Validation of a Respiratory Gating System for Automated Delivery of the Deep Inspiration Breath-hold Technique

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VALIDATION OF A RESPIRATORY GATING SYSTEM FOR AUTOMATED DELIVERY OF THE DEEP INSPIRATION BREATH-HOLD TECHNIQUE

A Thesis

Submitted to the Graduate Faculty of the Louisiana State University and Agricultural and Mechanical College in partial fulfillment of the requirements for the degree of Master of Science in

The Department of Physics and Astronomy

by

Michael G Stock
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Table of Contents

Acknowledgments ........................................................................................................ ii

List of Tables ............................................................................................................. iv

List of Figures ............................................................................................................. v

Abstract .................................................................................................................... vii

Chapter 1. Introduction ............................................................................................... 1
  1.1. Background and Significance ........................................................................... 1
  1.2. Research Motivation ....................................................................................... 11
  1.3. Hypothesis and Specific Aims ........................................................................ 13

Chapter 2. Methods and Materials .......................................................................... 15
  2.1. Aim 1: Output and Energy Constancy .......................................................... 15
  2.2. Aim 2: Treatment Delivery Accuracy ............................................................ 18
  2.3. Aim 3: Latency ............................................................................................... 26

Chapter 3. Results ................................................................................................... 37
  3.1. Aim 1: Output and Energy Constancy .......................................................... 37
  3.2. Aim 2: Treatment Delivery Accuracy ............................................................ 39
  3.3. Aim 3: Latency ............................................................................................... 44

Chapter 4. Discussion .............................................................................................. 50
  4.1. Aim 1: Output and Energy Constancy .......................................................... 50
  4.2. Aim 2: Treatment Delivery Accuracy ............................................................ 51
  4.3. Aim 3: Latency ............................................................................................... 52

Chapter 5. Summary and Conclusions ................................................................... 57
  5.1. Limitations and Future Work ......................................................................... 57
  5.2. Conclusions .................................................................................................... 58

References ............................................................................................................... 59

Vita .............................................................................................................................. 63
List of Tables

Table 2.1. Linear Accelerators Utilized in this Work.................................................................18
Table 2.2. Treatment Plan Characteristics..................................................................................22
Table 3.1. Gamma Passing Rates (3%, 3mm) for Non-Gated Treatment Deliveries.................40
Table 3.2. Gamma Passing Rates (3%, 3mm) for Gated Treatment Deliveries............................40
Table 3.3. Respiratory Motion Monitoring System Component Latencies.................................44
Table 3.4. Linear Accelerator Latency .........................................................................................45
Table 3.5. Displacement vs. Velocity Fitting Parameters ............................................................48
Table 3.6. Comparison of End-to-End Latency and Component Totals .......................................49
List of Figures

Figure 1.1. From (Keall et al., 2006), thoracic CT images of the same patient acquired during (a) free breathing (b) exhale .......................................................... 2

Figure 1.2. From (Choi and Seong, 2018), illustrating various respiratory motion management methods used in radiation therapy ......................................................... 5

Figure 1.3. From (Wiant et al., 2015), heart contours during free-breathing (purple), partial inspiration (turquoise), and deep inspiration breath-hold (yellow) ............... 6

Figure 1.4. From (Saito et al., 2018), system latencies during the (a) first gated segment (b) subsequent gated segment ................................................................. 12

Figure 2.1. Gating control software ........................................................................ 15

Figure 2.2. Tracking software .................................................................................. 20

Figure 2.3. Dynamic phantom .................................................................................. 20

Figure 2.4. Representative breathing waveform ....................................................... 21

Figure 2.5. Experimental setup used for treatment plan delivery ............................... 23

Figure 2.6. Analysis software ................................................................................... 25

Figure 2.7. Experimental setup used to quantify the tracking camera’s streaming latency ...... 28

Figure 2.8. Plot depicting linear accelerator latencies ............................................... 30

Figure 2.9. Experimental setup used to quantify end-to-end latencies of the gating system. 32

Figure 2.10. Cross sectional view of a cylindrical insert housing a 3 cm diameter spherical target .......................................................... 32

Figure 2.11. Images used to quantify end-to-end beam-on latency ......................... 35

Figure 2.12. Displacement between the reference position and the target .................. 36

Figure 3.1. Central-axis output constancy ................................................................ 38

Figure 3.2. TPR and PDD constancy ...................................................................... 38

Figure 3.3. Average gamma pass rates for treatment plans delivered using a (3%, 3mm) gamma criteria .......................................................... 41
Figure 3.4. Average gamma pass rates for treatment plans delivered using a (2%, 1mm) gamma criteria.................................................................41

Figure 3.5. Average difference in gamma pass rates between non-gated and gated modes of operation. ..............................................................................42

Figure 3.6. Percent dose difference comparison .................................................................................................................................43

Figure 3.7. Repeat delivery of a lung VMAT treatment plan ........................................................................................................43

Figure 3.8. End-to-end beam-on displacements ..............................................................................................................................46

Figure 3.9. End-to-end beam-off displacements ............................................................................................................................47
Abstract

Purpose: To validate the performance of a respiratory gating system for the automated delivery of the deep inspiration breath-hold (DIBH) technique.

Methods: The gating system utilized an automatic gating interface (Elekta Response) which connected a marker-based respiratory motion monitoring system to the linear accelerator control system. The gating system was characterized dosimetrically and temporally using two distinct approaches. Central-axis output and energy constancy were evaluated across 8 beam-matched linear accelerators. Additionally, a representative set of 5 treatment plans were delivered both non-gated and gated to a 2D diode array (MapCHECK). The respiratory motion monitoring system optically tracked a reflective marker that was attached to a dynamic phantom (QUASAR). The phantom was programmed to replicate a typical DIBH breathing waveform. The passing rates between these modes of operation were evaluated using gamma analysis and a percent dose difference comparison. Modular and end-to-end approaches were used to quantify system latencies. The modular components evaluated were the streaming latency of the tracking camera, sampling rate of the tracking software, signal travel time, and latency of the linear accelerator. The end-to-end approach involved measuring the displacement of a target moving at known velocities during the gating process.

Results: Output and energy constancy were both within ± 0.5% for each beam energy and linear accelerator investigated. The average differences in passing rates between non-gated and gated modes of operation were within ± 0.4% using gamma analysis (2%, 1mm). Average passing rates between modes of operation were greater than 99% using a percent dose difference comparison (1%). The first gated segment was found to have significantly ($p = .02$) longer beam-
on latency compared to the subsequent gated segment. End-to-end beam-on and beam-off latency for the subsequent gated segment was found to be 1.49 and 0.34 seconds, respectively, which was consistent with measured component totals.

Conclusion: The gating system was able to achieve dosimetric operating characteristics that are desirable for accurate delivery of the DIBH technique. The methodology presented can be generalized to other respiratory gating systems that utilize the automatic gating interface studied in this work.
Chapter 1. Introduction

1.1. Background and Significance

1.1.1. Patient Motion

Radiation therapy for cancer treatment uses ionizing radiation to destroy or damage cancerous cells within the body. The goal is to deliver a tumoricidal dose to the target volume while also minimizing the radiation toxicity to the surrounding healthy tissues. To achieve this goal, personalized treatment plans are created for each patient. The treatment planning process begins with a radiation therapy treatment simulation. During simulation, images are taken of the patient to obtain a representation of their anatomy in the treatment position at a particular point in time. Immobilization devices are often used during simulation and treatment delivery to ensure a reproducible setup and limit the patient’s movement throughout imaging and treatment. The planning images obtained during treatment simulation are typically acquired via helical computed tomography (CT) and reconstructed to create a 3D image set of the patient. The image set is used to design the custom treatment plan with the assumption that the planning images will accurately represent the patient’s anatomy throughout the course of treatment.

Patient motion that occurs during treatment simulation can cause artifacts in the CT images used to plan the patient’s treatment. In general, a CT artifact refers to any systematic discrepancy between the CT numbers in the reconstructed image and the true attenuation coefficients of the imaged anatomy (Tsai et al., 2011). Motion artifacts are a specific type of CT artifact which result from the CT reconstruction algorithm’s assumption that the patient’s anatomy will be stationary during image acquisition (Keall et al., 2006). Motion artifacts can appear as blurring, streaking or a distortion of the imaged anatomy, as shown in Figure 1.1. Ultimately, motion artifacts that are present in the images used for treatment planning can affect the accuracy of the dose calculation or can make it difficult to delineate critical anatomic structures when designing the plan (Balter et
Typically, safety margins are added to critical structures during treatment planning to account for their positional uncertainty due to patient motion (Landberg et al., 1999). Choosing the appropriate margins to apply to these structures can be difficult if significant patient motion occurred during treatment simulation because of the resulting motion artifacts that can present in the planning images (Keall et al., 2006).

Figure 1.1. From (Keall et al., 2006), thoracic CT images of the same patient acquired during (a) free breathing (b) exhale to reduce respiratory motion during image acquisition. If not accounted for, patient motion that occurs during image acquisition can cause motion artifacts in the reconstructed images.

Separate from its ability to produce motion artifacts during treatment simulation, patient motion can also impact treatment delivery by causing a material change in the patient’s anatomy or geometric arrangement relative to how the treatment was planned. Interfraction motion refers to changes that occur between treatment fractions which typically result from the finite ability to perfectly reproduce the treatment position. Deviations in the patient’s anatomy or positioning relative to the planning CT have been shown to cause systematic dose delivery errors (Erridge et al., 2003). If significant interfraction changes are observed, a new treatment plan can be created to account for the anatomic changes that have occurred.
In contrast, intrafraction motion is any movement that occurs during the treatment fraction and could be the result of the respiratory, skeletal muscular, cardiac, or gastrointestinal motion (Keall et al., 2006). Respiratory motion in particular has been shown to affect all tumor sites in the thorax and abdomen and is the most extensively studied cause of intrafraction motion in radiation therapy (Keall et al., 2006). Respiratory motion presents challenges, especially for advanced techniques like intensity modulated radiation therapy (IMRT), which seeks to deliver highly conformal radiation doses to the target volume using relatively tight margins to spare the surrounding healthy tissues (Brandner et al., 2017). In the presence of respiratory-induced tumor motion these margins may need to be expanded to ensure that the target volume receives adequate dose coverage. However, simply increasing these margins to account for the full range of tumor motion isn’t always feasible as it can lead to excessive healthy tissue exposure.

Previous work has shown that respiratory-induced tumor motion can vary widely between patients and that the magnitude of this motion is not predictable by the tumor’s size, its location, or the patient’s pulmonary function (Stevens et al., 2001). Therefore, it is recommended that tumor motion be assessed prior to treatment if respiratory induced-tumor motion is expected (Keall et al., 2006). Imaging studies are often used to quantify the magnitude of respiratory motion. CT scans acquired at inhale and exhale can define the range of tumor motion in three dimensions and fluoroscopy studies have been used to observe two-dimensional anatomic motion with respect to time (Malone et al., 2000). Another technique, known as four-dimensional computed tomography (4DCT), provides a compromise between the time resolution of fluoroscopy and the 3D spatial resolution of a CT study. 4DCT correlates the CT scan with the patient’s breathing so that image sets can be reconstructed at particular phases or amplitudes of the respiratory cycle.
study shows respiratory-induced tumor motion that exceeds 5 millimeters (mm) it is recommended that methods be employed to manage the patients respiratory motion (Keall et al., 2006).

The goal of respiratory motion management is to mitigate the effects of respiration on the radiation therapy process. Motion encompassing methods accomplish this by increasing the margins around the target volume to account for the full range of respiratory motion. Forced shallow breathing techniques seek to limit the amplitude of respiration through compression of the patient’s abdomen with a ridged frame. Real-time tumor-tracking methods continuously reposition the radiation beam to follow the path of the target volume. Additionally, there are several motion management methods that utilize beam gating. Beam gating is a broad term used in radiation therapy to describe treatments that synchronize delivery of the radiation beam to a particular stage of the patient’s breathing cycle, referred to as the gating window (Saito et al., 2018). Free-breathe respiratory gating is a specific type of beam gating in which the patient’s treatment is delivered at preselected phases or amplitudes of their natural breathing cycle. Free-breathe respiratory gating can decrease the range of target motion while the beam is being delivered, potentially reducing the needed margins around the target volume (Keall et al., 2002). Each of these respiratory motion management methods are illustrated in Figure 1.2.

1.1.2. Deep Inspiration Breath-Hold

Deep inspiration breath-hold (DIBH) is another respiratory motion management method used in radiation therapy which utilizes beam gating. The DIBH technique involves delivery of the patient’s treatment while they are holding their breath at deep inspiration and has been shown to offer improved efficiency when compared to free-breathe respiratory gating (Berson et al., 2004). Typically, the patient will hold their breath for approximately 15-25 seconds while the beam is delivered. Then, treatment delivery is paused, allowing the patient to breathe freely until they
are ready to begin the next DIBH. This process is repeated until the treatment fraction is fully delivered.

There are two major benefits derived from the DIBH technique. The first major benefit is that DIBH can create a more favorable anatomical arrangement for treatment of certain disease sites. During deep inspiration, the heart is displaced posteriorly, inferiorly, and medially, as depicted in Figure 1.3. For the treatment of left-sided breast cancer, this displacement can reduce the radiation dose that the heart receives throughout the course of treatment (Pedersen et al., 2004). This is especially beneficial because major coronary events have been shown to increase linearly with the mean dose to the heart by 7.4% per gray (Darby et al., 2013). Therefore, the DIBH technique provides a valuable tool to reduce the likelihood of major coronary events later in life for patient’s receiving radiation therapy for left-sided breast cancer.

Figure 1.2. From (Choi and Seong, 2018), illustrating various respiratory motion management methods used in radiation therapy: (A) motion encompassing methods (B) breath-hold (C) forced shallow breathing (D) free-breathe respiratory gating (E) real-time tumor tracking.
The other major benefit of the DIBH technique is the reduction in respiratory motion that occurs during a breath-hold. Often times this technique is employed for the treatment of thoracic tumors, such as lung cancer, to reduce tumor motion during delivery. The reduction in tumor motion allows for decreased margins around the target volume which can reduce the radiation exposure to the surrounding healthy tissues (Hanley et al., 1999). While left-sided breast and lung are among the most common treatment sites that employ the DIBH technique, a variety of other sites such as the liver and pancreas have also been shown to benefit from the respiratory motion mitigation derived from this technique (Zeng et al., 2019; Dawson et al., 2001).

![Heart contours during free-breathing](image)

Figure 1.3. From (Wiant et al., 2015), heart contours during free-breathing (purple), partial inspiration (turquoise), and deep inspiration breath-hold (yellow). Axial, sagittal, and coronal views of the same patient are shown from left to right. During deep inspiration, the heart is displaced posteriorly, inferiorly, and medially relative to its position during free breathing.

Treatments that utilize DIBH are planned using an image set acquired during treatment simulation with the patient in the breath-hold position. Typically, audio or visual feedback is used to synchronize the helical CT scan with the DIBH so that image acquisition only occurs while the patient is in the DIBH position (Keall et al., 2006). Throughout treatment simulation, a respiratory motion monitoring system is typically used to display a respiratory trace of the patient. The amplitude of the respiratory trace should be within a predefined range during each DIBH to ensure
that the image set acquired during treatment simulation will adequately represent the patient’s anatomy when they are performing a DIBH during treatment.

The treatment position used during simulation is reproduced prior to each fraction of treatment and a respiratory motion monitoring system is typically used to demonstrate that patient’s respiratory trace during each DIBH is consistent with the amplitude levels used during treatment simulation. Once the patient’s respiratory trace enters the predefined amplitude range, the radiation beam is activated, either manually by the therapist or automatically if the linear accelerator is equipped with an interface that connects the respiratory motion monitoring system to the linear accelerator control system. This implementation of the DIBH technique has the advantage of constant respiratory monitoring which allows the beam to be paused, or held, automatically, should the patient exit the breath-hold position prematurely.

One of the most widely studied respiratory motion monitoring systems used in radiation therapy is an abdominal marker-based optical tracking system (Real-Time Position Management, Varian Medical Systems, Palo Alto, CA). The system utilizes an infrared camera to track a reflective external marker typically placed on the patient’s abdominal surface; the marker’s displacement is used as the surrogate for the patient’s respiratory motion. This system relies on the assumption that internal tumor motion will correlate with the motion of the external marker surrogate throughout the course of treatment (Ionascu et al., 2007). Previous works have shown that tumor motion can be correlated well with the motion of an external marker placed on the patient’s abdominal surface (Beddar et al., 2007; Vedam et al., 2003; Gierga et al., 2005). However, interfraction changes in the tumor-surrogate relationship have been observed, therefore it is recommended that the tumor is imaged directly throughout the course of treatment to verify the accuracy of this relationship (Hoisak et al., 2004). One of the major benefits of the abdominal
marker as a respiratory motion surrogate is that they can be used for the majority of radiation therapy patients because of their non-invasive nature (Keall et al., 2006).

1.1.3. Automatic Gating Interface

Recently, an automatic gating interface has been commercially released by a major linear accelerator manufacturer (Response, Elekta Oncology Systems, Stockholm, Sweden) that is capable of automating the delivery of the DIBH technique. The gating interface utilizes a digital relay inserted into the linear accelerator’s Pulse Repetition Frequency (PRF) interlock chain to perform the automated beam gating (Evans et al., 2010). When the PRF relay is open, pulses of radiation cannot be generated, causing a beam-hold. Third party respiratory motion monitoring systems can be used to generate the binary gating signal which triggers the gating interface to open or close the digital relay at the appropriate times during gated treatment deliveries.

1.1.4. Dosimetric Considerations

Saito et al. (2018) validated the gating interface for use with a commercially available respiratory motion monitoring system (Abches, APEX Medical Inc, Tokyo, Japan). To help establish dosimetric accuracy, they measured central-axis output and beam energy with and without the use of beam gating. Gated measurements utilized beam-on durations ranging from 1.1 to 7 seconds and beam-holds of 2 seconds in between each beam-on period. For both output and beam energy, the percentage difference of the gated measurements relative to non-gated measurements was found to be less than 1%.

Freislederer et al. (2015) used the gating interface to investigate the impact that the duty cycle had on the accuracy of gated treatment deliveries. The duty cycle refers to the percentage of time that the radiation beam is on during treatment delivery. Their results showed that the dose difference between gated and non-gated deliveries increased when low duty cycles were utilized. Previous work has shown that the transient temperature of the electron gun filament and magnetron
can cause unstable beam production upon start-up (Fujimoto et al., 2013). One of the risks of beam gating with low duty cycles is the increase in the amount of start-up processes that occur during a treatment delivery, negatively impacting dosimetric accuracy. However, for duty cycles of at least 30%, the differences in dose delivery for static open fields were found to be less than 1% (Freislederer et al., 2015).

Noto et al. (2014) assessed the accuracy of delivering volumetric modulated arc therapy (VMAT) treatment plans in multiple gated segments to simulate the DIBH technique. A gated segment refers to the portion of the treatment that is delivered in between each beam-hold during a gated treatment delivery. Each plan was delivered with gated segment durations ranging from 10 to 40 seconds and beam-holds of 5 seconds in between each gated segment. In this study, they manually pushed the start and interrupt buttons to deliver and hold the beam, respectively. They found that the dose delivery was stable and accurate when the gated segment durations were 15 seconds or greater. However, the manual beam gating performed in this study may not adequately replicate the operating characteristics encountered with use of an automated gating interface.

Jermoumi et al. (2017) used a linear accelerator (Elekta Synergy, Elekta Oncology Systems, Stockholm, Sweden) that was equipped with the gating interface to study the impact of automated beam gating on delivery accuracy. This study delivered two separate stereotactic body radiation therapy (SBRT) treatment plans using a variety of beam-on and beam-off durations during delivery. One combination in particular was intended to simulate the DIBH technique and involved beam-on durations of 12 seconds followed by beam-holds of 6 seconds after each beam-on period. These gated treatment deliveries exhibited passing rates of 100% using gamma analysis with a passing criteria of (1%, 1mm). Their work has shown that the gating interface is capable of facilitating the accurate delivery of gated SBRT treatments. However, they may not have used
beam-holds that were long enough to be representative of the DIBH technique that their work was investigating.

1.1.5. Temporal Considerations

Another important characteristic to assess when validating the performance of a system used for automated beam gating is latency. Beam-on or beam-off latency refer to the time delay that occurs prior to beam start-up or a beam-hold, respectively. Beam-on latency can decrease the efficiency of treatment delivery by reducing the number of monitor units (MU) that can be delivered during each gated segment. Beam-on latency is especially of concern for systems intended for free-breathe respiratory gating as time delays prior to beam start-up can exceed the short gating windows that are often utilized with this type of beam gating. In this case, the patient’s respiratory surrogate would exit the gating window before beam delivery can begin. Alternatively, beam-off latency can result in the patient’s respiratory surrogate exiting the gating window before the beam is held, which could lead to a geometric miss of the target (Smith and Becker, 2009).

Snyder et al. (2017) quantified both beam-on and beam-off latencies using a linear accelerator (Elekta Versa HD, Elekta Oncology Systems, Stockholm, Sweden) equipped with the gating interface and a commercially available respiratory motion monitoring system (AZ-733VI, Anzai Medical Systems, Tokyo, Japan). This work used the electronic portal imaging device (EPID) mounted on the linear accelerator to capture images of a ball bearing phantom. The EPID is a flat panel detector that can record exposures using the linear accelerator’s therapeutic MV beam. Reference images were acquired with the phantom kept stationary at predefined amplitudes. Experimental images were also collected with the phantom moving at known velocities and set to trigger at the same amplitudes used for the reference images. The displacement of the ball bearing between the reference and experimental images was divided by its known velocity to calculate the beam-on and beam-off latencies.
Woods and Rong (2015) used a similar approach as Snyder et al. (2017) to quantify latency. In their work, the average temporal delay was found by plotting the measured positional displacements against the known target velocities. The slope of a linear fit line was used to quantify the end-to-end latencies of the system. They argued that using a linear fit is more accurate than averaging the time delay calculated at each velocity, because low velocity data points may be largely affected by the measurement uncertainty and the size of pixels in the images used to measure the displacements.

Saito et al. (2018) evaluated latency using a multichannel oscilloscope. This method allowed for various component latencies of the system including the gating interface and linear accelerator (Elekta Synergy, Elekta Oncology Systems, Stockholm, Sweden) to be quantified separately. They observed considerably larger beam-on latency during the first gated segment compared to subsequent gated segments, as shown in Figure 1.4. Because of the differences they observed, this study did not include the first gated segment in their analysis. To our knowledge, no work has investigated the temporal characteristics of both the first and subsequent gated segments during automated delivery with this gating interface.

1.2. Research Motivation

Before implementing a system used to facilitate automated beam gating, it is recommended that a baseline of dosimetric and temporal accuracy be established (Klein et al., 2009). In particular, systems intended for use with the DIBH technique should be able to deliver the planned treatment accurately in the presence of extended beam-holds. These extended beam-holds should not significantly alter important dosimetric characteristics like output or energy. When evaluating the temporal characteristics of such a system the main concern is the beam-off latency, as time
delays prior to a beam-hold can pose a safety concern during treatment delivery (Smith and Becker, 2009).

No previous studies have reported on validation of this gating interface for use with an abdominal marker-based optical tracking system. There are commercially available respiratory motion monitoring systems that utilize optical tracking, but these systems directly track the patient’s surface in three dimensions. Currently, there are no systems that are compatible with this particular gating interface which are designed to optically track an external marker placed on the patient’s abdominal surface. Historically, most of the knowledge and clinical data regarding motion monitoring for radiation therapy has utilized optically tracked markers on the patient’s abdominal surface as the surrogate for respiratory motion. The body of knowledge existing for these types of respiratory motion monitoring systems provides motivation for their implementation with this gating interface.

![Figure 1.4](image)

Figure 1.4. From (Saito et al., 2018), system latencies during the (a) first gated segment (b) subsequent gated segment. Oscilloscope output is shown for the respiratory motion monitoring system (blue), gating interface (green), and linear accelerator monitor chamber (orange).
Woods and Rong (2015) evaluated the performance of a gating system across multiple linear accelerators (TrueBeam, Varian Medical Systems, Palo Alto, CA). The gating system studied in their work utilized a similar respiratory motion monitoring system (Real-Time Position Management, Varian Medical Systems, Palo Alto, CA) as the one studied in this work. However, they mainly focused on temporal characteristics of the gating system during their validation efforts. There is motivation to also evaluate the dosimetric aspects of gating performance across multiple linear accelerators.

The goal of this work was to validate the temporal and dosimetric characteristics of a respiratory gating system. The gating system consisted of an automatic gating interface that connected a marker-based respiratory motion monitoring system to the linear accelerator control system. Dosimetric characteristics were evaluated across multiple linear accelerators to allow for the intercomparison of its gating performance. Because of the intended clinical use of the gating system, this work focused on validating the system for automated deliveries that utilize DIBH.

1.3. Hypothesis and Specific Aims

The hypothesis of this work was that the respiratory gating system can achieve the desired operating characteristics during automated deliveries that are representative of the DIBH technique. These operating characteristics include output and energy constancy within 2%, an average decrease in gamma passing rates between the non-gated and gated modes of operation that is less than 2%, and end-to-end beam-off latency that is within 300 milliseconds (ms). The three specific aims used to address this hypothesis are as follows:

Aim 1: Evaluate output and energy constancy using the gating interface to facilitate automated beam-holds
Aim 2: Deliver a set of treatment plans to assess delivery accuracy resulting from use of the gating system

Aim 3: Quantify the end-to-end beam-on and beam-off latency of the gating system as well as the individual components that contribute to these overall latencies
Chapter 2. Methods and Materials

2.1. Aim 1: Output and Energy Constancy

Central-axis output and beam energy were measured with and without the use of automated beam-holds. In this work, deliveries with and without the use of automated beam-holds are referred to as the gated mode of operation and non-gated mode of operation, respectively. Constancy between these modes of operation was quantified to assess the impact that automated beam-holds had on these dosimetric quantities.

2.1.1. Gating Control Software

For this aim, gating control software provided by the manufacture (Response Service Tool, Elekta Oncology Systems, Stockholm, Sweden) was used to generate the binary gating signal. The software allows the user to create gating cycles with custom beam-on and beam-off combinations to test the functionality of the gating interface (Jermoumi et al., 2017). To incorporate beam-holds that were representative of the DIBH technique, a gating cycle was created with a 33 second period and 70% duty cycle, as shown in Figure 2.1. This gating cycle was looped continuously while gated measurements of central-axis output and beam energy were collected.

![Figure 2.1. Gating control software displaying a gating cycle with a 33 second period and 70% duty cycle.](image)
2.1.2. Output

Linear accelerator output was measured along the beam central axis using a PTW 30013 or a PTW 30006 farmer ionization chamber (PTW, Freiburg, Germany). The ionization chamber was connected to a CNMC model 206 electrometer (CNMC Company Inc., TN, USA) which was used to measure the charge liberated during each delivery. The raw electrometer reading was converted to dose to water using protocol recommended by Task Group 51 of the American Association of Physicists in Medicine (Almond et al., 1999). The resulting dose was divided by the number of monitor units (MU) delivered to yield the output with units of cGy/MU. Measurements collected under the non-gated mode of operation were delivered using 100 MU and those collected under the gated mode of operation were delivered using 600 MU to ensure that approximately 3-4 beam-holds occurred during each gated measurement. Photon output was evaluated using reference conditions of 100-centimeter (cm) source-to-axis distance (SAD), 10 cm depth in solid water, 10x10 cm² field size at isocenter, and 11 cm of backscatter. Electron output was evaluated using reference conditions of 100 cm source-to-surface distance (SSD), 2.4 cm depth in solid water, 10x10 cm² field size at the surface, and 11 cm of backscatter.

2.1.3. Beam Energy

In this work, a tissue phantom ratio (TPR) was used as the surrogate for photon beam energy. TPR was calculated using the equation

$$\text{TPR} = \frac{D_d}{D_{d0}},$$

(2.1.)

where \(D_d\) is the absorbed dose in a phantom at a depth \(d\) and \(D_{d0}\) is the absorbed dose at a reference depth \(d_0\) with both measurements sharing the same source-to-point distance. TPR was evaluated at a reference depth \(d_0\) of 10 cm and a depth \(d\) of 5 cm in solid water using a 10x10 cm² field size at isocenter.
A percent depth dose (PDD) was used as the surrogate for electron beam energy. PDD was calculated using the equation

\[
PDD = \frac{D_d}{D_{d0}} \times 100\% ,
\]

where \(D_d\) is the absorbed dose at a depth \((d)\) and \(D_{d0}\) is the absorbed dose at a reference depth \((d_0)\) with both measurements sharing the same SSD. Electron PDD was evaluated at a reference depth \((d_0)\) of 2.4 cm and a depth \((d)\) of 4.1 cm in solid water using a 10x10 cm² field size at the surface. Measurements of PDD and TPR both utilized 100 MU for non-gated deliveries and 600 MU for gated deliveries.

2.1.4. Constancy

In this work, constancy was used to quantify the change in central-axis output and beam energy resulting from the use of automated beam-holds during delivery. Constancy was calculated using the equation

\[
\text{Constancy} = \frac{\overline{\text{Gated}} - \overline{\text{Nongated}}}{\overline{\text{Nongated}}} \times 100\% ,
\]

where \(\overline{\text{Gated}}\) and \(\overline{\text{Nongated}}\) refer to average value of the measurements collected in each mode of operation.

Output constancy and energy constancy were evaluated on each of the linear accelerators listed in Table 2.1, which were all beam-matched. For this aim, constancy was evaluated for each photon energy available on the linear accelerator and a single representative 10 MeV electron energy. Three measurements of output and beam energy were collected under non-gated and gated modes of operation for each of the energies utilized. The uncertainty associated with output and energy constancy was calculated using the equation
\[
\text{Constancy Uncertainty} = \sqrt{\left(-100\frac{Gated \cdot \delta_{\text{Nongated}}}{(\text{Nongated})^2}\right)^2 + \left(100\frac{\delta_{\text{Gated}}}{\text{Nongated}}\right)^2}, \tag{2.4.}
\]

where \(\delta_{\text{Nongated}}\) and \(\delta_{\text{Gated}}\) represent the standard deviation of the three measurements collected in each mode of operation. Equation (2.4.) was derived using the general propagation of errors methodology.

Table 2.1. Linear Accelerators Utilized in this Work

<table>
<thead>
<tr>
<th>Label</th>
<th>Make and Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>Accelerator # 1</td>
<td>Elekta Infinity</td>
</tr>
<tr>
<td>Accelerator # 2</td>
<td>Elekta Versa HD</td>
</tr>
<tr>
<td>Accelerator # 3</td>
<td>Elekta Infinity</td>
</tr>
<tr>
<td>Accelerator # 4</td>
<td>Elekta Infinity</td>
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<tr>
<td>Accelerator # 5</td>
<td>Elekta Infinity</td>
</tr>
<tr>
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<td>Elekta Infinity</td>
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<td>Accelerator # 7</td>
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<tr>
<td>Accelerator # 8</td>
<td>Elekta Synergy</td>
</tr>
</tbody>
</table>

2.2. **Aim 2: Treatment Delivery Accuracy**

Treatment plans from representative anatomic sites were delivered both non-gated and under conditions that mimicked the DIBH technique, using the gating system to facilitate automated beam-holds. The treatment delivery accuracy and reproducibility achieved with the gating system were evaluated using these measurements.
2.2.1. Gating System

The gating system studied in this work consisted of an automatic gating interface that connected a respiratory motion monitoring system to the linear accelerator control system. The respiratory motion monitoring system was created in-house at Mary Bird Perkins Cancer Center (MBPCC) and was designed to track the displacement of a reflective marker placed on the appropriate surrogate for respiratory motion. A reflective sphere embedded within the marker was optically tracked using a complimentary metal-oxide semiconductor (CMOS) camera with a USB interface.

The respiratory motion monitoring system utilized in-house tracking software to display a respiratory trace based on the reflective marker’s displacement, as shown in Figure 2.2. The tracking software was also used to define the position and width of an amplitude-based gating window. When the respiratory trace passed through the midline of the amplitude-based gating window the tracking software would signal to the gating interface to close the digital PRF relay, allowing beam delivery to begin. Delivery would continue continuously until the respiratory trace exited the gating window entirely. Then the tracking software would signal to the gating interface to open the PRF relay, causing an automated beam-hold. This process would repeat throughout the gated treatment delivery until the treatment fraction was fully delivered.

2.2.2. Dynamic Phantom

A dynamic phantom (Quasar Respiratory Phantom, Modus Medical Devices, Ontario, Canada), shown in Figure 2.3, was used to simulate the abdominal surface displacement of a patient undergoing the DIBH technique. This phantom consisted of an acrylic body attached to a programmable drive unit which could simultaneously move the phantom’s chest wall platform in the anterior-posterior direction and the phantom’s translation stage in the superior-inferior direction. Cylindrical inserts of various densities could be attached to the translation stage and
moved within the phantom’s body, simulating one-dimensional lung and tumor motion. The phantom could be operated under software control to reproduce one-dimensional motion of inputted respiratory waveforms.

Figure 2.2. Tracking software used the displacement of a reflective marker to display a respiratory trace (black). An amplitude-based gating window (red) and window midline (blue) were used as thresholds to trigger the beam off and on, respectively.

Figure 2.3. Dynamic phantom with chest wall platform, translation stage and attached cylindrical insert.
2.2.3. Representative DIBH Waveform

The abdominal displacement of a patient who was previously treated at MBPCC with the DIBH technique was used to create a representative DIBH breathing waveform. The patient’s abdominal displacement trace was imported into the phantom’s control software. Using the software’s wave editor tools, a portion of the trace was selected, which included a breath-hold and the free-breathe recovery period. This portion of the breathing trace was repeated to produce the final waveform used in this work, shown in Figure 2.4. The waveform had a breath-hold duration of approximately 24 seconds and a free-breathe duration of approximately 33 seconds, which are representative of patients with the DIBH technique.

![Representative DIBH Waveform](image)

Figure 2.4. Representative breathing waveform used to simulate the abdominal displacement of a typical patient undergoing the DIBH technique.

2.2.4. Treatment Plan Set

A set of five treatment plans from representative anatomic sites that commonly employ the DIBH technique were used to assess the change in delivery accuracy resulting from use of the gating system. Delivering the same set of treatment plans across all linear accelerators tested ensured that there was constancy across measurement sets, allowing for the intercomparison of
gating performance. The five treatment plans utilized in this work were taken from patients who were previously treated at MBPCC using the DIBH technique. Included in the set were VMAT and SBRT cases because gated delivery of these plan types presents challenges for the linear accelerator to deliver accurately due to varying gantry speeds and dose rates (Snyder et al., 2017). A 3D conformal radiation therapy (3D-CRT) case for left-side breast cancer was also included in the set because the DIBH technique is commonly employed for these plan types. Information about each of the treatment plans included in the set is shown in Table 2.2.

Table 2.2. Treatment Plan Characteristics

<table>
<thead>
<tr>
<th>Anatomic Site</th>
<th>Treatment Type</th>
<th>Number of Fields</th>
<th>Energy</th>
<th>Total MU</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lung</td>
<td>VMAT</td>
<td>2</td>
<td>6x</td>
<td>425</td>
</tr>
<tr>
<td>Chest Wall</td>
<td>VMAT</td>
<td>2</td>
<td>6x</td>
<td>477</td>
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<tr>
<td>Pancreas</td>
<td>VMAT</td>
<td>2</td>
<td>6x</td>
<td>482</td>
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<tr>
<td>Lung</td>
<td>SBRT</td>
<td>2</td>
<td>6x</td>
<td>1934</td>
</tr>
<tr>
<td>Left-Sided Breast</td>
<td>3D-CRT</td>
<td>2</td>
<td>6x</td>
<td>323</td>
</tr>
</tbody>
</table>

2.2.5. Treatment Plan Delivery

To mimic a typical treatment delivery that employs the DIBH technique, the experimental setup shown in Figure 2.5 was used to deliver treatment plans in the gated mode of operation. For these deliveries, the phantom was placed on its side with the chest wall platform removed. This orientation allowed the translation stage to move in the anterior-posterior direction which reduced off-axis motion and allowed for greater maximum displacement as discussed by Belanger et al. (2016). The reflective marker was attached to the translation stage of the phantom and centered on the treatment couch using the room’s sagittal laser. The phantom was operated under software
control and programmed to replicate a representative DIBH waveform, mimicking the abdominal
displacement of a typical patient being treated with this technique. The camera used to track the
fiducial maker was also centered using the room’s sagittal laser and positioned 80 cm from the
reflective marker. A two-dimensional diode array (MapCHECK, Sun Nuclear Corporation,
Melbourne, FL, USA) was positioned at isocenter and used to record the incident dose distribution
from each delivery.

Figure 2.5. Experimental setup used for treatment plan delivery. A tracking camera monitored the
displacement of a reflective marker. The reflective marker was able to move in the directions
specified to reproduce a representative DIBH waveform. A two-dimensional diode array at
isocenter recorded the incident dose distribution for each delivery.

In contrast to specific aim 1, where a vendor software tool was used to generate the gating
signal, the respiratory motion monitoring system generated the gating signal during each treatment
delivery in the gated mode of operation. The respiratory motion monitoring system’s tracking
software was calibrated using the known diameter of a reflective sphere embedded within the
marker, which allowed for the width of the gating window to be designated in mm rather than
pixels. In this work, the width of the amplitude-based gating window was approximately 8 mm for
the VMAT and 3D-CRT breast and thoracic cases and 5 mm for the SBRT thoracic case, which reflected typical values used clinically at MBPCC.

All treatment plan deliveries were carried out on linear accelerators #1 through 7 in Table 2.1. Linear accelerator #8 was not utilized for this aim of the project because the set of treatment plans were not deliverable on this linear accelerator due to an incompatible linear accelerator head. Each treatment plan was delivered under non-gated and gated modes of operation to the two-dimensional diode array. Typically, 3-6 beam-holds occurred during the delivery of each treatment plan in the gated mode of operation. The diode array was kept in the same position between non-gated and gated deliveries to allow for comparison between these modes of operation. In addition, the lung VMAT treatment plan was chosen from the set to be delivered 3 separate times under each mode of operation to assess delivery reproducibility. The dose distribution measured by the diode array during each delivery was saved and analyzed to assess the delivery accuracy and reproducibility of the gating system.

2.2.6. Analysis

Measurements of treatment plan deliveries acquired in both modes of operation were analyzed using commercially available analysis software (SNC Patient, Sun Nuclear Corporation, Melbourne, FL, USA), shown in Figure 2.6. For each of the treatment plans included in the set, a planned dose distribution was exported from the treatment planning system (TPS). The planned dose distribution represents the expected dose profile incident on the diode array during treatment delivery. Gamma analysis was used to quantify the agreement between the planned and measured dose distributions. Gamma analysis is a composite metric that takes into account the dose difference (%) and the distance-to-agreement (DTA) between the points in each distribution (Low et al., 1998). The percentage of points satisfying the dose difference and DTA gamma criteria is referred to as the gamma passing rate. In this work, gamma passing rates for each treatment
delivery were evaluated using gamma criteria of (3\% dose difference, 3mm DTA) and (2\% dose difference, 1 mm DTA). To assess the change in delivery accuracy resulting from use of the gating system, a metric referred to in this work as the passing difference was used. The passing difference is the difference in gamma passing rates between non-gated and gated deliveries of the same treatment plan, as shown by the equation

\[
\text{Passing Difference (\%)} = [\text{NonGated Passing Rate}] - [\text{Gated Passing Rate}]. 
\]  
\text{(2.5.)}

The average passing difference for the set of treatment plans and associated standard error were calculated for both gamma criteria utilized.

Figure 2.6. Analysis software used to compare the measured dose distribution (left) to the planned dose distribution (right) using gamma analysis.

A percent dose difference comparison was also used to assess the delivery accuracy of the gating system. Instead of comparing each measurement to its planned dose distribution, as was done with gamma analysis, the percent dose difference provided a method for comparing the non-gated and gated treatment deliveries directly to one another. This comparison sampled the same point from each distribution to test whether the two points were within a specified percentage dose difference of one another. The passing rate from this comparison was calculated for each treatment plan included in the set using passing criteria of 3\% and 1\%. Additionally, the average passing rate
for the set of treatment plans and associated standard error were computed for both passing criteria utilized.

To assess delivery reproducibility, passing rates for repeat deliveries of the lung VMAT treatment plan were calculated using gamma analysis with gamma criteria of (3% dose difference, 3mm DTA). The average passing rate and associated standard deviation were calculated for each mode of operation. The magnitude of the standard deviation for these repeat deliveries was used to evaluate delivery reproducibility in the non-gated and gated modes of operation.

2.3. Aim 3: Latency

End-to-end and component latencies were quantified during the first and subsequent gated segments of automated delivery to evaluate the temporal characteristics of the respiratory gating system. End-to-end beam-on or beam-off latency refers to the overall time delay associated with the gating system automatically turning on or off the radiation beam during gated treatment delivery, respectively. Component latency refers to the time delay of the individual elements of the gating system that contribute to these end-to-end latencies. The gating system studied in this work was modular in nature, which means that the latency of the individual components that make up the system were independent of one another and could be evaluated separately. The summation of the individual component latencies was compared to the end-to-end measurement to assess the agreement between these two approaches.

2.3.1. Modular Approach

The modular components that contributed to the latency of the respiratory motion monitoring system were the streaming latency of the camera used to optically track the reflective marker surrogate, the sampling rate of the tracking software, and the time it took for the gating signal generated by the tracking software to travel to the gating interface. Streaming latency
represents the time it takes for frames captured by the tracking camera to be displayed on a video monitor. To quantify the tracking camera’s streaming latency, a physical stopwatch was filmed using the camera. Frames of the stopwatch captured by the tracking camera were displayed on a video monitor. A separate recording device with a framerate of 120 frames per second (fps) was used to simultaneously record both the video monitor and the physical stopwatch, as shown in Figure 2.7. The streaming latency of the separate recording device was negligible because the physical stopwatch and the video monitor were both included in the same frame. The time displayed by the physical stopwatch was used as the reference timestamp. The difference between the reference timestamp and the video monitor timestamp was used to quantify the tracking camera’s streaming latency. Ten separate frames were used to calculate the average streaming latency of the tracking camera and associated standard deviation.

The sampling rate of the tracking software represents the frequency at which the software could update the position of the reflective marker surrogate. The tracking software was designed to produce log files listing reflective marker’s positions and associated timestamps during use of the software. The sampling rate of the tracking software was confirmed using these log files. The difference between 10 successive timestamps within the log file were used to calculate the average sampling rate and associated standard deviation.

To quantify the travel time of the gating signal from the tracking software to the gating interface, a software program was created which was capable of triggering the gating signal in the tracking software at a known time. The gating control software provided the time at which the gating interface received this gating signal. The difference between the timestamp in the gating control software and the triggered timestamp was used to quantify the signal’s travel time.
In addition to the components of the respiratory motion monitoring system, the beam-on and beam-off latencies of the linear accelerator were also evaluated. In this work, the linear accelerator beam-on latency was defined as the time between the PRF relay closing and the dose rate increasing to 80% of its maximum value. Alternatively, the linear accelerator beam-off latency was defined as the time between the PRF relay opening and the dose rate decreasing to 20% of its maximum value. These latencies were measured during the first and subsequent gated segments to investigate changes to the linear accelerator’s temporal characteristics during automated delivery using the gating interface.

The service graphing feature of the linear accelerator control system was used to observe the status of the PRF relay and the dose rate of the linear accelerator, simultaneously. The service graphing feature allows treatment control system variables to be plotted with respect to time on the same set of axes. Item 44 part 4 was used to observe the dose rate of the linear accelerator and item 2201 part 190 was used to observe the status of the PRF relay. Values of 0 and 1 for the PRF relay status corresponded to an open and closed relay state, respectively. The dose rate was scaled by a factor of 0.01 so that both variables could be viewed clearly on the same set of axes. These
variables were plotted simultaneously as a function of time to measure the beam-on and beam-off latencies of the linear accelerator.

Ten separate plots of these two variables were acquired during the first and subsequent gated segments using accelerator # 2 in Table 2.1. Plots generated during the first gated segment were collected by initializing beam delivery while the PRF relay was in an open state, which prevented beam delivery to begin. The service graphing feature was triggered to begin collecting data and was programmed to stop collecting when the dose rate of the linear accelerator decreased to zero. Once data collection began, the PRF relay was switched to a closed state using the gating control software. The beam was delivered for approximately 10 seconds before the PRF relay was returned to an open state, which caused the service graphing feature to stop data collection.

Plots generated during the subsequent gated segment were acquired by first delivering the beam for approximately 10 seconds. Then, the PRF relay was switched open using the gating control software and left in this state for roughly 10 seconds. The 10 second beam-hold duration was chosen to be representative of the DIBH technique. After this beam-hold, the service graphing feature was triggered and the beam was delivered with the same procedure used during the first gated segment. The plots generated from these measurements were used to measure the latencies of the linear accelerator during the subsequent gated segment.

The beam-on and beam-off latency of the linear accelerator was measured on each plot generated. Because of the 4 hertz (Hz) sampling rate of the service graphing feature, the time at which the PRF relay opened or closed could not be defined precisely. Therefore beam-on and beam-off latency were measured with the PRF relay status at values of 0 and 1 to establish the range of potential latencies, as depicted in Figure 2.8. The latencies measured from the set of plots generated were used to calculate average linear accelerator latencies and associated standard
errors. A two-tailed independent t-test with a significance level of 0.05 was used to test the null hypothesis that there exists no statistically significant difference in the latency of the linear accelerator during the first and subsequent gated segments.

Figure 2.8. Plot depicting linear accelerator latencies. Beam-on latency was measured as the time between the PRF relay status (blue) rising from 0 (dashed arrow) or reaching 1 (solid arrow) and the dose rate (red) rising to 80% of its maximum value. Beam-off latency was measured as the time between the PRF relay decreasing from a value of 1 (solid arrow) or reaching 0 (not shown) and the dose rate decreasing to 20% of its maximum value.

2.3.2. End-to-End Approach

A linear relationship is expected between the displacement of a target from a reference position and its velocity during the gating process. The rate of change between these two variables is directly related to the magnitude of latency present in the gating system. Therefore, the end-to-end latency of the gating system was computed as the slope the target’s displacement plotted as a function of velocity.
The linear accelerator’s EPID (iViewGT, Elekta AB, Stockholm, Sweden) was used to capture the position of the target during the gating process, as discussed by Woods and Rong (2015) and Snyder et al. (2017). A dynamic respiratory phantom was used to simultaneously move the target and a reflective marker in orthogonal directions as depicted in Figure 2.9. The target was a 3 cm diameter plastic sphere embedded in a cylindrical cedar insert, as shown in Figure 2.10. The insert was attached to the translation stage of the phantom. The reflective marker was secured to the chest wall platform of the phantom and monitored using the tracking camera, as shown in Figure 2.9.

Reference images were acquired to define the position of the target when the beam was turned on or off during the gating process under the idealized conditions of no latency present in the gating system. Beam-on and beam-off reference images of the target were acquired with the reflective marker positioned at the midline and the bottom threshold of the gating window since these marker positions trigger the tracking software to send the beam-on and beam-off gating signal, respectively. The amplitude of the chest wall platform and attached reflective marker were manually adjusted using the dynamic phantom until the respiratory trace in the tracking software reached the midline or bottom threshold of the gating window. Setting the reference position of the target using the position of the reflective marker was possible because of the reproducible and simultaneous motion of the phantom’s translation stage and chest wall platform. The linear accelerator was operated in service mode to deliver 100 MU to the target in each reference position using a 6 MV beam energy. The EPID was used to record these exposures using single exposure and maximum frame averaging settings.
Experimental images were compared to the reference images to measure the displacement of the target when driven at various velocities. The experimental images were acquired with the phantom and gating window midline were kept in the same positions that were used to collect the reference images. This ensured that deviations in the position of the target in the experimental
images from its position in the reference image would be the result of the end-to-end latencies present in the gating system. Separate procedures were employed to acquire the experimental images used to evaluate beam-on and beam-off latency during the first and subsequent gated segments.

For the experimental images used to evaluate beam-on latency during the first gated segment, the EPID was initialized to record a single exposure and the linear accelerator control system was programmed to deliver 5 MU using a 6 MV beam energy. The beam was initialized while the PRF relay was in an open state, which prevented beam delivery to begin. The dynamic phantom was operated under software control to drive the reflective marker from the bottom to the top of the gating window at constant velocity. Once the marker’s trace passed the midline of the gating window the tracking software would signal to the gating interface to close the PRF relay and begin beam delivery. During beam delivery, the EPID captured frames of the target which were averaged together to produce a single image of the target’s travel throughout the 5 MU exposure. Frame averaging can improve image quality by reducing image noise but can also make it difficult to delineate the target in the image due to motion blurring, as discussed by (Yip et al., 2014).

Experimental images used to evaluate beam-on latency during the subsequent gated segment were acquired with linear accelerator control system programmed to deliver 8 MU. After the first 3 MU of beam delivery, the tracking software was used to open the PRF relay, causing an automated beam-hold. During the beam-hold, the EPID was initialized to record a single exposure. After approximately 10 seconds of the beam being held, the reflective marker was driven from the bottom to the top of the gating window at constant velocity, triggering beam delivery to begin. The EPID recorded the position of the target throughout the remaining 5 MU of beam delivery.
Multiple experimental images were acquired during the first and subsequent gated segments using target velocities ranging from 0.5 mm per second to 8 mm per second.

For the experimental images used to evaluate beam-off latency during the first gated segment the EPID was initialized to record a single exposure. While the beam was being delivered, the reflective marker began inside the gating window and was driven out through the bottom threshold of the gating window with constant velocity. The position of the target was recorded throughout the exposure. To capture images during the subsequent gated segment, the beam was delivered without the EPID initialized. During delivery of the first gated segment the tracking software was used open the PRF relay, causing an automated beam-hold. During the automated beam-hold, the EPID was initialized and the PRF relay was closed. While the second gated segment was being delivered, the reflective marker was driven out of the gating window. Approximately 1-5 MU were delivered during each image acquisition. Multiple experimental images were acquired during the first and subsequent gated segments using target velocities ranging from 1 mm per second to 48 mm per second; the images and their corresponding velocities were saved for later analysis.

All image analysis was preformed using open-source image processing software (Schneider et al., 2012). The images used to quantify beam-on and beam-off latency during the first and subsequent gated segments were compiled into separate image sets using the software. The images in each set were window and leveled with the same settings to obtain consistent image quality. The target was contoured on the reference image and the contour was overlaid on all of the other images in the set, as shown in Figure 2.11. The known diameter of the target was used to calibrate the image processing software, allowing distances within each image to be measured in mm rather than pixels.
Beam-on displacements were measured on each experimental image from the reference position contour to the same corresponding point on the target. The direction of motion was taken into account when choosing the point on the reference position contour to measure from so that the displacement resulted from the target’s travel before the beam was turned on, excluding the target’s travel during beam delivery, as shown in Figure 2.12. For beam-off displacements, the measurement points were chosen to ensure that the displacement would include the target’s travel during beam delivery, as shown in Figure 2.12.

![Images used to quantify end-to-end beam-on latency. The direction of target travel is specified by the arrow, shown in red. The contour of the reference position (yellow) and the target are shown for target velocities of: (a) 0 mm per sec (reference image) (b) 1.5 mm per second (c) 4.5 mm per second (d) 7.5 mm per second.](image-url)
Figure 2.12. Displacement between the reference position and the target. Direction of target travel is specified by the arrow, shown in red on each image. Displacements were measured from the reference position contour to the corresponding point on the target. (a) Beam-on displacement with the target moving at 7.5 mm per second. (b) Beam-off displacement with the target moving at 32 mm per second.

Average displacements were calculated from the set of repeat measurements collected at each velocity. Beam-on and beam-off displacements were plotted separately against the known velocities of the target. A linear fit was applied to each graph using a weighted least squares fitting routine. The slope of linear fit was used to quantify the end-to-end latency of the gating system. A two-tailed independent t-test with a significance level of 0.05 was used to test the null hypothesis that there exists no statistically significant difference between the gating system’s end-to-end latencies and the summation of its individual component latencies.
Chapter 3. Results

3.1. Aim 1: Output and Energy Constancy

Central-axis output and beam energy were measured with and without the use of automated beam-holds. The constancy and associated uncertainty for these dosimetric quantities were calculated using equation (2.3.) and equation (2.4.), respectively.

3.1.1. Output Constancy

The results for central-axis output constancy are shown graphically in Figure 3.1 for each of the linear accelerators utilized, numbered 1 through 8. Flattening filter-free (FFF) beams were also studied using linear accelerator # 2 for the 6 MV and 10 MV beam energies, as shown in Figure 3.1. For every beam energy investigated, output constancy was within ± 0.25% across all linear accelerators studied. The error bars in Figure 3.1 represent the standard deviation in each output constancy and were calculated using equation (2.4.); these standard deviations ranged from 0.021% to 0.167%.

3.1.2. Energy Constancy

In this work, TPR and PDD were used as the surrogates for photon and electron beam energy, respectively. The results for beam energy constancy are depicted in Figure 3.2 for each of the linear accelerators studied, numbered 1 through 8. A 6 MV and a 10 MV FFF beam were also studied using linear accelerator # 2, as shown in Figure 3.2. For each of the photon energies investigated, TPR constancy was within ± 0.3% across all linear accelerators studied. The error bars in Figure 3.2 represent the standard deviation in each energy constancy and were calculated using equation (2.4.); the standard deviations in TPR constancy ranged from 0.047% to 0.197%. The PDD constancy for a 10 MeV electron beam was found to be within ± 0.5% across all linear accelerators studied. The standard deviations in PDD constancy ranged from 0.068% to 0.406%.
Figure 3.1. Central-axis output constancy for (a) 6 MV photons (b) 10 MV photons (c) 15 MV photons (d) 10 MeV electrons.

Figure 3.2. TPR and PDD constancy for (a) 6 MV photons (b) 10 MV photons (c) 15 MV photons (d) 10 MeV electrons.
3.2. **Aim 2: Treatment Delivery Accuracy**

A set of treatment plans from representative anatomic sites were delivered to linear accelerators numbered 1 through 7. Each treatment plan was delivered under non-gated and gated modes of operation. The gated mode of operation involved use of the gating system under conditions that simulated automated delivery of the DIBH technique. Analysis was performed on the measured dose distributions obtained in each mode of operation to assess changes in the delivery accuracy resulting from use of the gating system.

3.2.1. Analysis

The passing rates for each treatment plan using a gamma criteria of (3% dose difference, 3mm DTA) are shown in Table 3.1 and Table 3.2 for the non-gated and gated modes of operation, respectively. All treatment deliveries in both modes of operation achieved passing rates between 94% and 100%. Additionally, the passing rates for each plan included in the set were averaged across all linear accelerators studied. The average passing rate for each treatment plan is depicted in Figure 3.3 and Figure 3.4 using gamma criteria of (3% dose difference, 3mm DTA) and (2% dose difference, 1mm DTA), respectively.

The passing difference was calculated for each plan included in the set using equation (2.5.). The average passing difference and standard error for the set of treatment plans delivered to each linear accelerator is shown graphically in Figure 3.5. Average passing differences on these linear accelerators were found to be within ± 0.2% and ± 0.4% using gamma criteria of (3% dose difference, 3mm DTA) and (2% dose difference, 1mm DTA), respectively.
Table 3.1. Gamma Passing Rates (3%, 3mm) for Non-Gated Treatment Deliveries

<table>
<thead>
<tr>
<th>Linear Accelerator #</th>
<th>Lung VMAT</th>
<th>Pancreas VMAT</th>
<th>Chest Wall VMAT</th>
<th>Breast 3D-CRT</th>
<th>Lung SBRT</th>
<th>Average Passing Rate of Set (%)</th>
<th>Standard Deviation</th>
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<td>0.7</td>
</tr>
</tbody>
</table>

|                           | 99.0      | 99.4          | 98.6            | 97.6          | 100.0    |                                |                   |
| Average Passing Rate of Plan (%) | 1.0      | 0.9           | 1.4             | 1.6           | 0.0      |                                |                   |

Table 3.2. Gamma Passing Rates (3%, 3mm) for Gated Treatment Deliveries

<table>
<thead>
<tr>
<th>Linear Accelerator #</th>
<th>Lung VMAT</th>
<th>Pancreas VMAT</th>
<th>Chest Wall VMAT</th>
<th>Breast 3D-CRT</th>
<th>Lung SBRT</th>
<th>Average Passing Rate of Set (%)</th>
<th>Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>99.6</td>
<td>99.7</td>
<td>99.2</td>
<td>96.5</td>
<td>100</td>
<td>99.0</td>
<td>1.4</td>
</tr>
<tr>
<td>2</td>
<td>99.3</td>
<td>99.7</td>
<td>99.5</td>
<td>96.6</td>
<td>100</td>
<td>99.0</td>
<td>1.4</td>
</tr>
<tr>
<td>3</td>
<td>99.6</td>
<td>97.6</td>
<td>99.3</td>
<td>94.2</td>
<td>100</td>
<td>98.1</td>
<td>2.4</td>
</tr>
<tr>
<td>4</td>
<td>100</td>
<td>99.9</td>
<td>99.3</td>
<td>99.3</td>
<td>100</td>
<td>99.7</td>
<td>0.4</td>
</tr>
<tr>
<td>5</td>
<td>97.1</td>
<td>99.9</td>
<td>95.5</td>
<td>99.2</td>
<td>100</td>
<td>98.3</td>
<td>2.0</td>
</tr>
<tr>
<td>6</td>
<td>98.9</td>
<td>99.1</td>
<td>98.3</td>
<td>98.5</td>
<td>100</td>
<td>99.0</td>
<td>0.7</td>
</tr>
<tr>
<td>7</td>
<td>98.9</td>
<td>98.8</td>
<td>98.8</td>
<td>98.8</td>
<td>100</td>
<td>99.1</td>
<td>0.5</td>
</tr>
</tbody>
</table>

|                           | 99.1      | 99.2          | 98.6            | 97.6          | 100.0    |                                |                   |
| Average Passing Rate of Plan (%) | 1.0      | 0.8           | 1.4             | 1.9           | 0.0      |                                |                   |

| Standard Deviation       | 1.0      | 0.8           | 1.4             | 1.9           | 0.0      |                                |                   |
Figure 3.3. Average gamma pass rates for treatment plans delivered using a (3%, 3mm) gamma criteria. Non-gated deliveries are shown in grey and gated deliveries are shown in blue.

Figure 3.4. Average gamma pass rates for treatment plans delivered using a (2%, 1mm) gamma criteria. Non-gated deliveries are shown in grey and gated deliveries are shown in blue.
Figure 3.5. Average difference in gamma pass rates between non-gated and gated modes of operation.

The results for the percent dose difference comparison are depicted in Figure 3.6 for linear accelerators numbered 1 through 7. Average passing rates of 99% or greater were observed using a passing criterion of 1%. The standard error associated with each average passing rate was used to calculate the error bars shown in Figure 3.6.

To assess delivery reproducibility, the same lung VMAT treatment plan was delivered three times in each mode of operation. The gamma passing rate for each delivery was calculated using gamma analysis with gamma criteria of (3% dose difference, 3mm DTA). The average passing rates and associated standard deviations of these deliveries are shown in Figure 3.7 for each linear accelerator investigated. Linear accelerator #5 exhibited a standard deviation of 0.4% in both modes of operation, which was the largest standard deviation observed for any of the linear accelerators studied. Linear accelerators #4 and #7 were the only linear accelerators that showed different standard deviations between non-gated and gated modes of operation. In each case, the
standard deviations observed in the gated mode of operation were larger by just 0.1% relative to the non-gated mode of operation.

Figure 3.6. Percent dose difference comparison between non-gated and gated treatment deliveries using passing criterion of 3% (grey) and 1% (blue).

Figure 3.7. Repeat delivery of a lung VMAT treatment plan. The average gamma passing rate and standard deviation for non-gated (grey) and gated (blue) modes of operation.
3.3. Aim 3: Latency

3.3.1. Modular Latency Components

The modular components that contributed to the overall latency of the gating system were each evaluated separately. The components of the respiratory motion monitoring system included the streaming latency of the tracking camera, the sampling rate of the tracking software, and the travel time of the gating signal from the tracking software to the gating interface. The magnitude of these various components and their associated uncertainty are listed in Table 3.3. The total latency of the respiratory motion monitoring system was found to be 200 ms. Additionally, the range of linear accelerator beam-on and beam-off latencies during the first and subsequent gated segments are tabulated in Table 3.4. A statistically significant difference in the linear accelerator’s beam-on latency was observed t(9)=2.8, p=.02, between the first (M=2.75, SD= 0.25) and subsequent (M=1.35, SD= 0.25) gated segments. A statistically significant difference in the linear accelerator’s beam-off latency was not observed t(9)=0.17, p=.87, between the first (M=0.37, SD= 0.25) and subsequent (M=0.29, SD= 0.25) gated segments.

Table 3.3. Respiratory Motion Monitoring System Component Latencies

<table>
<thead>
<tr>
<th>Component</th>
<th>Latency (Seconds)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tracking Camera Streaming Latency</td>
<td>0.15 ± 0.02</td>
</tr>
<tr>
<td>Tracking Software (Time between samples)</td>
<td>0.052 ± 0.011</td>
</tr>
<tr>
<td>Signal Travel Time</td>
<td>3.7 x 10^{-5} ± 2 x 10^{-5}</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>0.20 ± 0.02</strong></td>
</tr>
</tbody>
</table>
### Table 3.4. Linear Accelerator Latency

<table>
<thead>
<tr>
<th>Component</th>
<th>Latency Range (Seconds)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Linear Accelerator Beam-on Latency (First Gated Segment)</td>
<td>2.50 - 2.75 ± 0.25</td>
</tr>
<tr>
<td>Linear Accelerator Beam-on Latency (Subsequent Gated Segment)</td>
<td>1.10 - 1.35 ± 0.25</td>
</tr>
<tr>
<td>Linear Accelerator Beam-off Latency (First Gated Segment)</td>
<td>0.12 - 0.37 ± 0.25</td>
</tr>
<tr>
<td>Linear Accelerator Beam-off Latency (Subsequent Gated Segment)</td>
<td>0.04 - 0.29 ± 0.25</td>
</tr>
</tbody>
</table>

#### 3.3.2. End-to-End Latency

Plots of target displacements used to quantify end-to-end beam-on latency are shown in Figure 3.8 for the first and subsequent gated segments. Plots used to quantify end-to-end beam-off latency are shown Figure 3.9 for the first and subsequent gated segments. The parameters obtained from the linear fits shown in each plot are tabulated in Table 3.5. End-to-end beam-on latency was found to be 3.18 seconds and 1.49 seconds during the first and subsequent gated segments, respectively. Alternatively, the end-to-end beam-off latency was found to be 0.35 seconds and 0.34 seconds during the first and subsequent gated segments, respectively. End-to-end latency measurements are compared to the component totals in Table 3.6. The component totals shown in Table 3.6 include the latency of the respiratory motion monitoring system and the linear accelerator. Two separate component totals are shown for each entry in Table 3.6 to account for the range of linear accelerator latencies measured. In each case, no statically significant differences were found between the end-to-end latency of the gating system and its measured component totals.
Figure 3.8. End-to-end beam-on displacements during the (a) first gated segment (b) subsequent gated segment.
Figure 3.9. End-to-end beam-off displacements during the (a) first gated segment (b) subsequent gated segment.
Table 3.5. Displacement vs. Velocity Fitting Parameters

<table>
<thead>
<tr>
<th></th>
<th>Beam-On (First Segment)</th>
<th>Beam-On (Subsequent Segment)</th>
<th>Beam-Off (First Segment)</th>
<th>Beam-Off (Subsequent Segment)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Degrees of Freedom</td>
<td>13</td>
<td>14</td>
<td>31</td>
<td>31</td>
</tr>
<tr>
<td>Slope (Seconds)</td>
<td>3.18</td>
<td>1.49</td>
<td>0.35</td>
<td>0.34</td>
</tr>
<tr>
<td>Standard Error of Slope</td>
<td>0.05</td>
<td>0.03</td>
<td>0.01</td>
<td>0.01</td>
</tr>
<tr>
<td>Y-Intercept (mm)</td>
<td>-0.51</td>
<td>-0.34</td>
<td>-0.39</td>
<td>-0.72</td>
</tr>
<tr>
<td>Standard Error of Y-Intercept</td>
<td>0.12</td>
<td>0.14</td>
<td>0.14</td>
<td>0.10</td>
</tr>
<tr>
<td>( R^2 )</td>
<td>0.997</td>
<td>0.994</td>
<td>0.981</td>
<td>0.994</td>
</tr>
<tr>
<td>Chi-Squared</td>
<td>0.83</td>
<td>0.93</td>
<td>1.03</td>
<td>0.82</td>
</tr>
<tr>
<td>Component Total</td>
<td>End-to-End</td>
<td>t-value</td>
<td>p</td>
<td></td>
</tr>
<tr>
<td>-----------------</td>
<td>------------</td>
<td>---------</td>
<td>----</td>
<td></td>
</tr>
<tr>
<td><strong>Beam-On Latency (Seconds)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>First Gated Segment</td>
<td>2.70 ± 0.25</td>
<td>3.18 ± 0.05</td>
<td>t(df=13)=1.59</td>
<td>.14</td>
</tr>
<tr>
<td></td>
<td>2.95 ± 0.25</td>
<td>3.18 ± 0.05</td>
<td>t(df=13)=0.76</td>
<td>.46</td>
</tr>
<tr>
<td>Subsequent Gated Segment</td>
<td>1.30 ± 0.25</td>
<td>1.49 ± 0.03</td>
<td>t(df=14)=0.67</td>
<td>.51</td>
</tr>
<tr>
<td></td>
<td>1.55 ± 0.25</td>
<td>1.49 ± 0.03</td>
<td>t(df=14)=0.21</td>
<td>.83</td>
</tr>
<tr>
<td><strong>Beam-Off Latency (Seconds)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>First Gated Segment</td>
<td>0.33 ± 0.25</td>
<td>0.35 ± 0.01</td>
<td>t(df=31)=0.10</td>
<td>.92</td>
</tr>
<tr>
<td></td>
<td>0.58 ± 0.25</td>
<td>0.35 ± 0.01</td>
<td>t(df=31)=0.87</td>
<td>.39</td>
</tr>
<tr>
<td>Subsequent Gated Segment</td>
<td>0.24 ± 0.25</td>
<td>0.34 ± 0.01</td>
<td>t(df=31)=0.40</td>
<td>.69</td>
</tr>
<tr>
<td></td>
<td>0.49 ± 0.25</td>
<td>0.34 ± 0.01</td>
<td>t(df=31)=0.56</td>
<td>.58</td>
</tr>
</tbody>
</table>
Chapter 4. Discussion

4.1. Aim 1: Output and Energy Constancy

The output and energy constancies reported in this work were all within ± 0.5%, which was generally consistent with the findings of Saito et al. (2018), who reported constancies within 0.5% when the beam-on durations were 2 seconds or greater. They also performed these measurements with gating cycles consisting of 1.1 second beam-on durations followed by 2 second beam-holds and reported larger output constancies approaching 1% for these deliveries, but an increase in output constancy is expected when utilizing low duty cycles during gated delivery (Freislederer et al., 2015).

To our knowledge, this is the first work to assess the output and energy constancy of an electron beam resulting from automated beam gating. The output constancies and associated uncertainties observed using electrons were similar to those observed using photons. In contrast, the uncertainties associated with electron PDD constancy tended to be larger than those observed for photon TPR measurements. This may be attributed to the steeper dose falloff at depth that electrons exhibit relative to photons which could result in larger variations in the repeated measurements of electron PDD compared to photon TPR.

Output and energy constancy were evaluated across 8 beam-matched linear accelerators using a variety of beam energies. This allowed for the intercomparison of gating performance across linear accelerators and beam energies. Ultimately, the results reported in this work support the hypothesis that the gating system can achieve output and energy constancy within 2% during automated deliveries that are representative of the DIBH technique.
4.2. **Aim 2: Treatment Delivery Accuracy**

In this work, gamma analysis was used to evaluate the agreement between the measured and planned dose distributions for each treatment delivery. The passing difference represented the change in gamma passing rates between non-gated and gated modes of operation. This metric could be misleading if the passing rates for both non-gated and gated treatment deliveries were similar but demonstrated poor agreement to the planned dose distribution. This situation would yield a small passing difference, even though the treatment plan was not accurately delivered in each mode of operation. This was not a concern of this work because of the relatively high gamma passing rates achieved in each mode of operation. Our findings support the hypothesis that the gating system could facilitate gated treatment deliveries with passing differences within 2% on average.

Delivery accuracy was also evaluated using the percent dose difference comparison. While the passing difference compared the two modes of operation indirectly using the planned dose distribution, the percent dose difference comparison allowed for deliveries in each mode of operation to be compared directly to one another. Noto *et al.* (2014) reported passing rates of 100% using a percent dose difference comparison with a 1% passing criterion. However, the manual beam-holds that were employed in their work may not adequately replicate the delivery characteristics encountered with use of an automatic gating interface. In our work, average passing rates ranged from 99% to 100% depending on the linear accelerator being studied. These results demonstrate a high level of agreement between treatments delivered in the non-gated and gated modes of operation.

A single lung VMAT treatment plan was delivered three times in each mode of operation. The standard deviation in the gamma passing rates of these repeat deliveries was used to assess delivery reproducibility. The largest standard deviation observed was just 0.4%, demonstrating
that in each mode of operation, the treatment plan was able to be delivered in a reproducible manner. Additionally, the differences in the standard deviations between modes of operation suggest that the gating system could facilitate automated delivery of the DIBH technique with a level of reproducibility that is comparable to the non-gated mode of operation.

4.3. Aim 3: Latency

4.3.1. Modular Latency Components

This work presented a modular approach to quantify the latencies of a gating system. One of the benefits of a modular approach is that relevant latencies can be substituted for gating systems that are composed of different components than the ones studied in this work. Of the component latencies reported for the respiratory motion monitoring system, the latency of the tracking camera was greatest. The total latency of the respiratory motion monitoring system, including the latency of the tracking camera, was found to be 200 ms. This finding was consistent with the latency of other respiratory motion monitoring systems that utilize optical tracking as discussed by Freislederer et al. (2015), who reported values ranging from 162 to 262 ms.

The other major contributor to the overall latency of the gating system was the latency of the linear accelerator (Freislederer et al., 2015). In this work, the beam-on or beam-off latency of the linear accelerator was defined as the time delay between the PRF relay changing to the appropriate state and the dose rate of the linear accelerator rising to 80% of its maximum value or decreasing to 20% of its maximum value, respectively. During the first and subsequent gated segments, the beam-on latency of the linear accelerator was found to be larger than its beam-off latency. This is expected, because beam startup involves magnetron tuning and electron gun ramp-up whereas initiating automated beam-holds simply require interrupting the linear accelerator’s digital PRF interlock (Saito et al., 2018). Based on our results, the respiratory motion monitoring
system is the major contributor to the gating system’s beam-off latency and the linear accelerator is the major contributor to the gating system’s beam-on latency, which is consistent with the findings of Saito et al. (2018).

A statistically significant difference in the linear accelerator’s beam-on latency was observed between the first and subsequent gated segments. Saito et al. (2018) attributed this to the electron gun remaining at its operating level during beam-holds, allowing quicker start up during the subsequent gated segments compared to the first segment of gated delivery. The electron gun-hold time is an adjustable parameter on certain linear accelerators (Elekta Oncology Systems, Stockholm, Sweden) that determines the amount of time that the electron gun will remain at its operating level following a beam-hold. Previous work has shown that when the duration of a beam-hold exceeds the electron gun-hold time, the gun will enter a stand-by state which requires additional ramp-up time to return to its operating level before beam delivery can resume (Cui et al., 2014). In this work, the beam-holds that occurred prior to the subsequent gated segment were purposefully chosen to exceed the maximal gun-hold time achievable on the linear accelerator of 6.5 seconds. Even with the additional ramp-up time resulting from the electron gun entering a stand-by state, a statistically significant difference in beam-on latency was still observed between the first and subsequent gated segments. Therefore, the observed difference in beam-on latency is likely the result of linear accelerator startup processes that only occur during the first gated segment as well as quicker electron gun ramp-up times that occur prior to the subsequent gated segment compared to the first gated segment. No statistically significant difference in the linear accelerator’s beam-off latency was observed between the first and subsequent gated segments. To our knowledge, this is the first study that has quantified the latency of the linear accelerator during the first and subsequent gated segments using this gating interface.
4.3.2. End-to-End Latency

Negative offset values were observed for each of the linear fits performed. Ideally each offset should be zero, because at zero velocity there should be no displacement of the target from the reference position. This suggests a systematic error in our experimental methodology, which could have resulted from incorrectly setting the marker surrogate at the appropriate gating thresholds when establishing the reference position. However, this type of systematic error is not expected to impact the latencies reported in this work because the resulting offsets would not affect the slopes of the linear fits performed. Woods and Rong (2015) performed similar measurements and also observed negative offset values, which they discussed was likely the result of measurement uncertainty and finite pixel size. In our work, the observed offsets were within the length of a single pixel used to track the marker surrogate.

The end-to-end beam-on latency of the gating system was found to be 3.18 seconds and 1.49 seconds for the first and subsequent gated segments, respectively. Others in the literature have reported considerably smaller beam-on latencies than those reported in this work (Saito et al., 2018; Snyder et al., 2017). However, these works were evaluating the latencies associated with free-breathe respiratory gating, which utilizes shorter beam-holds than those encountered with the DIBH technique. Therefore, the linear accelerator’s electron gun was able to remain at its operating level throughout each beam-hold, which is likely the reason for the considerably shorter beam-on latencies that these works have reported. The end-to-end beam-on latency reported in this work for the subsequent gated segment of 1.49 seconds was consistent with the findings of Cui et al. (2014), who utilized a similar respiratory motion monitoring system. They reported beam-on latencies of 1.38, 1.44, and 1.49 seconds using a continuously variable dose rate and beam-holds that exceeded the electron gun-hold time, both of which were conditions utilized in this work.
For beam-off latency, measured displacements were found to deviate from the linear fit at high target velocities. This is likely the result of the target passing the gating threshold before it accurately reached its programmed velocity. To assess the impact that these data points had on the reported beam-off latencies, the linear fits were also performed without data from target velocities greater than 24 mm per second. For both the first and subsequent gated segments, excluding these data points changed the resultant latency values by less than 30 ms. Therefore, the inclusion of these high velocity measurements did not impact the major findings reported in this work. End-to-end beam-off latency was found to be 350 ms and 340 ms for the first and subsequent gated segments, respectively, which exceeded the hypothesized value of 300 ms. Moving forward, the tracking software could be adjusted to compensate for the gating system’s beam-off latency by signaling to the gating interface to initiate a beam-hold prior to the surrogate exiting the gating window. However, this adjustment could prematurely terminate the breath-hold cycle if the patient trended toward, but was able to remain within, the gating window. Another option would be to replace the tracking camera with one that has less streaming latency, as the tracking camera’s streaming latency was found to be the largest contributor to the gating system’s beam-off latency.

In every case tested, no statistically significant difference was observed between end-to-end latency derived from the linear fit and the summation of the gating system’s measured component latencies. This agreement lends credibility to both the modular approach and the methods used for end-to-end measurements. The methods described in this work to quantify latency are generalizable to other respiratory gating systems that utilize this gating interface. However, some degree of variability in the latencies measured for other systems are expected. Within the control system of the linear accelerator, several parameters can be adjusted to alter the
temporal characteristics of gated delivery, particularly with regards to beam-on latency (Cui et al., 2014; Snyder et al., 2017).

It is expected that the beam-on latencies reported in this work could be reduced by altering several parameters within the control system of the linear accelerator. However, there is some tradeoff between the time it takes for the linear accelerator to begin producing radiation pulses and the dosimetric stability of those pulses during beam startup. A potential consequence of altering the control system configuration of the linear accelerator to diminish the gating system’s beam-on latency is dosimetric degradation, which was not studied in this particular work. For our application, we didn’t need to decrease the beam-on latency of the gating system given the duration of the breath-hold intervals used clinically. For systems that are intended for free-breathe gating applications, optimizing the control system configuration of the linear accelerator to minimize the gating system’s beam-on latency may be necessary.
Chapter 5. Summary and Conclusions

5.1. Limitations and Future Work

One of the limitations of this work is the different number of monitor units that were delivered in the non-gated and gated modes of operation to measure central-axis output and energy constancy. Measurements collected in the gated mode of operation were more likely to be affected by electrometer leakage because of the longer delivery times utilized in the gated mode of operation compared to the non-gated mode of operation. While it is expected that the contribution of the electrometer leakage on the measured constancies was negligible, future work could address this by utilizing comparable delivery times for measurements made in each mode of operation.

Another limitation is the use of measured displacements to quantify end-to-end latencies, which relied on visual identification to define the appropriate edge of the target in each image. Motion blurring and poor image quality at MV beam energies makes this method susceptible to error. Additionally, the service graphing feature provided a convenient method for quantifying the latency of the linear accelerator, but its sampling frequency limited the achievable precision of this technique. Future work could evaluate the temporal characteristics of the respiratory gating system using methods described by Wiersma et al. (2016), which would provide higher temporal resolution when measuring beam-on and beam-off latencies.

It was shown that during automated delivery, there was a statistically significant difference in the linear accelerator’s beam-on latency between the first and subsequent gated segments. Often times, treatments involving the DIBH technique are delivered by manually activating and interrupting the radiation beam at the appropriate times. Future work is needed to determine if the increased beam-on latency observed during the first gated segment occurs after each manual beam-hold. If so, automated delivery of DIBH treatments using this gating interface may increase the delivery efficiency compared to manual delivery. Regardless, the differences in beam-on latency
observed in this work between the first and subsequent gated segments are not clinically meaningful for gating systems intended for DIBH applications, given the duration of the breath-hold intervals used clinically. Future work is needed to investigate these beam-on latencies when the linear accelerator control system is optimized for free-breathe respiratory gating, as latency differences between the first and subsequent gated segments could have a clinically meaningful impact for this type of application.

5.2. Conclusions

This work characterized the performance of a respiratory gating system both dosimetrically and temporally. The gating system utilized an automatic gating interface to connect a marker-based respiratory motion monitoring system to the linear accelerator control system. To our knowledge, this is the first work to extensively validate this automatic gating interface for use with an abdominal marker-based optical tracking system. Dosimetric characteristics associated with use of the gating system were evaluated across multiple linear accelerators which allowed for intercomparison of its gating performance. It was shown that the gating system can achieve the desired dosimetric operating characteristics during automated delivery of the DIBH technique, which included output and energy constancy within 2% and an average decrease in gamma passing rates between the non-gated and gated modes of operation that was less than 2%. The hypothesis of this work was not fully supported, as the gating system was not able to achieve all of the desired operating characteristics, in particular, the system’s beam-off latency was found to exceed 300 ms. However, now that a baseline has been established, adjustments can be made to compensate for these measured delays. The methodology presented in this work can be used by others to validate the dosimetric and temporal characteristics of their own respiratory gating systems that utilize this automatic gating interface.
References


Choi S H and Seong J 2018 Stereotactic Body Radiotherapy: Does It Have a Role in Management of Hepatocellular Carcinoma? *Yonsei Medical Journal* 59 912


Pedersen A N, Korreman S, Nyström H and Specht L 2004 Breathing adapted radiotherapy of breast cancer: reduction of cardiac and pulmonary doses using voluntary inspiration breath-hold *Radiotherapy and Oncology* 72 53-60


Smith W L and Becker N 2009 Time delays in gated radiotherapy *J Appl Clin Med Phys* 10 2896


Tsai H-Y, Chen M-C, Tsai I C and Chen C C-C 2011 Partial ring artifact on cardiac CT: image presentation and clinical implication *The International Journal of Cardiovascular Imaging* 27 689-93


Yip S, Rottmann J and Berbeco R 2014 The impact of cine EPID image acquisition frame rate on markerless soft-tissue tracking *Medical Physics* 41 061702

Vita

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