Validation of the Spatial Accuracy of the ExacTrac® Adaptive Gating System

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Abstract

Stereotactic body radiation therapy (SBRT) is a method of treatment that is used in extracranial locations, including the abdominal and thoracic cavities, as well as spinal and paraspinal locations. At the McGill University Health Centre, liver SBRT treatments include gating, which places the treatment beam on a duty cycle controlled by tracking of fiducial markers moving with the patient’s breathing cycle. Respiratory gated treatments aim to spare normal tissue, while delivering a dose properly to a moving target.

The ExacTrac® system (BrainLAB AG Germany) is an image-guided radiotherapy system consisting of a combination of infra-red (IR) cameras and dual kilovoltage (kV) X-ray tubes. The IR system is used to track patient positioning and respiratory motion, while the kV X-rays are used to determine a positional shift based on internal anatomy or fiducial markers.

In order to validate the system’s ability to treat under gating conditions, each step of the SBRT process was evaluated quantitatively. Initially the system was tested under ideal static conditions, followed by a study including gated parameters. The uncertainties of the isocenters, positioning algorithm, planning computed tomography (CT) and four dimensional CT (4DCT) scans, gating window size and tumor motion were evaluated for their contributions to the total uncertainty in treatment.

The mechanical isocenter and 4DCT were found to be the largest sources of uncertainty. However, for tumors with large internal amplitudes (>2.25 cm) that are treated with large gating windows (>30%) the gating parameters can contribute more than 1.1 ± 1.8 mm.
**Abrégé**

La radiochirurgie stéréotaxique corporelle (RCSC) est une modalité de traitement utilisée sur les lésions extracraniales, tels que les cavités abdominales et thoraciques, ainsi que les lésions situées à l’intérieure ou l’extérieure de l’épine dorsale. Au Centre Universitaire de Santé de McGill (CUSM), les traitements de la foie par RCSC se basent sur la synchronisation respiratoire qui permet de contrôler le faisceau de photons à l’aide de marqueurs référencés du cycle respiratoire du patient. L’objectif des traitements par synchronisation respiratoire est de limiter le dommage au tissu normal tout en transmettant la dose appropriée à la cible en mouvement.

Le système ExacTrac® (BrainLab AG Allemagne) est un système de radiothérapie guidée par image comprenant une caméra infrarouge (IR) et deux tubes à rayons-x à l’échelle des kilovolts (kV). Le système IR suit le positionnement des patients et le mouvement respiratoire, alors que les rayons-x kV déterminent la variation des positions basées sur l’anatomie interne ou les marqueurs référencés. Afin de valider la capacité d’opération du système sous les conditions de synchronisations respiratoires, chaque étape du RCSC a été évaluée quantitativement. Initialement, le système fut testé sous les conditions statiques idéales, suivie par une étude incluant les paramètres de synchronisations. Les incertitudes de l’isocentre, l’algorithme de positionnement, la planification de la tomodensitométrie (CT) et les balayages par tomodensitométrie à quatre dimensions (4DCT), la période de synchronisation, et le mouvement de la tumeur furent évalués et la contribution de chacun des facteurs à l’erreur totale du traitement déterminée. L’isocentre mécanique et le 4DCT s’avèrent être les sources d’incertitudes majeures. Cependant, pour les tumeurs à large amplitude interne (>2.25 cm) qui sont traitées avec de larges périodes de synchronisations (>30%), les paramètres de synchronisations peuvent aussi avoir des contributions supérieures à 1.1 ± 1.8 mm.
# Table of Contents

Acknowledgements .............................................................................................................. ii  
Abstract ............................................................................................................................... iii  
Abrégé .................................................................................................................................... iv  
List of Figures ......................................................................................................................... viii  
List of Tables ........................................................................................................................ xi  
Chapter 1: Introduction ........................................................................................................ 1  
  1.1 Stereotactic Radiosurgery and Radiotherapy ............................................................... 1  
    1.1.1 History ..................................................................................................................... 1  
    1.1.2 General SRS Requirements ................................................................................... 3  
  1.2 General SBRT Concepts and Process .......................................................................... 3  
    1.2.1 Organ Motion ......................................................................................................... 4  
    1.2.2 Dose Prescription and Fractionation ....................................................................... 5  
    1.2.3 Clinical SBRT Process at the MUHC ..................................................................... 5  
      1.2.3.1 Internal Fiducial Markers ............................................................................... 6  
      1.2.3.2 Patient Immobilization and Organ Motion ..................................................... 7  
      1.2.3.3 Four-Dimensional Computed Tomography Scan ......................................... 8  
      1.2.3.4 Treatment Planning and Dose Prescription .................................................... 9  
      1.2.3.5 Patient Positioning ........................................................................................... 10  
      1.2.3.6 Treatment Delivery .......................................................................................... 11  
      1.2.3.7 Image Guidance ............................................................................................... 12  
      1.2.3.8 Daily Quality Assurance and Calibration ..................................................... 12  
  1.3 Uncertainty Tests: A Literature Review ....................................................................... 13  
    1.3.1 Phantom Studies .................................................................................................... 14  
    1.3.2 Patient Studies ....................................................................................................... 17  
  1.4 Purpose and Organization of the Thesis ..................................................................... 17  

Chapter 2: Theory .............................................................................................................. 23  
  2.1 Radiation Therapy Target Volumes ........................................................................... 23  
  2.2 Theory of Uncertainties ............................................................................................... 24  
  2.3 Isocenters .................................................................................................................... 28  
  2.4 Respiratory Gating ...................................................................................................... 30
2.4.1 The Breath Trace and Surrogate Correlation ........................................ 30
2.4.2 Respiratory Gating Benefits and Issues ............................................. 32
2.5 Imaging Modalities ................................................................................. 33
  2.5.1 Computed Tomography ................................................................. 33
  2.5.2 BrainLAB ExacTrac® 6D Image Guidance System ......................... 35
2.6 Combination of Uncertainties and Summary of Theory ........................... 39

Chapter 3: Materials and Methods ................................................................. 42

3.1 Equipment ............................................................................................... 42
  3.1.1 CT and RPM ..................................................................................... 42
  3.1.2 Treatment Planning Software ............................................................. 43
  3.1.3 Linac .................................................................................................. 43
  3.1.4 ExacTrac® and Accessories ................................................................. 43
  3.1.5 Phantoms .......................................................................................... 44
  3.1.6 Film QA ............................................................................................ 48

3.2 Isocenter Evaluation Experiments .......................................................... 50
  3.2.1 IR Tracking of Couch Movement ....................................................... 50
  3.2.2 ExacTrac® X-ray Tracking of Couch Movement ............................... 51
  3.2.3 Daily Winston Lutz Check ................................................................. 51
  3.2.4 The Hidden Target Test ..................................................................... 53
  3.2.5 Coordinate Deviation of the ExacTrac® System ................................. 54

3.3 Image Fusion Experiments ...................................................................... 55
  3.3.1 Relative Comparison of Fusion Methods on a Static Phantom .......... 56
  3.3.2 Relative Comparison of Fusion Methods on a Static Anthropomorphic
      Phantom ................................................................................................. 56

3.4 Computed Tomography Experiments ..................................................... 57
  3.4.1 Effect of Slice Thickness Using a Static CT ...................................... 57
  3.4.2 Effect of Slice Thickness on Four-Dimensional CT ............................ 58

3.5 Respiratory Gated Experiments .............................................................. 58
  3.2.10 Effect of Gating Window Size and Tumor Motion .......................... 58

3.6 Patient Study ........................................................................................... 59

Chapter 4: Results and Discussion .................................................................. 63
4.1 Isocenter Evaluation ................................................................. 63
  4.1.1 IR Tracking of Couch Movement ....................................... 63
  4.1.2 X-ray Tracking of Couch Movement ................................. 64
  4.1.3 Evaluation of Daily WL Films ........................................... 66
  4.1.4 Coordinate Deviation of the ExacTrac® System .................... 68
4.2 Image Fusion ........................................................................... 68
  4.2.1 Relative Comparison of Fusion Methods on a Static Phantom .... 69
  4.2.2 Comparison of Image Fusion Methods for a Static Gating Phantom . 71
  4.2.3 Comparison of Fusion Methods on a Static Anthro. Phantom .... 72
4.3 Computed Tomography ............................................................. 77
  4.3.1 Effect of Slice Thickness Using a Static CT .......................... 77
  4.3.2 Effect of Slice Thickness on a Four-Dimensional CT .......... 78
4.4 Respiratory Gated Effects ........................................................ 79
  4.4.1 Comparison of Treatment Plans using End-to-End Test ............ 79
  4.4.2 Tumor Amplitude ............................................................... 82
  4.4.3 Effect of Gating Window Size and Tumor Motion .................. 83
  4.4.4 Dose Profiles ................................................................... 84
4.5 Patient Study and Summary ...................................................... 85
  4.5.1 Patient Study ................................................................... 86
  4.5.2 Comparison of Spatial Accuracy on a Phantom vs. Patient .... 87

Chapter 5: Conclusion ................................................................. 91
  5.1 Summary of Thesis ................................................................ 91
  5.2 Future Work ......................................................................... 92

BIBLIOGRAPHY ........................................................................... 97
List of Figures

Figure 1-1: ITV treatment (A) versus gated treatment (B) including a setup margin. The red area denotes the treatment field. ................................................................. 4
Figure 1-1: Two implanted fiducials for a liver SBRT patient. The fiducials are chosen to give high contrast on an X-ray of the patient. .............................................. 7
Figure 1-2: Vacuum bag used for placing patient in a reproducible position on both the planning CT and the treatment couch. ................................................................. 8
Figure 1-3: The Varian® RPM system consists of an IR camera and screen (A), which relays the respiratory information to the RPM interface (B). .................. 9
Figure 1-4: The reference star is attached to the side of the couch, allowing the system to track the relative motion of the patient. .................................................. 11
Figure 1-5: Setup for WL test done during machine calibrations. The ball bearing is mounted to the end of the couch, and a piece of radiochromic film is mounted to the linac head using a clamp behind the bearing in line with the treatment beam. ........................................................................................................ 13
Figure 2-1: Target volume definitions, as described by the International Commission on Radiation Units and Measurements ............................................ 24
Figure 2-2: Allowable tolerance of spatial uncertainties through the course of a typical treatment using the ExacTrac® system for positioning .................. 25
Figure 2-3: Flowchart describing the method used to check the different isocenters and coordinate systems. The radiation isocenter is compared to the laser isocenter, which is then used to set up the IR coordinate system. The IR system is then used to check that the ExacTrac® X-ray isocenter is accurate. Finally, the X-rays are then compared to the original WL pointer to check that the radiation isocenter and the ExacTrac® isocenter coincide within tolerance. ...... 28
Figure 2-4: Diagram of a ‘star pattern’ exposure, useful in determining the variability of the radiation isocenter. ................................................................. 29
Figure 2-5: Typical breath trace of a patient with a consistent rhythm over a 60 second period. .................................................................................................... 31
Figure 2-6: The effect of moving 12 mm spherical target on a standard axial scan (A) versus images acquired during a 4DCT scan (B). (C) shows the axial slices of a 4DCT of a moving spherical target. The spiral pattern results from the reconstructions over a full rotation of the CT, and show a decreased density around the outer surface, due to averaging effects. Reproduced from [42]............ 33
Figure 2-7: Effect of partial voluming at the border of two types of tissues. The white line represents the structure border and causes the detectors to average the tissues into a single value. .......................................................................................... 34
Figure 2-8: Cranial Array, Patient Mask, Reference Star and Grid (Reproduced from ExacTrac® Clinical User Guide) ................................................................. 36
Figure 2-9: BrainLAB ET Isocenter Phantom. The five IR markers are placed with a known configuration and point in space, allowing for the determination of an IR coordinate system ........................................................................................................ 36
Figure 2-10: Determination of the location of an IR marker using dual IR cameras. When the location of the focal spot is known, virtual lines can be used to locate the position of single IR reflectors (A). Ambiguities can appear for certain
setups (B) with multiple intersections, leading to the detection of ‘ghost markers.’

Figure 2-11: X-ray tube and housing box used for kV imaging (Reproduced from ExacTrac® Clinical User Guide).

Figure 3-1: Both the cranial array (A) and the reference array (B) are outfitted with six IR markers.

Figure 3-2: PMMA phantom. Each slice can contain a number of slots or internal markers, allowing for a unique configuration specific for the user’s purpose.

Figure 3-4: The ET Gating Phantom. The longitudinally moving platform (A) and the vertical marker plate (B) simulate the internal and external motion of a SBRT patient.

Figure 3-3: Anthropomorphic head phantom used for testing positioning fusion algorithms. The lighter colored pegs placed in an array could be removed and replaced with markers, dosimeters, or different density inserts.

Figure 3-5: Conversion of scanned film after image enhancement and median filter.

Figure 3-6: Preparation for alignment of WL pointer (A) with lasers. Precision positioning knobs (B) are used to shift the pointer along the lateral and vertical axes. A properly aligned WL pointer can be seen on the right.

Figure 3-7: Stereotactic cones (A) used for treatment and calibration procedures. Cones are placed into a mount (B) directly beneath the linac head.

Figure 3-8: X-ray image of WL pointer from Tube 1. The embedded sphere and expected center are compared in the ExacTrac® software.

Figure 3-9: Treatment planning portion of the liver SBRT patients. Figure A shows the contouring of the target volumes (GTV – pink, CTV – purple, PTV – red). Figure B shows the beam configuration around the targeted area, as well as the heart (orange), small bowel (yellow), and the healthy liver (blue).

Figure 3-10: ExacTrac® images of a liver SBRT patient. Internal fiducials (A) are used to complete precise positioning, while IR markers (B) are used to track the patient’s movement and breath trace.

Figure 4-1: Comparison of the infrared detected position as a function of the couch angle.

Figure 4-2: Comparison of total deviation detected by the two image fusion methods. The blue points represent the internal marker fusion, while the pink shows the bony fusion method.

Figure 4-3: Scatter plot for the daily Winston-Lutz film exposures during the month of October 2010. The blue points represent a couch position of 0°, and multiple gantry angles. The red points represent a gantry angle of 0° and couch angles of 45° and 315°.

Figure 4-4: Scatter plot of the daily Winston-Lutz exposures for the month of October 2010. The points have been organized by the measured deviation at the individual couch or gantry angles and rotated into the couch’s frame of reference.

Figure 4-5: Comparison of Internal Marker, Bony Automatic, and Manual fusion methods available on the ExacTrac® software.
Figure 4-6: Comparison of the Internal Marker, Bony Automatic, and Manual fusion methods using a separate calibration of the ExacTrac® system.

Figure 4-7: Plot showing the total detected offset from planning position on a static anthropomorphic phantom. The pink shows the results of the bony fusion, while the blue shows the results of the internal marker.

Figure 4-8: Comparison of the IR, internal marker, and auto fusion positioning methods in the presence of lateral shifts.

Figure 4-9: Comparison of IR, internal marker, and auto fusion positioning modalities in the presence of SI shifts.

Figure 4-10: Comparison of IR, internal marker, and auto fusion positioning modalities in the presence of vertical shifts.

Figures 4-11 and 4-12: End-to-end results of variation of window size for a tumor amplitude of 2 cm. Figures include the 0.5 mm offset programmed into the treatment plan, resulting in overall deviations of greater than 1 mm.

Figure 4-13: Comparison of window size (10% on the left, 50% on the right) on the dose blurring effect.

Figure 4-14: Overall plot of comparison of gating window sizes for a tumor amplitude of 2 cm.

Figure 4-15: Plot showing the relative offset of the hidden target as a function of tumor amplitude.

Figure 4-16: Comparison of dose profiles for a target moving over a 5-second breath cycle.

Figure 4-17: Two patient positioning X-rays of the same patient on different treatment days. A well contrasted image (A) shows the implanted markers clearly on a background of the patient anatomy. An improperly warmed up X-ray tube (B) can result in a salt-and-pepper images that can cause difficulty in localizing the implanted markers.

Figure 4-18: Two patient positioning X-rays of same patient on different treatment days, showing effect of beam energy. Correctly chosen beam parameters (kV, ms, mA) will result in a contrasted image, where implanted markers can be detected easily (A). Improperly selected parameters can result in an oversaturated image, where markers are not readily visible against the anatomy of the patient.
List of Tables

Table 2-1: Relationship between the confidence interval, and the confidence level. ................................................................. 27
Table 4-1: Table summarizing the detected offset by IR marker, internal marker, and auto fusion from the actual position for shifts along any of the three couch axes. ..................................................................................................................... 71
Table 4-2: Summary of the anthropomorphic phantom’s average deviations from its actual position based on magnitude of the shift from planned isocenter. ........ 74
Table 4-3: The detected deviation calculated using WL exposures for various slice thicknesses used for SBRT. ................................................................. 78
Table 4-4: Positional Accuracy of the ExacTrac® Adaptive Gating System vs. CT Slice Thickness ......................................................................................... 78
Table 4-5: Positional Accuracy of the ExacTrac® Adaptive Gating System vs. Gating Parameters ........................................................................................................... 84
Table 4-6: Comparison of detected shifts for Liver SBRT patients at the MUHC hospital from 2010-2011. ..................................................................................... 86
Chapter 1

Introduction

The Canadian Cancer Society estimates in 2010 that 76,200 deaths will occur due to cancers, as well as approximately 173,800 new cases. Based on current rates, 40% of Canadian women and 45% of Canadian men will develop cancer at some point in their lifetime, of which an estimated one out of four is expected to die [1]. Cancer treatments typically include chemotherapy, radiation therapy (RT), or surgery, and the method of choice will vary based on the location and type of cancer.

External beam radiation therapy deals with energy deposited within a patient originating from a radiation source that is located outside the patient. When delivering the prescribed dose to a localized lesion which is defined in a three-dimensional (3D) plane, the technique is referred to as external beam stereotactic irradiation.

1.1 Stereotactic Radiosurgery and Radiotherapy

Stereotactic techniques in radiation therapy require the precise location of the target to be treated within the body in a 3D coordinate system, and can be further categorized by dose fractionation; either the total dose is delivered in a single fraction, stereotactic radiosurgery (SRS), or over the course of multiple fractions, referred to as stereotactic radiotherapy (SRT). Due to the nature of SRS, patients must be immobilized with invasive frames, whereas in SRT, non-invasive setups are preferred. These setups can vary greatly in variety, but the goal of properly imaging, planning, and treating a patient’s maladies remains the same.

1.1.1 History

SRS first came about in the late 1940’s as an attempt by Lars Leksell to treat localized areas in the brain using 200 kVp x-rays [2]. He defined the
technique as a single high dose of radiation, stereotactically directed to an intracranial region of interest. Over the next decade it became possible to use more penetrating beams to better deliver the high dose required for treatments. Initially proton beams produced in a cyclotron were used, and in 1968 Leksell released a paper describing the use of focused Cobalt-60 gamma rays for a thalamotomy [3]. Stereotactic radiosurgery has since expanded from the use of the original orthovoltage x-rays to specialized treatment equipment including the GammaKnife®, CyberKnife®, TomoTherapy® machines, and linear accelerators. The GammaKnife® uses 201 cobalt-60 sources arranged in a circular array above a patient’s head. The sources can be individually directed to target lesions in a patient’s brain. The CyberKnife® uses a 6 MV linac mounted on a robotic arm, which can be maneuvered in three-dimensions about the treatment site. TomoTherapy® accomplishes stereotactic treatments by using a rotating radiation source about the target area, and treating the patient in millimeter sized slices. Particularly relevant to this thesis, linear accelerators can accomplish stereotactic treatments by utilizing the rotation of the gantry head in conjunction with the treatment couch.

Stemming from the SRS techniques, stereotactic body radiation therapy (SBRT) is a relatively recent innovation in radiotherapy that is effective in controlling oligometastatic cancers in extracranial locations, such as the abdominal and thoracic cavities, as well as spinal and paraspinal locations. SBRT treatments began in the early 1990’s, with the first clinical patient results being published in 1995 [4]. Up until 2003, studies were concerned primarily with lung and liver tumors, when the first presentation of spinal lesion patients was published [5].

As opposed to typical radiation therapy treatments, which will deliver up to 3 Gy per fraction, for anywhere between 10-30 fractions, SBRT attempts to deliver a larger dose over a shorter period of time. Normal treatments will require 6-30 Gy per fraction over the course of 1-5 fractions. Due to the nature of SBRT requiring a fewer number of fractions, treatment accuracy must be carefully monitored to take into account the organ motion.
1.1.2 General SRS Requirements

According to AAPM Report No. 54 [6], the basic requirements for SRS are accurate localization, mechanical precision, accurate and optimal dose distribution, and patient safety. The accurate localization requires that the stereotactic system be able to determine the coordinates of a well-defined object within a framed coordinate system to within 1 mm for angiography and 2 mm for CT and MRI. A pointer or ball bearing is commonly used as the target object during the localization procedure.

The element of mechanical precision relies on the alignment of the frame-based coordinate system with the coordinate system of the treatment linac. The isocenter of the couch, gantry and collimator need to coincide within a 1 mm radius sphere for all possible angles.

1.2 General SBRT Concepts and Process

SBRT treatment setup accuracy requirements are equivalent to those of SRS, but also entails a strictly enforced maintenance of high spatial targeting accuracy for the entire treatment, through the use of immobilization and patient position monitoring. SBRT also requires the highest need for respiratory motion management as well as redundancy in geometric verification [7]. Treatments are typically applied using one of two approaches: treatments which irradiate during the entire breathing cycle (see Fig. 1-1A), over the entire internal target volume (ITV) plus a setup margin, or gated treatments (see Fig. 1-1B), which limit the irradiation of the target to a certain area of the breath cycle.
While ITV treatments are more common due to less planning and treatment complications, gated treatments are becoming more available to treat mobile tumors.

1.2.1 Organ Motion

In addition to internal organ and tissue motion, the target can also change shape and size. These changes will happen during a single fraction (intra-fractional) and over the entire course of treatment (inter-fractional) [8]. Respiratory effects in particular will be discussed in Section 2.4.1.

Both intra- and inter-fractional motion can have an adverse effect on the treatment. The issues can involve:

1. The imaging modality on which the treatment plan is based will be less than accurate, and give a false representation of the anatomy.
2. A larger field could be needed than the size of the tumor and its microscopic extensions would seem to require, causing more healthy tissue to become irradiated.
3. Fields may be designed too small if the extent of the tumor motion is not fully recognized, leading to underdosing of the target [8].
The patient’s internal anatomy, specifically the target volume, will move and change volume throughout the treatment, as well as over the course of the entire treatment. The extent of this motion varies greatly based on the location of the lesion [9], but can be as large as 50 mm in lung [10] and liver [11] cases. This motion occurs primarily in the SI direction for abdominal tumors, with movement along the other axes being less than 2 mm [12].

Because invasive immobilization is not an option, frameless setups are becoming more common, and treatment setups and imaging are fast being improved to accommodate the need for specialized treatments such as SBRT and gated treatments. These improvements include four-dimensional computed tomography (4DCT) and image guided radiation therapy (IGRT), which will be discussed later this chapter.

1.2.2 Dose Prescription and Fractionation

According to the report of AAPM Task Group 101, when prescribing dose, two conditions are considered [7]:

1. A limited volume, containing the gross tumor and its close vicinity, is targeted for treatment through exposure to a very high dose per fraction, and hotspots within the target are often deemed to be acceptable [7, 13].

2. The volume of normal tissue receiving high doses outside the target volume should be minimized to limit the risk of treatment toxicity. Thus, the gradient describing the dose fall-off outside the target should be sharp [7].

Following these conditions, dose prescriptions are often given at low isodoses (for example 80% isodose) with small margins for beam penumbra at the target edge. This is done in order to improve the dose fall-off and improve normal tissue sparing [7]. This will also increase dose heterogeneity inside the target, which has been surmised to be beneficial in eliminating radioresistant hypoxic cells [14].

1.2.3 Clinical SBRT Process at the MUHC

In 2007, McGill University Health Centre began treating patients using stereotactic body radiation therapy. The addition of the Novalis TX radiosurgery
system in 2010 has added many options for improvements in the treatment of these patients.

While the array of treatment plans and tumor locations may vary, the clinical process remains much the same for each different case. This section will cover the daily quality assurance and calibration, the implantation of internal fiducial markers, the patient immobilization, the four-dimensional planning computed tomography scan, the treatment planning, the final patient positioning, and the treatment delivery.

1.2.3.1 Internal Fiducial Markers

Soft tissue tumors are not readily visible using MV portal imaging, and even using modern image-guidance with kV capability, localization can be difficult. In order to provide a surrogate position of the target, a fiducial marker can be surgically implanted into the patient. Gold or platinum markers are commonly used, such as the Visicoil™ (Core Oncology, Santa Barbara, CA), which is a linear fiducial soft tissue marker (see Fig. 1-1). These markers can be implanted in numerous sites, but can cause a high risk of pneumothorax if injected by needle-point into the lung [15]. The marker is allowed to settle in the target for a few days in order to provide a rigid link between the markers movement and the target’s motion.
**Figure 1-1:** Two implanted fiducials for a liver SBRT patient. The fiducials are chosen to give high contrast on an X-ray of the patient.

### 1.2.3.2 Patient Immobilization and Organ Motion

In order to properly treat a patient through SBRT, it is necessary to gain an accurate image of the patient anatomy that can be used throughout the planning, setup, and treatment. Unlike SRS, SBRT commonly requires the use of alternative methods of pre-treatment imaging. For typical cranial stereotactic treatments, the patient is immobilized through the use of frames or masks. For lesions located extracranially, simply immobilizing the patient through these means is not necessarily feasible. Patient motion is more pronounced in these locations, and requires specialized equipment. Precautions must be taken to immobilize areas even adjacent to the treatment area. At the McGill University Health Centre, Elekta BodyFIX® vacuum bags (see Fig. 1-2) are used to comfortably place the patient in the treatment position on a daily basis.
Figure 1-2: Vacuum bag used for placing patient in a reproducible position on both the planning CT and the treatment couch.

Other immobilization options can include compression plates or respiratory belts that attempt to restrict the patient’s maximum breath intake, thereby minimizing target shifts.

**1.2.3.3 Four-Dimensional Computed Tomography Scan**

With the use of respiratory gating treatments, 4DCT or respiration-correlated CT has become the most widely used method and is the current method of gaining a planning image set at the MUHC. This method oversamples at each couch position to gain multiple images for each slice. The correlation between each of the image sets and the corresponding respiratory phase is determined using an external tracking system, such as the Varian® Real-time Positioning Management (RPM) System (Varian Medical Systems, Palo Alto, CA) (see Fig. 1-3).
Figure 1-3: The Varian® RPM system consists of an IR camera and screen (A), which relays the respiratory information to the RPM interface (B).

This system reports the phase of the motion from an infrared reflector box located on the patient’s abdomen, from 0% (initial inspiration), 50% (exhalation) to 100% (full inhale). The box’s position is tracked by an infrared (IR) camera located at the end of the couch. This information is then used to correlate the breathing phase with the CT image sets. The Varian software allows for either phase or amplitude based image binning, which refers to either the time windows during the breath cycle or the displacement of the patient’s abdomen during the scan, respectively. Once a suitable phase window has been chosen, typically centered about 50% at the MUHC, the image set is exported to the treatment planning system. This window is chosen to minimize motion artifacts, although other image sets, such as the 0% window or the maximum intensity projection (MIP) can be uploaded as well in order to estimate tumor motion. The MIP allows the user to superimpose the position of the target over several phases of the breathing cycle, allowing for visualization of the extent of the tumor motion.

1.2.3.4 Treatment Planning and Dose Prescription

The contoured CT image set is then imported into the Eclipse™ (Varian Medical Systems, Palo Alto, CA) treatment planning system. From here the data
can be imported into the BrainLAB iPlan® RT Treatment Planning Software (BrainLAB AG, Feldkirchen, Germany) in the Digital Imaging and Communications in Medicine (DICOM) formatting. Once the import wizard has begun, the user can select which image series they would like to use, depending on the type of treatment modality. This is where the user would choose to use the gated images or MIP for planning.

The images are then viewed and confirmed to be suitable, after which localization can be started. The localization defines a set of markers used to determine a frame of reference for the image set. This can be done using either the CT body markers or a localization box for the case of cranial lesions. A physician is then placed in charge of contouring the treatment target and the relevant critical structures.

Beam configurations, conformal shapes, prescribed dose, and all other treatment parameters are then selected before the plan can be approved by a medical physicist and a physician. Five to ten nonoverlapping beams are generally required to provide the high dose gradient required for SBRT. In addition to being exported to the treatment delivery system, image sets need to be loaded into the ExacTrac® computer for patient positioning purposes.

### 1.2.3.5 Patient Positioning

The patient is positioned on the treatment table using the immobilization device, similarly to the planning step. IR markers are placed on the patient’s chest or abdomen to track the patient’s movement and respiration pattern. An IR reference star (see Fig. 1-4) is attached to the couch to provide a reference frame for the moving IR markers on the patient’s chest. Using these markers the patient is roughly positioned at the treatment isocenter using a weighted marker algorithm, which will be described in Chapter 2. Orthogonal x-ray tubes located in the floor of the treatment room are then used to image the internal anatomy of the patient.
Figure 1-4: The reference star is attached to the side of the couch, allowing the system to track the relative motion of the patient.

These images are compared to digitally reconstructed radiographs (DRRs) of the patient from the planning CT. For the case of gated procedures, which will be discussed in detail in Section 2.4, internal markers are used to determine the offset of the patient from the planned position. This shift is then applied by the system and checked by the therapists.

1.2.3.6 Treatment Delivery

At the MUHC, SBRT treatments can be delivered on the Novalis TX™ linac (Varian Medical System, Palo Alto, CA). The 6MV linac is equipped with a 2.5 mm HD120 high-definition multi-leaf collimator (MLC). Once the patient is in the correct treatment position, the physician will chose a reference level in the patient’s breath trace and an amplitude-based window around which irradiation will occur. This will typically be chosen at the maximum inhale or exhale phase
to minimize residual tumor motion. Additional imaging levels will generally be chosen to confirm that the internal markers stay within tolerances throughout the treatment for the entirety of the gating window. The gating window is chosen by the physician, but is not necessarily patient specific [16]. The ExacTrac® computer provides a beam hold interlock to the treatment computer when the patient is breathing outside the chosen window. This ensures the irradiation only occurs when the target is in the proper position. Because this limits the beam on time, SBRT sessions can take upwards of 60 minutes, depending on the treatment and the patient’s respiration. Throughout each fraction of the patient’s treatment, this process is repeated.

1.2.3.7 Image Guidance

Image guided radiation therapy can be defined as the use of imaging to plan and initiate radiotherapy treatments. Image guidance originated with the use of planar and volumetric X-ray imaging, which were used for diagnostic and planning purposes. The modalities have since expanded to include the use of megavoltage (MV) portal images, as well as kilovoltage (kV) imaging which can be used to align the patient with the treatment machine at the beginning of each fraction [17]. The Novalis TX™ at the MUHC is equipped with image guidance in the form of both the ExacTrac® IGRT system, and an on-board imager, which allows combined MV and kV imaging and cone-beam CT scans. Both systems can be used for patient setup and pretreatment target localization and to check for intra-fractional target movement. At the MUHC, the ExacTrac® system is currently used to position and monitor the patient before and during gated SBRT treatments.

1.2.3.8 Daily Quality Assurance and Calibration

Each morning, several parameters need to be checked on the treatment machine to confirm that the system will perform within suggested tolerances to give accurate treatments. The first check performed verifies that the radiation isocenter does not vary more than 0.7 mm from the laser isocenter. In what is known as the Winston-Lutz (WL) test [18], a ball bearing attached to the end of
the treatment couch is placed by a therapist at the crosshairs of the room lasers. A piece of radiochromic film is placed behind the bearing, and attached to the linac head via a mounting arm. This film is then irradiated with 600 MU at a combination of gantry head angles (0°, 90°, 180°, 270°) and couch angles (0°, 45°, 135°).

If the bearing has not drifted outside of the treatment field at any of these positions, the WL pointer is then used to confirm that the external stereoscopic X-ray system is also properly lined up with the rest of the system. An infrared coordinate system is then established by placing an IR calibration phantom on the couch in line with the room lasers. Another calibration phantom is placed on the couch and autopositioned at the calculated isocenter using the IR camera. This phantom is then imaged using X-rays, checking the IR coordinate system.

1.3 Uncertainty Tests: A Literature Review

The documentation of the Novalis Treatment system has been covered in detail in many papers, each covering a different aspect of the system. Studies concerned with the positioning accuracy use different methods of comparison,
using different components on the machine to self-check. A popular method of quantifying positional uncertainties to give an overall end-to-end value is known as the hidden target test.

The hidden target test uses a marker placed within a phantom at a known stereotactic location. A radiation detector, typically a sheet of radiosensitive film, is placed in line with the linac head and the target. A single field is used to irradiate the target for any given direction, which will also irradiate the film behind. The field is chosen to be large enough to show a positional shift of the target from the center of the field, while keeping the target within the boundaries. The treatment will result in a shadow of the target in the middle of the field on the film. For a perfectly positioned target, the shadow will lie directly in the center of the film, whereas any deviations can be detected through a two-dimensional vector shift of the centers of the field and target.

1.3.1 Phantom Studies

The overall static accuracy of the ExacTrac® system has been evaluated by several groups, with results depending on the parameters used in the tests. The validation of the original version of the ExacTrac® system, which used separate IR cameras (Qualisys) requiring clinical calibration, was described by two groups. Verellen et al. [19] studied the IR and stereoscopic X-ray imaging systems in 2003. The group looked at the effects of different deviations in the presence of shifts and rotations using internal markers and DRR fusion. The overall uncertainty was determined relative to the accuracy of the IR positioning markers, and was found to give a three-dimensional vector shift of $0.41 \pm 0.92$ mm and $0.28 \pm 0.36$ using the bony anatomy and internal marker as fusion methods, respectively. Yan et al. [20] detailed the agreement of deviations determined by the IR cameras and those calculated though the use of DRR/kV image fusion. They found that the CT slice thickness was a non-negligible source of uncertainty, with 3-D shifts of up to $1.4 \pm 0.6$ mm for slice thicknesses up to 5 mm, while the image fusion methods were comparable. The couch’s frame of reference was chosen as the absolute coordinate frame to which each component was calculated.
However, this study did not conduct absolute measurements comparing the radiation isocenter and treatment isocenter, and did not include errors due to gantry or couch rotations.

The current version of the ExacTrac® implements an upgraded IR camera geometry (Polaris), which comes at a precalibrated distance. One of the initial studies of this system was done in 2009 by Hayashi et al. [21]. Hayashi combined several uncertainties, including those of the ExacTrac® coordinate deviation, the CT slice thickness, the registration error of the verification system, the marker detection. These uncertainties were combined in quadrature to give a total geometric uncertainty of 1.36 ± 0.32 mm.

Takakura et al. [22] reported an overall geometric uncertainty of 0.31 ± 0.77 mm using a static anthropomorphic head and neck phantom, but evaluated only the effects of couch accuracy, the mechanical isocenter, and the difference in position of the mechanical isocenter with the ExacTrac® isocenter. Wurm et al [23] determined an overall system accuracy of a static phantom using frameless positioning to be 1.04 ± 0.47 mm, with an average translational error of 0.31 ± 0.26 mm in any given direction.

In 2011, Kim et al. [24] extended the static tests to verify localization accuracy in an anthropomorphic pelvic phantom. The system accuracy was checked using a pair of orthogonal portal images and was found to be 1.15 ± 0.49, and was found to be comparable to using OBI cone-beam CT for image-guided localization. This comparison was also studied with similar results by Ma et al. [25] in 2009, showing a root mean square difference of less than 0.5 mm in a static phantom.

Several of these studies have gone on to break the system uncertainty into components, the first being the IR tracking system. Wang et al [26] first described the feasibility of using IR markers and cameras as a tracking method. Wang concluded that using weighted subsets from at least 5 markers, the IR detection system is capable of accuracy better than 0.3 mm. However, using a hidden target test, Verellen et al. [19] determined that the average deviation of the Qualisys type camera system with respect to the treatment isocenter was 0.24 ± 0.33, 0.45
± 0.55, and -0.49 ± 0.59 mm in the vertical, longitudinal, and lateral directions, respectively. Uncertainty in CT marker positioning and marker repositioning also contribute to these results. The effect of this positioning uncertainty is relatively unimportant, as the final position in the ExacTrac® IGRT system is determined by the X-ray images and fusion the internal markers or bony fusion of the corresponding DRR images.

Phantom tests of uncertainty in the bony anatomy image fusion can depend on several factors, including the type of anatomy, CT slice thickness [20], and the X-ray contrast [21]. Variation of these dependencies can result in uncertainties of up to 1.3 mm [19, 20]. For SBRT, internal fiducial markers are used for patient positioning, which allows for a direct calculation of the correction vector. Using internal markers results in smaller translational errors and standard deviations [19]. This method will be discussed in chapter 2.

In stereotactic radiosurgery, using the smallest slice thickness for planning CT scans will also result in the most accurate treatment. SBRT scans are typically done with 2 mm slice thickness for the sake of time and convenience for the patient. Yan et al. determined that slice thicknesses between 2 and 5 mm were suitable for SBRT treatments.

For non-static SBRT treatments, moving phantoms have been implemented for use with the ExacTrac® Gating System. The treatments are gated based on the simulated breathing motion of the phantom. In addition to the sources of uncertainty for static cases, the overall geometrical accuracy of gated treatments is primarily affected by the speed of the target, and the size of the gating window [27, 28]. The linac latency also affects the ability of the system to accurately target a moving tumor and is calculated during commissioning.

The shape of the patients breath trace and at what point in the cycle the patient is being treated can also affect the accuracy of the treatment. Most patients are modeled using either a sinusoidal breath trace, or more accurately a parametric characterization of breathing induced organ motion [29] which will be described in Section 2.4.1. Both models have amplitudes at which treating a patient would be more beneficial, as the target will tend to spend more time at the
endpoints. By placing the gating level closer to the end exhale position when gating can maximize the amount of time a target stays within the treatment beam and minimize latency errors from a quickly moving target. Even for patients with less reproducible breathing patterns, a gating level should still be chosen near the end exhale where it will consistently be crossed by the breath trace [27].

1.3.2 Patient Studies

The accuracy of the treatments in the clinic can vary from measurements done on phantoms, due to skin shift, patient motion, irregular breathing, and tumor motion or shrinkage. Inter- and intra-fractional patient set-up shifts can be more than 30 mm along any given axis, and image guidance allows these errors to be minimized. Initial tests with the ExacTrac® Adaptive Gating System show that the average marker positional deviation from the expected position can be reduced to less than 4 mm using verification imaging [30].

1.4 Purpose and Organization of the Thesis

SBRT has become an emerging radiotherapy method capable of increased tumor control for early stage primary and oligometastatic tumors in thoracic, abdominal, spinal, and paraspinal regions. This is accomplished through the delivery of large doses in a few fractions. The reduction in fractionation calls for a high level of confidence in the accuracy of the treatment delivery procedure, to ensure that healthy tissue complications are minimized.

In order to properly treat a lesion, the patient needs to be accurately positioned in relation to the treatment isocenter before and during each fraction of their therapy. The ExacTrac® IGRT system aims to give this confidence through the use of intrafractional X-ray imaging, coupled with an IR tracking system.

A main purpose of this work was to simulate an extracranial treatment as closely as possible to obtain a clinical understanding of the accuracy of the ExacTrac® positioning system. In order to do this, a quantitative method of determining the accuracy needed to be calculated through each step of the
treatment process. This included the daily machine calibration, CT scanning, treatment planning, patient positioning using the ExacTrac® X-ray image guidance system, and the treatment delivery. Parameters affecting the system accuracy were first determined for a static setup, which were then extended to include gated treatment parameters, such as gating window amplitude and tumor motion.

The second chapter of this thesis will deal with the theory behind the hardware used, as well as mathematical descriptions of methods used. The third chapter will go through the experiments and measurements used to determine the system accuracy, which will then be discussed with the results in the fourth chapter. Lastly, the conclusions of the work will be covered, as well as future work.
REFERENCES


Chapter 2
Theory

This section will deal primarily with the theory and background behind SBRT treatments and the experiments done for this thesis. First, the target volumes used for treatment will be defined and discussed, followed by the theory of uncertainties and isocenters. The last few sections of this chapter will deal with respiratory gating and the imaging modalities required.

2.1 Radiation Therapy Target Volumes

Just as with conventional radiation therapy, SBRT applies the target definitions described in ICRU reports 50 and 62. 

(1) The Gross Tumor Volume (GTV) is the gross palpable or visible/demonstrable extent and location of malignant growth.

(2) The clinical target volume (CTV) is the tissue volume that contains a demonstrable GTV and/or sub-clinical microscopic malignant disease, which has to be eliminated. This volume thus has to be treated adequately in order to achieve the aim of therapy, cure or palliation.

(3) The internal target volume (ITV) includes the CTV plus an internal margin for expected physiological movements and temporal variations in size, shape and position of the CTV. This target volume is most often determined using a time dependant imaging study such as a 4DCT.

(4) The planning target volume (PTV) is a geometrical concept, and it is defined to select appropriate beam arrangements, taking into consideration the net effect of all possible geometrical variations, in order to ensure that the prescribed dose is actually absorbed in the CTV [31].

Radiation therapy refers to the PTV as the primary target, which includes the visible tumor, microscopic extensions of the tumor, and a margin accounting for tumor motion throughout treatment and delivery uncertainties. Typically
minimal SBRT margins separating the CTV and PTV will be 5 mm in the axial planes, and 10 mm in the inferior/superior directions [7], although the final margin is decided by the radiation oncologist. Changes in the CTV’s shape and position due to the effects of respiration or organ filling are included into a margin which will make up the internal target volume (ITV) [31]. The ITV is included into the definition of the PTV, and its magnitude depends on the measures taken to compensate for this motion during treatment.

Figure 2-1: Target volume definitions, as described by the International Commission on Radiation Units and Measurements.

For SBRT, including metastatic lung, liver, and paraspinal cases, the GTV and CTV are considered identical [7, 32, 33]. Starting from the GTV, margins are added to account for various complications, and by understanding certain clinical uncertainties, more accurate, minimized margins can be used.

2.2 Theory of Uncertainties

Throughout a patient’s path during a typical radiation therapy treatment, there can be many sources of uncertainty and error (See Fig. 2-2).
The initial diagnosis depends on evaluating the patient histology and stage, but even when properly dealt with results in some uncertainty in the diagnosis. Imaging of the patient constitutes another source of uncertainty, due to artifacts, spatial distortions, or even misinterpretation. Once the planning images have been taken, the target and organs at risk need to be delineated. The prescription and treatment plan development bring about another level of uncertainty. Finally, the patient handling and actual treatment delivery also have an intrinsic level of uncertainty associated with them due to patient and organ motion, positioning, or machine configurations. It is impossible to completely eradicate the uncertainty, but through a proper examination of their source and cause, one can minimize them to a clinically acceptable level, or at least incorporate the uncertainty into the best possible result [8].

The goal of validating the ExacTrac® system was to determine the ability of the system to accurately and precisely position the patient for a gated treatment on a day-to-day basis. In order to properly understand the evaluation of this system, one must be familiar with the terminology and theory behind each of the measurements.

When measurements are repeated over a large sampling, the resulting frequency plot will typically result in a Gaussian distribution, as uncertainty values will generally be spread around a mean value. The variation in this
distribution is the result of what is known as random error. A systematic error occurs when bias occurs in these measurements. This materializes when the mean of the measurement set differs from the actual value of the variable in question. SBRT requires localization that is both accurate and precise. These two concepts correspond with each of the two types of error: accuracy with systematic error, and precision with random error.

The International Organization for Standardization (ISO) and the National Institute of Standards and Technology (NIST) recommend referring to these values as type A and type B uncertainties. Type A uncertainties refers to numerical values which are evaluated by statistical methods, and type B are those which are evaluated by other means [34]. While commonly used interchangeably, error and uncertainty refer to different properties of the measurement. When a measurement is made, an error can be made inadvertently. The magnitude of this error can then be evaluated and expressed as an uncertainty of the measured value. By this logic, a measurement with a large uncertainty can in turn have a negligible error.

A distribution of uncertainties is known as the probability density function. When measurements are made up of predominately random errors, the PDF will be represented by a Gaussian, as per the central limit theorem. The mean of the measurement ($\mu$) refers to the central axis of the curve, or the average of the function. The shape of the Gaussian is dependent on the variance ($\sigma^2$), which describes the dispersion of the measurements. For a limited sample size, these values can be approximated as a function of the individual measurements.

$$\mu \approx \bar{x} = \frac{\sum_{i=1}^{n} x_i}{n}$$

(2-1)

Where $n$ is the total number of measurements, and $x_i$ is the value corresponding to the individual measurement.

$$\sigma^2 \approx s^2 = \frac{1}{n-1} \sum_{i=1}^{n} (x_i - \bar{x})^2$$

(2-2)

The square root of the variance is also known as the standard deviation (SD or $\sigma$).
Using the standard deviation, a confidence interval (CI) can be described. The CI is used to evaluate the reliability of a measurement estimate. For example, a CI with 95% likelihood would give a range of measurements of $\bar{x} \pm 2\sigma$. This can also be interpreted in that 95% of measured values would lie within two standard deviations of the mean of the function. Other useful confidence intervals can be seen in Table 2.1.

Table 2-1: Relationship between the confidence interval, and the confidence level.

<table>
<thead>
<tr>
<th>Confidence Interval</th>
<th>Confidence Level (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\bar{x} \pm 0.675\sigma$</td>
<td>50</td>
</tr>
<tr>
<td>$\bar{x} \pm \sigma$</td>
<td>68.3</td>
</tr>
<tr>
<td>$\bar{x} \pm 1.64\sigma$</td>
<td>90</td>
</tr>
<tr>
<td>$\bar{x} \pm 2\sigma$</td>
<td>95</td>
</tr>
<tr>
<td>$\bar{x} \pm 3\sigma$</td>
<td>99.7</td>
</tr>
</tbody>
</table>

As is the case with the final accuracy of the ExacTrac® System, there are a number of contributing factors. In order to calculate the combined standard uncertainty, NIST recommends combining the standard deviations using the law of propagation of uncertainty, also known as the root-sum-of-squares.

$$U_{\text{combined}} = \sqrt{U_1^2 + U_2^2 + \ldots + U_n^2} \quad (2-3)$$

Where $n$ is the total number of sources of uncertainty. This formula assumes that independence of all of the factors.

When combining uncertainties of different types, the confidence levels must be the same. The type A uncertainties can then be combined in quadrature, separately from the type B uncertainties being combined in quadrature. The type A and type B uncertainties can then be combined to create the combined standard uncertainty [8, 34-36].
2.3 Isocenters

In minimizing the positional uncertainties associated with the ExacTrac® System, one must gain an understanding of the different isocenters that play a role in the calibration of the machine and the treatment of the patient. An isocenter can be described as the point or volume at the focal point of a specific coordinate system. There are four isocenters that need to coincide in a properly calibrated treatment machine equipped with the ExacTrac® System.

![Flowchart describing the method used to check the different isocenters and coordinate systems.](image)

**Figure 2-3:** Flowchart describing the method used to check the different isocenters and coordinate systems. The radiation isocenter is compared to the laser isocenter, which is then used to set up the IR coordinate system. The IR system is then used to check that the ExacTrac® X-ray isocenter is accurate. Finally, the X-rays are then compared to the original WL pointer to check that the radiation isocenter and the ExacTrac® isocenter coincide within tolerance.

The mechanical isocenter is defined as the intersection point of the axis of rotation of collimator and the axis of rotation of the linac gantry. Uncertainty in this isocenter results from mechanical limitations and slight shifts in the gantry frame due to its weight [37].

28
The radiation isocenter differs slightly from the mechanical isocenter, in that it is the point at which the central radiation beam passes for all gantry angles, rather than the optical beams and hardware. During a thorough QA, film can be placed on the treatment table, which is then irradiated at regular angular intervals of the treatment table and gantry. The collimator is narrowed to a slit, which will produce a star pattern on the film (See Fig. 2-4)

Figure 2-4: Diagram of a ‘star pattern’ exposure, useful in determining the variability of the radiation isocenter.

The laser isocenter is the coincidence of three sets of precision lasers located on the walls and ceiling of the treatment room. During the monthly QA for conventional linacs, the laser isocenter is lined up with the mechanical isocenter. For linacs being used for SRS, the coincidence of the laser isocenter and radiation isocenter is checked daily through the use of a Winston-Lutz test, when the WL pointer is lined up to the laser isocenter, and then irradiated at the specified gantry and couch angles.

The ExacTrac® system operates externally from the gantry frame, and therefore requires a separate calibration to overlay the isocenter of the stereographic X-ray tubes located in the treatment room floor to the other isocenters. This isocenter is defined during the commissioning of the system as the intersection of the central beams from the dual X-ray tubes to the center of the
detectors mounted to the treatment room ceiling. The isocenter is a function of immobile components, and therefore will ideally not move. The coincidence of this isocenter with the mechanical, radiation, and laser isocenters is paramount to the usefulness of IGRT, and is checked daily. Before the ExacTrac® isocenter can be inspected, the infrared coordinate system must be created. This is done by lining up a phantom outfitted with IR markers at the laser isocenter. Another phantom equipped with internal markers is then moved to the newly defined origin of the IR coordinate system. The phantom is imaged using the dual floor mounted X-ray tubes, checking that the IR coordinate system and the X-ray tubes are aligned within tolerance.

The daily variation in the ExacTrac® isocenter is defined directly by the ability to line a phantom up with the laser isocenter.

2.4 Respiratory Gating

Respiratory gating refers to the process of irradiation during a specified portion of the patient’s breathing cycle. The duty cycle of the radiation will typically be between 30% and 50% [17] and chosen to be around a point which will minimize tumor motion. The internal tumor position can be inferred from an external respiration signal or directly through internal marker imaging. This section will discuss the theory behind the patient’s breath trace and the benefits and issues with gated treatments.

2.4.1 The Breath Trace and Surrogate Correlation

The patient breath trace summarizes the patient’s movement along the anterior/posterior (AP) direction. This is typically done using external trackers placed on the patient’s abdomen. During the inhale phase, the patient’s lungs expand with the contraction of the diaphragm. This, in turn, causes the chest wall to elevate in the AP direction. A breath cycle is not perfectly symmetric, and the inhalation phase is typically longer than the expiration phase. The shape of this curve is typically modeled using a sinusoid, but can more accurately be modeled using a parametric characterization of breathing induced organ motion [29]:
\[ y(t) = y_0 - b \cos^{2n} \left( \frac{\pi \cdot t}{\tau} - \phi \right) \] (2-4)

Where \( b \) is the peak-to-peak amplitude, \( \tau \) is the breathing period, \( \phi \) is the starting phase, and \( n \) is an asymmetry factor.

Other considerations for pulmonary tumors include the hysteresis of the lung [38], which causes the intra-lung pressure during the inhalation to be less than that of the expiration, leading to a phase shift between the external surrogate motion and the internal motion.

![Typical breath trace of a patient with a consistent rhythm over a 60 second period.](image)

Each maximum corresponds to the end inhale point, while the minima represent the end exhale section of the patient’s respiration (see Fig. 2-5). On the day of treatment, a reference level is specified, around which a gating window for treatment is selected. This gating window accounts for residual target motion and a setup margin chosen by the physician. The idea of individualizing gating windows for more patient-specific treatment has been looked at using the
patient’s breathing pattern and the allowable residual target motion as a guide. By using a 4DCT to determine the amount of target motion, the physician can choose a gating window that will correspond to the setup margin, in order to maximize the efficacy of the treatment [16].

There are many methods of physically reducing the intra-fractional motion, such as abdominal compression, breath-holding, visual biofeedback, and active breathing control. These methods strive to produce a more reproducible breath trace, and maximize the area in which the target can be treated. In a gated treatment, residual movement of the target will still occur within the treatment window. The ExacTrac® imaging system allows for imaging during this window, to check that the internal markers do not deviate beyond the motion accounted for in the treatment planning. The system also comes equipped with a visual-feedback system, allowing the patient to follow their own breathing pattern. The visual feedback can be coupled with audio, and has shown to reduce treatment times from 1.7 ± 0.6 min / 100 monitor units (MU) to 0.9 ± 0.2 min / 100 MU [39].

2.4.2 Respiratory Gating Benefits and Issues

The goal of respiratory gating is to reduce the probability of delivering dose to normal tissue and also underdosing the target [7]. However, accomplishing these goals requires the use of many specialized techniques and types of equipment. Gating is heavily affected on the patient’s ability to have a reproducible breath trace throughout the treatment, as well as day to day. Several reports have stated that the benefits of gated beam delivery do not outweigh the increase in treatment time and complexity for patients with motion amplitudes less than 20 mm [7, 40, 41]. In these cases, it has been suggested to apply treatment to the entire ITV that has been outlined on the planning CT, rather than treating only during a fraction of the breathing cycle. For all cases, patient-specific tumor motion assessment is recommended, as it can be used to determine if the patient would benefit from gated treatment, to quantify the residual motion, and design margins for treatment planning, and to quantify and account for phase shifts between the tumor motion and respiratory signal [7].
2.5 Imaging Modalities

As is suggested by its name, image guided radiation therapy (IGRT) requires the use of several imaging modalities throughout a patient’s treatment course. This section will cover the issues and theory of imaging used in SBRT, including computed tomography, on-board imagers, and the ExacTrac® System.

2.5.1 Computed Tomography

As mentioned in the previous chapter, 4DCT is the most common method of obtaining a set of planning images for SBRT. Other methods for accounting for respiratory motion exist, such as slow CT scanning or inhale and exhale breath-hold CTs, but this section will deal primarily with the theory and issues in dealing with the 4DCT.

No matter the method of scanning chosen, imaging artifacts will be present in some form or another. Metal implants or other objects located in the scan area with a high atomic number can result in artifacts on the resulting image set. Motion artifacts during the simulation imaging are caused primarily by respiration, cardiac function, peristaltic activity, and organ filling and emptying [7]. The transaxial rotation of the CT tube and the motion of this internal anatomy are asynchronous, which will result in improper depictions of the organ and tumor volumes.

Figure 2-6: The effect of moving 12 mm spherical target on a standard axial scan (A) versus images acquired during a 4DCT scan (B). (C) shows the axial slices of
a 4DCT of a moving spherical target. The spiral pattern results from the reconstructions over a full rotation of the CT, and show a decreased density around the outer surface, due to averaging effects. Reproduced from [42].

AAPM Task Group-101 recommends that the typical scan length should extend at least 5-10 cm superior and inferior beyond the treatment field borders. This scan length may be extended to 15 cm beyond the borders in order to properly model the patient’s internal anatomy, but can be chosen using a scout view image before scanning. This scan length should include all possible organs at risk, so as to account for these structures during the treatment planning and evaluate them with the use of a dose-volume histogram (DVH). While CT slice thickness can affect the overall accuracy of the patient setup, 1-3 mm is generally taken as an acceptable value. This value should not exceed 5 mm, as shown through preliminary phantom studies for the Novalis TX treatment machine [20]. When a particular voxel contains more than a single type of tissue, partial volume effects can be seen (See Fig. 2-7). Increasing the resolution by reducing the slice thickness can lessen this effect. If a target falls within particular detectors during the initial positioning and scan, partial voluming will be less visible.

![Figure 2-7: Effect of partial voluming at the border of two types of tissues. The white line represents the structure border and causes the detectors to average the tissues into a single value.](image)
The scan requires a stable patient breathing pattern. Irregularities in the patient respiration signal from breath to breath factor into the residual motion artifacts [42].

During this time, the system continuously takes projection data at the first couch position for a time known as the cine duration. The duration is determined by the total length of the patient breathing cycle plus the rotation time of the CT tube. For a couch with a transit time of one second between positions, the total scan length ($T_{\text{scan}}$) is given by:

$$T_{\text{scan}} = \left( \frac{N_{\text{slices}}}{N_{\text{detector rows}}} \right) \times (T_{\text{cine}} + 1 \text{sec}) - 1 \text{ s}$$ (2-5)

and $N_{\text{slices}}$ is the longitudinal length of the scan, $N_{\text{detector rows}}$ is the number of detector rows, and $T_{\text{cine}}$ is the duration at each couch position given by the motion period plus the time for a full CT gantry rotation. At each of these couch positions, a number of images determined by the cine duration and the time interval between couch positions are reconstructed. These images are then retrospectively sorted based on the phase of the respiration cycle, as described in Section 1.2.3, typically with a 10% tolerance. This tolerance also contributes to motion artifacts in the final image set [42].

### 2.5.2 BrainLAB ExacTrac® 6D Image Guidance System

The initial step in a typical patient setup procedure requires that the patient be positioned according to the infrared (IR) body markers that reference the treatment isocenter.

The IR component of the ExacTrac® System utilizes two precalibrated IR tracking cameras mounted to the ceiling at the base of the patient couch. These cameras are used to triangulate the position of IR body markers placed on the patient. The body markers are 15 mm plastic spheres coated with an IR reflective surface. The patient can also be outfitted with several accessories (See Fig. 2-8) depending on the treatment to aid in precise positioning.
The process of using the IR markers begins with the calibration of the system at the beginning of the day. The BrainLAB ET Isocenter Phantom (See Fig. 2-9) is lined up with the room lasers, giving the ExacTrac® System an infrared coordinate reference with regards to the patient couch and linac.

Figure 2-9: BrainLAB ET Isocenter Phantom. The five IR markers are placed with a known configuration and point in space, allowing for the determination of an IR coordinate system.
If the focal point of the IR camera is known, a single camera can only
determine a virtual line along which a marker is located. Therefore, to determine
the actual superposition of the IR marker, a second camera simultaneously
overlays a second virtual line, which intersects at the position of the marker (see
Fig. 2-10A).

![Diagram](image)

**Figure 2-10: Determination of the location of an IR marker using dual IR
cameras. When the location of the focal spot is known, virtual lines can be used to
locate the position of single IR reflectors (A). Ambiguities can appear for certain
setups (B) with multiple intersections, leading to the detection of ‘ghost markers.’

Depending on the geometric shape of the markers, this process can lend
itself to ambiguities, such as several markers being aligned along the same virtual
line, undeterminable rotations of the patient, or markers hidden from view.
ExacTrac® has several methods to minimize the possible ambiguities, which
compare the IR setup to the true marker positions, known through the original CT
data set or initial patient setup. This comparison can minimize the effect of ‘ghost
markers’, which are created when there are other possible intersections of virtual
lines (See Fig. 2-10B).

In addition to positioning ambiguities, the marker shift resulting from
positioning reproducibility, patient motion, or skin shift can lend itself to
isocenter variability. This problem is resolved by taking subsets of three or more
markers (the minimum requirement to define the orientation of a rigid structure).
These subsets are then subsequently weighted to determine a proper isocenter.
\[
[X \ Y \ Z] = \frac{\sum w_s \cdot [X \ Y \ Z]_s}{\sum w_s}
\]

(2-6)

\[w_s = \frac{n_s}{\|d_s\|}\]

Where \([X \ Y \ Z]\) are the isocenter coordinates, \(w_s\) is the weighting factor, and \(\|d_s\|\) is the Euclidian norm of the deviation between the reference isocenter and subset isocenter. This method results in a minimization of localization errors versus the unweighted strategy [38].

Once the patient is initially setup using the IR camera system, the target position is checked. The imaging component of the ExacTrac® system incorporates two kV X-ray units located in the floor on either side of the linear accelerator (See Fig. 2-11).

![X-ray tube and housing box used for kV imaging](Reproduced from ExacTrac® Clinical User Guide).

The relationship between the kV X-ray images and the DRRs is established using the pinhole camera model:

\[
u = \frac{p_{21}X + p_{22}Y + p_{23}Z + p_{24}}{p_{31}X + p_{32}Y + p_{33}Z + p_{34}}
\]

\[
u = \frac{p_{21}X + p_{22}Y + p_{23}Z + p_{24}}{p_{31}X + p_{32}Y + p_{33}Z + p_{34}}
\]

(2-7)
Where \((x, y, z)\) are the three-dimensional coordinates of a point in the object being imaged, and \((u, v)\) are the two-dimensional coordinates of the kV X-ray projections onto the detectors. \(P_{ij}\) are unknown parameters, which are determined daily through the use of the X-ray calibration phantom, which comes embedded with implanted markers with known relative positions. The unknown parameters can then be determined mathematically \([43]\) to allow for coordinate conversion for different markers and objects.

2.6 Combination of Uncertainties and Summary of Theory

In the validation of the ExacTrac® Adaptive Gating System for use with SBRT, the uncertainties needed to be broken down into individual components. The system was evaluated under ideal static conditions to determine the effects of the isocenters \(E_{\text{Iso}}\), positioning algorithm \(E_{\text{Fusion}}\), and CT \(E_{\text{CT}}\) and treatment planning uncertainties. For the static phantom case, these components could be added in quadrature to determine an overall uncertainty.

\[
E_{\text{total}}(\text{Static}) = \sqrt{E_{\text{Iso}}^2 + E_{\text{Fusion}}^2 + E_{\text{CT}}^2} \quad (2-8)
\]

For gated SBRT, a static CT can no longer be used, and the term must be replaced with a 4DCT term \(E_{4\text{DCT}}\). In addition to replacing this term, gating window size and tumor motion \(E_{\text{Gating}}\) affect the overall accuracy of the system. This leads to an updated formula for gated procedures.

\[
E_{\text{total}}(\text{Gated}) = \sqrt{E_{\text{Iso}}^2 + E_{\text{Fusion}}^2 + E_{4\text{DCT}}^2 + E_{\text{Gating}}^2} \quad (2-9)
\]

An end-to-end test can give the total uncertainty, but by limiting the control elements, the individual uncertainties can be systematically determined. This process is not exact, but by assuming that experiments can be done to isolate each term, a conservative estimate of the uncertainties can be obtained.
REFERENCES


Chapter 3

Materials and Methods

Chapter 3 details the physical components and the method of executing the experiments accomplished. Stereotactic body radiation therapy requires the use of many different types of equipment, starting with the imaging equipment, the linac, the gating devices, and in-house and commercial software. The first tests were done to describe mechanical accuracy of the linac and couch. The next steps involved quantifying uncertainties for the case of a static phantom, and then a phantom under gated conditions. During the course of the phantom studies, patient introduced errors could not be taken into account, so in order to quantify the ExacTrac® system’s ability under gated conditions, a patient study was completed.

3.1 Equipment

This section will specify the various equipment used in a typical SBRT treatment, as well as the specific tools used for all the experiments completed.

3.1.1 CT and RPM

All planning CTs were done on a Philips Brilliance CT Big Bore (Koninklijke Philips Electronics N.V., Amsterdam, Netherlands) which was equipped with the Varian® Real-time Position Management™ (RPM) system (Varian Medical Systems, Palo Alto, CA). The seventh generation CT was capable of 16-slice per revolution acquisition, up to a 60 cm scan field of view, and respiratory correlated imaging. In conjunction with the RPM software, the system was able to retrospectively sort all of the CT images into corresponding phase or amplitude bins. Unless noted otherwise, 4DCT scans were done using the default value of 3 mm slice thickness, with a 0.44 second tube rotation time, and phase binning.
The RPM software sampled the IR marker box placed on the phantom at a rate of 20 Hz.

3.1.2 Treatment Planning Software

Treatment planning for each of the experiments was done using the BrainLAB iPlan® Treatment Planning Software (BrainLAB AG, Feldkirchen, Germany). Initial stereotactic localization was done in RT_Image to provide a reliable coordinate system, while contouring of internal structures and dose prescription was completed in RT_Dose. Localization of the internal structures was completed using a combination of auto-segmentation on a zoomed area around the target and manual painting. All plans, unless otherwise noted, used a single conformal beam in the AP direction, as complex beam arrangements were unnecessary.

3.1.3 Linac

The linac used for each of the experiments was the Novalis Tx™ linac (Varian Medical Systems, Palo Alto, CA). Mounted under the collimator jaws was the Varian HD120 MLC (Millennium Multileaf Collimator) comprised of 2.5 mm leaves. The MLC was capable of delivering static or dynamic treatments with a maximum field size of 15 cm x 15 cm. Dose rates of up to 1000 MU/min were possible for specialized treatments, such as SRS and SBRT, in order to minimize treatment times.

3.1.4 ExacTrac® and Accessories

The experiments required several accessories necessary for stereotactic and gated treatments. For stereotactic treatments, a special head and neck base plate and couch extension are used. A special IR cranial array with six asymmetrically spaced markers is rigidly attached to this base plate for localization purposes.

For extracranial procedures, a reference array is used to provide a reference frame for the markers placed on the patient’s skin. This array is horseshoe shaped with six asymmetrically placed markers which are meant to be
placed slightly above the patient’s abdomen without touching. The array attaches to the side of the couch through a screw assembly and allows for adjustment above the couch.

![Figure 3-1: Both the cranial array (A) and the reference array (B) are outfitted with six IR markers.](image)

### 3.1.5 Phantoms

The ExacTrac® system and its corresponding accessories were designed for use with live patients, but were tested and evaluated using phantoms designed to replicate treatment conditions.

**PHANTOM CONSTRUCTION**

The first phantom used was composed of 14 polymethyl methacrylate (PMMA) slices of 1 cm thickness. Slices could contain a number of internal marker types, including spherical tungsten or integrated steel strips. The constructed PMMA block was then attached to the BrainLAB gating phantom platform through the use of securing screws at the base of the phantom.
Figure 3-2: PMMA phantom. Each slice can contain a number of slots or internal markers, allowing for a unique configuration specific for the user’s purpose.

The ET Gating Phantom (BrainLAB AG, Feldkirchen, Germany) was set up by mounting the 14 PMMA onto a moving platform. Embedded in the phantom were two 4.5 mm tungsten ball bearing markers. Three of the slices included slight recesses with enough room to place film for dosimetric measurements. One of the film plates was placed directly below the upper tungsten marker.
The ET Gating Phantom was made up of two moving platforms: a vertical marker plate and a horizontal drive plate (See Fig. 3-4). The drive plate was a longitudinally shifting platform which held the PMMA plates and simulated internal motion due to respiration. The vertical was outfitted with a flexible gating sheet and body markers to mimic the external respiratory motion.

The next series of static tests used a Rando® anthropomorphic head phantom. The Rando® phantom consisted of a human skull cast inside a soft-tissue equivalent material. This urethane material was designed to mimic the effective atomic number and mass density of muscle tissue with randomly distributed fat. The head was made up of 10 slices of approximately 2.5 cm (1”) thickness, with each slice containing pockets designed to accept 0.5 cm (3/16”) implants. The implants typically used are fiducial markers, thermo-luminescent dosimeters, or bone, air, and tissue-equivalent inserts.
2.5 mm lead ball bearings were used as the internal fiducial markers, and were inserted into tissue-equivalent inserts. Three of these markers were placed behind the nasal cavity in the phantom, approximately around the center of the skull in the 3rd and 4th cranial slices.

![Anthropomorphic head phantom used for testing positioning fusion algorithms. The lighter colored pegs placed in an array could be removed and replaced with markers, dosimeters, or different density inserts.](image)

**Figure 3-3: Anthropomorphic head phantom used for testing positioning fusion algorithms. The lighter colored pegs placed in an array could be removed and replaced with markers, dosimeters, or different density inserts.**

**PHANTOM IMAGING AND PREPARATION**

A thermoplastic mask was needed for the phantom to be treated framelessly. The mask was created using no spacers between the base and the face of the mask in order to minimize unexpected shifts. Once the mask had set, the phantom was ready for the planning CT. The phantom was placed into the H&N Base plate and onto the Imaging Couch Top Frameless Extension, both parts of the BrainLAB Head & Neck Treatment System. During fitting for masks, the plastic will shrink slightly, and requires the use of spacers. Plastic spaces of 2 mm thickness were used with the anthropomorphic phantom. The H&N Localizer and
Target Positioner box was then attached to the base plate and secured. Markers were also placed directly onto the mask for use with a 3-point setup, in the event of the localizer box not being used. The slice thickness was then chosen, and the total images limited to no more than 300, which was the software limitation for the treatment planning system. The images were then exported to the iPlan® treatment planning system. The phantom slices were not taken apart at any point during the measurements to minimize marker and slice shifts.

The gating phantom did not require the use of a stabilizing mask, and could be placed directly onto the couch for both the CT planning scan and treatment irradiation. Scanning parameters were chosen for a 4DCT, and the phantom was imaged as described in Section 1.2.3. Using the control software for the phantom (ExacTrac® Gating Phantom 1.0.0), the displacement of the platforms could be set, as well as the cycle length, pattern, and phase differences between the two platforms.

3.1.6 Film QA

The film used for all hidden target and WL experiments was the GAFCHROMIC® RTQA2 (International Specialty Products, Wayne, NJ) film. The film was self-developing with a dynamic range of 0.02 Gy to 8 Gy. The film was chosen for its robust nature in regards to storage and ready availability. For experiments requiring higher sensitivity, such as the high tumor speed or the field profile scans, GAFCHROMIC® EBT2 (ISP, Wayne, NJ) film was used. This film is designed for use from 1 cGy to 10 Gy, but can be used at higher doses by switching color channels. The films were scanned in on an Epson Expression 1680 Professional scanner (Seiko Epson Corporation, Nagano, Japan) as an 8-bit grayscale image at 200 dpi.

ANALYZATION SOFTWARE

Once the films were scanned into a digital format, a MATLAB® script was used to quantitatively analyze the results. In preliminary versions of the script, a routine named Fiximg.m was used to modify the contrast and image
intensity. A 2D median filter using a 9x9 pixel array was then used to reduce the ‘salt-and-pepper’ noise of the image, while retaining edges. Another routine, Drawing.m, overlaid the film exposures with circles corresponding to the radiation field and the radio-opaque spheroid. A correction factor of $\frac{8}{7.5}$ was applied to account for magnification from the cone to the film plane.

![Figure 3-5: Conversion of scanned film after image enhancement and median filter.](image)

Secondary versions of the software included a graphical user interface (GUI) to streamline analysis. The script utilized an image contrast window to allow the user to manually adjust the visualization of the film exposure, and allowed the input of individual couch or gantry angles. The difference in center points between each of the circles was calculated and converted into a millimeter value.

These values then needed to be converted in to the couch’s frame of reference, in order to give a correction.

\[
\begin{align*}
    r &= \sqrt{x^2 + y^2} \quad (3-1) \\
    \phi &= a \tan 2(y, x) \quad (3-2) \\
    \phi &= \phi_0 + \phi_1 \quad (3-3) \\
    x &= r \cos(\phi) \quad (3-4) \\
    y &= r \sin(\phi) \quad (3-5)
\end{align*}
\]
Where $\phi$ is the angle formed with the lateral axis and $\varphi$ is the couch angle.
The coordinates were then corrected on the film for their position relative to the gantry angle.

\[
x' = x \cos(\theta) \quad (3-6)  
y' = y \quad (3-7)  
z' = -x \sin(\theta) \quad (3-8)
\]

Where $x'$ is lateral position, $y'$ is superior-inferior direction, and $z'$ is vertical direction.

### 3.2 Isocenter Evaluation Experiments

Before examining the accuracy associated with a gated treatment, experiments were done on the treatment machine itself, in order to map out the deviations associated with the mechanics of both the treatment couch and the rotation of the gantry head. The tracking of the couch movement was designed to give an estimate of the mechanical capability, while the overall uncertainty with regards to the isocenters (the $E_{iso}$ term of Equations 2-8 and 2-9) was obtained by combining the mechanical and coordinate deviation uncertainties in quadrature.

#### 3.2.1 IR Tracking of Couch Movement

For traditional treatments, the rotation of the couch and gantry must keep the machine isocenter within a 2 mm sphere, so mechanical shifts occurring during the treatment can be effectively ignored. In stereotactic treatments, this isocenter sphere is reduced to 1 mm, and requires strict maintenance. By monitoring the deviations of the couch using an external tool, a better understanding of the uncertainties seen in treatments can be obtained.

The IR markers and camera system can provide an external modality for viewing relative sub-millimeter shifts, so when zeroed at a particular couch position, the mapping of the couch’s movement as it rotates can be acquired.

Prior to the experiment, the ExacTrac® system was calibrated using the daily procedure as described in Section 1.2.3.8. The Rando® phantom was placed
into the H&N Base plate which was connected to a treatment couch extension. The cranial IR array was attached directly to the base plate.

Once the assembly was zeroed at the center couch position (0°), 15° steps were taken in either direction to ±90°. At each position the relative shift of the IR markers was noted. This experiment was repeated five times to gain a standard deviation for each position.

Initially, the cranial array did not sit properly over the phantom and mask. The caudal side of the array pressured the shoulders of the anthropomorphic phantom, causing approximately a 5 mm shift in the vertical direction, and a 1 degree offset in the longitudinal and latitudinal angles. This was remedied by removing the shoulder slice from the phantom and creating a new PMMA base. This required the creation of a new thermoplastic mask to accommodate the changes in physiology.

3.2.2 ExacTrac® X-ray Tracking of Couch Movement

At the same time as the IR marker shifts were noted as a function of couch angle, two other modalities were used to check the relative couch shift. Two algorithms used in the ExacTrac® software utilize different aspects of the target phantom (the embedded internal markers and the bony anatomy of the phantom) to determine a patient correction vector.

Shifts were noted at every 15° up to ±90°.

3.2.3 Daily Winston Lutz Check

The morning quality assurance using the Winston-Lutz check allows for a qualitative check of the accuracy of the radiation isocenter. All of these tests are stored, and can be evaluated to give a quantitative assessment of the deviation of the radiation isocenter as a function of couch and gantry angle.

During the morning calibration of the ExacTrac® and Novalis TX™ treatment system, the coincidence of several isocenters are checked. The film-based Winston-Lutz test compares that the radiation beam agrees with the laser alignment. In order to perform the test, the BrainLAB phantom pointer was attached to the couch mount. The pointer was equipped with a radio-opaque
sphere centered under the targeting cross-hairs. Using the precision positioning knobs (See Fig. 3-6) on the mount, the cross-hairs on the pointer were aligned to the laser isocenter.

![Figure 3-6: Preparation for alignment of WL pointer (A) with lasers. Precision positioning knobs (B) are used to shift the pointer along the lateral and vertical axes. A properly aligned WL pointer can be seen on the right.](image)

A film mount was attached to the gantry head, with a strip of radiochromic film clamped behind the phantom pointer. A stereotactic conical collimator of 7.5 mm diameter was attached to the gantry accessory tray slot. The film was then irradiated with 600 MU, in order to produce a projection of the sphere within the stereotactic radiation field. The field size was chosen to allow for a simple tolerance check, where the machine would be considered acceptable when the spheroid did not come into contact with the edge of the radiation field.

Films taken at gantry angles of 0° and 180° were used to determine shift deviations in the coronal plane, while gantry angles of 90° and 270° used to evaluate the sagittal plane. Couch angles of 0°, 45°, and 315° were also used with the 0° gantry angle. During monthly quality assurance checks, a total of eight exposures were taken, which included these angles as well as a two combinations of couch and gantry rotation. All exposures during the study were collected by technicians at the MUHC.
Winston-Lutz films from the month of October 2010 were evaluated using the developed in-house MATLAB script mentioned in Section 3.1.6. A total of 71 exposures were used over the 31 day period. Films taken at gantry angles of 0° and 180° were used to determine shift deviations in the coronal plane, while gantry angles of 90° and 270° used to evaluate the sagittal plane. Couch angles of 0°, 45°, and 315° were also used with the 0° gantry angle.

Because the central axis of the radiation beam should not change with relation to the gantry head for any given angle, the Winston-Lutz exposures can be used to track the head sag and isocenter shifts of the gantry and couch, giving the mechanical deviation component of the $E_{iso}$ term.

### 3.2.4 The Hidden Target Test

Similar to a Winston-Lutz test, the hidden target test is designed to determine uncertainties involved with a system. However, the hidden target test is primarily used for end-to-end tests, which include all aspects of a treatment, from CT scanning to beam alignment with the target.

The ‘hidden target’ refers to a small object that is placed inside a phantom to be the center of the treatment process. The target is localized in the treatment planning system after the planning CT. A single beam shaped by a stereotactic cone (See Fig. 3-8) is used to treat the target and is aligned to the center of mass of the target visualization.
A sheet of radiosensitive film is placed in line with the radiation beam behind the target. For moving targets, the film is placed in a stationary position outside the phantom, while for static cases, the film can be placed directly inside the phantom. After the film has been irradiated during the treatment, an image of the radiation field will be present, as well as the shadow of the ‘hidden target’. The end-to-end accuracy of the system in question can be evaluated from the deviation seen between the center of the field and the center of the shadow of the target.

3.2.5 Coordinate Deviation of the ExacTrac® System

The accuracy of the ExacTrac® System relies on consistent calibration throughout its use. The alignment of the laser with the calibration phantom and
WL pointer define the coordinate system around the isocenter and inherently the final accuracy of the patient position. An option in the ExacTrac® calibration software allows the user to align the WL pointer with the lasers.

A calibration image (70 kVp, 70 mA, 100 ms) was acquired, after which the software automatically delineates the embedded sphere and calculates the deviation from the center of the detector (See Fig 3-9). By assuming that the X-ray tube and detectors remain stationary throughout the measurement and repositioning the WL pointer as per the daily calibration, the calculated numbers can show the deviation resulting from the laser positioning. A comprehensive look at the coordinate deviation could be determined by repeating the alignment 20 times.

![X-ray image of WL pointer from Tube 1. The embedded sphere and expected center are compared in the ExacTrac® software.](image)

**Figure 3-8:** X-ray image of WL pointer from Tube 1. The embedded sphere and expected center are compared in the ExacTrac® software.

### 3.3 Image Fusion Experiments

The next step after evaluating the isocenter uncertainty component was to look at the ability of the software’s different fusion algorithms to correctly position the patient. For SBRT treatments, particularly gated treatments for use
with this thesis, internal marker fusion is used to obtain the $E_{\text{Fusion}}$ term from Equations 2-8 and 2-9.

### 3.3.1 Relative Comparison of Fusion Methods on a Static Phantom

The simple cubic PMMA phantom was used for the comparison of fusion methods. The phantom was placed at the isocenter with the use of the lasers, and confirmed through the use of the ExacTrac® stereoscopic X-ray system. The X-ray tubes used modifiable presets depending on the body part being imaged to gain the best possible image contrast for the fusion. Thorax Standard (120 kV, 100 mA, 160 ms) was chosen for the PMMA phantom.

On the ExacTrac® console computer, the tungsten spheroid marker was localized on the planning CT, for comparison with the ExacTrac® DRR. The internal marker was used for the initial isocenter positioning, as this method is generally accepted as more representative of the actual patient position. Because the internal marker was rigidly positioned relative to the isocenter, the isocenter shift could be calculated directly. Once this shift was calculated, the couch was automatically positioned to the proper coordinates.

A manual shift was then applied to the phantom, using the couch coordinates and the laser projections onto the graph paper to roughly measure the offset, and the IR system to confirm the shift. Shifts of up to 5 cm were applied along a single axis for each measurement. Once the shift was applied to the phantom, three different fusion algorithms were used to calculate this shift. Because the internal marker was used to initially place the phantom, the automatic and manual bony fusion methods were compared to this.

### 3.3.2 Relative Comparison of Fusion Methods on a Static Anthropomorphic Phantom

The anthropomorphic phantom was placed on the couch using Frameless Radiosurgery Positioning Array. The array was clamped to the couch and outfitted with graph paper along each face. The setup was moved to the planning isocenter based on the infrared markers on the array. The localized internal
markers were used to define the phantom shift from the rough isocenter defined by the initial infrared marker position.

The couch was systematically moved along each axis up to 5 cm, which was confirmed using the graph paper and IR system. Automatic positioning was not used. At each point, the ExacTrac® system was used to take two stereoscopic images using the ‘Thorax Std.’ preset. Automatic detection of the internal markers was used, and the markers were then fused to the planning CT position. Then, using the same image set, Automatic Bony Fusion was used to compare phantom positioning.

Each position was imaged three times before moving to the next. After each set of shifts along one axis, the phantom was returned to isocenter using automatic positioning. The positioning was then confirmed by imaging the internal markers.

3.4 Computed Tomography Experiments

The pretreatment imaging also constitutes a component of the uncertainty, and this section deals with the determination of the effects of the CT scans, for both static and gated cases. The experiments here were used to quantify the $E_{CT}$ and $E_{4DCT}$ terms.

3.4.1 Effect of Slice Thickness Using a Static CT

A total of nine CT scans were done on the ExacTrac® Gating Phantom, distributed between slice thicknesses of 2, 3, and 5 mm. Prior to each scan the ExacTrac® Gating Phantom was placed on the CT couch absent of any phantom rotations. The phantom was shifted each time in the AP direction to randomize the starting position of the scan, and introduce variations in the visualization of the target. The X-ray tube also began each scan at a random angle.

The field of view and scan length were chosen based on a scout image to include the entire PMMA portion of the phantom. Default settings for a standard scan were used.
Auto-segmentation was used to delineate the markers inside the phantom. Objects within the specified area with Hounsfield Units (HU) between 1700 and 3000 were accepted as the internal markers. A single element conformal beam plan was used with a 7 mm margin. The margin was set large enough to allow for the stereotactic cone to define the field.

After treatment planning, a hidden target test was required. The ExacTrac® system was calibrated and three WL exposures were taken to determine the baseline deviation. Each plan was executed a total of three times.

3.4.2 Effect of Slice Thickness on Four-Dimensional CT

In order to test the effect of CT slice thickness for the case of 4DCT scans, multiple CT scans were planned. Three scans were taken at three slice thicknesses (2, 3, and 5 mm) for a total of nine scans. Positioning of the phantom was done similarly to the setup of the static CT.

The ExacTrac® Phantom was set to oscillate sinusoidally in the AP direction with a maximum internal shift of 3 cm over a period of 5 seconds. No phase shifts were present between the internal motion of the phantom and the breath trace recorded by the IR camera system. The scan length was chosen as to include the minimum and maximum positions throughout the breath cycle.

3.5 Respiratory Gated Experiments

The last component of uncertainty to be evaluated concerns the effects seen during respiratory gated treatments ($E_{\text{Gating}}$). The main effects to be examined include the tumor motion and the gating window size.

3.2.10 Effect of Gating Window Size and Tumor Motion

Window size and the motion of the tumor constitute a large portion of the uncertainty in targeting in SBRT treatments. For typical treatments, the gating window and tumor amplitude can be as large as 50% and 3 cm, respectively. The ExacTrac® System was calibrated, and 3 WL exposures were taken prior to any measurements.
The treatment plan was created using a 3 mm slice thickness CT scan. Another set of plans was created with an intentional 5 mm offset from the center of mass of the imaged marker. This was done to observe the residual shift in the final position of the target with respect to the treatment beam. A hidden target test was done for a case of a static phantom to give a baseline offset. The phantom was then set to oscillate at 1 cm amplitude using a 5 second period sinusoidal breath trace. The phantom was repositioned using the gating procedure and then irradiated using window sizes of 10%, 20%, and finally 30%. These window sizes were repeated for internal motions of 2 cm and 3 cm.

Each combination of window size and amplitude was repeated three times, and evaluated as both an end-to-end test, and for individual component of window size and amplitude. The individual components were collected by subtracting out the deviation from a static HTT using the same initial positioning.

Lastly, using the 10% window size, a separate piece of GAFCHROMIC® EBT2 film was irradiated on the top of the moving phantom to visualize the effect of dose blurring as a function of tumor speed. The films were irradiated to 250 cGy. The films were then scanned using 24-bit color and dose profiles were taken along the SI axis using FilmQA™ (3cognition USA, Wayne, NJ).

### 3.6 Patient Study

Once the system had been evaluated using phantoms under ideal conditions, the next step was to compare these results with the positional shifts seen during a real patient treatment. This study would help delineate the uncertainties arising from the treatment system itself, and those uncertainties resulting from patient movement or changes in anatomy.

Data was collected from four liver SBRT patients treated on the Novalis Tx™ linear accelerator at the Montreal General Hospital over the previous two years. After treatments, all patients were removed from the ExacTrac® computer and backup on a remote server. These patients could be restored to view correction shifts on any individual treatment day. Fractions varied between three
and ten, with the total dose being between 30 and 50 Gy. A total of 23 correction shifts were collected from the patients.

Prior to treatment, internal fiducial markers were surgically implanted into the patient’s liver, with each patient containing between two and seven. For the CT, each patient was outfitted with six external, radiopaque body markers on their abdomen, and placed on the scanning table using the BodyFix immobilization system. The scans were done using the retrospective gating mode using 3 mm slice thickness.

The target volumes were defined within the iPlan® treatment planning software. Important structures such as the heart and bowels were also included in the dose volume histogram.
Figure 3-9: Treatment planning portion of the liver SBRT patients. Figure A shows the contouring of the target volumes (GTV – pink, CTV – purple, PTV – red). Figure B shows the beam configuration around the targeted area, as well as the heart (orange), small bowel (yellow), and the healthy liver (blue).

The liver lesions were contoured, as well as each of the important surrounding structures, including the heart, bowel, and healthy tissues (See Fig. 3-10A). The CTV and PTV were outlined around the GTV. Beam configurations implemented limited couch and gantry angles to minimize treatment times. Up to 10 conformal beams (See Fig. 3-10B) were chosen for treatment, as well as a boost if it was necessary.

On each day of treatment, patients were repositioned on the couch using the gating procedures in the ExacTrac® software. With the exception of Patient 3, a single point was used to define each of the internal fiducial markers center of mass. Patient 3 used three long markers, defined by their endpoints. Once the patient’s breath trace had stabilized, the reference level and typically two additional imaging levels were chosen. Corrections were based only on the
reference level images. Initial positioning was then accomplished using the external markers location in relation to the reference array. Following this procedure, the dual X-ray tubes were used to image the internal fiducials for precise positioning (See Fig. 3-11). If the determined offset was less than the tolerable 5 mm, treatment continued as planned. If not, the calculated shift was applied to the system and checked again with the X-ray tubes. This process would be repeated until the offset was reduced below tolerance levels.

![Figure 3-10: ExacTrac® images of a liver SBRT patient. Internal fiducials (A) are used to complete precise positioning, while IR markers (B) are used to track the patient’s movement and breath trace.](image)

After the treatment was completed, shift data from the software was exported from the computer to an archived hard disk. This data was used to determine the ability of the system to initially position the patient and then correct the position based on internal information. These values can then be compared directly with the results from the phantom experiments to quantify the patient component of uncertainty.
This chapter will review the experiments described in Section 3.2 of this thesis. The first sections will deal with the isocenter shifts, concerning the couch and linac. The next sections will continue to discuss the uncertainties associated with the ExacTrac® system and typical SBRT treatments with regards to a static setup, and then under gated conditions. All uncertainties are given to one standard deviation.

4.1 Isocenter Evaluation

The mechanical and radiation isocenter deviations were evaluated using experiments implementing the ExacTrac® equipment and film exposures. The coordinate deviations of the ExacTrac® system were also included in the study.

4.1.1 IR Tracking of Couch Movement

The experimental method described in Sections 3.2.3 and 3.2.4 was intended to give an understanding of the deviations associated with a typical couch rotation. These deviations were tracked using two external validation techniques making up the ExacTrac® system, the IR camera system, and the stereoscopic X-rays.
Figure 4-1: Comparison of the infrared detected position as a function of the couch angle.

The smallest detected deviation was around the couch angle of 0°. The deviation detected here was largely due to the daily fluctuations of the calibration. As the couch rotated to larger angles, the detected shifts increased in both magnitude and uncertainty. The largest component of the detected shifts appeared in the lateral direction. Including the error bars, which were taken from multiple measurements of the same point, the deviation at each point in the couch’s rotation was less than the 1 mm required for stereotactic radiosurgery.

4.1.2 X-ray Tracking of Couch Movement

At the same time as the IR readings were being taken of the phantom position, stereoscopic X-rays were being recorded. These measurements could be compared directly to the results from the previous section.
Figure 4-2: Comparison of total deviation detected by the two image fusion methods. The blue points represent the internal marker fusion, while the pink shows the bony fusion method.

Similar to the infrared detection, the detected phantom shifts increase at larger couch angles. Both methods show that the mechanical aspect of the linear accelerator is capable of accuracy of 1 mm. The ExacTrac® system is capable of detecting these mechanical shifts, which is important in determining the final correction vector.
4.1.3 Evaluation of Daily WL Films

Figure 4-3: Scatter plot for the daily Winston-Lutz film exposures during the month of October 2010. The blue points represent a couch position of 0°, and multiple gantry angles. The red points represent a gantry angle of 0° and couch angles of 45° and 315°.

The individual exposures were rotated into the couch’s frame of reference as described in Section 3.1.6, and the average shift and standard deviation seen in the exposures was calculated to be 0.3 ± 0.3 mm, -0.3 ± 0.4 mm, and -0.2 ± 0.4 mm in the lateral, superior-inferior, and vertical directions, respectively. These measurements were used in the subsequent laser calibration to reduce the overall deviations seen at these various gantry and couch angles. The WL pointer was placed at the current laser isocenter, and then moved to the newly calculated
position. The lasers were then realigned to the center of the pointer. Over the next month, the daily WL films were again analyzed for shifts. The resulting values were $0.3 \pm 0.4$ mm, $-0.1 \pm 0.4$ mm, and $-0.2 \pm 0.3$ mm for the lateral, SI, and vertical directions, respectively. Based on the standard deviations, the residual uncertainties tend to be on the order of 0.3-0.4 mm for any given direction.

Figure 4-4: Scatter plot of the daily Winston-Lutz exposures for the month of October 2010. The points have been organized by the measured deviation at the individual couch or gantry angles and rotated into the couch’s frame of reference.

The points along the plane of motion of the couch can be compared to the results of Sections 4.1.1 and 4.1.2: the IR and X-ray tracking of couch movement. Discrepancies can be seen between different angles for the couch versus what was seen in Figure 4-4. At 0 longitudinal shifts of up to 1 mm are visible, but the detected shifts at couch angles of 315° (-45°) are on the same order, whereas the IR system showed a differences of 0.4 mm between the two angles.
Comparison of the lateral directions between the IR and WL measurements shows similarity in both direction and magnitude. Vertical measurements for the couch angle of 0° show deviations less than 0.5 mm in the negative direction, due in part to the sagging of the linac head as it rotated about the couch.

4.1.4 Coordinate Deviation of the ExacTrac® System

When using the $E_{\text{iso}}$ term to define the total uncertainty, both the mechanical deviation of the linac and the coordinate deviation of the ExacTrac® system contribute.

The ExacTrac® component of the $E_{\text{iso}}$ variable (Section 2.6) was determined by averaging the alignment coordinates and taking the standard deviation over the course of 20 measurements. The three-dimensional coordinates determined were $0.1 \pm 0.2$ mm, $0.2 \pm 0.1$ mm, and $0.2 \pm 0.1$ mm in the lateral, SI, and AP directions, respectively. This gives a total offset vector of $0.3 \pm 0.2$ mm.

The coordinate deviation and the mechanical deviation can then be combined in quadrature to give a value of $0.6 \pm 0.7$ mm for the isocenter term. Independence is assumed because the two systems are separately mounted, although this value should be viewed as a conservative estimate.

4.2 Image Fusion

Once the isocenter uncertainty component was determined, the effects of the image fusion and the system’s ability to position the patient needed to be established. While internal markers are used for SBRT positioning, this modality was compared with the ExacTrac® system’s other positioning tools: the bony anatomy fusion and the IR system.
4.2.1 Relative Comparison of Fusion Methods on a Static Phantom

In the relative comparison, the two main positioning modalities, the internal marker (IM) and bony anatomy fusion were also checked against a manual fusion. Measurements were done with separate calibrations, with the first set taken relative to each other, while the second was taken to show absolute deviation for each method.

![Image Fusion Comparison](Image)

Figure 4-5: Comparison of Internal Marker, Bony Automatic, and Manual fusion methods available on the ExacTrac® software.

The measurements are from a single calibration, with the standard deviation taken from the entire set. The total deviation is shown with respect to the average internal marker deviation because the internal marker was used to initially position the phantom. The internal marker fusion and manual fusion had an average difference of less than 0.1 mm, while the automatic fusion position differed from the internal marker by an average of 0.3 mm. Each modality exhibited a standard deviation of less than 0.2 mm.
Figure 4-6: Comparison of the Internal Marker, Bony Automatic, and Manual fusion methods using a separate calibration of the ExacTrac® system.

The internal marker was used to initially position the phantom, but in this plot the different methods are presented in absolute deviation rather than with respect to the initial positioning method. The manual fusion again shows the smallest offset (0.2 ± 0.1 mm), while the bony fusion method differs by 0.6 ± 0.1. All methods had a standard deviation of less than 0.2 mm.
### 4.2.2 Comparison of Image Fusion Methods for a Static Gating Phantom

Once the fusion methods were compared for points at the isocenter, the fusion methods needed to be compared for their ability to detect and correct for known shifts.

Table 4-1: Table summarizing the detected offset by IR marker, internal marker, and auto fusion from the actual position for shifts along any of the three couch axes.

<table>
<thead>
<tr>
<th></th>
<th>Infrared Marker</th>
<th>Internal Marker</th>
<th>Auto Fusion</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lat (mm)</td>
<td>SI (mm)</td>
<td>AP (mm)</td>
</tr>
<tr>
<td></td>
<td>Lat (mm)</td>
<td>SI (mm)</td>
<td>AP (mm)</td>
</tr>
<tr>
<td></td>
<td>Lat (mm)</td>
<td>SI (mm)</td>
<td>AP (mm)</td>
</tr>
<tr>
<td>Lateral Shift</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10 mm</td>
<td>Average</td>
<td>0.1</td>
<td>-0.2</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>20 mm</td>
<td>Average</td>
<td>-0.1</td>
<td>-0.3</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>50 mm</td>
<td>Average</td>
<td>-0.1</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>Std Dev.</td>
<td>0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>SI Shift</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10 mm</td>
<td>Average</td>
<td>0.0</td>
<td>-0.2</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td>20 mm</td>
<td>Average</td>
<td>0.1</td>
<td>-0.2</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>50 mm</td>
<td>Average</td>
<td>-0.1</td>
<td>-0.4</td>
</tr>
<tr>
<td></td>
<td>Std Dev.</td>
<td>0.0</td>
<td>0.1</td>
</tr>
<tr>
<td>AP Shift</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10 mm</td>
<td>Average</td>
<td>0.0</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>20 mm</td>
<td>Average</td>
<td>0.0</td>
<td>0.1</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.1</td>
<td>0.1</td>
</tr>
<tr>
<td>50 mm</td>
<td>Average</td>
<td>-0.1</td>
<td>0.2</td>
</tr>
<tr>
<td></td>
<td>Std Dev.</td>
<td>0.1</td>
<td>0.1</td>
</tr>
</tbody>
</table>
The IR system showed an accuracy of $0.2 \pm 0.2$ mm for shifts up to 5 cm. The largest deviations from the actual position were seen in the vertical and SI directions, with the largest total deviation arising for the largest shifts, except in the lateral shift direction. These lateral shifts remained within the stated accuracy of the IR system. However, in clinical SBRT treatments the accuracy of the IR markers depends on several factors, including the repositioning of the markers on the patient at treatment time, as skin sagging. Results from this experiment give a baseline estimate for a rigid phantom setup.

The use of internal markers also gave an accuracy of $0.2 \pm 0.2$ mm, while the fusion of the bony anatomy was $0.5 \pm 0.2$ mm. These values can be used in determining the total uncertainty of the system, as the $E_{\text{Fusion}}$ term (Section 2.6), depending on the positioning modality chosen.

4.2.3 Comparison of Fusion Methods on a Static Anthropomorphic Phantom

To allow for a more robust evaluation of the $E_{\text{Fusion}}$ term, an anthropomorphic phantom replaced the PMMA phantom, which contained bony anatomy in addition to the internal fiducial markers.
Figure 4-7: Plot showing the total detected offset from planning position on a static anthropomorphic phantom. The pink shows the results of the bony fusion, while the blue shows the results of the internal marker.

The internal markers resulted in a more consistent and smaller offset than the automatic fusion algorithm for the anthropomorphic phantom. The internal marker offset was determined to be $0.1 \pm 0.2$ mm, while the automatic fusion was $0.2 \pm 0.3$ mm. The second calibration gave comparable results for each method, showing shifts in the same direction and within 0.1 mm of each other. The first calibration, however, showed a discrepancy between the two modalities, with the automatic fusion giving an average offset that was $0.3$ mm larger than the average shift determined by the internal markers. This shift should not be considered significant as the internal markers were used for the initial positioning, and is comparable to the relative difference seen in Section 4.2.1.

The same phantom was used in Table 4-2, but the table is broken into detected offsets based on the shift applied to the system.
Table 4-2: Summary of the anthropomorphic phantom’s average deviations from its actual position based on magnitude of the shift from planned isocenter.

<table>
<thead>
<tr>
<th>Shift (mm)</th>
<th>IR (Patient Positioning)</th>
<th>Internal Marker (X-ray)</th>
<th>AutoFusion (X-ray)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>AP (mm)</td>
<td>Lat (mm)</td>
<td>SI (mm)</td>
</tr>
<tr>
<td>10</td>
<td>Average</td>
<td>0.1</td>
<td>0.2</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.1</td>
<td>0.2</td>
</tr>
<tr>
<td>20</td>
<td>Average</td>
<td>0.2</td>
<td>0.3</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.2</td>
<td>0.1</td>
</tr>
<tr>
<td>50</td>
<td>Average</td>
<td>0.2</td>
<td>0.2</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>0.2</td>
<td>0.1</td>
</tr>
</tbody>
</table>

The average detected shifts seen in Table 4-2 tend to be larger than on the static gating phantom (See Table 4-1), with the IR and IM average offsets both increasing to $0.4 \pm 0.3$, respectively, and the AutoFusion increasing to $0.8 \pm 0.3$. The internal anatomy of the phantom became more complex, and IR markers were no longer directly fastened to the phantom, but rather on the cranial array. These are suitable to use in the $E_{Fusion}$ term in calculating the total uncertainty, for shifts up to 5 cm. For use in calculating the total uncertainty in section 4.5.2, the average internal marker shift was chosen, as this is the modality used in SBRT.
Figure 4-8: Comparison of the IR, internal marker, and auto fusion positioning methods in the presence of lateral shifts.

The system showed a consistent response for each positioning modality for detecting shifts in the lateral direction. The internal markers and IR showed comparable average offsets (within 2%) (0.23 ± 0.14 and 0.24 ± 0.14 mm) from the actual position, while the image fusion modality demonstrated a larger offset of 0.49 ± 0.13 mm, with a maximum of 5% error for the 1 cm shift.
Figure 4-9: Comparison of IR, internal marker, and auto fusion positioning modalities in the presence of SI shifts.

Longitudinal shifts were all reported within 5% of the actual position using each of the positioning modalities. Both the infrared and internal marker methods experienced a slight reduction in accuracy for shifts of 50 mm. The detected shifts in this direction were the largest of the three axes, but not significantly.
In the anterior-posterior direction the internal marker method showed the smallest overall shift from the actual position. All three modalities detected shifts up to 50 mm to within 0.4 mm, and each showed standard deviations between 0.1 mm and 0.2 mm.

4.3 Computed Tomography

This section deals with the uncertainty associated with the pretreatment imaging. These experiments were used to determine the $E_{CT}$ term for static cases, and the $E_{4DCT}$ term for treatments under gated conditions (Section 2.6).

4.3.1 Effect of Slice Thickness Using a Static CT

The experiment described in section 3.4.1 was completed over three series of dates, and was combined into a single table (see below).
Table 4-3: The detected deviation calculated using WL exposures for various slice thicknesses used for SBRT.

<table>
<thead>
<tr>
<th>Slice Thickness</th>
<th>Lat (mm)</th>
<th>SI (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2mm</td>
<td>0.0 ± 0.2</td>
<td>0.1 ± 0.5</td>
</tr>
<tr>
<td>3mm</td>
<td>0.2 ± 0.4</td>
<td>0.4 ± 0.7</td>
</tr>
<tr>
<td>5mm</td>
<td>0.1 ± 0.4</td>
<td>0.1 ± 0.6</td>
</tr>
</tbody>
</table>

The static CT showed an overall accuracy ranging from 0.1 ± 0.5 mm to 0.5 ± 0.9 mm. Due to the makeup of the CT, vertical resolution determined to be the same as in the lateral direction. Uncertainties associated with these directions was assumed to be similar, and were unaffected by the changes in slice thickness. Errors seen in the longitudinal direction tended to have a larger standard deviation associated with them. This is due to the resolution along the SI axis being directly affected by the slice thickness. Yan et al. recommended using slice thicknesses up to 5 mm for use with highly accurate treatments, and the positional errors up to this slice thickness remained within usable tolerances [20].

4.3.2 Effect of Slice Thickness on a Four-Dimensional CT

The experiment from the previous section was repeated, but using a moving phantom in conjunction with a 4DCT. In addition to the hidden target test, the internal marker matching and auto-fusion were used to check the final offset of the phantom from isocenter.

Table 4-4: Positional Accuracy of the ExacTrac® Adaptive Gating System vs. CT Slice Thickness

<table>
<thead>
<tr>
<th>Slice Thickness</th>
<th>Internal Marker</th>
<th>Auto Fusion</th>
<th>Hidden Target</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lat (mm)</td>
<td>SI (mm)</td>
<td>AP (mm)</td>
</tr>
<tr>
<td>2 mm</td>
<td>0.4 ± 0.2</td>
<td>0.6 ± 0.2</td>
<td>0.4 ± 0.2</td>
</tr>
<tr>
<td>3 mm</td>
<td>0.3 ± 0.2</td>
<td>0.7 ± 0.2</td>
<td>0.4 ± 0.2</td>
</tr>
<tr>
<td>5 mm</td>
<td>0.4 ± 0.2</td>
<td>0.6 ± 0.2</td>
<td>0.3 ± 0.5</td>
</tr>
</tbody>
</table>
The use of the 4DCT allows the user to pull a section of the breathing cycle for treatment, rather than treating over the entire internal target volume. While there is still residual motion during this window, the resulting scan can be used for highly accurate treatments, such as SBRT. The largest average offsets appeared in the SI direction, similarly to the static CT. These values result from the target motion along this axis in addition to the effect of the slice thickness. The detected offset in the lateral direction increased slightly from the static case, due possibly to slight rotational errors in the setup being compounded by the phantom movement.

The 4DCT slice thickness constitutes one of the larger sources of uncertainty for the final positioning, and can be integrated into total uncertainty formula as the $E_{4DCT}$ term (Section 2.6). The value can be chosen based on which slice thickness is to be used with the treatment.

### 4.4 Respiratory Gated Effects

The final term ($E_{Gating}$) described in Section 2.6 refers to the uncertainty effects created under gated conditions during treatment. This section will deal with the determination of these effects in regards to the tumor motion and gating window size.

#### 4.4.1 Comparison of Treatment Plans using End-to-End Test

Section 3.2.10 discusses the creation of multiple treatment plans to observe the ExacTrac® system’s ability to position in the presence of imprecise localization of the target.
Figures 4-11 and 4-12: End-to-end results of variation of window size for a tumor amplitude of 2 cm. Figures include the 0.5 mm offset programmed into the treatment plan, resulting in overall deviations of greater than 1 mm.
The average difference for all window sizes in the end-to-end uncertainty between the offset treatment plans and the aligned treatment plans was 0.3 ± 0.4 mm, with the largest difference seen in the 50% window. The 50% window produced larger dose blurring in the film exposures than the smaller windows, resulting in a larger uncertainty. The intrinsic shift of 0.5 mm falls within the error bars of the average difference.

Figure 4-13: Comparison of window size (10% on the left, 50% on the right) on the dose blurring effect.

The example on the left shows a target being treated using a 10% window, while the example on the right utilizes a 50% window. The larger window increased the uncertainty in deducing the center of mass of the target shadow.
The end-to-end results of the tumor amplitude comparison show that the 10% and 20% window sizes gave similar average offsets, with the standard deviation for the first three window sizes (10, 20, 30 %) being ± 0.2 mm, and ± 0.3 mm for a 50% window. As the window size increased, the average offset also increased to greater than 1.4 mm for both the 30 % and 50 % windows. The ExacTrac® software uses 30 % as the default window size for SBRT treatments.

4.4.2 Tumor Amplitude

In addition to the effects of the size of the gating window, the motion of the tumor being targeted plays a large role in the accuracy of the system. The following section looks at the system’s ability to target and deliver a treatment beam under gated conditions with varying tumor speeds.
The plot shows that the overall relative offset is proportional to the amplitude of the tumor motion. The largest contribution was due to the shifts detected in the SI direction. These shifts are in due to the increasing target speed along this axis, affecting the targeting of the system. This is similar to the increasing effect of the target motion during CT scanning. The lateral offset decreases with increased tumor speed, the standard deviation increases to a maximum of ± 0.3 mm for 3 cm tumor amplitude. The standard deviations of the individual directions were all within ± 0.2 mm, while the overall standard deviation increased from ± 0.3 mm to ± 0.4 mm.

4.4.3 Effect of Gating Window Size and Tumor Motion

After looking at the window size and tumor motion components individually, these terms were expanded to a wider range of clinical values. Depending on the treatment being given, the $E_{\text{Gating}}$ term’s (Section 2.6) contribution to the total uncertainty can be determined from Table 4-5.
Table 4-5: Positional Accuracy of the ExacTrac® Adaptive Gating System vs. Gating Parameters

<table>
<thead>
<tr>
<th>Gating Window</th>
<th>10% Lat (mm)</th>
<th>10% SI (mm)</th>
<th>20% Lat (mm)</th>
<th>20% SI (mm)</th>
<th>30% Lat (mm)</th>
<th>30% SI (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tumor Amplitude (cm)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>0.2 ± 0.2</td>
<td>0.2 ± 0.2</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>0.75</td>
<td>0.2 ± 0.2</td>
<td>0.1 ± 0.2</td>
<td>0.1 ± 0.2</td>
<td>0.1 ± 0.2</td>
<td>0.1 ± 0.3</td>
<td>0.1 ± 0.3</td>
</tr>
<tr>
<td>1.5</td>
<td>0.1 ± 0.2</td>
<td>0.0 ± 0.2</td>
<td>0.1 ± 0.3</td>
<td>0.1 ± 0.3</td>
<td>0.1 ± 0.2</td>
<td>0.4 ± 0.4</td>
</tr>
<tr>
<td>2.25</td>
<td>0.0 ± 0.2</td>
<td>0.2 ± 0.3</td>
<td>0.1 ± 0.2</td>
<td>0.1 ± 0.3</td>
<td>0.3 ± 0.4</td>
<td>0.8 ± 0.6</td>
</tr>
<tr>
<td>3</td>
<td>0.3 ± 0.4</td>
<td>0.3 ± 0.3</td>
<td>0.2 ± 0.2</td>
<td>0.1 ± 0.5</td>
<td>0.1 ± 0.3</td>
<td>1.1 ± 1.7</td>
</tr>
</tbody>
</table>

The contribution of the window size and tumor amplitude to the overall uncertainty was less than 1 mm for all amplitudes paired with up to a 20% gating window. However, for large tumor amplitudes with the 30% window, the detected overall offset increased to a maximum of 1.1 ± 1.8 mm. Even after changing to a more sensitive film, a residual shift was detected. The ExacTrac® Adaptive Gating system showed itself to be adequate for smaller window sizes and tumor amplitudes, but for tumor amplitudes of greater than 3 cm, external precautions would be necessary to reduce the uncertainty. The next section will discuss the more of the effects of large tumor motion on the system’s ability to properly deliver a treatment.

4.4.4 Dose Profiles

The system’s ability to target and properly irradiate a target located internally was checked by looking at the dose profiles along the axis of motion (SI). Dose blurring was more readily visible along this axis.
Figure 4-16: Comparison of dose profiles for a target moving over a 5-second breath cycle.

The figure shows different dose profiles of a 10 mm field treated to 250 cGy in the bottom 10% of the breath trace. Cross-sections were done along the SI axis. Doses at the center of the field were all within ± 5% of the prescribed dose. As the speed of the target increased, dose blurring increased along the axis, particularly in the inferior direction. This shift in dose is possibly due to the slight lag of the treatment beam with respect to the triggering mechanism.

4.5 Patient Study and Summary

The initial aim of this thesis was to quantify the sources of error associated with the ExacTrac® system during an SBRT treatment. The system has been evaluated under ideal conditions, using a static phantom as well as a moving phantom reproducing the movements of a patient. However, such conditions are impossible to replicate when a living patient is involved. This section deals with
the study described in Section 3.6, and the discrepancies between the phantom studies and actual patient treatments.

4.5.1 Patient Study

The 23 correction shifts were combined into a single table, divided into the individual patients from whom the data was taken.

Table 4-6: Comparison of detected shifts for Liver SBRT patients at the MUHC hospital from 2010-2011.

<table>
<thead>
<tr>
<th></th>
<th>Pretreatment Shift</th>
<th>Corrected Shift</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Lat (mm)</td>
<td>SI (mm)</td>
</tr>
<tr>
<td>Patient 1</td>
<td>Avg.</td>
<td>-0.8</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>2.2</td>
</tr>
<tr>
<td>Patient 2</td>
<td>Avg.</td>
<td>3.3</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>1.3</td>
</tr>
<tr>
<td>Patient 3</td>
<td>Avg.</td>
<td>5.7</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>-</td>
</tr>
<tr>
<td>Patient 4</td>
<td>Avg.</td>
<td>-2.2</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>1.6</td>
</tr>
<tr>
<td>Overall</td>
<td>Avg.</td>
<td>2.1</td>
</tr>
<tr>
<td></td>
<td>St. Dev.</td>
<td>2.8</td>
</tr>
</tbody>
</table>

Each patient consisted of several fractions, except for patient 3, during which only a single fraction was positioned with the ExacTrac® system. After the initial positioning, the correction vector calculated was $6.5 \pm 4.4$ mm, larger than the $5$ mm used for treatment at the MUHC. After applying the detected shift, the new correction vector was decreased by an average of $3.2$ mm to a final offset of $3.3 \pm 2.5$ mm. Uncertainty was largest along the SI axis, both before and after correction, due to contributions from the CT planning and tumor motion.
4.5.2 Comparison of Spatial Accuracy on a Phantom vs. Patient

The end-to-end film study results from Section 4.4.1 show ExacTrac® system is capable of consistent positioning within 2 mm of the intended treatment isocenter. This corresponds well with the combination of the individual component uncertainty in quadrature described in Section 2.6.

When using typical clinical settings (3 mm slice thickness, 30% window size) on a typical patient tumor (5 second breath cycle, 1.25 cm tumor amplitude), the formula in Section 2.6 can be used to estimate the spatial accuracy of the system. This calculation uses overall correction vectors, and is not specific to a particular direction.

\[ E_{total} = \sqrt{E_{iso}^2 + E_{Fusion}^2 + E_{4DCT}^2 + E_{Gating}^2} \]

\[
E_{iso} = 0.6 \pm 0.7 \text{ mm} \quad \text{(Section 4.1.4)} \\
E_{Fusion} = 0.4 \pm 0.3 \text{ mm} \quad \text{(Section 4.2.3)} \\
E_{4DCT} = 0.9 \pm 0.3 \text{ mm} \quad \text{(Section 4.3.2)} \\
E_{Gating} = 0.4 \pm 0.5 \text{ mm} \quad \text{(Section 4.4.3)}
\]

\[ E_{total} = 1.2 \pm 1.0 \text{ mm} \]

Even looking at the corrected positions of the patients, the detected offset from the isocenter is larger, being 3.3 ± 2.5 mm. The difference in positioning ability can be attributed to patient motion and internal shifts of the fiducial markers, but could have been reduced for some cases, as in Figures 4-17 and 4-18. It should be noted that the largest observed shifts for both the phantom and patients were in the SI direction, which has been attributed mainly to the resolution of the CT scan and the primary axis of motion of the target. However, the shifts in the AP and lateral directions on the patient were also non-negligible. Due to the limitations of the axis of motion of the phantom, a direct comparison of tumors moving along multiple axes was not possible, although these directions...
should not be neglected. As Chapter 1 discussed, motion in these directions can still occur.

Figure 4-17: Two patient positioning X-rays of the same patient on different treatment days. A well contrasted image (A) shows the implanted markers clearly on a background of the patient anatomy. An improperly warmed up X-ray tube (B) can result in a salt-and-pepper images that can cause difficulty in localizing the implanted markers.

Figure 4-18: Two patient positioning X-rays of same patient on different treatment days, showing effect of beam energy. Correctly chosen beam parameters (kV, ms, mA) will result in a contrasted image, where implanted
markers can be detected easily (A). Improperly selected parameters can result in an oversaturated image, where markers are not readily visible against the anatomy of the patient.

The uncertainty in determining the center of mass of the internal markers can partly be attributed to the CT scan and the resolution used, but due to less rigidly structured markers, localization may vary from patient to patient and over the treatment period. The user input plays a large role in the final determined shift, as fiducials can be much easier to delineate in a phantom. The center of mass is chosen by the user, and may be difficult to determine for markers that have a less than rigid structure.

However, even with slight user input fluctuations, the ExacTrac® system shows itself to be a suitable tool for accurately positioning for treatments.
REFERENCES

Chapter 5

Conclusion

5.1 Summary of Thesis

An in-depth look at overall positional accuracy of the ExacTrac® Adaptive Gating system was presented by simulating the clinical extracranial treatment process. A method of evaluating the individual components’ contributions to the uncertainty was based on the built-in software accompanying the equipment and external verification through the use of a radiochromic film study. The clinical process for SBRT treatments includes contributions from the mechanical aspects of the linac, the CT scanning, the positioning algorithm, and the gating parameters chosen for treatment. Each of these steps was evaluated individually to give a quantitative assessment of the accuracy.

The mechanical isocenter was checked using the daily WL exposures over the course of a month and the WL kit used for calibration. The effect of the CT slice thickness was tested for both static CT scans and 4DCT scans. This was accomplished by planning multiple treatments on scans of different slice thicknesses. The internal marker fusion was compared to the automatic bony anatomy fusion, and IR markers as different methods of patient positioning by checking their ability to detect known shifts from the treatment isocenter. The comparisons were done using two static phantoms, the ExacTrac® Gating Phantom, and an anthropomorphic head phantom, both allowing for internal marker inserts. Chosen because of its capacity to track internal tumors directly, the system’s ability to determine position was adequate for the use of SBRT. Lastly, the ability of the ExacTrac® Adaptive Gating system to irradiate a moving stereotactic target was tested. The internal tumor amplitude and the treatment window size were the parameters tested through the use of the hidden target test.
The relationship between each of these parameters and the total uncertainty in a gated treatment can be described by combining each term in the formula from Section 2.6 in quadrature.

Each of the parameters can be chosen from tables based on the corresponding steps taken during the treatment process to create a unique estimate for the total uncertainty.

All films used were interpreted by an in-house software script written in MATLAB® to evaluate the Winston-Lutz assessment and the hidden target test, allowing for quantification of past data taken, as well as giving a non-commercial method of checking the system. The script allowed for enhancement beyond qualitative checks and a precise calculation of the target’s offset with respect to the treatment field. Many of the exposures, such as the larger window sizes and faster moving targets would not have been readily visible without the use of software enhancement.

The largest sources of uncertainty were determined to be the 4DCT and the mechanical isocenter, however for tumors with large internal amplitudes (>2.25 cm) that are treated with large gating windows (>30%) the gating parameters can contribute 1.1 ± 1.8 mm or greater.

The interpreted results from the various experiments done on the ExacTrac® system show that it can adequately track and target a moving stereotactic target. The system itself includes an intrinsic amount of uncertainty, but even when combined with the added complication of patient motion and internal tumor shifts can accurately target within 5 mm.

5.2 Future Work

The script used for analysis of the film can be continued to be used in the analysis of Winston-Lutz films, particularly for the daily QA checks. Rather than checking that the system is merely within specifications, quantitative evaluation of WL films can provide useful information regarding the calibration of the lasers. If a systematic shift in isocenter is visible over a long period of time, a correction
vector can be provided by the software script to more precisely position the lasers with respect to the radiation isocenter.

The effect of inconsistent breathing can directly translate into inferior treatments, which can be prolonged and may require repositioning mid-treatment if breathing becomes too erratic. Patients have the option of using feedback through the form of goggles providing a visualization of their own breath trace in the ExacTrac® Adaptive Gating System. A study by Linthout et al. has looked at the effect of using audio-visual feedback on treatment time optimization [39]. In addition to increasing treatment time, an irregular breathing pattern can reduce correlation between the internal and external markers and could have a direct effect on the accuracy of the system. Fiducial markers may shift and contort internally, and a beneficial study with the ExacTrac® Adaptive Gating System could be done using a deformable phantom with a programmable breath trace. A hidden target test could then be done to evaluate the system accuracy under these conditions.

Lastly, only liver SBRT patients were evaluated, due to a relative lack of gated SBRT patients at the McGill University Health Centre. A further patient study could be extended to include other important areas of treatment, particularly the lung. The software script and phantom studies can be applied to different treatment parameters, and a comparison of treatments in the lung would be useful knowledge.

The ExacTrac® system has proved to be a useful device for accurate gated SBRT procedures, and has many opportunities to be utilized for years to come. The completed experiments have tested the accuracy of the system under certain circumstances, but expanding this work would surely benefit hospitals and patients worldwide.
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BIBLIOGRAPHY


