, jaw positioning and collimator angle. The system functions were run on the same environment, which uses MATLAB/Simulink programming, and the hardware is an Intel Core i5-2400 3.10GHz CPU with 8 GB RAM and a 64-bit operating system. The time measurement method uses Tik-Toc function in Matlab. The total time calculation of the entire system is approximately 0.3412 seconds. The results imply that 3 frames per second is recommended in the case of real-time system implementation in this hardware and software environment.

Table 4.7 The time measurement of the real-time dosimetry and geometry system.

<table>
<thead>
<tr>
<th>System Process</th>
<th>Calculation time (second)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Load EPID image</td>
<td>0.0234</td>
</tr>
<tr>
<td>Synchronisation</td>
<td></td>
</tr>
<tr>
<td>MLC positioning</td>
<td>0.0314</td>
</tr>
<tr>
<td>Cosine similarity</td>
<td>0.0042</td>
</tr>
<tr>
<td>Dose comparison</td>
<td></td>
</tr>
<tr>
<td>Individual comparison</td>
<td>0.0741</td>
</tr>
<tr>
<td>Cumulative comparison</td>
<td>0.0832</td>
</tr>
<tr>
<td>Jaw positioning</td>
<td>0.0287</td>
</tr>
<tr>
<td>Collimator angle</td>
<td>0.0962</td>
</tr>
<tr>
<td>Sum</td>
<td>0.3412</td>
</tr>
</tbody>
</table>
Chapter 5
Discussion

We have shown the feasibility of real-time dose delivery verification and geometric verification using continuous comparison of EPID image frames (cine-mode) with predicted EPID images. The methods presented in this thesis aim to detect gross dose errors which would result in patient over-dose or MLC leaf errors which can have similar effects. Additionally, the geometric verification is able to detect such factors as jaw position, collimator angle and MLC leaf position. It must be noted that even though there are existing techniques for pre-treatment verification, accidents or gross errors during irradiation can unfortunately still occur (Bogdanich 2010). Also, with the increasing move to hypofractionated treatment, errors cannot generally be corrected in subsequent fractions. We demonstrated that the system was able to detect similar gross dose errors immediately. The system could provide almost instantaneous warning to the treating therapist to interrupt the treatment and investigate the source of error. To date, we have simulated the system with Simulink/MATLAB, which simulated the clinical environment using actual linac acquired image frames input at the rate of image acquisition. However, jaw and collimator have a secondary readout that is present for gross errors and malfunction of primary readout.

The current system synchronizes an EPID image with the closest predicted image based on an integration of geometric (leaf positions) and dosimetric (image values) index values, using cosine similarity (Fraass, Lash et al.). A more direct synchronization would use an algorithm similar to gamma to identify the closest predicted image. However, the CS approach has several advantages. It is much faster than gamma calculation (see table 4.4), and it also provides a separate MLC position verification. This enables MLC position errors to be detected independently of gamma comparisons and allows separate classification of geometric errors and monitor unit errors. The synchronisation method is robust for a large range of differences in the number of predicted vs. measured images. We classified the synchronisation error in terms of control points, as these are directly related to dose delivery increments. In general, the method can determine the control point that the measured EPID image frame corresponds to with typical errors of less than 2 control points. This represents only very small changes in dose and MLC shape for sliding window MLC deliveries.
As the number of CPs is usually less than the number of EPID images measured in cine mode, unnormalized dose comparisons performed at every CP are prone to systematic differences. In order to circumvent this issue, the system can either perform a gamma comparison for every synchronised frame acquired, or the number of predicted frames for comparison can be chosen to represent the approximate frame rate set by the imager service monitor. The main aim of this publication was to find gross errors during radiation delivery, so only large deviations from the treatment plan were considered (Figures 4.4 and 4.5). The system allows the user to define the pass-rate threshold, and in this study we applied gamma pass (4%, 4mm) thresholds of 50% for individual and 60% for cumulative comparisons. Figure 4.4 and 4.5 show how the individual dose comparisons respond quicker to errors (approx. 0.1 seconds) compared to the cumulative dose comparisons (few seconds delay). However, the detection sensitivity should be investigated to provide the appropriate threshold for minimising errors detectable across a wide range of clinical cases.

There are two potential approaches to improve the system speed: 1) use of hardware such as a graphic-based processing unit (GPU) (Gu, Jia et al. 2011), and 2) a revised dose comparison algorithm (Bakai, Alber et al. 2003; Jiang, Sharp et al. 2006), as described in chapter 2, dose comparison tool section.

The system is designed to detect both dose error and geometrical error. In IMRT treatment, jaw position, gantry angle, and collimator angle are static. The system is able to detect the jaw position by analysing the 2nd frame of the EPID image. With the limitations of automatic jaw positioning, it can only analyse the beginning of the acquired images, due to the scatter issue. The MLC leaves normally move from the left to right side, and the first image shows less scatter. The algorithm performs better with less scatter, and accuracy is within less 0.1 cm. Nevertheless, the improvement of automatic jaw positioning algorithm should be considered in regard to the scatter issue. In the future, the system which allows the jaw to move during IMRT/VMAT to achieve optimal shielding, such as Smart LMC for Varian should be investigated with the method.

The system verifies the collimator angle with the plan and sounds an alarm if there is an error of more than 1 degree. The system performs the collimator angle detection only for the first 10 seconds, approximately the first 40 frames, because it is static during the treatment. The MLC verification can be performed independently of dose verification and provides more information on the cause of error. For example, the MLC leaves stop moving or are in an incorrect position. The combination of real-time dosimetry and real-time geometry potentially increases the quality and accuracy level, and the system can be easily implemented in the clinical environment. The algorithms perform as fast as real-time, and an error can be detected in 100 ms. Even though the MLC leaf position, jaw, and collimator angle can be evaluated
using the information contained in the dynamic log file, or “Dynalog files” in the case of Varian machine (Varian Medical System, Palo Alto, CA). These files contain leaf position and dose fraction information recorded every 50ms. However, these dynalog files are not full independent of the linac and cannot be accessed in real-time.
Conclusion and future work

We have studied the possibility of using an EPID in cine-mode as a tool for real-time patient-specific IMRT quality assurance. The system utilizes a sophisticated physics-based model to generate a predicted time-lapse series of EPID images during irradiation. As EPID images are recorded during delivery, these are synchronized to the predicted image data set using a new method that combines both geometrical and, when needed, dosimmetrical comparisons. Following synchronization, the measured images are compared to the predicted images using the gamma comparison. Automatic MLC leaf positioning is introduced and used for two purposes, cosine similarity synchronisation and geometric verification. In terms of real-time geometry, jaw position and collimator angle are verified in the beginning of the treatment.

We have shown that our technique is able to detect dose errors, including overdose and MLC leaf mispositioning. The system was simulated using Simulink/MATLAB, and 2 case studies were tested: 1) gross dose error and 2) MLC leaf failure. The system was able to detect these errors in real-time (within 0.1s). The technique is fast and can be easily applied in a clinical environment.

The system was applied to dynamic sliding window IMRT fields, as these are commonly used. The use of VMAT is increasing rapidly, and the system should be easily adapted to VMAT deliveries. In this case, a similar synchronization method could be applied, or synchronization based on the known gantry angle of the image applied. We are currently interfacing the MATLAB software with a frame-grabber/PC system connected to the linac, which receives each image frame so that we can demonstrate the system during patient treatment as a next implementation step. This will enable further optimisation of the system for patient verification, including establishment of sensitivity thresholds and classification of delivery errors.

Real-time 3D dose verification can be extended from real-time treatment verification using EPID. The 3D dose verification requires real-time dose delivery measurements to reconstruct the dose that is delivered to the current patient model (CT data). Based on the high performance of computers nowadays, it is possible to develop the real-time adaptive
radiotherapy that recalculates the predicted image in real-time based on the feedback of real-time treatment, such as tumour motion, patient motion, or systematic error.

In addition, the current system can also be used for data collection to be used for error treatment analysis. This uses a machine learning model to recognise the type of error based on the information collected from the system. The optimised model can classify the type of error in real-time that led to the root of problem. As a result, radiation therapists or physicists can improve the quality of treatment to be more precise and accurate.

Bakai, A., M. Alber, et al. (2003). "A revision of the gamma-evaluation concept for the

imaging device for multileaf collimator quality control and calibration." Physics in Medicine
and Biology 50(7): 1377-1392.


Boellaard, R., M. Essers, et al. (1998). "New method to obtain the midplane dose using portal in


Chang, J. and C. C. Ling (2003). "Using the frame averaging of aS500 EPID for IMRT


