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PATIENT MOTION MANAGEMENT WITH 6DOF ROBOTICS FOR FRAMELESS
AND MASKLESS STEREOTACTIC RADIOSURGERY

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For my mother

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ABSTRACT

Stereotactic radiosurgery (SRS) is a treatment modality capable of delivering high radiation dose to one or more cranial lesions. Current approaches include frame-based and frameless techniques. The former involves the use of a rigid metal frame to ensure patient positioning, and the latter uses a thermoplastic mask and patient motion monitoring to determine when patient movement exceeds clinical thresholds. Recent studies have also shown that the rotational motion of the patient can hinder target coverage and degrade the treatment plan, particularly in cases where lesions are a distance away from the treatment isocenter, such as when multiple targets are treated with a single isocenter. Our goal is to achieve sub-millimeter and sub-degree patient head motion control for patients undergoing frameless and maskless intracranial SRS using an approach which improves upon conventional treatment protocols. Our solution is to monitor and compensate for intrafraction head motion in 6 degrees-of-freedom (6DOF) in real-time using an in-house designed and built parallel kinematics robotic motion compensation system. To that end, we present the research and set of experiments required to reach this goal in this dissertation.

The research to achieve this objective is presented in six chapters, focusing first on the theory behind parallel robot control and the design and construction of a 6DOF robotic system, then the development of a general calibration methodology and first use of the robot as a phantom to evaluate clinical motion tracking systems, and finally the creation and implementation of a motion compensation algorithm for use with human subjects. Additional experiments help support these goals, which include the collection and analysis of both volunteer and real patient motion to better inform the control algorithms and system design, and target dose coverage improvements when using the robot for real-time head motion stabilization for a single-isocenter multi-target SRS treatment.

This work presents a novel approach to SRS treatments through the development of a robotic motion compensation system. By integrating real-time robotic head motion correction into frameless and maskless SRS, it may be possible to improve upon the current SRS

treatment regimes by streamlining clinical workflow, minimizing patient discomfort, and achieving target dose coverage with accuracy as good or better than traditional frame-based intracranial SRS.

CHAPTER 1

INTRODUCTION

1.1 Dissertation overview

patient motion can occur along the left-right (LR), superior-inferior (SI), and anterior-posterior (AP) directions, it is also important to consider rotational motion along the pitch (around LR), roll (around SI), and yaw (around AP) axes. [36, 35, 37, 71] Taking all 3 translational and 3 rotational directions into account during patient motion monitoring is termed 6 degree of freedom (6DOF) tracking

Patient motion during radiation therapy can negatively impact dose delivery, reducing target coverage and increasing undesired exposure to healthy tissues. For highly conformal treatments of brain disorders, patient motion can occur along the left-right (LR), superior-inferior (SI), and anterior-posterior (AP) directions, it is also important to consider rotational motion along the pitch (around LR), roll (around SI), and yaw (around AP) axes, degrading the treatment. Taking all 3 translational and 3 rotational directions into account during patient motion monitoring is termed 6 degree of freedom (6DOF). In this work, we present the development of a robotic system capable of stabilizing patient head motion during high-dose radiotherapy in 6DOF by considering translational and rotational degrees of freedom. Specifically, we describe the system's development by first discussing the control theory of parallel kinematics robots, as well as the design and fabrication of the system. Next, we construct a calibration algorithm for use with the robot and patient motion tracking systems. The robot is then used to perform quality assurance on these tracking systems. Finally, a robotic motion compensation algorithm for brain radiotherapy is created, and experimentally evaluated using phantoms as well as volunteers. These components of developing a 6DOF robotic head stabilization system for stereotactic radiosurgery (SRS), as well as project motivations and background information, are presented in this dissertation.

The remainder of Chapter 1 explores the history of medical robotics and automation, as

well as developments radiation therapy, with a focus on brain disorder treatments. The chapter continues by motivating the research presented in this dissertation, specifically discussing patient motion management techniques, and the role that robotics may play for real-time non-invasive immobilization.

Chapter 2 presents the control theory behind parallel kinematics robots, one class of robots which has particular advantages for real-time head motion stabilization during SRS. Both inverse kinematics and forward kinematics are described in this chapter, though inverse kinematics formed the foundation for the majority of robotic control throughout the research presented in this dissertation.

Chapter 3 describes the design process of the parallel kinematics robot, as well as its fabrication. The robot's control accuracy performance is explored, and its 6DOF workspace based on design choices are shown. Chapter 3 also includes simulations of the 6DOF robot using the inverse kinematics approach, which helped refine the robot's design. Finally, this chapter presents an evaluation of the robot's ability to reproduce input 6DOF motion, which included previously recorded cranial movements.

Chapter 4 presents the development of a comprehensive and generalizable calibration procedure. The theoretical foundation behind calibration in 6DOF is established, as well as the geometry of a clinical setup with the robot, the target position, and the motion tracking system. After presenting the theory, the calibration protocol used throughout this project is described.

Chapter 5 builds on the inverse kinematics and calibration formalisms, and uses the parallel kinematics robot to evaluate the tracking capabilities of three clinically-used patient motion monitoring systems. Another related study into the motion of patients immobilized using an open-face mask during whole-brain radiation therapy treatments is also included in this chapter.

Chapter 6 returns to performing real-time head motion stabilization by presenting a 6DOF motion compensation control algorithm. An earlier generation 4DOF (XYZ + pitch)

robot is first described, and the improvements on a control algorithm for motion compensation are presented. A 6DOF compensation algorithm is then discussed in detail, and an example control response is presented.

Chapter 7 brings together the previous chapters to evaluate the 6DOF head motion stabilization system for SRS. The system performance is evaluated in both phantom and volunteer studies. Methods to evaluate the 6DOF head motion compensation system performance are then described, and results from research into the dose target coverage in a single-isocenter multiple-target SRS treatment are presented.

Chapter 8 presents future work to introduce the head motion compensation system into the clinic in a frameless and maskless SRS setting, as well as concludes the dissertation.

1.2 Background

1.2.1 Medical robotics

The role robotics has played in the developed world and everyday life has evolved from myth to mainstream use, particularly in the last half century. Until recently, robots have existed solely in the realm of science fiction, as instruments which simplify or complicate different areas of human life. Fiction transitioned into function toward the end of the 1950s, when General Motors developed the Unimate system to aid in automobile production. [3] After their initial adoption in 1961, the use of robots in a variety of applications has grown substantially, with the overall goal of reproducing and improving upon human functions.

This evolution of automation has impacted the development and direction of numerous fields. From manufacturing and medicine to toys and household vacuums, robotic systems have become important fixtures in the modern world. In the healthcare industry, the first application of robotics is considered to have taken place in 1985 with PUMA 560, which is used in brain biopsy, and in 1988 with PROBOT, used in prostate surgery. [4] In 1990, a robotic arm and associated software platform, named Aesop and Hermes, respectively,

were the first systems for surgery assistance. Aesop was used for laparoscopic gynecological surgery, and was found to both reduce surgical time and improve performance. [5] Additions were made to this system to create the Zeus surgical system, which featured a design intended to mimic the motions of human surgeons. [4] In the early 2000s, the da Vinci robot (da Vinci Surgery, Intuitive Surgical Inc.) began aiding in surgeries, and has seen widespread media attention for the proposed advantages it has over conventional surgery techniques [6].

Technological advances and publicity have further propelled robotics from surgical operating rooms into the radiation oncology clinics around the world. Modern radiation therapy treatments use multiple robotic systems which work together to improve target coverage and overall patient care. The use of rotating linear accelerator gantries and robotic patient couches, for example, have become the standard-of-care. Recently, advancements have improved upon patient couches with 6DOF (translations and rotations) robotic couches, capable of accurately positioning the patient prior to treatment. [7, 8] In addition to rotating gantries, systems such as CyberKnife (Accuray, Sunnyvale, CA) use industrial robotics to provide more flexibility during radiotherapy. This system features a lightweight linear accelerator fitted to a robotic arm to provide 6DOF movement of the beam. [9] Figure 1.1 shows the CyberKnife treatment system and a 6DOF patient positioning couch. With an overall motivation of improving the accuracy of radiation therapy, the expanding use of robotics can have a profound impact on how patients are treated. [10]

1.2.2 Radiation therapy

The goal of radiation therapy is to deliver ionizing radiation to a target in order to damage the DNA of cancerous tissues, leading to cell death, while minimizing dose to healthy tissues. [11] Since the early uses of radiation in therapeutic applications, many advancements have been made which attempt to achieve this objective, which can generally be considered as increasing dose conformality. From square fields aimed at blocking the beam delivery to regions outside the target, to 3D conformal radiation therapy (3DCRT), and



Figure 1.1: CyberKnife robotic radiation delivery system, and a 6DOF robotic couch. The CyberKnife system features a linear accelerator attached to a 6DOF robotic arm. (Photo courtesy of Accuray Inc. of Sunnyvale, CA. Copyright 2017, Accuray Incorporated)

finally to stereotactic radiosurgery (SRS), intensity modulated radiation therapy (IMRT), and volumetric modulated arc therapy (VMAT), the past two decades of radiation therapy approaches have consistently improved the delivery of dose to a target while maintaining dose sparing features. [12, 13, 1, 14, 15, 16] These advancements in treatment modalities have allowed radiation oncology teams to improve disease treatment and reduce damage to secondary healthy tissues, enhancing patient quality of life.

In particular, stereotactic radiosurgery (SRS) is a radiation therapy treatment modality frequently used to treat benign and malignant intracranial lesions in a single or few high dose fractions with highly precise dose placement, and its clinical use has been growing in recent years. [83, 53, 9, 84] Because of the high-dose nature of the treatment, high accuracy is necessary for proper dose conformality, and patient motion can significantly degrade target coverage. [58, 37]

Progress from early radiotherapy techniques to high-dose and high-conformality approaches have involved developments both in software and hardware. In order to accurately deliver dose to a target, both aspects of radiation therapy are necessary. The software side of radiotherapy includes treatment planning and the advancements in computation that have occurred over the past several decades. For IMRT, the development of solutions in the form of inverse planning have formed the backbone of many radiation therapy techniques. [1, 17] Modern treatment planning software continues to grow, and is highly dependent upon advancements in computation capabilities. Furthermore optimization techniques continue to advance and improve rapidly, giving clinical teams and commercial developers a wide array of options for solving treatment planning problems. As early as the 1960s, computer-based optimization has been used, though modern techniques for IMRT, VMAT and SRS use state-of-the-art approaches. [18, 19, 20, 21, 22, 23, 24] In the high-dose and high-conformality case of SRS, accurate treatment planning techniques are critical.

On the hardware side, both beam delivery and patient immobilization techniques have undergone substantial changes. Returning to SRS and IMRT, the theoretical inverse planning

methods of the 1980s only became clinically practical with the development of metallic compensators and then with multi-leaf collimators (MLC), shown in Figure 1.2. [25] The latter technology is a mechanical beam intensity modulator; binary-type are used with tomotherapy techniques to modulate the beam and generate fluence patterns, while LINAC-based MLCs are positioned within the treatment head to modulate dose intensity. MLCs more generally are motor-driven collimators used to shape the radiation field after beam generation and before interacting with the target. [26, 1] Eventually, dynamic use of MLCs were incorporated into intensity-modulated treatments, creating the first MLC-based treatments, which are still in clinical use today. [13] When treating cranial lesions with a LINAC-based SRS approach, MLCs help shape the beams in order to achieve the necessary conformality. Due to the strict accuracy requirements for SRS, precisely-machined dose shaping hardware is essential for adherence to the treatment plan; these are particularly important when small lesions, or multiple targets, are being treated.

Patient motion management has also undergone substantial change throughout the history of radiation therapy. This topic will be discussed in greater detail in the following sections, but the overall goal of motion management and immobilization is to limit patient motion during treatment in order to ensure radiation dose is delivered to the intended regions. Respiratory motion has been the most widely studied because it is difficult to determine and control, its impact on several sites, and the prevalence of lung cancers. [27] However, the motion of targets and organs at risk (OAR) of all treatment sites have the potential to degrade the treatment efficacy. These effects become increasingly important when high-conformality approaches such as SRS are used for the treatment of lesions.

In summary, recent developments in computing, treatment planning, patient motion management, and dose delivery hardware have prompted the rapid improvement in tumor coverage and dose conformality. By improving target dose delivery during therapy and avoiding unnecessary exposure to normal tissue and nearby organs at risk (OARs), clinical outcomes have continued to improve. [11, 28, 23] In this dissertation, we focus on the



Figure 1.2: A multi-leaf collimator (MLC) is used in IMRT and 3DCRT to shape the radiation beam in the treatment head. MLCs consist of several leaves, typically made of tungsten for its high absorptive quality for radiation in the treatment energy range. The leaves can move dynamically during IMRT and VMAT, producing intensity-modulation rather than fluence-modulation. (Photo courtesy of Varian Medical Systems of Palo Alto, California. Copyright [1999-2005], Varian Medical Systems[1])

combination of accurate patient motion monitoring and immobilization devices that have permitted the development of high dose, few fraction treatment modalities such as SRS which improve clinical workflow and have been shown to either match or improve upon traditional multi-fraction treatment regimens. [29, 30] By maintaining confidence in dose coverage accuracy, it is possible to further incorporate recent advances in medical physics and related fields, such as control engineering and robotics, into the clinic to continue to improve upon the standard of care.

1.3 Motivation

1.3.1 Patient motion management

One of the most important objectives of dose delivery in radiation therapy is the consistent and accurate dose deposition to a delineated volume contouring the lesion. This volume is established by the radiation oncologist after a CT scan is performed, and both the prescription and treatment regimen are constructed around this volume delineation and the radiobiology of the particular site. A clinical target volume (CTV) is capable of undergoing both translational and rotational motions, which may degrade dose coverage of the target. [31, 32, 28, 33, 34, 35, 36, 37]. In addition to proper treatment planning and precision in beam generation and delivery, patient motion must also be effectively managed to ensure good clinical outcomes. Patient motion which is unaccounted for can negatively affect the CT scanning stage, delineation during treatment planning, and ultimately the radiation delivery. Although this dissertation is primarily focused on minimizing the effects of motion in the third aspect, it is important to note that patient motion affects nearly every stage of radiotherapy treatments. In terms of accounting for these effects during treatment, both patient setup and intrafraction (during treatment) motion can degrade treatment efficacy.

Patient motion management can be classified into two categories: immobilization techniques which restrict movement of the target and/or surrounding structures, and motion

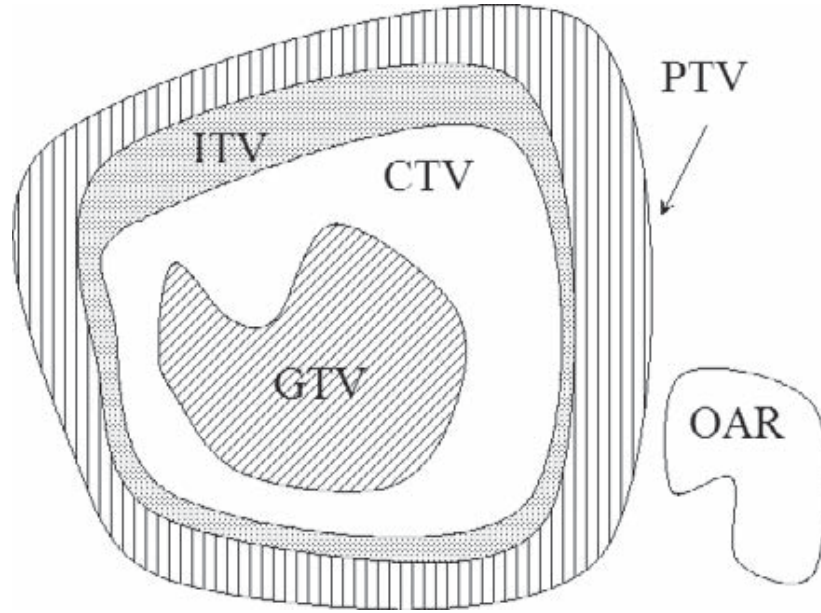


Figure 1.3: Target definitions from ICRU Reports 50 and 62, representing target structures used during treatment planning. [2]

monitoring, which tracks patient motion to determine if measured movements exceed preset thresholds and responds accordingly with radiation beam control and/or therapist intervention. In many treatments, both facets are used, though the extent to which they are implemented clinically varies substantially with the treatment site and radiotherapy protocol. Research into respiratory motion management has been dominant, though cardiac, skeletal muscular, gastrointestinal, and head-and-neck motions can also degrade various radiotherapy treatments for particular sites.

Intrafraction motion can also cause problems during the CT imaging scan stage, producing artifacts and hindering treatment planning. [38, 39] Solutions involving treatment planning in 4D (volumetric and temporal) have been used to reduce motion effects at the CT stage; 4DCT scans for treatment planning can also lead to unique solutions with 4D radiation delivery. [40, 41] In the case of SRS, minimal head immobilization during a CT scan is sufficient to prevent severe impediments to the development and delivery of an effective radiotherapy treatment.

When motion occurs during the imaging stage, proper delineation of the gross tumor

volume (GTV) can be hindered during treatment planning. The GTV defines the visible extent of malignant growth. Additional volumes may be defined for the purposes of target contouring, including the CTV, which expands the GTV for microscopic disease, the internal target volume (ITV), which further expands the visible target to account for organ motion, and finally the planning target volume (PTV), which expands the margin for setup uncertainties and other physical limitations (Figure 1.3). [2] When delineating volumes for SRS treatments, the target or targets are well-defined, and small margins for sub-clinical disease may or may not be included. As a result of the high-accuracy requirements of SRS, the treatment goal is to deliver dose to the GTV with as small of a margin as possible, thereby enabling radiotherapy with high dose. Consequently, margin expansions such as the ITV do not play a role in SRS treatments.

Returning to motion management using immobilization during dose delivery, many approaches have been used to minimize or account for the effects of patient movement throughout the history of radiotherapy. In early radiotherapy, simple immobilization instruments were used, including head cups and neck rolls, or tape to hold the patient in place. Because CT-based treatment planning and beam shaping were limited, the minimal level of immobilization was not necessarily the greatest factor for target coverage. As CT resolution, treatment planning optimization, and delivery machines became more accurate, solutions for patient motion also evolved. Plaster molds, bite blocks, skin marks, vacuum molds, and other techniques began entering clinics in the 1960s and 1970s; laser alignment, thermo-plastic masks, and polyurethane foam casts emerged in the 1980s. Many of these solutions continue to be used clinically for various treatment sites and radiotherapy protocols.

The immobilization techniques listed above are often coupled with motion monitoring, which tracks the motion of the target, either directly or through the use of surrogates. In the case of respiratory motion for example, beam gating, which determines the timing for beam ON and beam OFF based on motion monitoring, can use implanted fiducials near or in the target to track tumor motion, or an external marker placed on the surface. In the latter case,

tracking an external surrogate is assumed to be correlated to internal motion, though this assumption does not always hold. [42] When these recorded motions exceed thresholds or are outside preset windows, the radiation beam is halted. Treatment resumes after the target returns to the desired position, or with therapist intervention. In image-guided radiation therapy (IGRT), patient motion tracking techniques include those which monitor in 2DOF, such as with kilovoltage (kV) fluoroscopy, 3DOF or in full 6DOF, with the goal of preventing improper dose coverage to the patient. [43, 44]

Although the deleterious effects of poor translational localization of the CTV are well known, recent studies have shown that tumor rotational localization error can also significantly affect the target dose coverage. [35, 36] This is particularly concerning in the case of highly conformal radiation therapy such as SRS and IMRT. In these cases, even tumors which are well-positioned in the X (leftright, LR), Y (superiorinferior, SI), and Z (anteriorposterior, AP) axes can still produce significant sources of dosimetric uncertainty, with respect to efficacy of intended dose distribution as well as localization uncertainty at off-axis distances from isocenter. These potential complications are especially pronounced when targets are significantly nonspherical or in the single-isocenter multi-target regime, wherein multiple sites are undergoing treatment with a single, shared isocenter. [32, 45]

The ability to track and correct for target motion in 6 degrees-of-freedom has become more achievable with the introduction of new motion monitoring techniques and patient positioning devices. External 6DOF patient tracking has been performed, commonly with either infrared (IR) marker tracking or 3D optical surface imaging. [46, 47, 48] Internal 6DOF tracking of targets within the patient has been implemented using stereoscopic kV imaging, with single planar kV images taken to estimate 6DOF motion in real-time or magnetic fiducial tracking. [45, 49, 9, 50] Devices capable of positioning patients in 6DOF include robotic patient couches such as the Elekta Hexa-POD system (Elekta, Stockholm, Sweden) and the Varian TrueBeam 6D couch (Varian, Inc., Palo Alto, CA). [51] Such robotic systems are becoming more popular clinically, and while these devices are able to position patients

at the start of each treatment fraction to achieve improved 6DOF conformality, they are currently unable to perform real-time intrafraction correction as patients move during the course of treatment. Previous work which characterized such 6DOF technology had been limited to performing well-defined reference motions along only the translation axes XYZ or along XYZ and the yaw axes. [52]

In the high-dose high-conformality regime of SRS, full 6DOF tumor localization and patient motion control is even more critical to minimize normal tissue dose and guarantee target coverage. [53, 52, 54] When patients have multiple metastatic lesions within the brain, head rotations can have an even more profound impact because the treatment isocenter is a distance from the center of each target. [55] This effect is more pronounced the further these sites are from the treatment isocenter. Furthermore, there is growing evidence to support treating more than 3, even upwards of 15, sites at once in the brain, and rotational errors will be more pronounced in such cases when the lesion distances increase from isocenter. The common approach to ensuring proper dose coverage at the required clinical levels is to physically lock the patient’s skull into a rigid metal frame for treatment planning and treatment, to ensure that they are in the proper position (Figure 1.4, left). Both linear accelerator-based (LINAC) and GammaKnife (Elekta AB, Stockholm, Sweden) use frame-based SRS to treat brain disorders with high conformality. The high accuracy of this frame-based SRS is offset by potential drawbacks of patient discomfort and clinical workflow restrictions, as the patient must remain on-site with the frame attached until treatment is complete.

One solution is frameless SRS, in which the head ring is removed, and patient motion is monitored during treatment. [56] Improvements in image guidance during treatment have allowed clinics to treat brain disorders in a less invasive approach compared to frame-based SRS. A thermoplastic mask, which is heated to become pliant and molded to the patient’s head before cooled, is used as the primary immobilization tool instead of the stereotactic frame (Figure 1.4, right). Although this approach improves patient comfort, it has the potential to reduce treatment accuracy due to less-rigid immobilization techniques utilized

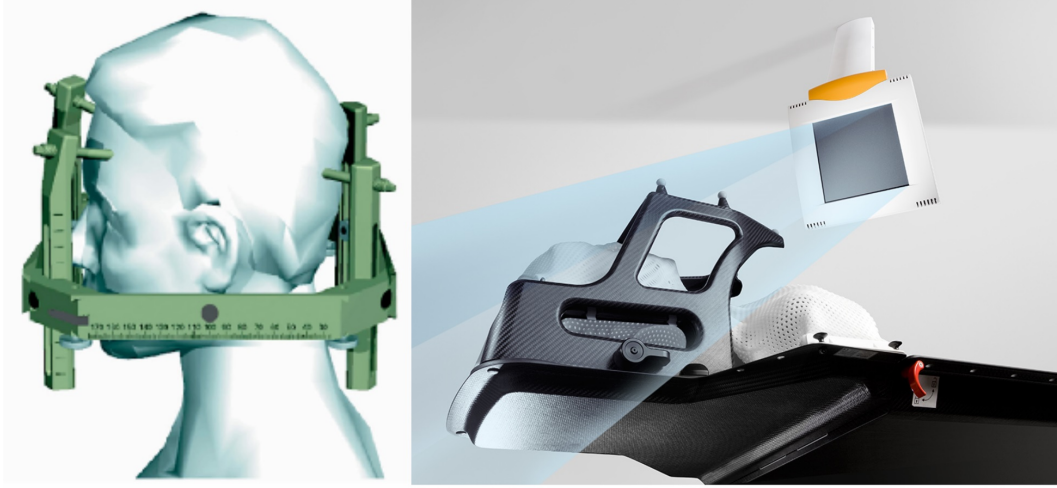


Figure 1.4: Immobilization methods for stereotactic radiosurgery (SRS). Left: frame-based SRS, in which a rigid metal frame is fixed to the patient’s skull. Right: frameless SRS, in which a thermoplastic mask is used for immobilization and patient motion is monitored during treatment. (Photos courtesy of Brainlab AG, Munich, Germany; GammaKnife, Elekta AB, Stockholm, Sweden)

in this SRS regime. [57, 58, 59, 60] Both frameless and frame-based SRS approaches have significant limitations for effective treatment of brain disorders, and therefore there is room for innovative solutions to incorporate the benefits of both techniques while removing their disadvantages.

1.3.2 *Robotics for cranial SRS motion management*

The goal of this work is to make use of advances in robotics and 6DOF patient motion tracking systems to improve frameless SRS by performing real-time cranial motion compensation during treatment. By monitoring and responding to patient head motion while they are undergoing radiotherapy, the robotic system can ensure the target or targets remain in the proper 6DOF position and orientation. We build on previous work which focused on translation-only motion compensation for brain radiation therapy. [54] The use of 6DOF robotics during treatment has the potential to improve upon the accuracy limitations of current frameless SRS techniques while reducing the workflow inefficiencies and complications from patient discomfort in the frame-based treatment. In the case of treating multiple lesions,

treatment options include developing independent plans for each target and treating independently, or simultaneous treatment with a single isocenter. In the former case, each plan's isocenter is centered on the specific target, and rotational effects on dose coverage are typically minimal, though overall treatment time can be long. [61] In the latter single-isocenter case, treatment times are reduced, which may reduce deleterious intrafraction motion effects; however, this approach is not robust against rotational motion because one or more targets are displaced from the point of rotation. For such single-isocenter multiple-target treatments, the use of 6DOF motion compensation during radiotherapy may dramatically improve dose conformality and further enable single-isocenter plans to become standard of care for multi-target treatments. [62] In this way, our solution can help reduce treatment time, and its proposed flexibility may open the door for previously unworkable therapeutic options, including cases when frames cannot be used due to special patient conditions.

To that end, this dissertation presents research into the development of a frameless and maskless SRS solution using 6DOF robotics for real-time head motion stabilization. We construct a theoretical control foundation, then the design, development, and evaluation of a novel 6DOF robotic system, before applying the device to volunteers as a step toward clinical implementation. By incorporating highly precise robotics into SRS treatments, we aim to achieve or exceed the dose delivery accuracy of frame-based SRS while improving clinical workflow, minimizing patient discomfort, and reducing therapist intervention.

CHAPTER 2

6DOF PARALLEL KINEMATICS CONTROL THEORY

2.1 Introduction

Performing real-time head motion stabilization in full 6DOF (translations and rotations) requires the integration of several key systems, including those which are already implemented in the clinic, and others which must be developed in-house. Among the latter such systems is the robot which will ultimately be performing the head motion compensation. In the first step toward 6DOF compensation, a robotic system capable of performing translational and rotational movements accurately must be designed, before simulation, modeling, construction, evaluation, and ultimately application can be considered. There are several archetypes of robotic systems capable of performing 6DOF motion trajectories: robotic arms, stacked serial systems, and parallel kinematic systems are among the most commonly used, not only in medical physics and radiation therapy applications, but also manufacturing, surgical guidance, and motion simulators, among others. Due to the nature of the final application, only the latter two design archetypes were considered as candidates for motion compensation of cranial motion. Although robotic arms often have large workspaces and high-accuracy 6DOF control, a workspace of only several millimeters and degrees was expected, based upon earlier research into cranial motion during radiotherapy. [59] Additionally, the expected placement of the robot under the patient's neck, or nearby to this site, made compactness an important feature of the robotic system.

The first option, a stacked serial 6DOF system, traditionally features a combination of 3 independent translational axes of motion and three independent rotational axes. The six degrees of translational and rotational freedom are each operated separately from the other five, but controlling the resultant 6DOF motion requires precise knowledge of the interaction between each axis in order to decouple the effects of the underlying degrees of freedom which can impact the subsequent axes. In this way, while each axis is controlled independently, the

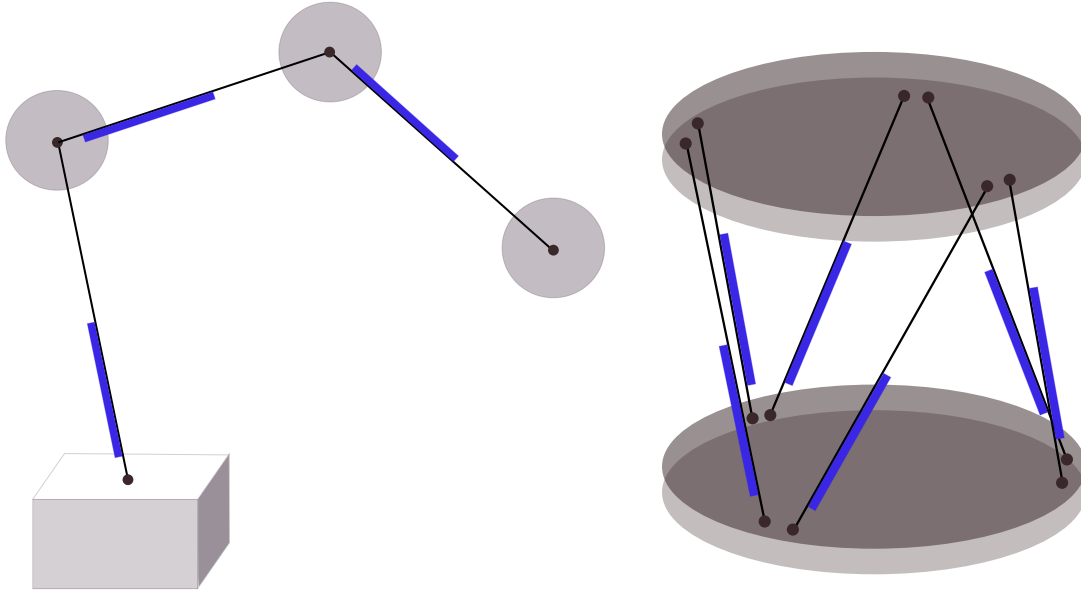


Figure 2.1: 6DOF stacked design (left) versus a parallel kinematics system (right). The 6DOF Stewart-Gough design features improvements in compactness, weight load ratio, and rigidity, as well as reduced complexity. Stacked serial designs, such as a model robotic arm shown here (left), have each control axis dependent upon the previous one in the stack configuration.

effect of on axis on the overall 6DOF motion is dependent upon all axes which exist beneath it. Robotic arms, such as the model shown in Figure 2.1, are also stacked serial systems, with each control axis dependent upon earlier controllers. Although many robotic systems feature stacked serial architecture, including those used in radiation oncology clinics, it is often more difficult to control them when their 6DOF trajectories are complex or when the motion is iterative and adaptive.

The second option was a parallel kinematics robotic system, in which the 6DOF motion of a mobile component is performed relative to a stationary component, which are connected to one another by a set of linear actuators. The most common design type of the parallel kinematics archetype is the Stewart-Gough platform, in which six linear actuators, often called struts or legs, separate the mobile top platform from the fixed bottom platform [63, 64]. In this design, the movements of the six linear actuators determine the translational and rotational movements of the top platform relative to the bottom platform. The general

designs of both the stacked serial system and parallel kinematics system are presented in Figure 2.1. Ease-of-use, compactness while maintaining a sizable range of motion, and a high load-to-weight ratio were among the most important reasons why the parallel kinematics architecture was chosen over the stacked serial design. One of the most well-studied and widespread parallel kinematics designs is the Stewart-Gough platform. The Stewart-Gough platform, consisting of six extensible legs in a precise geometrical configuration, offers several important characteristics for the cranial motion compensation application [63, 64]. In addition to the advantages of parallel kinematics robots in general, the Stewart-Gough architecture has minimized inertial mass, longevity, high stiffness, and a faster response time compared to other robotic systems.

The specific design considerations will be discussed in Chapter 3, but the first steps toward developing a control algorithm were first established and are presented in this chapter. Although the robot has been used in many applications, and the specific control algorithms have been modified to suit the situation, the fundamental control theory formed the foundation upon which all future algorithms were built.

2.2 Inverse kinematics control formulation

The two fundamental approaches to controlling the motion of the parallel kinematics platform are inverse kinematics and forward kinematics. Given a particular position and orientation, inverse kinematics attempts to solve a separate system of equations to find the strut (actuator) lengths needed for that position. Alternatively, given the strut lengths, forward kinematics attempts to solve for the position and orientation of the top platform. For our control purposes, the position of the six linear actuators is the control input, and the resultant 6DOF position of the top platform is the output. For motion compensation, the desired top platform position is known, and the linear actuator lengths required to achieve that position are the computational objective. Inverse kinematics is therefore the appropriate choice, and it is conveniently more straightforward computationally, compared

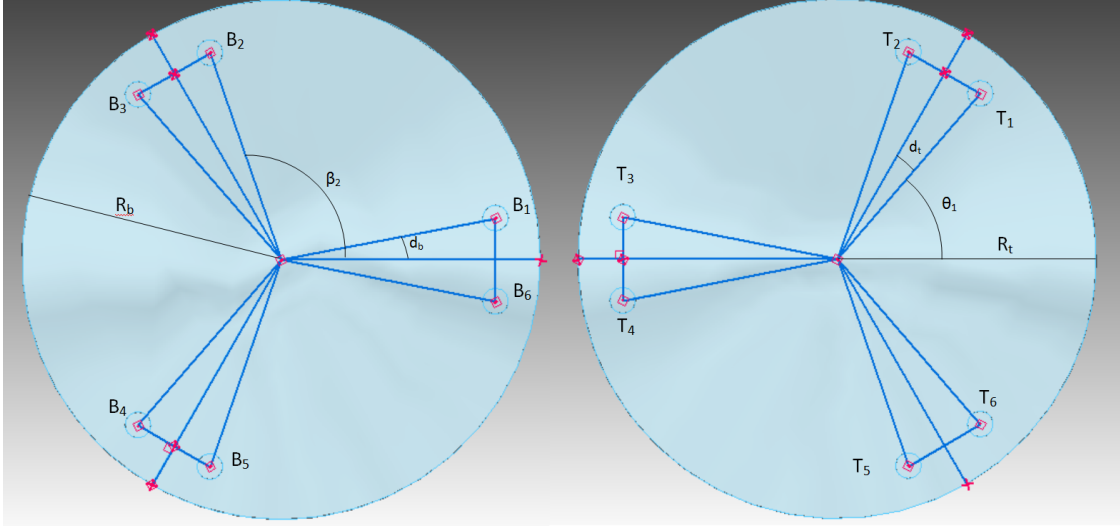


Figure 2.2: Geometry of the parallel kinematics system. The bottom platform geometry (left) corresponds to the top platform geometry (right) through the six linear actuators, or struts. Each B_i joint location on the base is connected to the corresponding T_i joint location on the top platform via a linear actuator.

to forward kinematics. With the configuration of the top and bottom platforms shown in Figure 2.2, the inverse kinematics control approach considers the XYZ translational and $\alpha\beta\gamma$ rotational position of the top platform relative to the bottom and calculates the linear actuator positions. $B_1, B_2, \dots, B_6$ and $T_1, T_2, \dots, T_6$ are the joint positions of the bottom and top platforms, respectively, which connect the platforms to the associated linear actuators. The vector representation of these points is given by equation 2.1

$$\vec{b}_k = R_b \begin{bmatrix} \cos(\beta_k) \\ \sin(\beta_k) \\ 0 \end{bmatrix} \quad \vec{t}_k = R_t \begin{bmatrix} \cos(\theta_k) \\ \sin(\theta_k) \\ 0 \end{bmatrix} \quad \forall k = 1 \dots 6 \quad (2.1)$$

Here, β_k and θ_k are the rotational component of the polar representation of the k^{th} joint contact point for the bottom and top platforms, respectively (Figure 2.2). The scalars R_t and R_b , which are the radii of the top and bottom platforms, respectively, are the radial components of the polar representation of these joint positions. The β_k and θ_k angles are given explicitly below in equation 2.2. The determination of appropriate values for the radii

and angular separation are dependent upon the robot's design considerations, discussed in the following chapter.

$$\begin{aligned}
\beta_1 &= d_b & \theta_1 &= 60 - d_t \\
\beta_2 &= 120 - d_b & \theta_2 &= 60 + d_t \\
\beta_3 &= 120 + d_b & \theta_3 &= 180 - d_t \\
\beta_4 &= 240 - d_b & \theta_4 &= 180 + d_t \\
\beta_5 &= 240 + d_b & \theta_5 &= 300 - d_t \\
\beta_6 &= -d_b & \theta_6 &= 300 + d_t
\end{aligned} \tag{2.2}$$

With this geometrical configuration established, if the robot is given a desired translational position of the top platform relative to the bottom, $\vec{r}$ and a rotational orientation of the top relative to the bottom platform $\vec{R}$, the k^{th} linear actuator length, s_k , required to achieve this 6DOF position is given by equation 2.3

$$s_k = |\vec{r} + \vec{R} \cdot \vec{t}_k - \vec{b}_k| \tag{2.3}$$

Using this inverse kinematics formulation, with a properly constructed $\vec{R}$ orientation matrix, one can begin simulating the control input to each linear actuator to achieve a desired 6DOF position of the device's top platform relative to the bottom platform.

2.3 Forward kinematics formalism

Forward kinematics is another control approach for 6DOF parallel robots, which determines a top platform position and orientation based upon the lengths of the six linear actuator legs. The forward kinematics approach is more difficult to solve than the inverse kinematics formulation because the forward problem is under-defined and has a high-degree non-linear formulation, compared to the relatively straightforward and geometric inverse problem. [65]

Consequently, analytic solutions to the forward problem are not tenable and multiple real and complex solutions to the problem may exist; as a result, optimization techniques are often employed to aid in solving the forward kinematics problem. Although the inverse kinematics formalism formed the backbone of the robotic control theory, forward kinematics can be used to validate that the robot performed the intended motion correctly. If this validation is sufficiently fast, when the 6DOF robot is used for real-time motion compensation, forward kinematics can provide instantaneous feedback on performance. In this section, we discuss the forward kinematics formalism, and a rapid calculation for use in 6DOF head motion compensation.

When controlling the parallel kinematics robot, defining the rotation matrix $\vec{R}$ from Equation 2.3 requires consistency and care; the order in which each axis of rotation (pitch α , roll β , and yaw γ) is included in the overall 3×3 rotation matrix is important. For this project, a 1-2-3 order was selected, in which pitch, then roll, then yaw are included in the matrix $\vec{R}$. Many approaches choose 3-2-1, but so long as one is consistent in the control, the choice may be arbitrary. Explicitly, the 1-2-3 rotation matrix is given by Equation 2.4 below

$$\vec{R} = \begin{bmatrix} \cos \beta \cos \gamma & \cos \alpha \sin \gamma + \sin \alpha \sin \beta \cos \gamma & \sin \alpha \sin \gamma - \cos \alpha \sin \beta \cos \gamma \\ -\cos \beta \sin \gamma & \cos \alpha \cos \gamma - \sin \alpha \sin \beta \sin \gamma & \sin \alpha \cos \gamma + \cos \alpha \sin \beta \sin \gamma \\ \sin \beta & -\sin \alpha \cos \beta & \cos \alpha \cos \beta \end{bmatrix} \quad (2.4)$$

The forward kinematics theory developed here for parallel kinematics robots uses this convention, and takes as inputs the linear actuator leg lengths, the spacing of the joints connecting the actuators to the platforms, the initial spacing condition between the top and bottom platforms, and the initial orientation of the top platform relative to the bottom.

The 6DOF position of the top platform can be described the position $\vec{r}$ and the Euler angles α , β , and γ , defined in the rotation matrix $\vec{R}$. The positions of the top platform

joints in the coordinate frame of the top platform are once again denoted $\vec{t}_k$, and the joint positions of the bottom platform in the bottom platform coordinate frame are $\vec{b}_k$, $\forall i = 1 \dots 6$. One can define $\vec{R} \cdot \vec{t}_k = \vec{h}_k$.

Equation 2.3 can be used to determine the six leg lengths s_k for $i = 1 \dots 6$. One can expand Equation 2.3 as

$$(\vec{r} + \vec{h}_k - \vec{b}_k)^\top (\vec{r} + \vec{h}_k - \vec{b}_k) = s_k^2 \quad (2.5)$$

The time derivative of the leg lengths can be represented as $\dot{s}_k$, shown in Equation 2.6

$$(\vec{r} + \vec{h}_k - \vec{b}_k)^\top (\dot{\vec{r}} + \dot{\vec{h}}_k) = s_k \dot{s}_k \quad (2.6)$$

The angular velocity of the top platform is ω , defined by Equation

$$\omega = \begin{bmatrix} \cos \beta \cos \gamma & \sin \gamma & 0 \\ -\cos \beta \sin \gamma & \cos \gamma & 0 \\ \sin \beta & 0 & 1 \end{bmatrix} \begin{bmatrix} \dot{\alpha} \\ \dot{\beta} \\ \dot{\gamma} \end{bmatrix} = P(\Psi) \dot{\Psi} \quad (2.7)$$

If one considers that $\dot{\vec{h}}_k = \omega \times \vec{h}_k$, as well as the property $a^\top (b \times c) = (c \times a)^\top b$, one can obtain

$$(\vec{r} + \vec{h}_k - \vec{b}_k)^\top \dot{\vec{r}} + (\vec{h}_k \times (\vec{r} - \vec{b}_k))^\top \omega = s_k \dot{s}_k \quad (2.8)$$

Consequently, Equation 2.9 establishes a system of equations for forward kinematics

$$ABv = C\dot{s} \quad (2.9)$$

where

$$v = \dot{x}, \quad x = \begin{bmatrix} \vec{r} \\ \Psi \end{bmatrix}, \quad A = \begin{bmatrix} \vec{h}_1 + (\vec{r} - \vec{b}_k) & \cdots & \vec{h}_1 + (\vec{r} - \vec{b}_k) \\ \vec{h}_1 + (\vec{r} - \vec{b}_k) & \cdots & \vec{h}_1 + (\vec{r} - \vec{b}_k) \end{bmatrix}^\top, \quad B = \begin{bmatrix} I \\ P(\Psi) \end{bmatrix}$$

$$s = [s_1 \cdots s_6]^\top, \quad C = \text{diag}(s).$$

Here, the Euler angles α , β , and γ are the elements of the vector Ψ , as defined in Equation 2.7; Equation 2.7 also defines $P(\Psi)$. With these relationships established, recall the objective of forward kinematics is to calculate the 6DOF position of the top platform relative to the bottom, x from a set of leg lengths ℓ . We define the function $f(x)$, which takes the 6DOF position x as an input and computes a corresponding set of leg lengths s_k using forward kinematics. We also define $J(x) = (AB)^{-1}C$. The forward kinematics problem then involves finding x such that $f(x) = \ell$.

Define the objective function $F(x) = \frac{1}{2}(f(x) - \ell)^\top(f(x) - \ell)$. Then the iteration of the gradient descent is given by

$$x^{(n+1)} = x^{(n)} - \mu^{(n)} \nabla F(x^{(n)}), \quad (2.10)$$

where $\nabla F(x^{(n)}) = J(x^{(n)})(f(x^{(n)}) - \ell)$. A suitable $\mu^{(n)}$ must be found such that $F(x^{(n+1)}) \leq F(x^{(n)})$. This can be done by line search.

When the error is sufficiently small, the forward kinematics algorithm records the top platform position and orientation based on the linear actuator lengths, and continues to the next set of linear actuator positions. This process is repeated for the entire recorded trajectory of the 6 linear actuators, thereby constructing the robot's top platform 6DOF trajectory using the leg lengths.

CHAPTER 3

DEVELOPMENT OF A 6DOF ROBOTIC MOTION PLATFORM

3.1 Introduction

This chapter focuses on the design and development of the 6DOF robot based on the parallel kinematics approach. Some of the advantages of this design were discussed in Chapter 2; compactness, longevity, and a high load-to-weight ratio were of particular importance for the purposes of this project.

Although several commercial parallel kinematics systems are available, none currently exist with specifications meeting our requirements for cranial motion compensation. Specifically, the combination of compactness, weight load, control accuracy, and the ability to input real-time adaptive trajectories were unmet by current commercial systems. As the ultimate application of this system is to be positioned under the patient's neck during intracranial treatment, the device should be able to hold the weight of a patient's head, approximately 4.5 kilograms, and fit under the treatment couch with enough room to prevent collisions with the rotating gantry: less than 30 cm in height and 15 cm in width. Additionally, accessing real-time controllers for commercial systems, including 6DOF treatment couches, would be difficult or impossible to implement. Finally, it is possible that the large distance between the control and target coordinate frames, such as in the configuration of 6DOF couches, can amplify small control errors to produce large motions at the treatment site; having a robotic system as near as possible to the treatment isocenter would minimize this leverage effect. Consequently, we were required to design and build a parallel kinematics platform in-house based on weight load, compactness, 6DOF workspace, and velocity considerations.

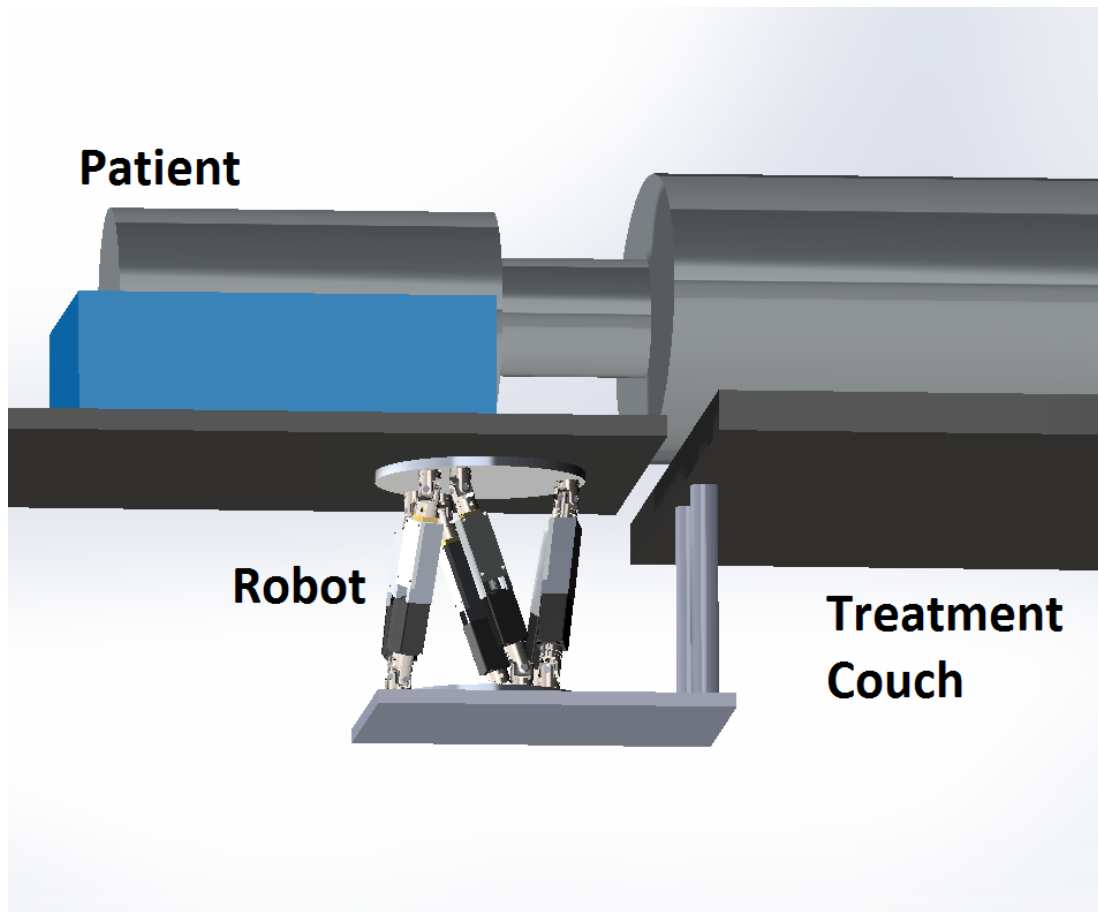


Figure 3.1: A model representing the proposed position of the robotic stage under the patient's neck during radiation therapy treatment. Note that the robot does not interact with the primary radiation beam or imaging beams with this setup. The importance of device compactness is also presented here, as the treatment head motion could yield collisions if the device were too large.

3.2 Design of the 6DOF parallel kinematics platform

The initial design consideration focused on the device size and assembly configuration. While various configurations are often used, most Stewart-Gough platforms feature circular top and bottom platforms with six linear actuators arranged at alternating obliquity angles which connect to the top and bottom platforms. Connecting the linear actuators to the immobile bottom platform and the mobile top platform are universal joints, which are couplings capable of transmitting rotary power of a shaft over a large array of angles. The contact points need to be well-known for motion control of the robotic stage, as well as close to the adjacent linear actuator contact for proper 6DOF control. The locations of these contact points for the top and bottom platforms are presented in Figure 2.2, given by vectors $B_i, T_i \quad \forall i = 1 \dots 6$. The choice of the angular spacing parameters d_t and d_b , for the top and bottom platform, respectively, was selected based on the size of the universal joints and width of linear actuators, which will be discussed in more detail in Section 3.3. The radii of the platforms, R_t and R_b were selected based upon estimated device size restrictions for downstream clinical implementation (Figure 3.1), and several radii were modeled and the resultant 6DOF workspaces were computed. The combination of compactness and 6DOF workspace limits motivated the choice of 75 mm as the radius for both platforms, measured from the center of the platform to the center of the universal joint contact points (see Figure 2.2).

Consequently, design of the prototype was motivated by the required 6DOF specifications to move a patient’s head accurately in response to movement, both in terms of workspace and accuracy, which is aimed to be at the sub-millimeter and sub-degree level. Sufficient torque and weight load capacities, as well as compactness and cost minimization, were among the primary considerations we addressed in the design of the robotic stage, specifically with regards to the linear actuator design. Given the lack of commercial linear actuators with the required specifications, several linear actuator designs were modeled in computer-aided design (CAD) software SolidWorks [2013, SolidWorks Corp.]. An open-face model of the

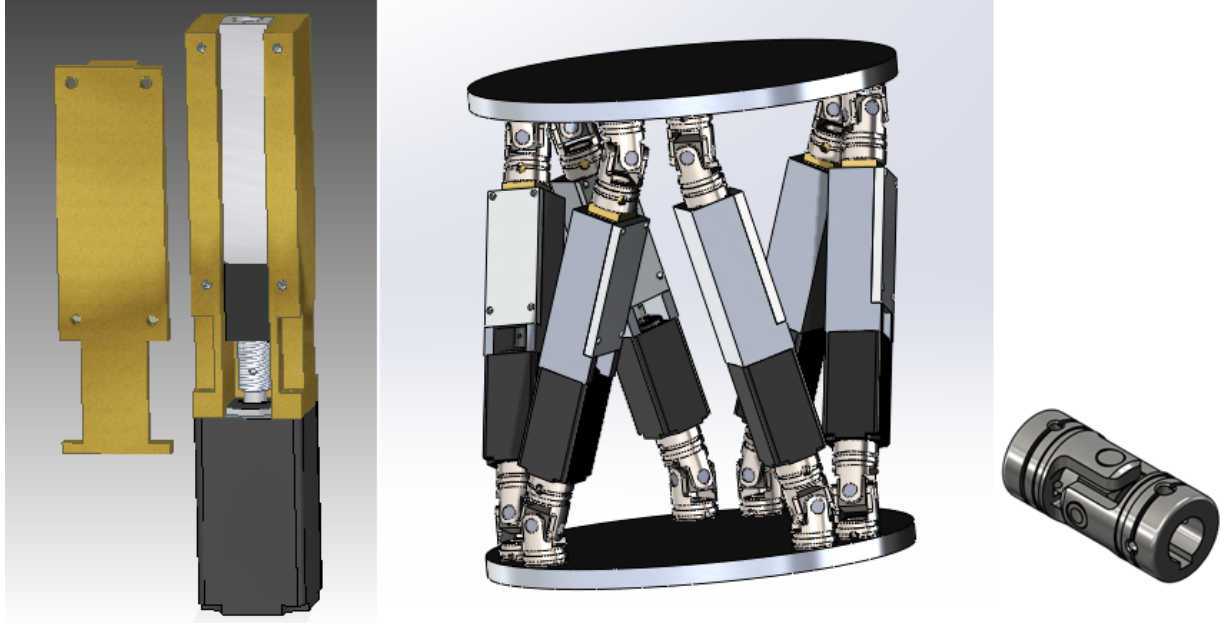


Figure 3.2: Prototype design of the linear actuator which would be used to control the motion of the parallel kinematics device (left). Six linear actuators based on this general design would be used in the robotic stage, and control of each actuator's length governs the 6DOF motion of the parallel kinematics robot (middle). The re-designed linear actuators are connected to the top and bottom platforms using universal joints for free rotational motion, which is necessary to prevent restrictions on the top platform's 6DOF motion. A model of the universal joints used in the assembly of the 6DOF robot are also shown (right). Twelve universal joints were used in the fabrication of the parallel kinematics robot: one to connect the actuator to the mobile top platform, and one to connect the actuator to the stationary bottom platform, for each of the six actuators.

linear actuator design is shown in Figure 3.2 (left). In this design, which is similar to the final actuator used in the device, a stepper motor is connected to a lead screw, which pushes a shaft in the direction of desired motion; an outer casing (shown in brass) provides the necessary anti-rotation element, to effectively convert the rotations to linear motion. The screw shaft, or sliding rod, is hollow to allow the lead screw to move in and out of it as the shaft moves. Shown as a black piece connected to the shaft in Figure 3.2 (left), a delrin tapped piece is fixed to the hollow shaft and matched to the lead screw pitch, such that as the lead screw rotates (given the motor rotation) the delrin piece moves linearly. The linear motion produced is due to the anti-rotation casing, and because of the connection between the delrin piece and the shaft, the shaft moves linearly within the casing. A model assembling the actuators with the top and bottom platforms is also presented in Figure 3.2 (middle).

Changes were then made to the actuator’s design to reduce overall weight and assembly cost, through simplification of the outer casing and changing materials. Additionally, models for the universal joints which connect the actuators to the top and bottom platforms were created based upon commercially available joints (Figure 3.2, right). These considerations added, the resultant finalized design for a prototype 6DOF parallel kinematics robot was established. This design is shown in Figure 3.2, modeled in SolidWorks.

3.3 Simulation of 6DOF robotic stage control

With the inverse kinematics formulation established in Section 2.2, motion simulations of the parallel kinematics robot could then be performed. To accomplish this, we modeled several control trajectories by converting the 6DOF top platform positions to the necessary six linear actuator positions, over the time span of the selected trajectory. The ultimate goal of these simulations was to further motivate the fabrication specifications of the device, and to determine what capabilities of the robot would potentially be required when used for motion compensation in frameless maskless SRS cases.

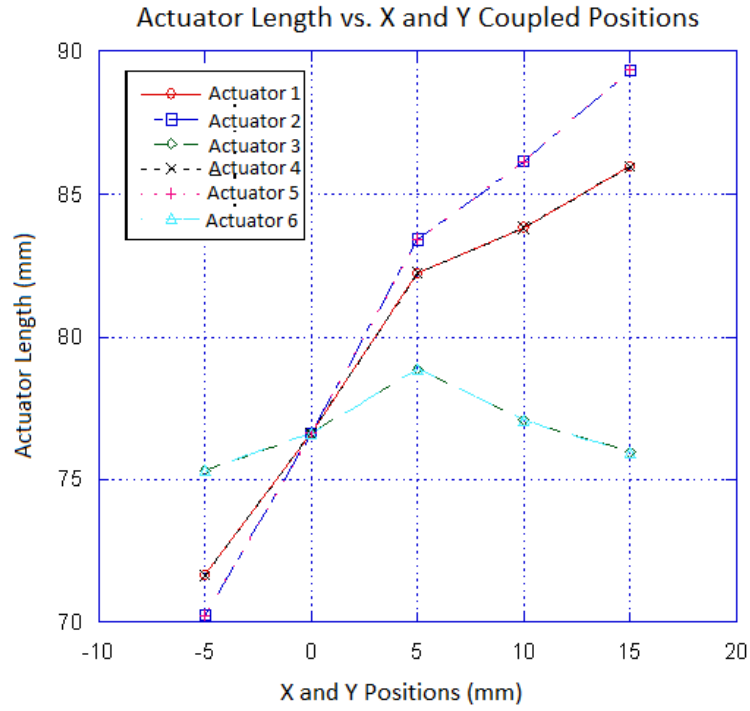


Figure 3.3: Simulated motion in simultaneous x- and y-direction are used to determine the required strut lengths that would produce this sample input motion. Note the transition at 5mm; here, the individual struts of the parallel kinematics device are shifting from pushing to pulling, as expected from the design configuration and coordinate axis definition. Coupling often occurs in simulated motion that involves fewer than 6 degrees of freedom.

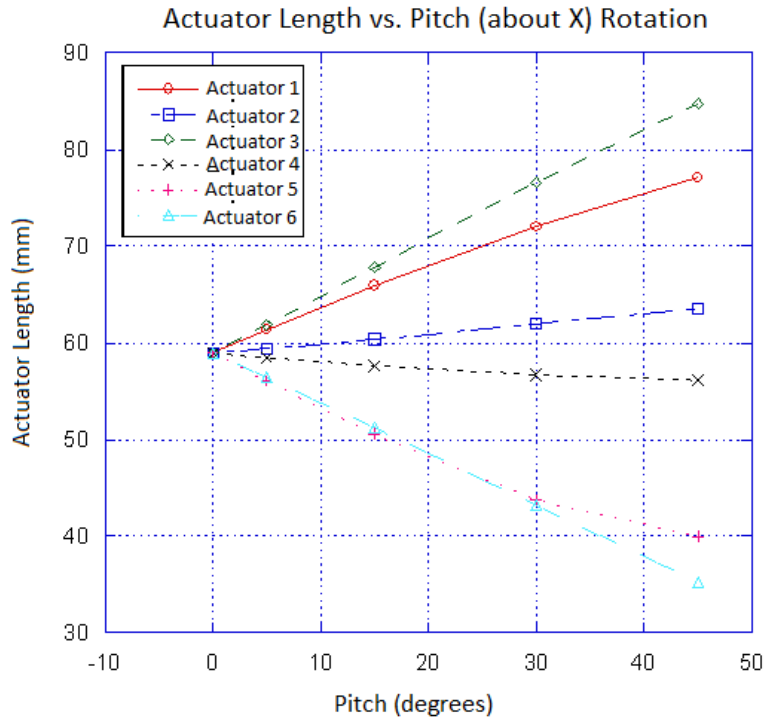


Figure 3.4: Simulated motion in the pitch (about X) degree of freedom is used to determine the required strut lengths that would produce this sample input motion. A simulated 45 degree rotation is performed.

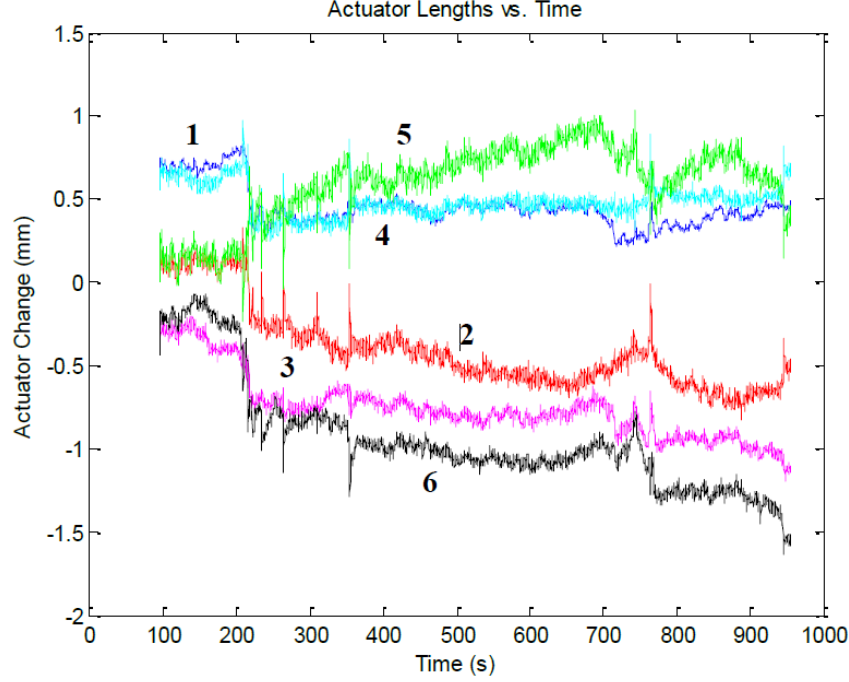


Figure 3.5: Simulated motion of the robotic motion stage required to compensate for real cranial motion. This 6DOF motion was collected on volunteers using an IR marker tracking system and converted to the six associated linear actuator trajectories required to compensate for this 6DOF motion.

Several trajectories were selected for simulation, in order to comprehensively compare the intended 6DOF top platform motion to the associated six linear actuator positions to achieve these translational and rotational positions. The actuator lengths required to obtain the desired positions were computed using MatLab [2012b, The MathWorks] code based on the inverse kinematics formulation. Using this approach, we first determined how the device would respond to inputs of simple linear motion in one or two dimensions. Figure 3.3 demonstrates the change in actuator length from an initial position, given the simulated input of a motion in the X (LR) and Y (SI) axes, defined as the two directions on the plane of the top platform (the Z, or AP, axis is that which is normal to the face of the top platform) as an example case. These axes are defined by the initial orientation of the top platform relative to the bottom. The simulated top platform moved to five positions in both X and Y, spanning from -5 mm to 15 mm, which was considered to be an appropriate test range based upon previous analysis of cranial motion [48]. As an additional example

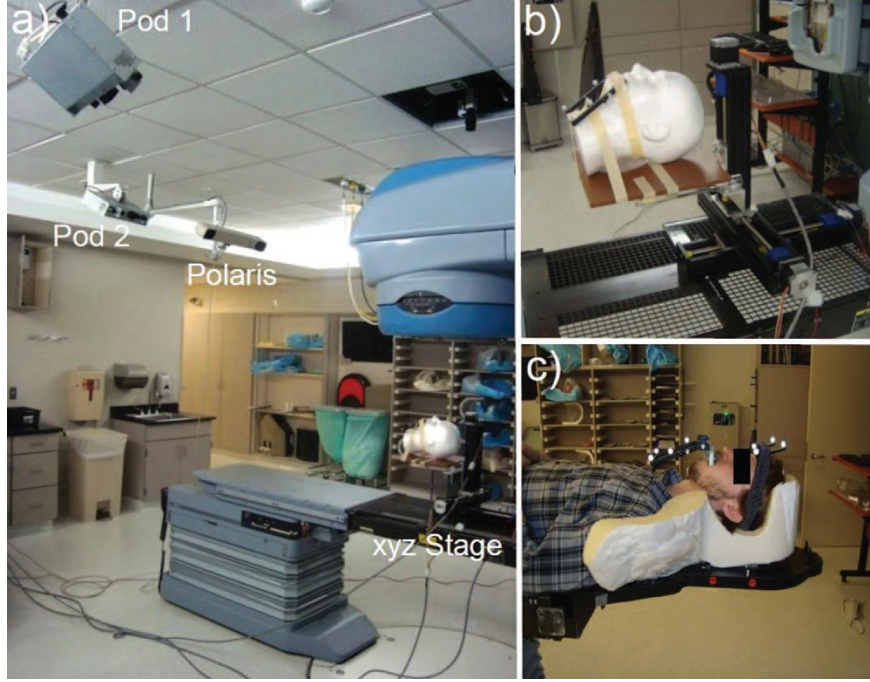


Figure 3.6: (a) LINAC vault external tracking camera configuration, with the infrared (Polaris) camera as well as the AlignRT optical surface mapping system cameras indicated. (b) Magnified view of a 3D XYZ motion stage capable of performing sub-millimeter translational movements of a head phantom. (c) Volunteer setup using a custom head cradle and IR dental and forehead markers.

of controlled motion, we simulated a 45 degree pitch (about X) rotation (Figure 3.4). From these simulations of basic top platform trajectories, it was clear that the corresponding linear actuator trajectories often followed nonlinear movements to reproduce the desired linear top platform motions. In addition to informing the design choices of the robot, these results later inspired deeper investigation into algorithms for optimal robot trajectories in the head motion compensation control application.

As the end goal of this work is to perform real-time cranial motion compensation, we also sought to simulate head motion using the inverse kinematics formulation for the parallel kinematics robot. Using a stereoscopic infrared (IR) marker tracking system (Polaris, Northern Digital, Inc., Waterloo, ON, Canada) with a 12 Hz sampling frequency, we collected cranial motion on volunteers over 15 minute tracking periods. This external tracking system functions by using the two IR cameras to track a set of spherical reflective markers (we used

four in our study) to define a coordinate system, both in position and orientation. These four reflective markers were attached to a plate which was fixed to a volunteer-specific bite block, similar to a mouth guard used in athletics, in order to rigidly connect the reflective markers to the skull and brain of the volunteer. This bite block method is often used clinically as a surrogate for internal target motions. These reflective markers define a coordinate system, established using a text-based tool file, which is created prior to motion monitoring to define the coordinate system as a point in the brain. This point is often referred to as the virtual point. With this setup established, shown in Figure 3.6, as the volunteer’s head moves in space, so do the reflective markers, and such motion is subsequently tracked and recorded by the Polaris system and an in-house developed LabVIEW code.

Several volunteers were used for the collection of this cranial motion, and their 6DOF head motion was used to calculate the corresponding six linear actuator trajectories which would be used to compensate for this head motion, using the inverse kinematics approach described in Section 2.2. These actuator leg lengths for one example volunteer head motion are shown in Figure 3.5. As described above, these results brought about modifications to the geometry of the robotic stage, such as the spacing between joint positions (d_t and d_b), and the radii of the top and bottom platforms (R_t and R_b). Performing these simulations on different geometry choices for the parallel kinematics robot helped finalize the design of the device by examining the relationship between linear actuator travel and the corresponding movements of the mobile top platform.

3.4 Fabrication and testing of parallel kinematics control in 6DOF

In this section, we discuss the fabrication and initial workspace evaluation of the robotic motion stage. Building on the simulations and geometrical consideration described previously, the robot was designed under the constraints of high compactness, weight load, and

precision. Specifically, the robot was developed to be sufficiently compact so as to maximize its utility in real-time head motion stabilization, selecting a platform radius of 75 mm and a stationary height of 183 mm. Custom linear actuators, together with the top and bottom platforms of the parallel kinematics device, were designed and built in-house (Figure 3.7, left). Each actuator was powered by a NEMA 8 stepper motor (OSM Technology) and controlled using an 8-axis stepper motor controller (PCI-7358 National Instruments [NI]).

The lead screw-motor system was selected to provide substantial vertical lift capabilities such that the device was capable of performing 6DOF motions with common anthropomorphic phantoms, such as a water-equivalent head phantom; theoretical expectations yield a proposed weight load capacity of 32 kilograms at a motor speed of 45 rotations per second following Equation 3.1, where d_m is the lead screw diameter (6 mm), l is the thread (0.64 mm, corresponding to a high-resolution 40 threads/inch (0.64 threads/mm)), and μ the coefficient of friction between the Kerkote TFE-coated lead screw and the Delrin anti-rotation collar (0.08 static, 0.09 dynamic).

$$T_{raise} = \frac{F \cdot d_m}{2} \left(\frac{l + \pi d_m}{\pi d_m - \mu l} \right) \quad (3.1)$$

Under the formulation, the robotic stage has the capacity to support a multitude of clinically used phantoms in order to provide controlled 6DOF motions to currently used phantoms. Additionally, the fine thread on the lead screw was selected to provide an exceptional accuracy of 3149 steps/mm, after tuning the stepper motor software to produce a control at the level of 2000 steps/rotation.

The universal joints which connect the linear actuators to the top and bottom platforms were not designed and built in-house, but were selected based on their rigidity, size, and range of motion. Their design specifications were extracted and modeled in SolidWorks [2013, SolidWorks Corp.] so the complete assembly could be visualized prior to full-scale device construction (Figure 3.7 (left)). A model of the universal joints used for the robot are shown in Figure 3.2 (right).

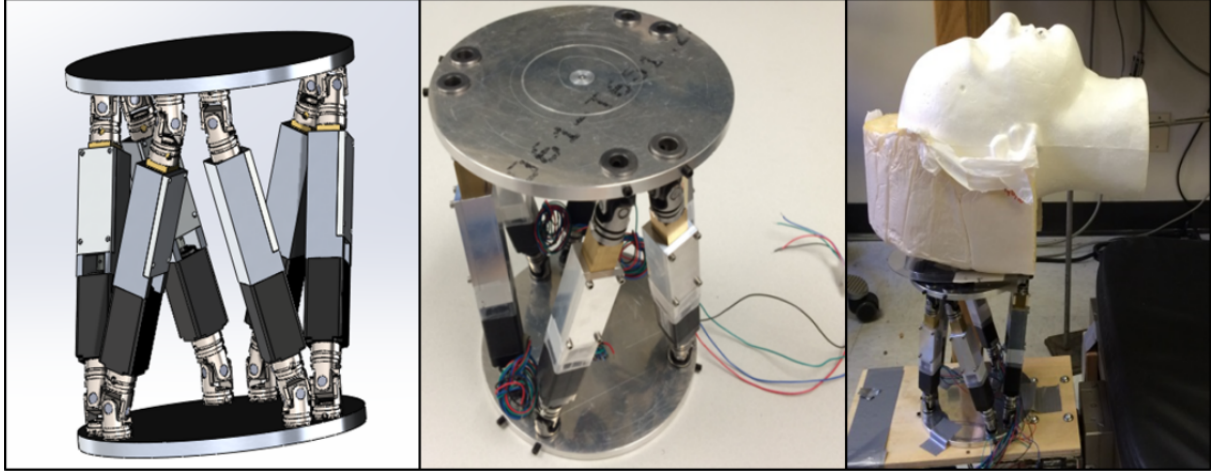


Figure 3.7: Design of the 6D robotic phantom (left), the completed device (middle), and the lab setup (right).

Couplings were designed to be able to join the linear actuators to the platforms, and modifications to the platforms were made so the universal joints could connect to them. The specialized couplings were individually built-in house to accommodate sub-millimeter variations among the 12 stainless steel die-cast universal joints. A prototype linear actuator was also initially built based on the model in Figure 3.2 (left), and its velocity and weight load limitations were assessed. The alloy 360 brass outer casing, which provides anti-rotation necessary to convert the rotational output of the stepper motor to linear motion of the drive shaft, was ultimately exchanged for aluminum to reduce weight, and the drive shaft was changed to brass in the final design. Because the outer casing and drive shaft have near-constant contact during driven motion, they could not be made of the same material, as doing so would produce shavings over time and erode the materials, reducing actuator efficacy and longevity. The now aluminum outer casing, brass drive shaft, and delrin coupling were built off-site based on these designs, while the Teflon-coated lead screw, aluminum top and bottom platforms, and other couplings were built in-house.

With every element complete, the robotic stage was assembled, shown in Figure 3.7 (center). The 6DOF workspace was then tested to determine the full range of motion capable; these values were computed for movements performed only in the direction reported since

	X	Y	Z	Pitch	Roll	Yaw
Minimum Position	-42.2 mm	-51.5 mm	-15.6 mm	-14.7 degrees	-15.0 degrees	-41.0 degrees
Maximum Position	42.2 mm	39.4 mm	15.5 mm	14.3 degrees	15.0 degrees	41.0 degrees
Maximum Speed (Theoretical)	55.8 mm/s	53.8 mm/s	29.3 mm/s	27.1 degrees/s	28.4 degrees/s	69.7 degrees
Maximum Speed (Experimental)	40.7 mm/s	38.1 mm/s	18.8 mm/s	17.4 degrees/s	18.1 degrees/s	44.6 degrees/s

Table 3.1: Summary of maximum workspace and maximum speed of the 6DOF robotic phantom in all six dimensions. Reported values correspond to 1D motions in the recorded direction.

specific 6DOF orientations of the top platform can yield a myriad of restrictions on the workspace for that particular configuration (Table 3.1). The relative directions selected for X, Y, and Z are arbitrary, but consistent following the geometry established in the design of the robotic stage: Z is the direction perpendicular to the top platform, and X and Y are in the plane of the top platform. The maximum velocities in each 6 directions were also theoretically calculated given the limits of the stepper motors and a computed actuator maximum speed of 31.75 mm/s. Experimentally evaluated using the prototype linear actuator, each leg of the device were found to be capable of linear speeds of up to 20.32 mm/s. Using the inverse kinematics formulation from earlier, we converted these limits on linear actuator velocity to limitations on each degree-of-freedom for the top platform, yielding the maximum 1D speeds in all 6 possible directions. The 6DOF workspace presented in Table 3.1 also includes the velocity limitations for the robotic stage.

3.5 Evaluation of parallel kinematics robot as a phantom for SRS applications

The goal of this section is to evaluate the 6DOF robot by using it as a robotic motion phantom for radiation therapy. The accuracy and reproducibility of the robotic motion stage was characterized through comparison of measurements of its 6DOF output trajectory to different intended 6DOF input trajectories. The input files included repeated motions

along each single axis to verify reproducibility and test the full 6DOF workspace, as well as more complicated trajectories to determine the full range of 6DOF capabilities of the device. Human 6DOF motion included prior recorded volunteer head motion over a 15 minute period (Wiersma 2010, Belcher 2012) and patient prostate motion over a 2 minute period (Tehrani 2013). [54, 66, 45] These input trajectories were then transmitted to a LabVIEW [2012, National Instruments] application, and subsequently communicated to the six motors through the NI PCI-7358 controller card and the two stepper motor power drives.

Output motion of the robotic stage was tracked using an infrared (IR) optical tracking camera [Polaris, Northern Digital Inc., Waterloo, Ontario, Canada]. Here, a 4-point reflective marker tool was fixed to the top platform of the device whose motion was monitored in real-time. The four markers on the tool define a set of vectors directed to a point in space; this point in space, called the virtual point, also has an inherent orientation as well as a position, and therefore has its own coordinate frame. For the purposes of this evaluation, this virtual point coordinate frame was established at the control coordinate frame of the Stewart platform, such that the resultant 6DOF motion of the device could be effectively compared to the intended trajectory. That is, the tool file defining the relationship between the four IR reflective markers and their virtual point coordinate frame was created such that this coordinate frame is located at the same origin and with the same orientation as the coordinate frame defining the control point of the robotic stage. The control coordinate frame is the coordinate frame used when controlling the robotic stage; when 6DOF motions are performed by the Stewart platform, it is this coordinate frame which moves in space. Defining the control coordinate frame for the robotic motion control as well as the virtual point coordinate frame are necessary for quantitative comparisons between intended and measured output motion of the device. A calibration technique was applied to connect these coordinate frames to the camera coordinate frame for measurement data alignment. [67]

To ensure proper alignment of the device coordinate frame, a transformation was performed between the coordinate frame of the Polaris camera system and the virtual point

frame, which was previously established as the Stewart platform frame of reference. [68] The approach used for calibration was later improved upon, and did not require the user to establish a virtual coordinate frame at the robot control frame prior to tracking, as was the case in this initial evaluation of the 6DOF robot for the phantom application. The calibration approach used for this evaluation required the robot to follow a translation-only trajectory, and the measured X, Y, and Z motion was compared to the expected motion singular value decomposition (SVD). [67] This methodology considers the two reference frames $\mathbf{x}_n$ and $\mathbf{y}_n$ with weight w_n between each pair, for $n = 1 \dots 8$ positions and attempts to find an orthogonal matrix $U = u_{ij}$ which minimizes equation 3.2 below, subject to the constraint in equation 3.3, where δ_{ij} are the elements of the unit matrix.

$$E = \frac{1}{2} \sum_n w_n (U\mathbf{x}_n - \mathbf{y}_n)^2 \quad (3.2)$$

$$\sum_k u_{ki}u_{kj} - \delta_{ij} = 0 \quad (3.3)$$

The elements of this orthogonal matrix U are found in equation 3.6 with the unit eigenvectors of $R^\top R$, $\mathbf{a}_k$ and $\mathbf{b}_k$ defined in equation 3.4, and the elements of R , r_{ij} defined in equation 3.5. Here μ_k are the associated positive eigenvalues of $R^\top R$.

$$\mathbf{b}_k = \frac{1}{\sqrt{\mu_k}} R\mathbf{a}_k \quad (3.4)$$

$$r_{ij} = \sum_n w_n y_{ni} x_{nj} \quad (3.5)$$

$$u_{ij} = \sum_n b_{ki} a_{kj} \quad (3.6)$$

This technique permits accurate tracking of the robotic stage in 6DOF, and excluding the inherent IR camera noise and minimal setup uncertainty, it is an error-minimizing procedure.

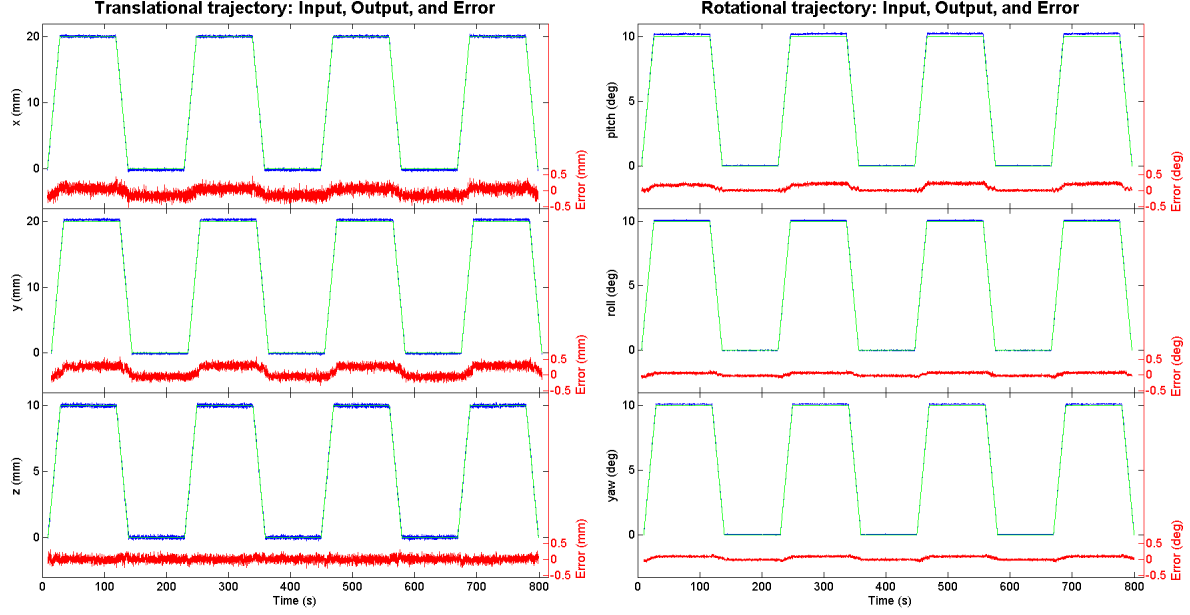


Figure 3.8: Comparison between input (green) and output (blue) translational motion for 20 mm in X (top left) and Y (center left), and 10 mm in Z (bottom left), with resultant RMSE values of 0.14 mm, 0.22 mm, and 0.08 mm, respectively. Rotationally, a comparison was made for 10 degrees in pitch around X (top right), roll around Y (center right), and yaw around Z (bottom right), with resultant RMSE values of 0.16 degrees, 0.06 degrees, and 0.08 degrees, respectively. Below the comparison plot is the difference between the intended input trajectory and the measured output trajectory in red.

As mentioned, a more robust calibration methodology for arbitrary virtual point positions and orientations was later developed, but the technique described above, combined with the overlapping of the virtual point and control coordinates frames is sufficient to evaluate the accuracy of this robotic stage.

As a first step in evaluation, an error simulation was conducted. This study took as inputs potential machining and assembly errors, and using the inverse kinematics formulation, translated these errors (dominated by linear actuator assembly errors) to potential resultant 6DOF error. The uncertainty in the 6DOF positional accuracy of the device was analyzed assuming conservative estimates of the initial actuator lengths and the center of the universal joint pivot points, which define the control point of the top platform [69]. Under the assumption of normally distributed initial 6DOF setup errors of the robotic stage in

the above elements, with standard deviations of 0.1 mm translationally in both the initial actuator length and universal joint pivot positions, we found the mean of the translational errors to be 0 with RMSE of 0.01 mm, 0.02 mm, and 0.01 mm in X, Y and Z, respectively, and the mean of the rotational errors to be 0 with RMSE of 0.01 degrees in all pitch, roll, and yaw. These values were obtained after 10,000 simulated points sampled in a workspace spanning 2 mm in X, Y, and Z, and 1 degree in pitch, roll, and yaw; this workspace covers the 6DOF span of both the cranial and prostate motion used in the experimental evaluation, and is large enough to cover 6DOF cranial motions commonly observed during 15-minute frameless SRS procedures. [70]

A comprehensive analysis was then performed to ascertain the 6DOF top platform position errors produced from the individual sources of uncertainty. Upon completing the calibration, several trajectories were transmitted to the stage and recorded using the IR marker tracking camera, and the root mean square error (RMSE) was computed over the entire duration of each motion to validate the efficacy of the 6DOF system. The results of the six independent 1D motion analysis are shown in Figure 3.8 for translational and rotational motion. We obtained RMSE values of 0.14 mm, 0.22 mm, 0.08 mm, 0.16 degrees, 0.06 degrees, and 0.08 degrees for X, Y, Z, pitch, roll, and yaw, respectively. As displayed in the figures, the 6DOF system exhibited excellent levels of reproducibility with minimal backlash over a substantial range of motion. As can be seen in the lower plot components in Figure 3.8, which characterizes the difference between the intended trajectory and the measured output trajectory, the error does not vary substantially over time, nor does it increase significantly at the transition point in the trajectory between linear 1D motion and rest, indicating low backlash levels and a high degree of reproducibility.

The recorded output motion in the cranial case was compared to the intended 6DOF trajectory, and over a 15 minute trajectory, we found RMSE values of 0.03 mm, 0.04 mm, 0.04 mm, 0.03 degrees, 0.02 degrees, and 0.04 degrees in X, Y, Z, pitch, roll, and yaw, respectively (Figure 3.9). For the 6DOF prostate motion, the IR-recorded output motion

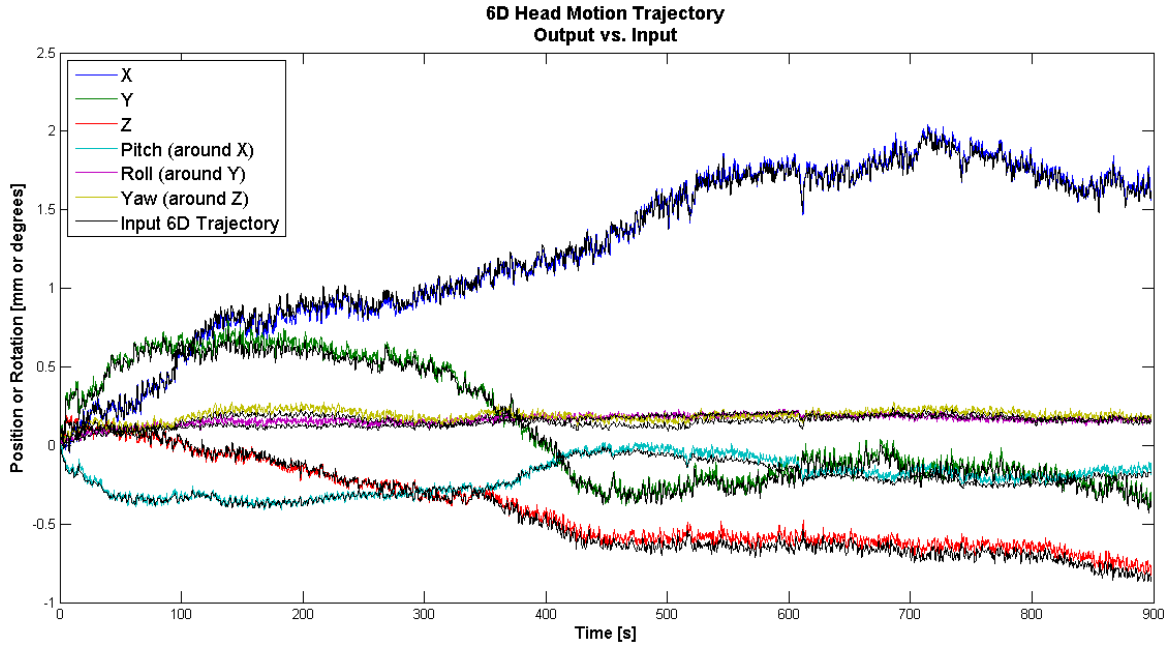


Figure 3.9: Comparison between input and output 6D head motion for X, Y, Z, pitch, roll, and yaw, with resultant RMSE values of 0.03 mm, 0.04 mm, 0.04 mm, 0.03 degrees, 0.02 degrees, and 0.04 degrees, respectively.

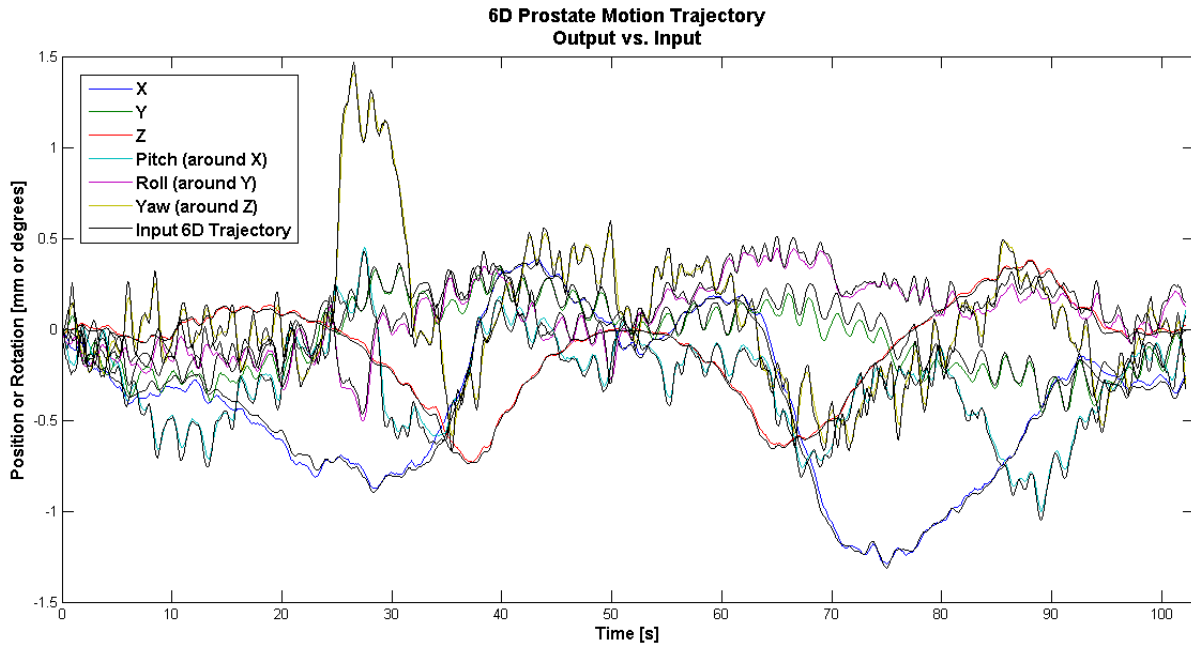


Figure 3.10: Comparison between input and output 6D prostate motion for X, Y, Z, pitch, roll, and yaw, with resultant RMSE values of 0.03 mm, 0.06 mm, 0.01 mm, 0.03 degrees, 0.04 degrees, and 0.04 degrees, respectively.

was once again compared to the desired trajectory over a 110-second trajectory, and we found RMSE values of 0.03 mm, 0.06 mm, 0.02 mm, 0.03 degrees, 0.04 degrees, and 0.04 degrees in X, Y, Z, pitch, roll, and yaw, respectively (Figure 3.10). For both the cranial and prostate cases, the difference between the intended trajectory and the measured trajectory, even over a 15-minute trajectory, does not vary substantially over time, and furthermore that this difference is within an envelope of ± 0.15 mm and ± 0.16 degrees for both cases over the entire trajectories.

The presented 6DOF robotic system has demonstrated a high level of accuracy in replicating input 6-dimensional trajectories, while maintaining many of the robust and important features of a parallel kinematics machine, such as high load-to-weight ratio, compactness, longevity, speed and reproducibility. Furthermore, validation of theoretical characteristics of the device was conducted experimentally, and a large 6DOF workspace was established. The device was also shown to have notable speed capabilities, with maximum translational and rotational speeds of up to 4 cm/s and 44 degrees/s, further supporting its multipurpose functionality. As presented in the RMSE results and in Figures 3.8 - 3.10, the robotic system is capable of rapid direction changes; the inherent noise in the Polaris tracking system increased the variability from set point to set point, demanding swift alteration of direction and sharp increases in instantaneous velocity. The response time and communication lag between the software and various hardware components were found to be sufficiently small, providing effective motion control even over irregular trajectories, such as those for tumors. Although such lesions are also capable of undergoing deformations, this work focused on the replication of rigid 6DOF motion, as deformable motion trajectories are substantially more complex and would require site-specific phantoms.

A versatile, accurate and effective 6DOF robotic motion phantom for use in radiation therapy QA and for research purposes was evaluated in terms of its ability to precisely replicate complex 6DOF motions, as well as its ability to respond to rapid changes in direction. In combination with the high degree of accuracy achievable by the device, and its compact

size, the robotic 6DOF motion system presented here has a high degree of flexibility and can support a wide range of applications for simulating the motion of several tumor locations and anatomical features. By first using the parallel kinematics robot as a motion phantom, we were able to demonstrate its feasibility for following real 6DOF tumor trajectories. Since the device was able to reproduce the cranial motions over 15 minutes, we were more confident in its ability to respond to similar motions over the same time period by compensating for head movements in real-time.

CHAPTER 4

6DOF CALIBRATION PROCEDURE

4.1 Introduction

One approach to managing patient motion involves monitoring intrafraction movements of the target or nearby surrogates. Many tracking systems are used clinically, and are capable of monitored patient motion translationally and rotationally. The ultimate goal of this thesis work is to develop a 6DOF robotic motion compensation system, which uses clinical patient motion tracking systems as its primary input for control. It is essential to consider a start-up calibration procedure for the robot and the camera system, and to connect these coordinate frames with the target position. This chapter explores such a calibration procedure for the set of applications of the robotic stage, including motion compensation.

To effectively interpret the raw camera measurements, care must be taken to connect the camera and control coordinate frames, particularly when translations and rotations are both monitored. For the purposes of this work, the coordinate system includes the camera coordinate frame and the six degrees-of-freedom (6DOF) robotic motion stage, as well as the target coordinate frame. This parallel kinematics device follows the design and construction characteristics described at length in Chapter 3. A theoretical formalism for the calibration procedure was first established, and then tested with an IR marker tracking system (Polaris, NDI). It is important to note that the theoretical approach is not restricted to the IR marker tracking system used throughout the majority of this thesis project; instead, any camera system capable of recording motion in 6DOF with adequate levels of accuracy (sub-millimeter and sub-degree), is sufficient for the purposes of calibrating the robot to the tracking frame.

In addition to the theoretical approach to calibrating the robot to a generic 6DOF motion tracking system, a calibration protocol was also developed. This protocol is a series of calibration trajectories which could be performed by the 6DOF robot. The camera tracking

system would track the motion of the robot, and the resultant 6DOF motion measurements would be interpreted to determine the rotation matrices and displacement vectors connecting the camera, robot, and virtual point coordinate frames. This rapid calibration was then performed prior to any subsequent use of the 6DOF robot. The overall calibration procedure is described below in Section 4.2.

It is also important to consider the start-up procedure for calibrating the robot to the treatment position if and when the 6DOF motion compensation system is used with patient radiotherapy. A theoretical treatment of this clinical procedure is later described in Chapter 8.

4.2 Theoretical 6DOF calibration of motion tracking coordinate systems

4.2.1 *Establishing coordinate frame geometry*

One must first consider the coordinate frame setup of the camera tracking system (Figure 4.1, left). This setup includes the inherent camera reference coordinate frame (**A**), the inherent robotic motion phantom control frame (**H**), and the inherent virtual point, or measurement, frame (**M**); these three un-primed coordinate frames are not fundamentally associated mathematically and can be considered as the initial conditions before the motion is performed. As the virtual point frame (**M**) represents the origin of the tracked coordinate frame, and the elements which are tracked by either the IR camera or the optical surface map are rigidly connected to the top platform of the robotic motion phantom, the coordinate frames **M** and **H** are also necessarily rigidly connected. The orientation of these coordinate systems are not initially aligned, but the translational vector connecting the origins of these two frames, which we call δ_{HM} , is immutable in magnitude.

The notation δ_{12} can be understood as the 3D vector connecting the origins of two arbitrary coordinate systems 1 and 2, in the frame of reference of 1. Similarly, we introduce

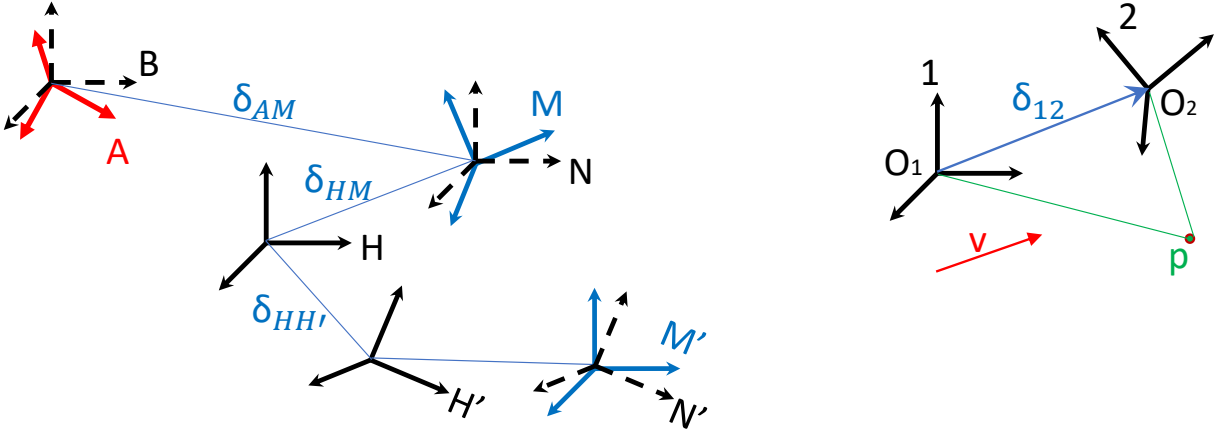


Figure 4.1: The assembly of coordinate systems considered for motion control, calibration and tracking evaluation (left), and the point and vector representation in two coordinate frames (right). Here, coordinate frames **A** and **B** are the camera frames, **M** and **N** are the measurement coordinate frames, **H** is the control frame for the phantom, and the primed analogs are these latter coordinate frames after motions were performed by the **H** control frame.

the 3x3 rotation matrix R_{12} which defines the orientation of coordinate frame 2 in reference to coordinate frame 1. With this established, it can be seen that some point p in coordinate frame 1 can be transformed to be represented in another coordinate frame 2 by equation 4.1 below (Figure 4.1, right). Further, a vector v represented in one coordinate frame can be represented in another frame simply through the application of the requisite rotation matrix (equations 4.2 and 4.3)

$$p_2 = \delta_{21} + R_{21}p_1 \quad (4.1)$$

$$v_2 = R_{21}v_1 \quad (4.2)$$

$$v_1 = R_{12}v_2 \quad (4.3)$$

4.2.2 Calibration theory

To aid in simplifying the derivation for a calibration approach, we can also define the **B** and **N** frames as coordinate systems which share an origin with the camera frame **A** and the virtual

point frame $\mathbf{M}$, respectively, but which are oriented to the inherent control coordinate frame $\mathbf{H}$ (Figure 4.1, left). Motions are performed by the robotic motion phantom ($\mathbf{H}$), moving this control coordinate system to the $\mathbf{H}'$ frame. As the $\mathbf{H}$ frame is rigidly connected to the $\mathbf{M}$ (and $\mathbf{N}$) frame, $\mathbf{H}'$ is necessarily connected to the resultant $\mathbf{M}'$ (and $\mathbf{N}'$) frames, and so as motions are performed, $\mathbf{H} \rightarrow \mathbf{H}'$, $\mathbf{M} \rightarrow \mathbf{M}'$, and $\mathbf{N} \rightarrow \mathbf{N}'$, with the orientation of $\mathbf{N}'$ continuing to match the orientation of $\mathbf{H}'$ by geometrical requirement. Subsequently, as trajectory inputs are fed into the robotic motion phantom, the measurement coordinate system $\mathbf{M}'$ is moved in reference to the camera system $\mathbf{A}$. In order to be able to compare the input motion of the phantom to the measured motion of the frame $\mathbf{M}'$, several geometrical relations must be established; to this end, a calibration methodology was employed, which served to primarily compute R_{HA} , R_{NM} and δ_{HN} . This methodology is presented below, under the assumption that motion control of the phantom was done so precisely; namely, that $\delta_{HH'}$ and $R_{HH'}$ are known.

In order to compute R_{HA} , one must utilize both the input intended motions, as given by $\delta_{HH'}$ and $R_{HH'}$, and the measurement output, as given by $\delta_{AM'}$ and $R_{AM'}$. An input trajectory, discussed in greater detail in the following section, was performed by the robotic phantom in a $1.5 \times 1.5 \times 1.5$ cm workspace, and monitored using the tracking system. Comparing the measured values for these points to the intended values using singular value decomposition yielded a rotation matrix R_{HA} , which shows the orientation relationship between these two coordinate systems. Furthermore, utilizing this matrix and the average values of both the intended and measured motions, one can extract the translational vector between the camera frame and the measurement frame, δ_{BM} using equation 4.4 below. Because the origins of frames $\mathbf{M}$ and $\mathbf{N}$ are the same, and the vector separating them from frame $\mathbf{B}$ is the same in the reference frame of $\mathbf{B}$; that is, $\delta_{BM} = \delta_{BN}$. Using the measured angular data $R_{AM'}$, which is uniform throughout this translation-only trajectory, the rotation matrix R_{AM} is also found.

$$\delta_{BM} = -(R_{HA} \cdot \text{mean}(\delta_{AM'})) + \text{mean}(\delta_{HH'}) \quad (4.4)$$

Then to determine δ_{HN} , the displacement vector between the origins of the robot and virtual point coordinate frames, a trajectory consisting of a set of six angles were performed by the robot: 5 degrees and 10 degrees in the pitch degree of freedom (around left-right, LR), 5 and 10 degrees in roll (around superior-inferior, SI), and 5 and 10 degrees in yaw (around anterior-posterior, AP). Similar to the previous translation-only trajectory, the measurement data were compared to the intended input angles and analysis was performed to construct the vector separating the robot **H** and virtual point **N** coordinate frames. In practice, the trajectory of six rotations was reproduced by the robot, and the measured movements of the virtual point relative to the camera, $\delta_{AM'}$, were recorded by the tracking system and used analytically to compute the unknown vector δ_{HN} using the theoretical approach outlined below. The measured pitch, roll, and yaw angles were first converted to a 3×3 rotation matrix and subsequently transformed to $R_{NN'}$ using equation 4.5 below. This equation exploits the fact that $R_{HA} = R_{NA}$ because coordinate frames **N** and **H** are oriented in the same way by definition, as well as the relationship $R_{MN} = R_{M'N'}$, which holds because the orientation of **M** relative to **N** is maintained even when the virtual point moves (**M** $\rightarrow$ **M'** and **N** $\rightarrow$ **N'**). Finally, in Equation 4.5 we use the definition of applying multiple rotation matrices in sequence to yield a composite rotation matrix: $R_{XY}R_{YZ} = R_{XZ}$ as well as the definition for the transpose of a matrix in this construction to be $R_{XY}^\top = R_{YX}$ for arbitrary coordinate frames **X**, **Y**, and **Z**.

$$(R_{HA}R_{AM'}) \cdot (R_{MA}R_{HA}^\top) = (R_{NM'})(R_{MN}) = R_{NM'}R_{M'N'} = R_{NN'} \quad (4.5)$$

With these relationships determined experimentally, it is necessary to show the theoretical calculation of δ_{HN} , presented in the series of equations below. Because coordinate frames **M** and **N** share an origin, so do **M'** and **N'**, which permits the construction of equation 4.6.

$$\delta_{BN'} = \delta_{BM'} = R_{BA}\delta_{AM'} = R_{HA}\delta_{AM'} \quad (4.6)$$

From geometry, one can also observe the relationship given by equation 4.7.

$$\delta_{NN'} = \delta_{BN'} - \delta_{BN} \quad \text{where} \quad \delta_{BN} = \delta_{BM} \quad (4.7)$$

An arbitrary point p in frame $\mathbf{N}'$ can be represented in frame $\mathbf{N}$ using now-known quantities by equation 4.8.

$$p_N = \delta_{NN'} + R_{NN'}p_{N'} \quad (4.8)$$

Exploiting the rigid connectivity between frames $\mathbf{H}$ and $\mathbf{N}$, we find

$$\delta_{HN} = \delta_{H'N'} \quad (4.9)$$

$$R_{HN'} = R_{NN'} = R_{HH'} \quad (4.10)$$

An arbitrary point p in frame $\mathbf{N}$ can then be represented in frame $\mathbf{H}$ by equation 4.11

$$p_H = \delta_{HN} + p_N \quad (4.11)$$

One can then find the relationship between the $\mathbf{N}'$ and $\mathbf{H}$ frames with equations 4.12 and 4.13

$$\delta_{HN'} = \delta_{HH'} + R_{HH'}\delta_{H'N'} = \delta_{HH'} + R_{HH'}\delta_{HN} \quad (4.12)$$

$$p_H = \delta_{HN'} + R_{HN'}p_{N'} \quad (4.13)$$

Equating equations 4.13 and 4.11, and using the relationship $R_{HN'} = R_{NN'}$, one finds

$$p_N = \delta_{HN'} - \delta_{HN} + R_{NN'}p_{N'} \quad (4.14)$$

Comparing to equation 4.8 yields

$$\delta_{HN'} - \delta_{HN} + R_{NN'}p_{N'} = \delta_{NN'} + R_{NN'}p_{N'} \quad (4.15)$$

Reducing and considering equation 4.12 yields the desired equation for δ_{HN} , evaluated using known quantities $R_{HH'}$, $\delta_{NN'}$ and $\delta_{HH'}$

$$\delta_{HN} = (R_{HH'} - \mathbb{I}^{3 \times 3})^{-1}(\delta_{NN'} - \delta_{HH'}) \quad (4.16)$$

As $\delta_{NN'}$ is known from the measured data and equation 4.7, and $R_{HH'}$ and $\delta_{HH'}$ are the intended input control motion, δ_{HN} can be effectively computed after the robotic motion stage performs a trajectory composed of a set of translational and rotational points, as described above. This calibration series was implemented in Matlab (The Mathworks, 2013b) and used for proper coordinate frame alignment prior to camera system evaluation or performing motion compensation, depending upon the particular application of the device. An L_2 optimization form was used to compute δ_{HN} from the set of input data, transformed to $\delta_{NN'}$.

4.2.3 Calibration protocol

The practical aspects of the calibration procedure were discussed briefly in the previous subsection in order to aid in describing the theoretical backbone of 6DOF calibration. Here, we delve into the functional elements of the calibration protocol for real-time motion control. Using the geometry established in Figure 4.1, which corresponds to a physical setup shown in Figure 4.2.

A Styrofoam head phantom had a IR marker block with four reflective markers fixed to the phantom's forehead, in a similar fashion to how the forehead block is attached to

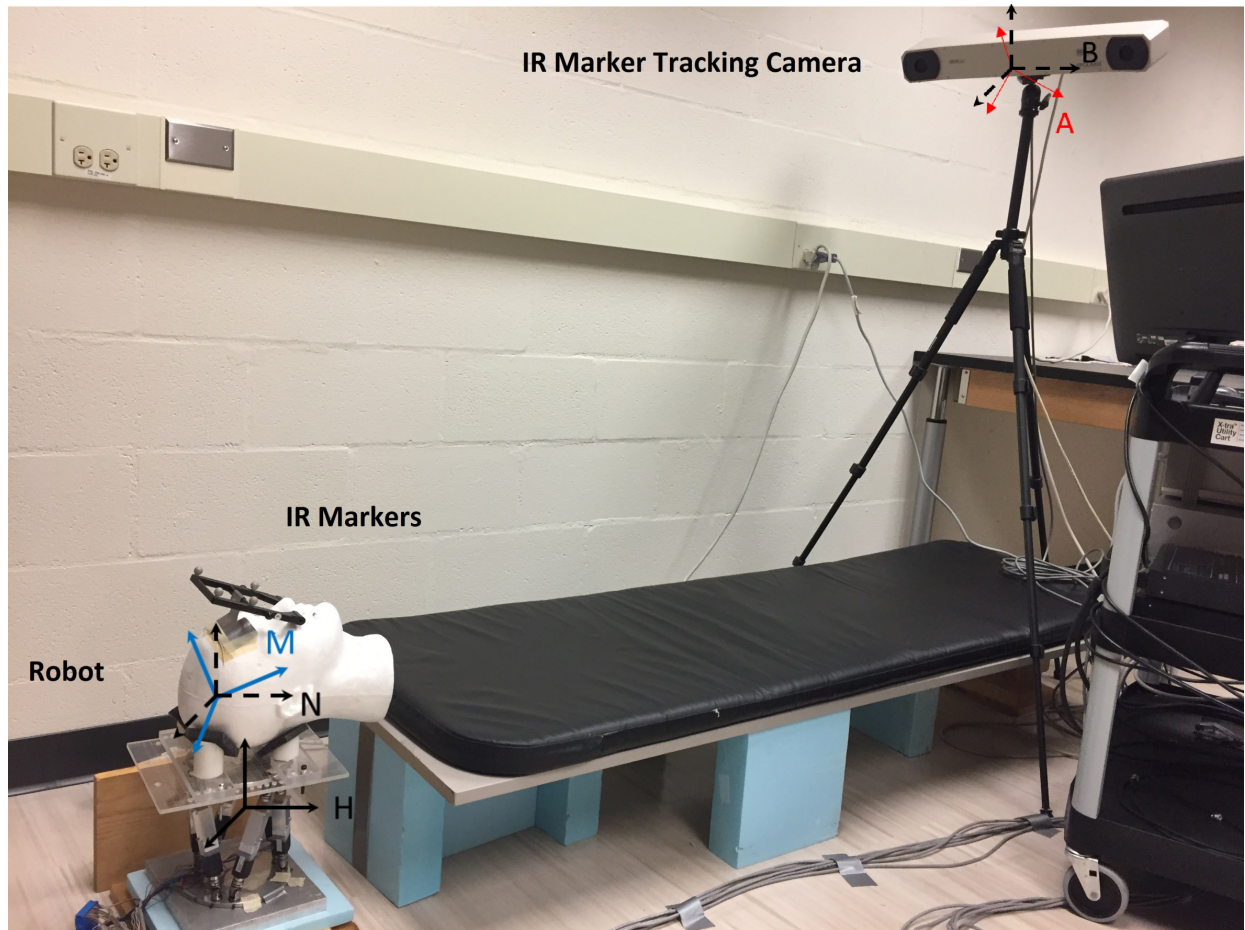


Figure 4.2: Experimental lab setup with the 6DOF parallel kinematics robot, the set of four reflective IR markers, and the IR marker tracking camera. The coordinate frames corresponding to each of these three components are also shown. The virtual point frame $\mathbf{M}$ ($\mathbf{N}$ in the orientation of the robot frame $\mathbf{H}$) was placed near the center of the brain, to better mimic the expected clinical setup, wherein the $\mathbf{M}$ frame is placed at the treatment position.

volunteers. The 6DOF robot was placed several meters away from the stereoscopic IR camera, to simulate a clinical setup, and a head rest consisting of four posts with foam surfaces was attached to the top platform of the robot (Figure 4.2). The head phantom was placed on top of the head rest, further mimicking the setup which was used with volunteers. A quick test was then performed to confirm that the four IR markers were within the field of view of the IR camera.

After setup, the robot performed the calibration trajectory, which included both a translational and a rotational component. In general, the translational component provided information about the orientation differences among coordinate frames, while the rotational component gave displacements between coordinate frame origins. The translational calibration consisted of an 15-point calibration trajectory wherein the robotic device moved across a 3D (translation-only) trajectory to 15 points in a 1.5cm^3 cube and remained stationary for 90 seconds (1000 data points at a Polaris frame rate of 12 Hz) to provide good statistics for averaging. A rotation-only trajectory was then performed, which consisted of a set of rotations about the **H** robot frame: 1 and 5 degree rotations in pitch, roll, and yaw. By moving the virtual point **M** frame about the **H** frame, one can extract information about the displacement vector separating these two frames, when used with information from the translational calibration. Using the theoretical calibration methodology described above, one can import the camera measurements and automatically evaluate the rotation matrices and displacement vectors separating the three coordinate frames.

Although the two calibration trajectories can theoretically be combined into one protocol of both translations and rotations, it was useful to perform them independently for the purposes of developing the calibration technique. When the robot is used clinically, a faster calibration methodology may be developed which achieves similar results. Also important to note is that the phantom geometry does not match the geometry of humans when the robot is used for real-time head motion compensation. As a result, the coordinate frame geometry calculated from the calibration approach becomes slightly incorrect when transitioning from

the phantom to humans. When the human replaces the phantom, the calibrated camera measurements capture a setup error, which is effectively the difference between the phantom and human head geometries. A rapid transformation is performed prior to motion monitoring with the human, which aims to zero-out this setup error. This ad-hoc transformation degrades the accuracy of the calibration slightly, but it is a necessary initialization component for real-time motion compensation and general motion monitoring for human subjects. Recent research into the effect of setup errors such as this calibration degrading initialization has shown that the motion compensation function of the parallel kinematics robot is actually quite robust to such errors. Experimentally shown in Chapter 7, this practical calibration protocol and initialization procedure for real-time head motion compensation is essential for stabilizing volunteer motion, and achieves the desired sub-millimeter and sub-degree performance.

Finally, to validate this calibration methodology, the calibration trajectory was performed using the parallel kinematics robot, and the resultant motion was tracked using the stereoscopic IR marker tracking camera with capture rate of 8 frames/second. In this setup, the origins of the camera and measurement coordinate frames were separated by several meters, which is similar to a clinical setup for patient tracking, and the orientations of the frames of reference were not initialized prior to the start of motion; that is $\delta_A M$ was non-zero, and R_{AM} , R_{HM} , and R_{HA} were not unit matrices. The four reflective markers shown fixed to the robotic motion phantom in Figure 4.2 defined a virtual point coordinate frame $\mathbf{M}$ which was tracked with the IR camera frame $\mathbf{A}$.

The maximal errors across the entire calibration procedure were less than 0.22 mm translationally and 0.25 degrees rotationally, and was therefore considered effective, as this is on the same order as the Polaris IR camera tracking noise. [48] Specifically, the translational error in calculating δ_{HN} was 0.14 mm, 0.22 mm, and 0.20 mm in LR, SI, and AP, respectively; the maximum angular error comparing intended angular motion to an appropriately transformed measured output was 0.20 degrees; and comparing $\delta_H H'$ to the transformed

$\delta_{AM'} \rightarrow \delta_{NN'}$ data using the theoretical calibration technique described in Section 4.2.2 yielded a maximum translational error of 0.23 mm after calibration.

CHAPTER 5

PATIENT MOTION TRACKING

5.1 Introduction

5.1.1 Overview of patient motion tracking systems

Patient motion monitoring systems are used to inform clinicians about the changes in patient positioning during treatment, and whether measured motion requires therapist intervention or radiation beam OFF gating. The goal of this chapter is to perform 6DOF quality assurance (QA) on clinical patient tracking systems using the 6DOF robot as a QA phantom.

Recent studies have shown that in addition to monitoring patient motion along the left-right (LR), superior-inferior (SI), and anterior-posterior (AP) directions, it is also important to consider rotational motion along the pitch (around LR), roll (around SI), and yaw (around AP) axes. [36, 35, 37, 71] Failure to consider all 6 degree-of-freedom (6DOF) movements can result in poor dose conformity during radiation therapy (RT). This is especially true for highly conformal modalities, such as frameless stereotactic radiosurgery (SRS), where removal of the frame in favor of light immobilization demands the need for continuous real-time 6DOF head motion tracking. [53, 52]

Patient motion monitoring systems can be either classified as internal or external in nature, and many clinical tracking systems now have the capacity to monitor both translational and rotational motion. Internal 6DOF tracking systems typically use kilo-Voltage (kV) imaging (Tehrani 2013, Shirato 2000) to track patient anatomy or markers which have been embedded into or a target site.[45, 43] However, due to the invasiveness of marker insertion and the potential high cost of imaging dose to the patient, use of kV tracking is often limited to occasional snap-shots or short periods of fluoroscopy.[45, 72, 73] Magnetic field imaging can also be used to track embedded transponders in 6DOF in patients, though such techniques are similarly invasive and cannot be used in many treatment modalities, including



Figure 5.1: Two external patient tracking systems, an AlignRT 3D surface mapping system (left) and a Polaris IR marker tracking system (right). (Image credit: VisionRT, London, UK and Polaris, Northern Digital, Inc., Waterloo, ON, Canada).

SRS.[35, 74] Radiotherapy treatments of cranial lesions do not require such invasive internal tracking.

Optical 6DOF tracking systems do not require the use of ionizing radiation and can provide continuous real-time patient tracking over long periods of time. [10, 75] Two such external optical tracking systems which have observed expanding clinical use in radiotherapy, including frameless SRS treatments, are infrared (IR) marker tracking and 3D optical surface imaging. [46, 76, 47, 48] Among these external tracking modalities, 3D surface image based imaging offers a potential ease-of-use advantage in that it does not require the addition of IR markers for tracking, which may accelerate patient setup and prevent marker reproducibility issues. The IR marker tracking system monitors the movement of reflective IR markers which are placed on the patient's surface near the treatment site, or in other useful positions on the treatment couch. 3D surface imaging projects an infrared speckle map on the patient's surface near the target, and monitors motion of the contour as a surrogate for internal motion.

5.1.2 *Quality assurance of 6DOF systems*

Many current radiotherapy treatment techniques rely on 6DOF tracking systems to perform critical functions such as intra-fractional patient motion monitoring or gating the radiation treatment beam [45, 72, 73, 35, 77, 10, 75, 46, 76, 48]. Failure of such devices to operate

within desired specifications could lead to improper dose coverage, poor clinical outcomes and patient injury. However, while 6DOF optical patient monitoring is widespread in radiotherapy, as of yet there are no quality assurance (QA) methods to fully quantify their full 6DOF operating performance. Previous research which has explored the capabilities of such systems has typically focused only on translational degrees of movement or has relied on limited reference standards for angular accuracy evaluation [36, 52, 77, 48, 78, 79, 80, 81]. We use the calibration technique described in Section 4.2 and the highly precise parallel kinematics robot to evaluate the performance of three 6DOF patient tracking systems: a reflective IR marker tracking system called Polaris (Polaris, Northern Digital, Inc., Waterloo, ON, Canada), which monitors the motion of four reflective IR markers that define a coordinate frame, and two 3D optical surface mapping systems by AlignRT (VisionRT, London, UK), presented in Figure 5.1. In this chapter, by coupling the calibration method with an evaluative technique using the robot, we enable the evaluation of multiple points located throughout a 6DOF workspace quickly and accurately.

5.2 6DOF assessment of external patient tracking systems

5.2.1 *Experimental setup*

To effectively evaluate the tracking accuracy of a 6DOF patient tracking system, care must be taken to connect the camera and control coordinate frames, particularly when translations and rotations are both monitored. The two coordinate systems are the camera coordinate frame $\mathbf{M}$ and the six degrees-of-freedom (6DOF) parallel kinematics robot coordinate frame $\mathbf{H}$, which was used to perform motions spanning a comprehensive 6DOF workspace. [70] The robot’s geometry, described in Chapters 2 and 3, followed a Stewart-Gough parallel kinematics archetype and was controlled using inverse kinematics. [69, 64] Previous chapters focused on the design, production, and evaluation of this system demonstrated the control capabilities to be less than 0.22 mm translationally and 0.16 degrees rotationally. [70] As

such, the highly precise robotic motion phantom control was considered as the reference; and this control point, or control coordinate system, was used as the reference baseline against which the camera point, or tracking coordinate system, was compared.

As an initial step to evaluate the performance of the optical tracking systems, the calibration methodology described in Chapter 4 was developed and verified in order to associate the camera and control coordinate frames effectively. [70, 67] The calibration was implemented in this study to accurately evaluate the tracking capabilities of the two clinical patient tracking systems chosen for this study: the Polaris IR tracking camera and two AlignRT optical surface mapping systems. Because of the generalized nature of the methodology, the same calibration protocol could be used with every 6DOF camera system evaluated in this study.

In the Polaris IR camera evaluation case, a set of four reflective IR markers were fixed to the top platform of the parallel kinematics robot. A tool file was created for the IR marker block, and the associated virtual point coordinate system was placed several centimeters away from the center of the marker block. This position was chosen to simulate a typical distance between marker block center and the treatment position, when the IR marker tracking system is used for patient motion monitoring. The robot and IR marker block system was positioned a distance away from the stereoscopic IR camera to further mimic a standard clinical setup. This setup is shown in Figure 5.2.

A similar setup was used when the AlignRT 3D surface cameras were evaluated. The configuration differed slightly to the IR camera case: the reflective IR marker block was replaced by a head phantom which was fixed to the mobile top platform of the 6DOF robot. As discussed above, the AlignRT system projects an IR speckle map on the tracked surface, and defines a relationship between the contoured point mesh generated from this patient surface and the pre-calibrated treatment isocenter. The AlignRT system relies on an initialization procedure which constructs a reference surface at the beginning of treatment, determines the vector displacement between points in the surface mesh and isocenter, and during treatment evaluates patient motion based on the changes between the real-time tracked surface and the



Figure 5.2: Experimental setup for the test of the calibration methodology using an IR marker tracking system. Shown are the robotic motion phantom with a set of four reflective IR markers fixed to the top platform, which were tracked using the stereoscopic IR camera (not shown). The arrangement of the four markers produce a virtual point for tracking, which is called the **M** frame of reference.

reference contour. Because of this difference, a head phantom was used to simulate a patient head. The robot was placed on the treatment couch, in full view of all three AlignRT camera pods (Figure 5.3, left). Also shown in Figure 5.3 (right) is the AlignRT software which provides therapists with the real-time 6DOF measurements of the surface contour’s movements. The AlignRT system requires the user to select a region of interest (ROI) on the tracking surface, so that the software does not monitor the motion of arbitrary structures within the camera fields of view. Although the phantom’s surface does not suffer from complications caused by a deformable surface, which exists when tracking human motion, the ROI was selected to mimic one which would be reasonably used in the clinic for a thermoplastic mask with an open face. Such an ROI covers the most rigid and non-deformable parts of the patient’s head. Previous work has shown that the ROI shown in Figure 5.3 (right), spanning in the SI axis from the nose bridge to the top of the forehead and in the LR axis from the left to the right temple, proved to be a reasonably effective region under these requirements. [48] This earlier research also explored the effect of ROI size on frame rate (fps) and absolute positional accuracy, concluding that the processing period ($1 / \text{fps}$) increased linearly as a function of ROI area, while tracking accuracy was independent of ROI size, provided a sufficiently large region with suitable 3D features was selected. As a result, we were confident that our choice of ROI shown in Figure 5.3 (right) would be adequate for the purposes of 6DOF accuracy evaluation with the parallel kinematics robot.

5.2.2 *Evaluation procedure*

The optical surface mapping system, HD AlignRT (VisionRT, London, UK; software version 5.0.1747) system was evaluated using a comprehensive 6DOF trajectory. This trajectory was composed of a set of points in a 3D cube about isocenter: a straight line in both the SI (superior-inferior) and LR (left-right) directions including ± 5 mm and ± 10 mm, and a line in the AP (anterior-posterior) direction including ± 4 mm and ± 8 mm. At each of these 12 points, and at isocenter, a series of seven angular motions were performed: 1 and 5 degrees



Figure 5.3: Experimental setup for the evaluation of the AlignRT system. A Styrofoam head was rigidly connected to the top platform of the robotic motion phantom (left), and a region of interest on the head, shown in purple across the forehead and nose bridge, was selected for tracking (right). The optical surface contour is tracked, and relate to tracking virtual point, which is located at room isocenter.

of rotation each for pitch (around LR), roll (around SI), and yaw (around AP), as well as the no-angle case. This 6DOF workspace was selected to cover a range of motions typically observed in intracranial SRS procedures, as well as other typical tumor trajectories. [52] As in the calibration procedure, the robotic motion phantom moved to each point in this trajectory and rested for 15 seconds to permit adequate averaging. A Styrofoam head was rigidly fixed to the top platform, and served as the surface region of interest (ROI) used for the purposes of motion tracking (Figure 5.3, left). The ROI was selected to cover a region spanning the forehead to the bridge of the nose, to provide a good balance of spatial and temporal resolution, while mimicking an ROI which could be clinically used for an open-face thermoplastic mask during brain disorder treatments. [48] The motion capture framerate was 4 frames/second with this ROI selection. This trajectory was performed twice under surveillance by the HD AlignRT system with room lights on and off in order to compare the effectiveness of 6DOF tracking with different ambient light conditions.

The measured output motion was recorded by the HD AlignRT system, and analyzed using Matlab (The Mathworks, 2013b). For each translational position and corresponding angular motion, a root-mean squared error (RMSE) analysis was performed for all six degrees

of freedom to evaluate both the translational and rotational precision within this workspace. To demonstrate the accuracy of the system, the 13 intended input points were compared directly to the calibrated measured points.

The same primary evaluation trajectories were also independently performed under the tracking of the IR camera system and an older generation of the AlignRT system. These results were analyzed in the same manner as above, considering both precision and accuracy, and compared quantitatively to the HD AlignRT system. In the case of the two AlignRT generations, the same collection software version was used to permit a fair comparison.

5.2.3 Accuracy and precision of patient tracking systems

In the AlignRT case, with the room lights ON, the RMSE analysis was performed on the appropriately transformed surface measurement data, compared to the intended motion determined by the 6DOF robotic motion phantom. At isocenter, RMSE was 0.29 mm in LR, 0.37 mm in SI, and 0.31 mm in AP, with rotational RMSE at 0.03 degrees in pitch (around LR), 0.06 degrees in roll (around SI), and 0.07 degrees in yaw (around AP). As the robotic motion phantom positioned itself distally along LR, SI, and AP independently, the RMSE analysis generally degraded, with RMSE for LR, SI, and AP reaching 0.46 mm, 0.49 mm, and 0.54 mm, respectively, for the ± 10 mm motions along the LR direction alone, RMSE of 0.34 mm, 0.52 mm, and 0.44 mm in LR, SI, and AP, respectively, for the ± 10 mm motions along the SI direction alone, and up to 0.46 mm, 0.61 mm, and 0.33 mm for the RMSE of LR, SI, and AP for the ± 8 mm motions along the AP direction alone (Table 5.1). Additionally, at points the eight extreme "corners" of the 6DOF workspace, where LR and SI were ± 10 mm and AP was ± 4 mm, the analysis showed RMSE values of up to 0.69 mm in LR, 0.75 mm in SI, and 0.62 mm in AP.

In a direct accuracy comparison between the intended input points and the transformed measured data points, the minimum translational errors occurred when no angular motions were performed: the average angle-free error was found to be 0.14 mm in LR, 0.27 mm in

	RMSE X	RMSE Y	RMSE Z	RMSE Pitch	RMSE Roll	RMSE Yaw	System
(0,0,0)	0.29	0.37	0.31	0.03	0.06	0.07	<i>HD AlignRT</i>
	0.32	0.20	0.33	0.21	0.09	0.04	<i>AlignRT</i>
	0.03	0.05	0.04	0.03	0.02	0.03	<i>Polaris</i>
(± 5 ,0,0)	0.31	0.31	0.45	0.06	0.07	0.07	<i>HD AlignRT</i>
	0.69	0.41	0.45	0.18	0.13	0.13	<i>AlignRT</i>
	0.05	0.05	0.05	0.03	0.04	0.04	<i>Polaris</i>
(± 10 ,0,0)	0.46	0.49	0.54	0.06	0.11	0.07	<i>HD AlignRT</i>
	0.51	0.62	0.46	0.22	0.17	0.18	<i>AlignRT</i>
	0.13	0.05	0.06	0.05	0.05	0.04	<i>Polaris</i>
(0, ± 5 ,0)	0.33	0.45	0.40	0.05	0.08	0.07	<i>HD AlignRT</i>
	0.42	0.31	0.40	0.21	0.13	0.12	<i>AlignRT</i>
	0.12	0.07	0.06	0.02	0.03	0.07	<i>Polaris</i>
(0, ± 10 ,0)	0.34	0.52	0.44	0.04	0.08	0.07	<i>HD AlignRT</i>
	0.46	0.51	0.47	0.28	0.17	0.16	<i>AlignRT</i>
	0.18	0.25	0.07	0.04	0.04	0.05	<i>Polaris</i>
(0,0, ± 5)	0.37	0.55	0.32	0.04	0.09	0.06	<i>HD AlignRT</i>
	0.48	0.65	0.39	0.21	0.19	0.14	<i>AlignRT</i>
	0.08	0.05	0.05	0.02	0.03	0.07	<i>Polaris</i>
(0,0, ± 10)	0.46	0.60	0.32	0.04	0.07	0.08	<i>HD AlignRT</i>
	0.50	0.74	0.36	0.23	0.18	0.15	<i>AlignRT</i>
	0.09	0.06	0.06	0.02	0.03	0.09	<i>Polaris</i>

Table 5.1: RMSE analysis for three optical tracking systems, for a set of points along the three primary axes in a 6DOF workspace, in mm for translations and degrees for rotations. Within each box, the RMSE values are displayed for the HD AlignRT system (top), the previous generation AlignRT system (middle), and the Polaris IR marker tracking system (bottom).

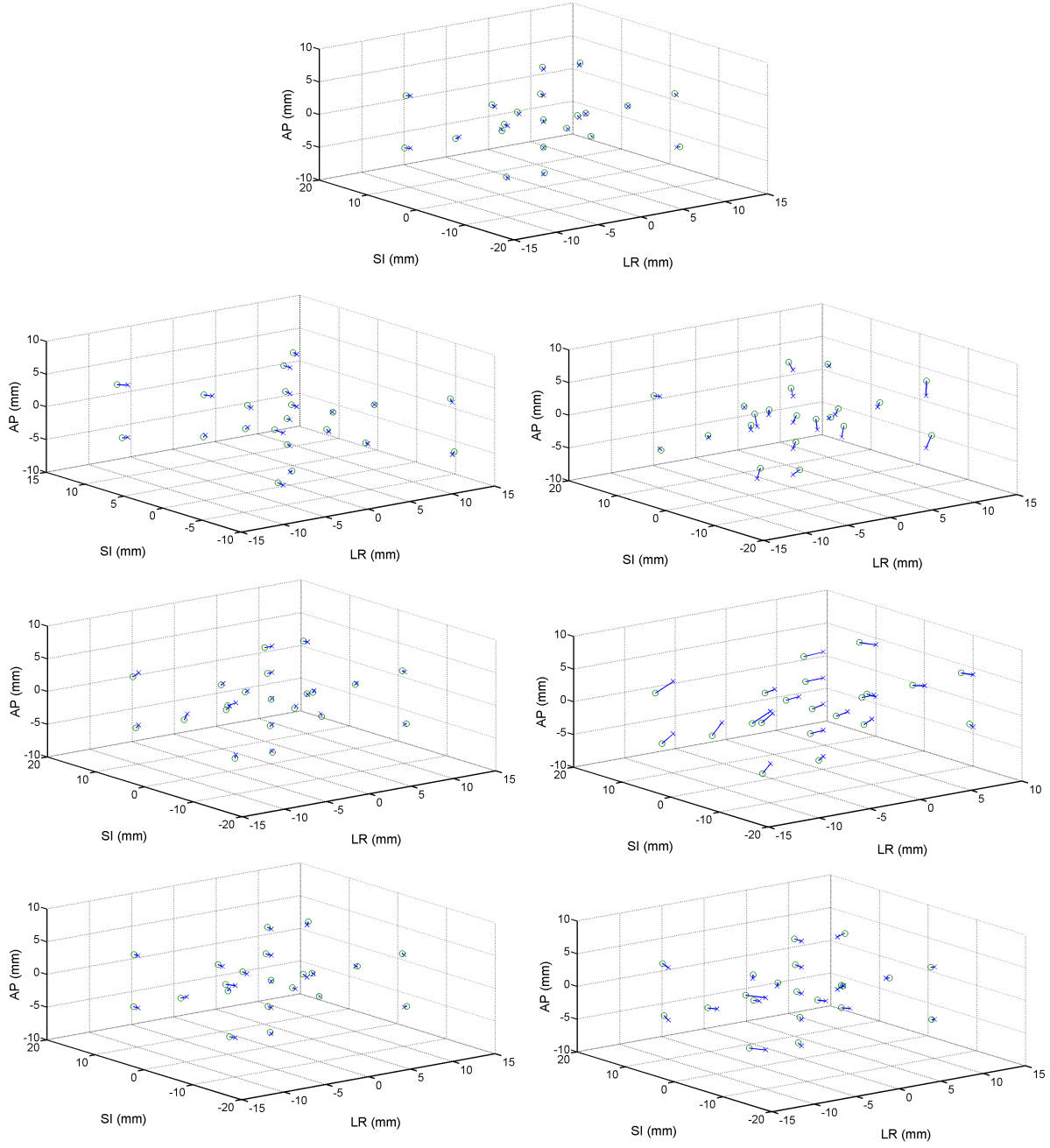


Figure 5.4: HD AlignRT comparison between intended (x) and measured (o) points in the 6DOF workspace, separated by angular motions performed at each of the 21 translational positions. No rotations (top), pitch (around LR) of 1 degree (top left) and 5 degrees (top right), roll (around SI) of 1 degree (middle left) and 5 degrees (middle right), and yaw (around AP) of 1 degree (bottom left) and 5 degrees (bottom right).

SI, and 0.15 mm in AP. When angular motions in the workspace were confined to 1 degree, these magnitudes increased slightly, to 0.24 degrees in LR when 1 degree roll was performed, 0.51 mm in SI when a 1 degree yaw was performed, and 0.27 mm in AP when a 1 degree roll was performed. However, once a 5 degree rotation was performed, in any orientation, the average discrepancies between the input and transformed measured data reached up to 1.1 mm in LR, 1 mm in SI, and 1 mm in AP. In Figure 5.4, the intended translational points are compared directly in 3D space to the transformed measured data, separated by the angular motions performed at each of these 13 points in the workspace.

Across this entire 6DOF workspace, analysis on the angular data produced average RMS errors of less than 0.1 degrees consistently, even at extreme translations. This analysis considered rotations of up to 5 degrees in pitch, roll, and yaw. For all positions and rotations, the maximal angular discrepancy was less than 0.3 degrees for this case with room lights ON. Important to note is that the largest observed angular RMSEs occurred for roll, around the SI axis, and the smallest angular error occurred for pitch, around the LR axis (Table 5.1).

Analysis was also performed for the HD AlignRT system when the treatment room lights were OFF, and the results were similar but slightly worse in most translational axes and rotational directions. At isocenter, the RMSE was 0.38 mm for LR, 0.71 mm for SI, 0.23 mm for AP, 0.12 degrees for pitch, 0.10 degrees for roll, and 0.08 degrees for yaw. These RMSE values reached up to 0.70 mm in LR, 1.62 mm in SI, and 0.66 mm in AP across the same evaluation trajectory as described above. Compared to the lights ON case, the angular results were slightly worse in almost all cases, excluding when a 5 degree roll was performed.

The IR marker system produced similar motion tracking results to the HD AlignRT system. The analysis for this case yielded maximal translational RMSE of 0.18 mm and maximal angular RMSE of 0.10 degrees. In a direct comparison between the tracking efficacies of the two systems, the HD AlignRT system had slightly improved angular resolution but worse spatial resolution, compared to the Polaris IR marker tracking system (Table 5.1).

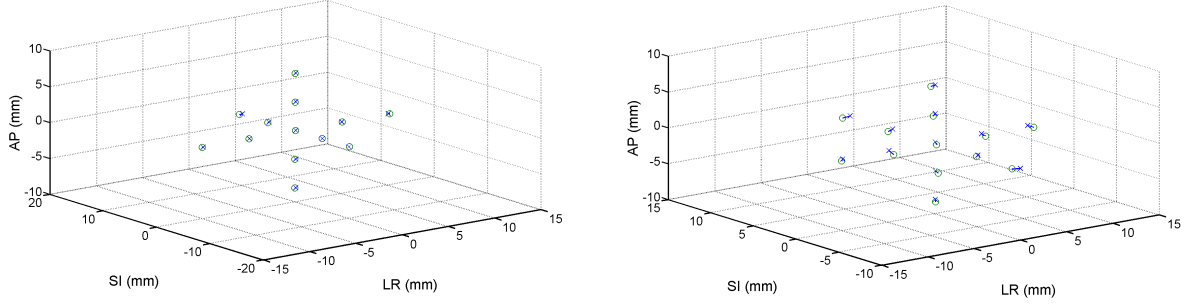


Figure 5.5: Direct comparison of the accuracy of the IR marker tracking system (left) and the older generation of the optical surface mapping system (right). Both plots show the intended translational position (x) compared to the transformed measured data (o) in the case when no angular motions were performed at each of the points in the 6DOF workspace.

Examining the true translational difference between the intended and measurement data, as a measure of camera precision, the Polaris system showed better translational tracking compared to the AlignRT systems (Figure 5.5, left).

Comparing the newer HD AlignRT system to the older AlignRT system showed improved tracking accuracy for the newer system. For the older generation AlignRT system, RMSE analysis found LR errors to range from 0.32 mm at isocenter and up to 0.69 mm at the most extreme translational movements, occurring with 5 degree rotations; SI errors ranged from 0.20 mm at and up to 0.74 mm at the more distal positions in the 6DOF workspace, when large angular motions were also performed; AP errors ranged from 0.33 mm at isocenter and up to 0.47 mm at the most extreme translational motions with large rotations (Table 5.1). Angular RMSEs were similar to that of the HD AlignRT system, with errors ranging from 0.21 degrees, 0.09 degrees, and 0.04 degrees at isocenter and up to 0.28 degrees, 0.19 degrees, and 0.18 degrees in pitch, roll, and yaw, respectively, at locations more distal to isocenter. A direct comparison between the intended position and the appropriately transformed measured position for the rotation-free case is presented in Figure 5.5 (right).

This work presents a novel method for validating the accuracy of external tracking systems for use in radiation therapy. The method is not inherently limited to external optical tracking systems, and may be used to evaluate the performance of any 6DOF patient

tracking system, such as stereoscopic x-ray imaging or CBCT. With increased use of 6DOF patient motion tracking devices in radiotherapy, this method provides a rigorous technique for performing QA in order to guarantee that such devices are operating within designed specifications. The methodology for camera system calibration and evaluation presented in this work has the potential to serve as a foundation for such an evaluation. The calibration methodology will be applied initially, and then using a robotic device as the motion control phantom, one can perform an exhaustive trajectory covering a desired 6DOF workspace, such as the one presented earlier in this work.

In a direct 6DOF comparison between the AlignRT systems and the Polaris IR marker tracking system, it was found that both 3D optical mapping systems had worse translational tracking capabilities and similar rotational errors to the latter, based on the quantitative RMSE results for precision and the accuracy results comparing the intended and measured outputs. The reason behind the difference between the IR and 3D surface map paradigms is potentially in the fundamental nature of how these systems track points during treatment. The virtual point coordinate system characterized by the reflective IR marker set may be more precisely defined by the careful marker configuration, compared to the array of approximately 500 vertex points. [48]

Compared to the previous generation AlignRT, the HD AlignRT system performed better in most cases. Given the overall end-to-end clinical standards of 1-2 mm accuracy for highly conformal cases such as SRS, it is best to maintain sub-millimeter camera accuracy such that its contribution to total error is small. [82] In most cases the HD AlignRT system was able to meet this clinical requirement, and was able to meet this requirement when motions were limited to 5 mm and 1 degree. This subset of points is more typical for motions observed during frameless SRS treatments. [52] For a highly conformal treatment modality such as SRS, failure to achieve submillimeter accuracy can result in improper dose coverage and significant normal tissue irradiation, due to the high-dose low-fraction characteristics of this technique. However, in several cases, both the HD and older optical surface mapping systems

reached a threshold of angular motion, beyond which the system was unable to effectively track the ROI contour, in either translations or rotations. This contributed substantially to limiting the accuracy and precision of the device, especially in the cases when RMS errors exceeded 1 mm.

As shown in Figure 5.4, when the phantom positioned itself further from isocenter, the viable 6DOF tracking workspace decreased noticeably; that is, as the phantom moved distally from isocenter, the threshold angular motion beyond which the HD AlignRT camera system could not effectively identify true motions decreased, when rotations were performed in pitch (around LR) and roll (around SI). For this HD AlignRT system, within the presented workspace, however, no such failures were observed before rotations in pitch and roll reached approximately 3 degrees, and none of these issues were present when yaw (around AP) rotations were performed, up to 10 degrees. In the older AlignRT system, the viable rotational workspace was similar but smaller still, and yaw rotations at the 10 degree mark experienced such failures. In the 6DOF volume examined, the Polaris IR camera did not observe issues of this nature. Such failure-to-track events are particularly concerning in a highly conformal case such as SRS, and monitoring of the clinical targets will still be required when observed motions begin to deviate over the course of treatment. However, under the sub-millimeter and sub-degree patient positioning conditions for SRS, deviations from isocenter over 1 mm or 1 degree, would not be observed clinically, and within this 6DOF workspace subset, no such failure-to-track errors were observed. Still, such events motivates the necessity of clinical oversight when using these systems.

A quantitative and qualitative comparison between the room lights ON and OFF cases was also performed, and in most cases, tracking accuracy was improved when the room lights were ON, which is the condition typically used with patients. Because of the nature of the 3D optical tracking paradigm, we explored the potential for ambient light to affect tracking accuracy. We found that the two conditions yielded similar results, while the room lights ON case, which is more common clinically, yielded slightly better tracking performance. In

addition to this examination, future work aims to evaluate the dynamic capabilities of 6DOF systems.

This section showed how the combined application of an effective calibration methodology and a precise 6DOF robotic motion phantom permitted the complete evaluation of three optical tracking systems. Quality assurance using this procedure offers the advantage of off-center evaluation, and complete examination of tracking capabilities in 6DOF. This QA procedure can be set up quickly and is highly automatic, and can therefore be used as a daily QA, or potentially immediately before a procedure. This exhaustive evaluation of the Polaris IR marker tracking system, and two AlignRT systems, was performed across a clinically-relevant 6DOF workspace, and the accuracy of the tracking was found to be within clinical tolerances for intracranial SRS in most cases. Based on this level of translational and rotational accuracy, effective motion monitoring during such highly conformal procedures may be possible at the sub-millimeter and sub-degree level with the systems explored in this work.

5.3 Collection and analysis of patient head motion using optical surface mapping

5.3.1 Clinical procedure and motion collection

The optical surface mapping system AlignRT (VisionRT, London, UK) projects an infrared speckle map onto a contour using a set of two stereoscopic camera "pods", creating a set of vertex points on the surface, which define a relationship between the surface and a coordinate frame, defined at treatment isocenter. This system is clinically used for monitoring intrafraction patient motion of several sites, including the head and breast, during radiotherapy. GateRT provides beam OFF and ON gating control for the AlignRT system, and determines when the patient's motion exceeds preset thresholds. The tracking accuracy of this system was evaluated in Section 5.2, and was found to be capable of full 6DOF tracking

at the sub-millimeter and sub-degree level when the contour remained within the camera system’s 6DOF workspace.

In this section, we make use of the AlignRT system for monitoring brain cancer patients while they undergo whole brain radiotherapy (WBRT) treatment. An IRB was secured to permit monitoring of cranial motion during their treatment for eight patients. During this treatment, patients were immobilized using an open-face thermoplastic mask. For WBRT, this level of immobilization exceeds clinical requirements due to the palliative nature of many whole brain treatments, but was chosen because of its similarity to frameless stereotactic radiosurgery (SRS) immobilization techniques.

Cranial motion was monitored using the AlignRT system, with the tracking region of interest (ROI) spanning the semi-rigid region over the eyes, forehead and nose bridge. The objective was to monitor patient cranial motion only during beam-on time by excluding patient setup and pre-treatment positioning. Translations and rotations were tracked throughout treatment, in reference to AlignRT calibrated isocenter. Eight patients were tracked across three treatment rooms, and tracking across both the HD AlignRT and older generation AlignRT systems, for a total of 58 usable full trajectories obtained during the course of monitoring. At the time of initial analysis, it appeared as though the 58 trajectories also included beam OFF times. As a result, we segmented the acquired trajectories into regions which we believed to only include beam ON times, yielding 119 beam ON-exclusive trajectories for analysis.

5.3.2 Analysis of patient head motion measurements

Quantitative analysis of the patient head motion was conducted using Matlab (Mathworks, 2013b). As mentioned earlier, trajectories were first sub-divided based on visual inspection and smoothed to remove inherent camera noise. The degree of smoothing was based on the camera frequency. Shown in Figure 5.6 is an example motion for one of the patients monitored in this study. Near the middle of the tracking period, a large jump in several

axes is observed. This deviation was determined to be an issue with camera calibration, and occurred when the gantry rotated during treatment, blocking one of the AlignRT camera pods. To monitor the surface effectively, the AlignRT system requires the surface contour ROI to be within the field of view of two camera pods. When the AlignRT system registers a camera blockage for one of the three pods in the treatment room, it rapidly switches to monitor the patient with the other available camera pod. The three cameras are placed such that the rotation of the gantry will only limit the view of one pod at any given time, so there would always be at least two cameras with line of sight on the patient's surface. However, if one or more of the cameras are not calibrated in such a way to match the others, switching between cameras after a blockage has occurred may disrupt the tracking and measure motions which do not reflect the physical motions performed by the patient undergoing radiotherapy. The timing of the jumps shown in Figure 5.6, as well as several other patient motion cases acquired in this study, corresponded to the LINAC gantry rotating during treatment. Consequently, analysis of patient motion for the remainder of this study involved visually segmenting the measurements from each session based upon these camera blockage-produced jumps. For example, in the case shown in Figure 5.6, analysis was performed on two segments, spanning 0-32 seconds and 45-82 seconds.

A cumulative distribution function representing the results from the set of patients tracked is presented in Figure 5.7. In all but one case, all motions were within 4 mm total translation during the course of beam ON motion tracking. Additionally, in almost all cases, only 10% of the time was translational motion greater than 0.75 mm. In terms of average motion across all patients, the mean in X (LR), Y (SI), Z (AP) were 0.01 mm, 0.21 mm, and 0.05 mm, respectively, with an overall variance of 0.26 mm translationally. These results from patient motion using an open-face thermoplastic mask have been used to inform the design of a second generation parallel kinematics robot, which we aim to use in a clinical setting for WBRT treatments and ultimately SRS therapy.

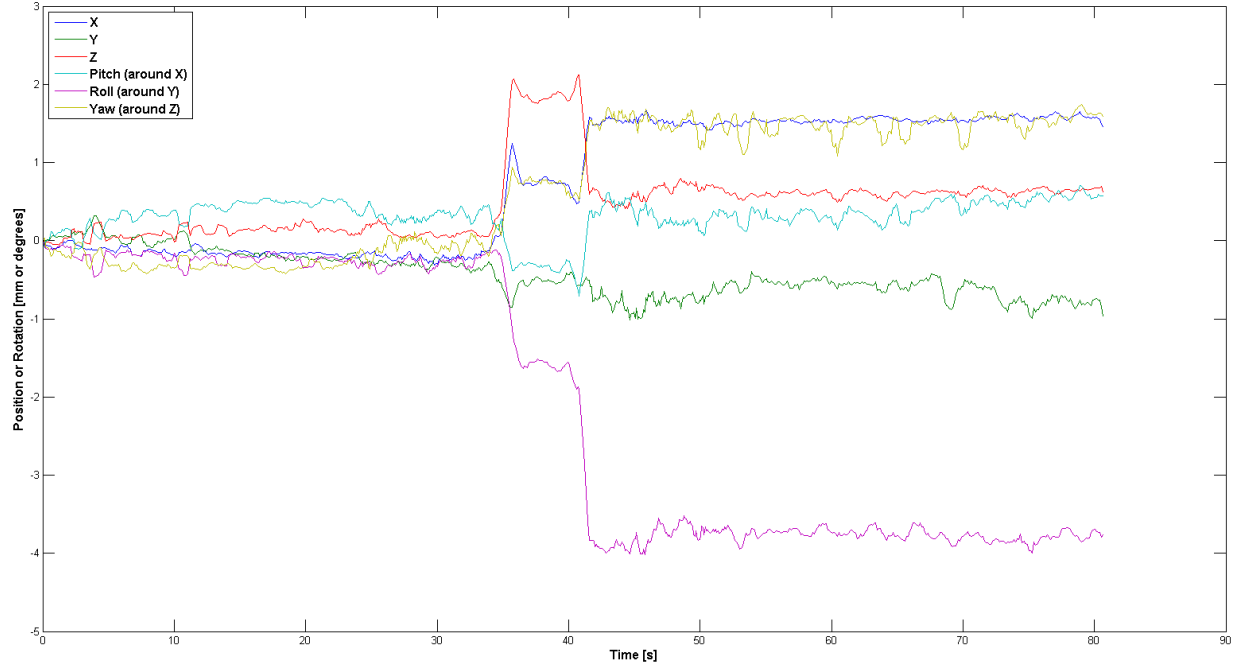


Figure 5.6: Example cranial motion for one of the eight patients monitored under the IRB protocol.

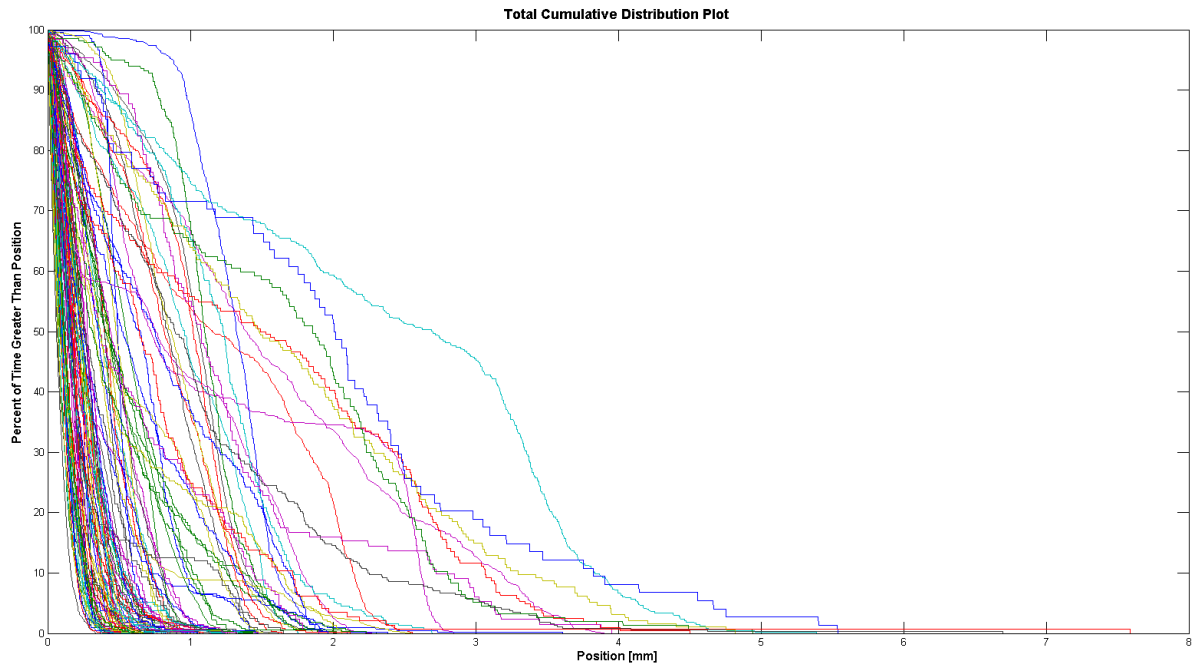


Figure 5.7: Cumulative distribution of all patient cranial motion, comparing the percent of time the each patient was outside the translational threshold indicated on the horizontal axis. Patients undergoing WBRT were monitored with the AlignRT system while immobilized using an open-face thermoplastic mask. Each tracking session is presented as its own line in the figure.

CHAPTER 6

DEVELOPMENT OF A MOTION COMPENSATION ALGORITHM

6.1 Introduction

Stereotactic radiosurgery can be used to treat intracranial lesions with high target coverage conformality, and two approaches to SRS are used clinically. Frame-based SRS uses a metal head ring rigidly fixed to the patient's skull and bolted to the treatment couch to provide strict mechanical immobilization. This approach minimizes patient head motion, but is uncomfortable to the patient and can hinder the clinical workflow. In frameless SRS, the metal frame is removed and replaced with a thermoplastic mask for head immobilization, and head motion is monitored during treatment using external surrogate tracking systems and kV imaging. Although patient comfort and clinical workflow are improved, setup time can still be substantial and systematic drifts of several millimeter have been reported. [60, 58] As a result, treatment interruption for intrafractional patient realignment can occur to ensure proper target coverage. Given these limitations of both the frame-based and frameless SRS approaches, there is a need to develop an improved methodology.

Our solution uses the parallel kinematics 6DOF robot, described in Chapters 2 and 3, for the purposes of performing real-time head motion compensation during SRS. The robot was previously shown to achieve the control capabilities necessary for the high-accuracy application in frameless and maskless SRS. [70] In this chapter, we describe the control algorithm which will be used with the 6DOF robot for the purpose of stabilizing human head motion during SRS treatments. We first describe the development of a robotic stage with control in 4DOF (3D translational and one rotational axis) as an initial step toward full 6DOF motion control due to similarities in control implementation. A 6DOF iterative closest-point algorithm was then developed, and the control scheme is described in this chapter. Finally, we explore an example scenario for the motion compensation algorithm,

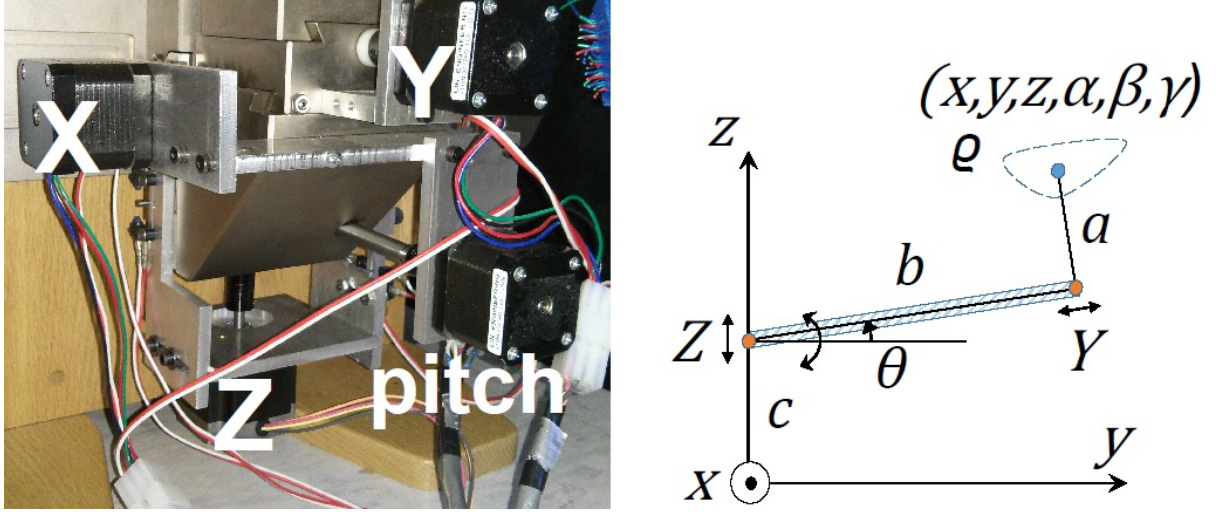


Figure 6.1: Left: 4DOF stage. Motors controlling the motion is all 4 axes are also shown. Right: The $Y - Z$ plane of 4DOF stage. A target was fixed on the top of the 4D stage, which position was specified by the 6DOF coordinates $(X, Y, Z, \alpha, \beta, \gamma)$. Tuning θ not only changes the pitch angle of the target, but also changes the Y and Z coordinates of its center ρ .

and run through several iterations of the control response.

6.2 Architecture of 4DOF motion compensation stage control

The development of a 6DOF algorithm for cranial motion compensation in frameless and maskless SRS began with advancements made on a 4DOF motion stage. In this section, we discuss research of a 4DOF robotic stage for real-time head stabilization for the purposes of using the 4DOF architecture as a step toward 6DOF algorithm development. The 4DOF stage was constructed by modifying a 3D robotic stage to include control in the pitch axis (α , about X (LR)) of approximately 2 degrees. The 4DOF stage is shown in Figure 6.1, left. Four stepper motors were used to govern the translation of the left-right (X), superior-inferior (Y), anterior-posterior (Z) axes, and the rotation of the pitch (about X axis) around a pivot point at the base of the platform. NEMA type 17 motors were used for X and Y axes, while a larger NEMA type 23 was used for z and pitch axes due to the weight of the patient's head.

Control of the 4DOF stage is first built upon the setup geometry, shown in Figure 6.1, right. Here, the position of a target can be specified by a translational vector $l = \{X, Y, Z\}$ and a rotational vector $\Psi = \{\alpha, \beta, \gamma\}$ (Euler angles: pitch α , roll β , yaw γ). When there is no relative motion between the target and the IR markers (rigid body), the target position can be calculated through the position of the IR markers. These two vectors can be represented in different coordinate frames. There are three coordinate frames in the 4DOF motion correction system: the camera frame, the normal stage frame, and the linear accelerator (LINAC) frame. The normal stage frame is a coordinate frame with its X, Y and Z axes parallel to the X, Y and Z motor axes, respectively, when all X, Y, Z and pitch θ motor axes are in their initial zero position. The LINAC frame is the linear accelerator reference frame with the origin at the radiation isocenter. For control purposes, the target position was represented in the normal stage frame, allowing easy computation of control, while for display and radiation delivery purposes, the target position was represented in the LINAC frame. Transformations between these frames involve a rotation matrix and a translational vector, which can be computed by Kabsch's algorithm, a least-squares fitting algorithm based on singular value decomposition. [67] More generally, this approach is a constrained orthogonal Procrustes problem.

Four stepper motors were used to control the translational position and pitch angle of the target. This is a multiple input and multiple output control system, where the rotation of the pitch motor will not only change the pitch angle of a target, but will also change the Y and Z coordinates making control design difficult. For example, consider a target with its center at ϱ as shown in Figure 6.1, right. Here, the target is rigidly fixated to the top of 4DOF stage, and its translational and pitch coordinates in the normal stage frame are $(x_\varrho, y_\varrho, z_\varrho, \alpha_\varrho)$

$$\begin{aligned}
x_{\varrho} &= X \\
y_{\varrho} &= (b + Y) \cos \theta - a \sin \theta \\
z_{\varrho} &= (b + Y) \sin \theta + a \cos \theta + c + Z \\
\alpha_{\varrho} &= \theta
\end{aligned} \tag{6.1}$$

where X , Y , Z , and θ are the displacements of the X , Y , Z , and pitch θ motor axes from their normal positions, respectively, and a and b are the variables specifying the position of ϱ with respect to the pivot of the pitch θ motor axis. It can be seen from Equation 6.1, changing θ will change y_{ϱ} , z_{ϱ} , α_{ϱ} . The coordinate x_{ϱ} is not coupled with θ , since the rotation axis of the pitch θ motor is parallel to the x axis. To control such a system, we first consider linear approximation of Equation 6.1 when θ is small. Let the normal position $(x_0, y_0, z_0, \alpha_0)$ of the center ϱ be the position where $X = Y = Z = \theta = 0$, i.e., $x_0 = 0$, $y_0 = b$, $z_0 = a + c$ and $\alpha_0 = 0$. Thus, if θ is small, the translational and pitch displacements of ϱ in the normal stage frame, $x_{\varrho} - x_0$, $y_{\varrho} - y_0$, $z_{\varrho} - z_0$, and $\alpha_{\varrho} - \alpha_0$ can be linearly approximated by

$$\begin{aligned}
x &= X \\
y &= Y - a\theta \\
z &= Z + b\theta \\
\alpha &= \theta
\end{aligned} \tag{6.2}$$

We then consider two feedback control configurations, with and without feed-forward. Figure 6.2 shows the signal flow diagram of feedback control on the displacement y , where y_r is the desired position, e_y is the error, u_y is the controller output, K_y is the feedback controller and H is the sensor. The effect of θ on the control of y can be viewed as disturbance as shown in the red channel 1 arrow in Figure 6.2, top. When feedback control without feed-

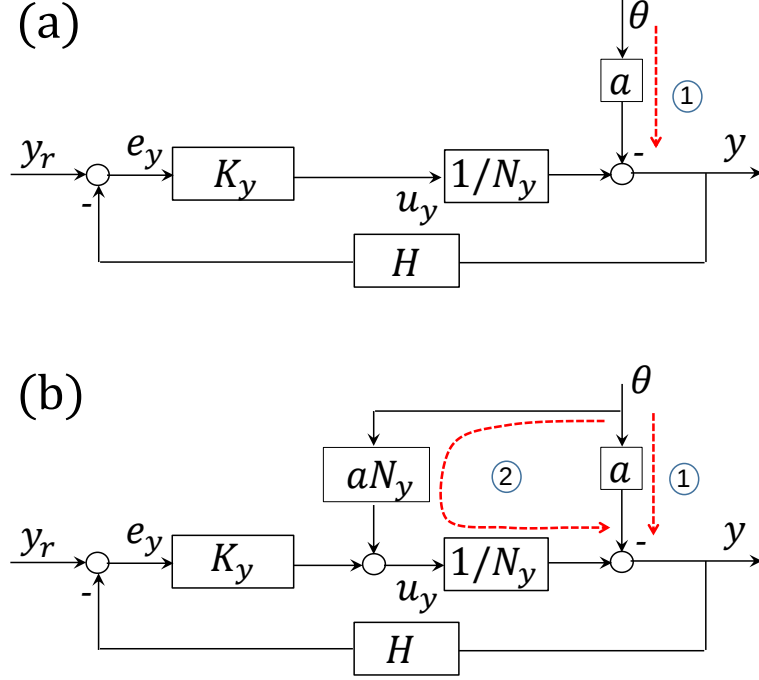


Figure 6.2: (a) Feedback control without feed-forward. The system responds after the effect of θ appears in y ; the effect is expressed as a red arrow labeled channel 1. (b) Feedback control with feed-forward. The system takes action before θ has effect on y , labeled channel 2.

forward is applied, the controller responds only after the effect of the disturbance shows up in the system output, and will finally cancel out the effect of disturbance after certain time. On the other hand, in the configuration with feed-forward (Figure 6.2, bottom), a forward channel labeled 2 is added, which means the disturbance is measured and responded to before it has time to affect the system. Whenever the pitch θ axis motor moves n steps, the y axis motor is controlled to move additional $N_y n a / N_\theta$ steps simultaneously, then the effect of θ on the displacement y will be entirely eliminated. This can also be seen in Figure 6.2 by the fact that following the two channels, their sum is equal to zero. Thus, in general, the feedback control with feed-forward will lead to more optimal correction and have better control performance. The displacement z can be controlled in the same way, and the method can also be extended to other coupled axis such as yaw and roll. Note that the parameters a and b as in Equation 6.2 and Figure 6.1 (right) correspond to the relative position between the target and the normal stage frame.

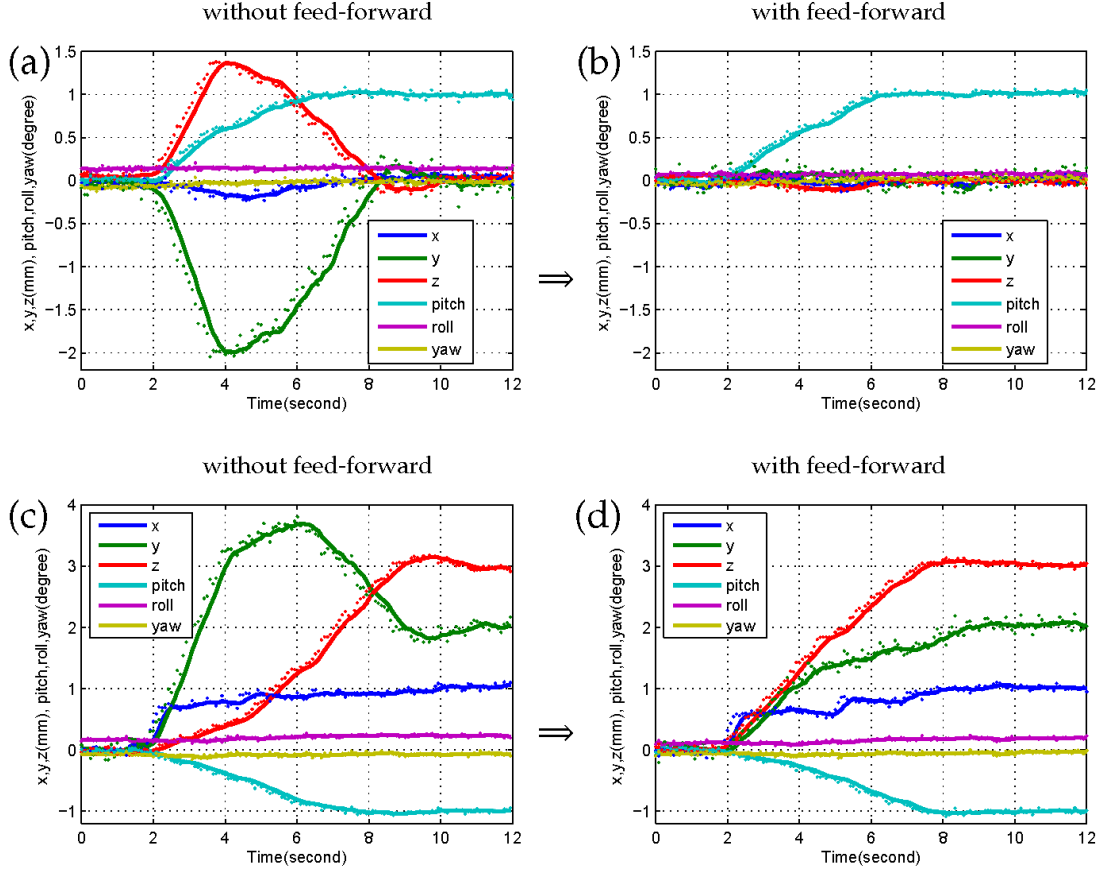


Figure 6.3: Comparison of feedback control without and with feed-forward. For all four cases, the target started at the zero position ($x = y = z = 0$ mm, $\theta = 0$ degrees), then controllers were engaged to move the target to the desired positions ($x=y=z=0$ mm, $\theta = 1$ degree) in (a and b), and ($x = 1$, $y = 2$, $z = 3$ mm, $\theta = -1$ degree) in (c and d). In (a) without feed-forward, when the pitch was tuned from 0 to 1 degree, y and z were away from the desired position at first, and then came back, while in (b) with feed-forward, the target almost remained in the desired translational position all the time when the pitch angle was adjusted. Without feed-forward, there is 80% overshoot on y in (c), while there is no overshoot and faster response with feed-forward (d).

By including a feed-forward algorithm in the 4DOF control, we were able to improve the response time of the 4DOF robotic stage and prevent the Y and Z axes from moving undesirably, as shown in Figure 6.3. The feed-forward approach to 4DOF stage control was an important step toward using robotics to perform real-time motion compensation for frameless maskless SRS. However, while the dominant axes for cranial motion are often LR, SI, AP, and pitch, rotations can occur in full 6DOF. Particularly in single-isocenter multi-target SRS protocols or distinctly non-spherical lesions can roll and yaw also impact target dose coverage and overall treatment accuracy. The framework for motion control in 4DOF helped to form the initial architecture of 6DOF motion control.

6.3 Design of a 6DOF motion compensation algorithm

The algorithm to control the robot for the purposes of real-time adaptive head motion compensation in full 6DOF is an essential step toward performing stabilization in the frameless maskless SRS regime. Building on the architecture of the 4D stage discussed previously, a novel 6DOF algorithm for use with the parallel kinematics robot was developed. Many versions of the control formulation were tested with the robotic system, but ultimately an algorithm was developed based on an iterative closest path approach. This technique makes use of the robot by responding to 6DOF errors detected in real time by an external camera tracking system. The algorithm uses the measured cranial motion and rapidly calculates a trajectory iteratively for each linear actuator in order to cancel the head movements translationally and rotationally. This iterative approach was necessary to correct for cranial motion in this non-rigid body coupling between the robot and the patient’s head. The complexities from rigidity are discussed in greater detail later in this section. The control algorithm used for motion correction is outlined in the control diagram in Figure 6.4 and in Algorithm 1.

This compensation system automatically transforms the camera-measured 6DOF cranial motion, determines the errors in each axis, and computes the linear actuator trajectories necessary to cancel the observed motion of the target position. Specifically, the camera

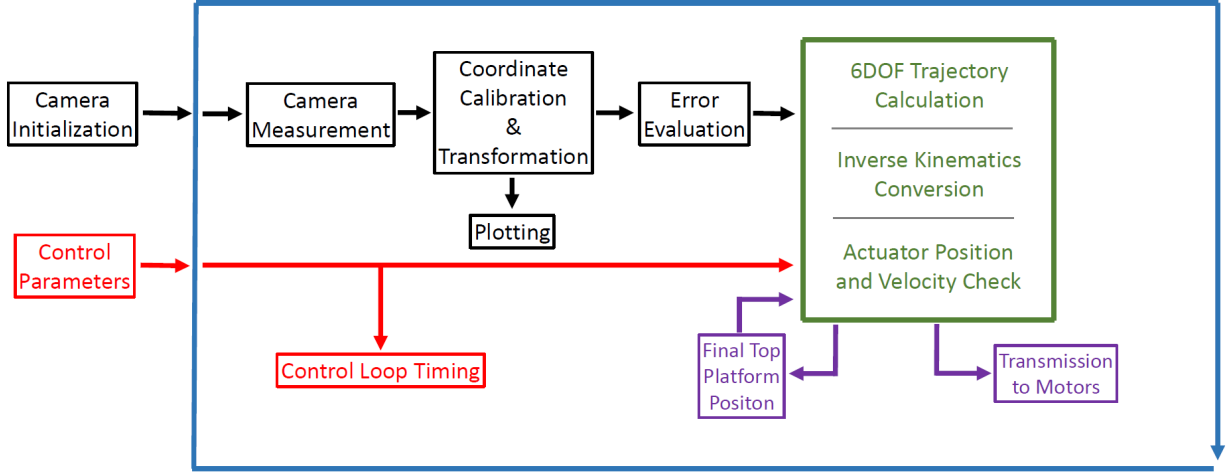


Figure 6.4: Control diagram for motion compensation. Camera inputs are automatically transformed using the calibration method, and 6DOF errors are determined. A cancellation trajectory is calculated, and the six linear actuator motions to accomplish the cranial compensation are transmitted to the motors. The 6DOF errors continue to be updated 12 times per second. The blue robotic control loop’s behavior is governed by the camera system inputs and measurement transformations (black), and the user’s parameter inputs (red). The inverse kinematics calculation (green) uses these inputs to determine the necessary linear actuator cancellation trajectories. These necessary movements are finally communicated to the linear actuator’s motors (purple), which are then recorded and used for subsequent loop iterations.

measurements track the translational and rotational movements a set of four reflective infrared (IR) markers, which are fixed to the forehead of the human subject and which define a coordinate frame within the individual’s head. In clinical practice, this coordinate frame, called the virtual point, would be positioned at the treatment isocenter. The coordinate system used for the automatic transformations, and upon which the algorithm’s control was built, was discussed in the earlier treatment of 6DOF calibration in Chapter 4. Specifically, the formalism described in Figure 4.1 is used for the theoretical development of the motion compensation algorithm.

Exploring the control design shown in Figure 6.4 and Algorithm 1 more closely, the first elements of the algorithm are initializing the camera system, and determining user input parameters to tune the algorithm. Although the Polaris (Northern Digital, Inc., Waterloo, ON, Canada) IR marker tracking camera was used for the majority of this project, the

camera can be any motion tracking system with real-time 6DOF output capabilities. The input parameters alter some aspects of the control algorithm, such as the degree to which it segments the head positioning error for overshoot prevention and the 6DOF threshold for robotic intervention.

After initializing, the motion stabilization is engaged, and the robot begins using the real-time 6DOF camera measurements to keep the head stable, keeping the virtual point at its initial position. For the purposes of clinical implementation, this corresponds to keeping the target's pre-defined coordinate frame at treatment isocenter. The blue control loop shown in Figure 6.4 is repeated for each new camera measurement; for the Polaris IR camera, this corresponds to 12 times per second, due to its 12 Hz refresh rate. The raw camera measurements are recorded, and the 6DOF error is evaluated. Because the initialization procedure zeroed the virtual point coordinate frame to the starting position, the 6DOF error is effectively the translational (X (LR), Y (SI), and Z (AP)) and rotational (α (about X), β (about Y), and γ (about Z)) real-time measurement from the camera.

When the measured cranial motion exceeds a threshold of 0.25 mm or 0.25 degrees, which was chosen to reduce response to noise from the Polaris camera, the robotic system is engaged and the necessary cancellation trajectory is performed automatically. First, the 6DOF head error is transformed using the calibration trajectory discussed in Chapter 4 into a shifted robot coordinate frame $\mathbf{H}'$ based on the measurement data for the shifted virtual point frame $\mathbf{M}'$. By performing this transformation, the cancellation trajectory can be more easily computed. Essentially, the transformed error evaluation determines the 6DOF difference between the $\mathbf{H}$ frame and the $\mathbf{H}'$ frame, assuming a rigid connection between the virtual point and robot coordinate frames. This $\mathbf{H}'$ corresponds to a 6DOF position of the robot if the virtual point shift from $\mathbf{M} \rightarrow \mathbf{M}'$ was performed by the robot rather than the undesired intrafraction human head motion. The motion compensation algorithm then uses the 6DOF difference between the artificial $\mathbf{H}'$ and true $\mathbf{H}$ robot coordinate frames to compute a head stabilization trajectory, which would return $\mathbf{M}' \rightarrow \mathbf{M}$.

An important and potentially subtle point is the assumption of rigidity between the $\mathbf{M}$ and $\mathbf{H}$ coordinate frames. In an ideal scenario with perfect rigidity, the virtual point would not shift relative to the robot, and because the robot is stable, there would be no head motion to stabilize. However, slippage between the robot and the patient’s head can occur, resulting in measured head positioning errors. If those errors exceed the 0.25 mm and 0.25 degree control thresholds, the compensation algorithm attempts to return the patient’s head to its starting position. For the robot to perform this action, perfect rigidity is once again assumed to exist between the robot and the virtual point. Although this assumption is not strictly correct, for a robotic cancellation trajectory to be generated, it is a helpful to assume during calculation that when the robot moves, the patient’s head also moves. Again, because of the non-rigid coupling between the 6DOF robot and the head, the patient does not perfectly respond to the robot’s intervention for stabilization.

Our solution to complications caused by the degradation of the rigidity assumption during control was to develop a cancellation trajectory based on an iterative closest-point algorithm. The iterative algorithm takes as an input the 6DOF error between the $\mathbf{H}'$ and $\mathbf{H}$ coordinate frames, and segments the error into several parts. The cancellation trajectory is calculated to move the robot $\frac{1}{n^{th}}$ of the way to its final position at coordinate frame $\mathbf{H}$, for n segments. This action is performed for every loop iteration (Figure 6.4), defined by the frame rate of the 6DOF tracking camera. After the segmented correction is performed, a new camera measurement is acquired and the 6DOF error is registered; if robotic intervention is deemed necessary based upon the thresholding, then a new segmented trajectory is computed, and the compensation is performed automatically. This approach does not rely as strictly on the assumption of rigidity during head stabilization to perform effective control, because the robot is attempting to correct the head positioning error quickly and in smaller steps. This iterative approach rapidly stabilizes the patient and also helps minimize the effects of overshoot, in which the robot over-responds to measured cranial motion when the compensation trajectory requires a large response.

Algorithm 1 6DOF Control Algorithm

```
1: procedure INITIALIZATION
2:   I. Initialize tracking camera and control hardware
3:   II. Establish control parameters
4: procedure CALIBRATION
5:   I. Run calibration procedure
6:   II. Update transformations
7: procedure MOTION COMPENSATION
8:   loop tracking camera records motion
9:     I. Transform measurement data based on calibration
10:    II. Evaluate target error in 6DOF
11:
12:    if  $ERROR_{XYZ} > 0.25\text{ mm}$  or  $ERROR_{\alpha\beta\gamma} > 0.25\text{ degrees}$  then
13:      engage motors for compensation ( $comp = true$ )
14:    else iterate & continue tracking
15:
16:    if  $comp = true$  then
17:      1. transform target error into robot control frame error:
18:         $\delta_{MM'} \rightarrow \delta_{HH'}$ 
19:         $R_{MM'} \rightarrow R_{HH'}$ 
20:      2. divide the errors  $\delta_{HH'}$  and  $R_{HH'}$  into  $n$  segments
21:      3. calculate response trajectory for a  $\frac{1}{n}$  error response:
22:        determine trajectory for  $\mathbf{H}' \rightarrow \mathbf{H}$ 
23:        transform the  $\mathbf{H}' \rightarrow \mathbf{H}$  trajectory into six linear actuator trajectories
24:      4. validate the response motion does not exceed robot capabilities:
25:        i. 6DOF workspace
26:        ii. linear actuator velocity
27:      5. transform linear actuator trajectories into stepper motor steps
28:      6. record final position of top platform for next iteration
29:
30:    III. Write recorded measurements and response trajectory to text file
```

Returning to Figure 6.4, the iterative compensation trajectory is calculated in the green box, which also makes use of the inverse kinematics formalism developed in Chapter 2 to determine the six linear actuator trajectories needed to produce the robot’s top platform 6DOF response trajectory. These six motions are transmitted to their respective linear actuators’ motors after a safety check is performed. This check evaluates the velocity and absolute position of each linear actuator required to perform the compensation trajectory within the timing of the control loop. If the measured patient motion changed by a large magnitude rapidly, the calculated response trajectory of the robot may exceed the design limitations; alternatively, if the patient’s systematic drift over the course of tracking is large, the continued motion compensation response may demand that the robot move outside its pre-determined 6DOF workspace. In these cases, the robot will no longer be able to effectively perform head motion stabilization. For the former situation, the linear actuator velocities which would exceed design specifications, the maximum permissible velocity is used instead, based on the experimental evaluation described in Chapter 3 (Table 3.1). In the latter case, the robot becomes unable to perform real-time head motion stabilization because the required motion is outside its functional workspace, and a warning is presented to the user.

After these checks are performed, the six linear actuators execute their corresponding trajectories, and the resultant top platform position is recorded. This book-keeping is an important step in iterative control, because the inverse kinematics formalism which governs the trajectory calculation uses a starting position and an ending position for the absolute 6DOF motion. In this way, for each loop iteration, the algorithm develops a response trajectory based upon the robot’s 6DOF top platform position of the previous iteration.

An example scenario for which the motion compensation algorithm can be used is presented in Algorithm 2. This example demonstrates four iterations in the control algorithm for the case when errors are segmented by $\frac{1}{6}$, meaning $n = \frac{1}{6}$ in Algorithm 1.

Algorithm 2 Example control response

- 1: **Iteration 1** ($time = t_0$)
 - 2: 1. Error evaluation
 - 3: $ERROR_{XYZ} = 0.15\text{ mm} < 0.25\text{ mm}$
 - 4: $ERROR_{\alpha\beta\gamma} = 0.05\text{ degrees} < 0.25\text{ degrees}$
 - 5: 2. Error below threshold, no response required
 - 6: **Iteration 2** ($time = t_0 + \Delta t$)
 - 7: 1. Error evaluation
 - 8: $ERROR_{XYZ} = 0.60\text{ mm} > 0.25\text{ mm}$
 - 9: $ERROR_{\alpha\beta\gamma} = 0.30\text{ degrees} > 0.25\text{ degrees}$
 - 10: 2. Error exceeds threshold, response required
 - 11: 3. Error transformed from $\mathbf{M}'$ to $\mathbf{H}'$ in 6DOF
 - 12: 4. Error segmented ($n = 6$):
 - 13: i. $ERROR_X = 0.30\text{ mm} \rightarrow ERROR_X = 0.05\text{ mm}$
 - 14: ii. $ERROR_Y = 0.20\text{ mm} \rightarrow ERROR_Y = 0.03\text{ mm}$
 - 15: $\vdots$
 - 16: vi. $ERROR_\gamma = 0.30\text{ degrees} \rightarrow ERROR_\gamma = 0.05\text{ degrees}$
 - 17: 5. Robot 6DOF compensation trajectory converted to linear actuator motions
 - 18: 6. Compensation trajectory executed by robot ($time = t_0 + \Delta t + t_r$)
 - 19: 7. Final position of robot recorded
 - 20: **Iteration 3** ($time = t_0 + 2\Delta t + t_r$)
 - 21: 1. Error evaluation
 - 22: $ERROR_{XYZ} = 0.30\text{ mm} > 0.25\text{ mm}$
 - 23: $ERROR_{\alpha\beta\gamma} = 0.15\text{ degrees} < 0.25\text{ degrees}$
 - 24: 2. Error exceeds threshold, response required
 - 25: 3. Error transformed from $\mathbf{M}'$ to $\mathbf{H}'$ in 6DOF
 - 26: 4. Error segmented ($n = 6$):
 - 27: i. $ERROR_X = 0.00\text{ mm} \rightarrow ERROR_X = 0.00\text{ mm}$
 - 28: ii. $ERROR_Y = 0.20\text{ mm} \rightarrow ERROR_Y = 0.03\text{ mm}$
 - 29: $\vdots$
 - 30: vi. $ERROR_\gamma = 0.12\text{ degrees} \rightarrow ERROR_\gamma = 0.02\text{ degrees}$
 - 31: 5. Robot 6DOF compensation trajectory converted to linear actuator motions
 - 32: 6. Compensation trajectory executed by robot ($time = t_0 + 2\Delta t + 2t_r$)
 - 33: 7. Final position of robot recorded
 - 34: **Iteration 4** ($time = t_0 + 3\Delta t + 2t_r$)
 - 35: 1. Error evaluation
 - 36: $ERROR_{XYZ} = 0.10\text{ mm} < 0.25\text{ mm}$
 - 37: $ERROR_{\alpha\beta\gamma} = 0.05\text{ degrees} < 0.25\text{ degrees}$
 - 38: 2. Error below threshold, no response required
-

CHAPTER 7

EVALUATION OF 6DOF HEAD MOTION STABILIZATION SYSTEM FOR STEREOTACTIC RADIOSURGERY

7.1 Introduction

Stereotactic radiosurgery (SRS) is a non-surgical approach to radiation delivery which features a low-fraction and high-dose technique with high accuracy to treat cranial tumors, arteriovenous malformations, trigeminal neuralgia, acoustic neuromas, and other disorders. In order to prevent complications to nearby critical structures such as nerves, the brain stem, or other vascular regions, well-planned SRS employs a steep dose fall-off around the target volume for high levels of conformality.

To achieve this steep dose gradient, strict mechanical, dosimetric, immobilization, and quality assurance tolerances are required. To adhere to the American Association of Physicists in Medicine Task Group Report, TG-42 overall targeting accuracy recommendation of 1-2 mm for intracranial SRS, a metal head ring is rigidly fixed to the patient's skull and then bolted to the treatment couch. [82] This frame-based SRS approach relies on fixing the head into a precise 3D stereotactic position, suppressing both voluntary and involuntary patient motion. Historically, frame-based SRS approaches were developed primarily to address the lack of image guidance available for treatment delivery at the time. Unfortunately, the discomfort and invasive nature from using the frame-based approach can limit the number of fractions delivered and/or the number of lesions that may be treated simultaneously, or prohibit its use for some patients entirely. [85] Finally, the clinical workflow is impeded compared to conventional radiotherapy, as imaging, treatment planning, and treatment delivery must occur on the same day. These issues, as well as a large setup time and limitations for fractionated or repeat treatments further highlight some of the current limitations of frame-based SRS.

To overcome the obstacles of frame-based SRS, frameless SRS, in which the head ring

is removed and replaced by a thermoplastic mask and a back of the head mold, has seen expanded clinical implementation. [86] Here, significant effort is required in order to ensure effective mask immobilization and, in certain cases, even with the mask, large systematic drifts of up to several millimeters from initial setup have been reported. [60, 58] As a result, treatment interruption for intrafractional patient realignment can occur to ensure proper target coverage. To address these issues, some solutions consider tracking the patient and altering the radiation source position to respond to their motion. [87, 9] Another widespread approach uses external camera tracking systems, often with occasional internal snapshots for position verification, and beam gating to ensure proper dose coverage. [49, 10, 76, 52]

Recently, our group has investigated the use of compact robotics to actively assist patients in maintaining a stable head position for long periods of time. [88, 89, 54] However, these earlier studies did not provide full 6DOF head motion control, and were limited to either 3DOF translational (XYZ) or 4DOF (XYZ + pitch) due to robot design and 6DOF trajectory control algorithm complexities. [51] As shown in previous studies, failure to provide full 6DOF motion control can lead to serious issues in the exposure of healthy brain tissues to high levels of radiation and poor target dose conformality. This is of particular concern in cases when lesions are non-spherical or located far from the isocenter as in single-isocenter multi-target treatments. [37, 34, 36]

In this chapter, we bring together the work of previous chapters, particularly the development of the parallel kinematics robot and the motion compensation algorithm, and present a novel real-time 6DOF robotic pre-clinical SRS system that is both frameless and maskless. The system can be used with any real-time 6DOF motion tracking device, and depending on the accuracy of the tracking device, is capable of matching or exceeding traditional frame-based SRS positional accuracies. System efficacy was first evaluated in a structured experimental setting, and then on several healthy volunteers as a step toward clinical implementation for the improvement of current intracranial SRS regimes. This robotic SRS system has the potential to improve dose delivery accuracy and patient comfort, while improving

the clinical workflow for intracranial SRS, especially for non-spherical lesions or in the single-isocenter multiple-target case. We also investigate the spatial dose coverage improvements when using the robot for head motion stabilization in a phantom case for single-isocenter multi-target SRS treatments.

7.2 Implementation of 6DOF real-time compensation framework

The theoretical 6DOF motion compensation algorithm was developed in Chapter 6, and its implementation using both software and hardware is discussed in this section. Because the required accuracy of the system is sub-millimeter and sub-degree, timing must also be highly accurate to prevent systematic errors in the robot's performance. Consequently, one controller card for all six linear actuators was required, and an 8-axis stepper motor controller (PCI-7358, National Instruments) was chosen. This controller from National Instruments communicates easily with LabVIEW software, also from National Instruments, and as a result the majority of the control algorithm was implemented in LabVIEW. This software can also easily read-in the measurements from the Polaris (Northern Digital, Inc., Waterloo, ON, Canada) IR camera, and so both the head motion inputs and the robot compensation outputs were controlled by the same LabVIEW program, further reducing the likelihood of timing discrepancies.

Although LabVIEW permits straightforward integration of the system's hardware, its graphical interface makes lengthy calculations cumbersome. A Matlab (2012b, The MathWorks) formula node was used within the LabVIEW framework for the control trajectory calculations. In Figure 6.4, the Matlab node corresponds to the green box, which takes as inputs the camera measurements and other control parameters, and calculates the six linear actuator trajectories to produce the robot's 6DOF head motion stabilization control. The LabVIEW program performs the remaining functions, which primarily consists of communicating with hardware, and providing users with real-time graphical representations of the 6DOF head motion and robot trajectory.

7.3 Application of parallel kinematics robot for motion compensation algorithm evaluation

7.3.1 *Experimental setup*

As a first step toward 6DOF motion compensation, we seek to use the parallel kinematics device and apply it to evaluate the effectiveness of the 6DOF compensation algorithm developed described in Chapter 6. [70, 63] The algorithm was implemented in LabVIEW (2012, National Instruments) and Matlab (The Mathworks, 2013b), and the control hardware is consistent with earlier descriptions. In order to effectively perform real-time motion compensation, an experimental setup consisting of the 6DOF robotic motion stage, a camera system, objects tracked by the camera, and the motion control hardware were first established. As before, four reflective markers define the virtual point coordinate system, both in position and orientation, which is tracked by the stereoscopic IR camera. With this setup established (Figure 7.1), as the reflective markers move in space, such motion is subsequently tracked and recorded by the Polaris IR camera system, and sent automatically to an in-house developed LabVIEW code, after the calibration procedure is performed.

7.3.2 *Evaluation of the 6DOF compensation control*

To initially evaluate the efficacy of the motion compensation algorithm, several disturbances were introduced to the IR marker block connected to the top platform. By adjusting the position of the four reflective markers, we shifted the virtual point coordinate frame $\mathbf{M}$ relative to the camera frame $\mathbf{A}$. The errors were registered, transformed to the robot control coordinate frame $\mathbf{H}$, and responded to automatically by the robot in full 6DOF. In Figure 7.2, we introduced shifts in 6DOF with the motion compensation routines initially turned off. When the routine is initiated, the stabilization algorithm registers the 6DOF error, and the robot begins to respond, cancelling this motion automatically translationally and rotationally. The errors observed in Figure 7.2 are larger than what is typically expected for

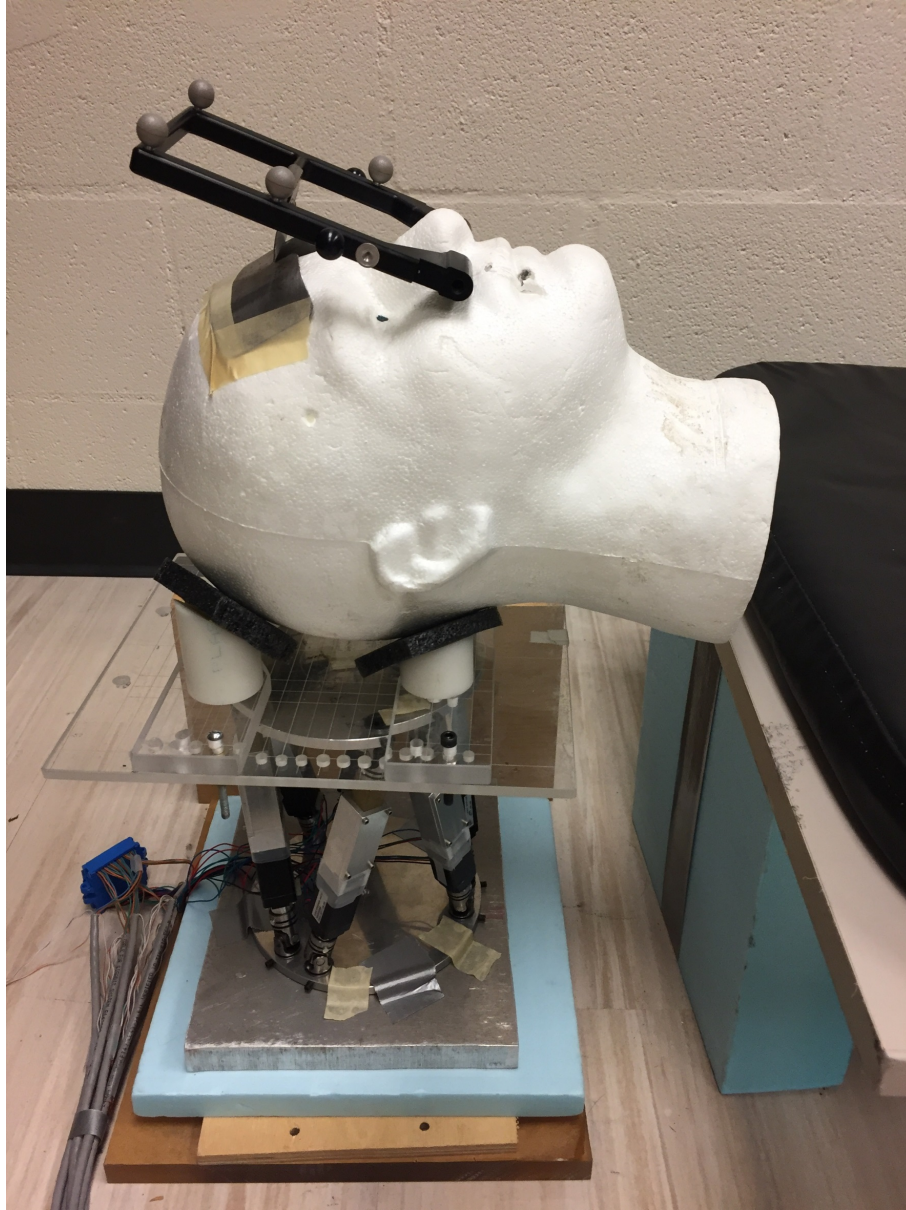


Figure 7.1: Experimental setup with the 6DOF parallel kinematics robot, a Styrofoam head phantom fixed to the top platform, and four reflective IR markers rigidly connected to the head phantom. Not pictured is the stereoscopic IR marker tracking camera.

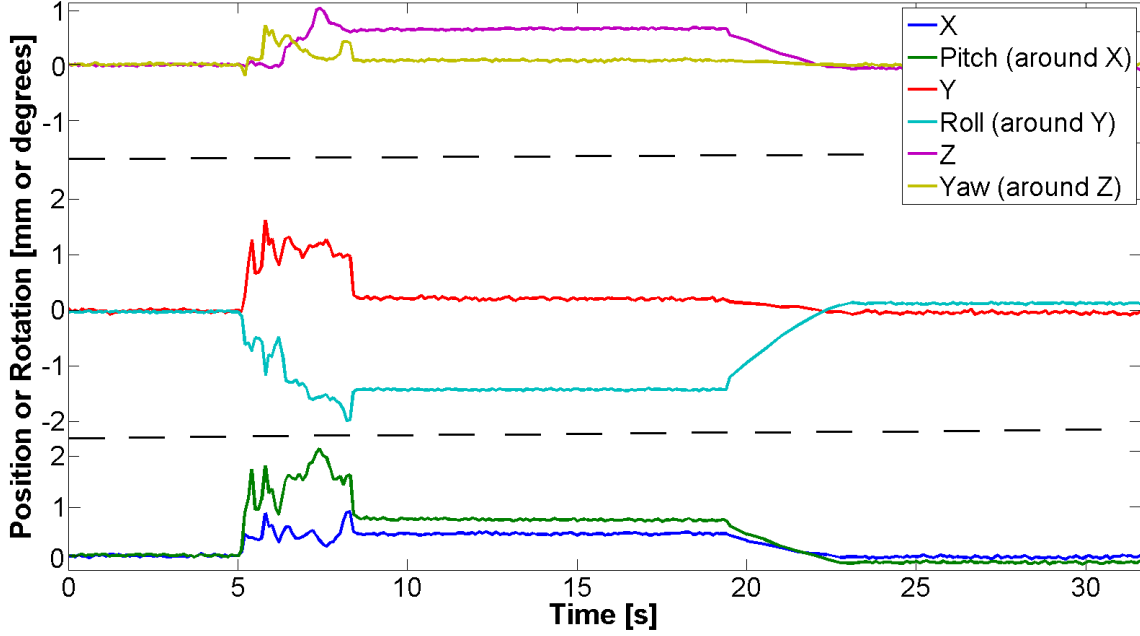


Figure 7.2: Motion compensation outputs from the IR marker camera, after calibration transformed the measurements to the virtual point coordinate system. A 6DOF motion is detected by the camera at the 5 second mark, which settles to a displaced position at 8 seconds. The motion compensation is enabled at 19 seconds, bringing the virtual point position to the desired position shortly thereafter.

cranial motion cases at any instantaneous point in time, based on the results of the patient motion tracking study in Chapter 5 and other earlier research, but over the course of a 15-minute treatment period, such gross error levels are possible. With motion compensation response enabled, the errors are cancelled such that the errors pre-correction are 0.5 - 2.0 mm and 0.5 - 2.0 degrees, and post-correction are less than 0.1 mm and 0.1 degree. In order to prevent continuous correction when the observed errors are within the IR camera noise, a lower bound threshold on correction of 0.25 mm or 0.25 degrees in any axis was instated. [48, 70]

In this evaluation of the 6DOF robotic motion compensation system, we introduced errors of several degrees and several millimeters to evaluate the capabilities of the robot in a motion compensation application. The accuracy of the robot had been previously evaluated, and our aim was to confirm the system's efficacy in responding to 6DOF motion in real-time by automatically generating stabilization trajectories, and executing them rapidly and faithfully.

By examining the system’s performance for this application, we could also help tune the algorithm’s parameters. As shown in the control flow diagram in Figure 6.4, some aspects of the algorithm are governed by user inputs, and by evaluating the system with simple 6DOF errors introduced, we could modify these inputs in a controlled setting, improving the response timing and minimizing overshoot.

After modifying the algorithm parameters in this experimental evaluation, the 6DOF motion cancellation system was found to be effective at compensating for translational and rotational motions in real-time. The system was able to respond to motions of several millimeters and degrees and return the virtual point (target) position with sub-millimeter and sub-degree accuracy, meeting current SRS immobilization requirements on patient positioning. In this section, we have presented the first evaluation of the system which can improve frameless SRS by automatically returning patients to isocenter with high 6DOF accuracy.

7.4 Implementation of 6DOF motion compensation device with volunteers

7.4.1 Volunteer recruitment and experimental setup

The study used six healthy volunteers to evaluate the efficacy of the 6DOF motion compensation system; volunteers were aged between 25 and 45, and included 4 males and 2 females. Due to volunteer availability, each volunteer contributed at least one motion-compensated data set and one non-compensated (uncorrected) set, though some volunteer 6DOF cranial movements were measured multiple times. Volunteers were recruited on IRB 14-0040.

The volunteers were requested to lay down on a simulated treatment couch, with their head positioned to maximize contact with the head rest and their comfort. The block of four reflective IR markers, which define the measurement coordinate frame, was fixed to their forehead for tracking; the marker block was attached to a curved metal plate to contour the forehead and attached with medical tape. The experimental setup is shown

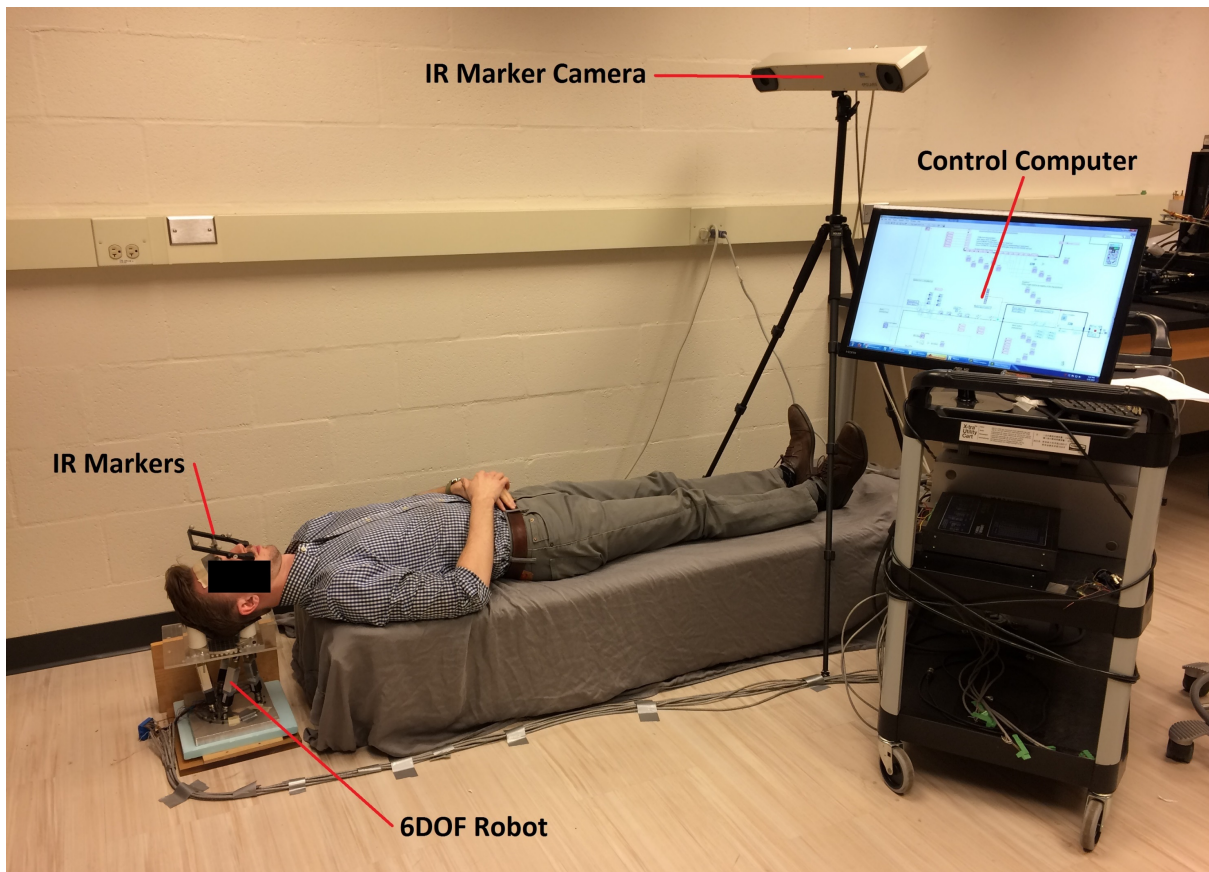


Figure 7.3: Volunteer setup with robotic device, control computer, and IR marker tracking system. The four reflective IR markers are fixed to the volunteer's head, to permit measurement of their 6DOF cranial motion. Because this setup is frameless and maskless, the robot and the head are weakly coupled, a complication which was overcome using an iterative compensation algorithm.

in Figure 7.3, with a human volunteer, the 6DOF parallel kinematics robot, the Polaris (Northern Digital, Inc., Waterloo, ON, Canada) stereoscopic IR marker tracking camera, and the control hardware. Volunteers were asked to breathe normally and relax in a supine position over the 15-minute monitoring duration, in order to mimic standard treatment conditions for intracranial SRS; volunteers were also resting in the same configuration for 2-3 minutes prior to tracking to allow them to become comfortable with the setup, similar to clinical procedures. No external immobilization was used, simulating a frameless maskless SRS setup. The choice to enable or disable the robotic compensation system was selected randomly prior to the start of tracking and the volunteer was not informed. This served to provide information about each volunteer's movement habits with and without correction, but also to prevent unintended bias from the volunteers. In the uncorrected cases, the volunteer's motion was simply monitored in 6DOF; in the corrected cases, the volunteer's cranial movements were measured and responded to in real-time by the parallel kinematics robot using the algorithm described in Chapter 6.

7.4.2 Experimental procedure

After the volunteers were in a comfortable position and the IR marker block was within view of the 6DOF camera, the head motion compensation procedure was initiated. The control algorithm used for motion correction was outlined in detail in Chapter 6. This compensation system automatically transforms the camera-measured 6DOF cranial motion, determines the errors in each axis for the "target" virtual point, and computes the linear actuator trajectories necessary to cancel the observed motion of the target position. In clinical practice the virtual point would be positioned at the treatment isocenter; but for use with volunteers, it was placed near the center of the head to help simulate a typical treatment condition.

When the measured cranial motion exceeds a threshold of 0.25 mm or 0.25 degrees, which was chosen to reduce response to noise from the Polaris camera, the robotic system is engaged

and the necessary head motion stabilization trajectory is performed automatically. This trajectory is calculated based on an iterative closest-point algorithm using inverse kinematics, which determines the movements of each linear actuator necessary to produce the desired 6DOF movement of the robot’s mobile top platform. The 0.25 mm / 0.25 degree movement thresholds were selected based upon the accuracy and noise of the camera system used in this study. [48] While the robotic system was actively compensating for the cranial motion, the 6DOF errors were continually being monitored and the trajectory was being updated to constantly respond to the patient’s motion.

The real-time 6DOF monitoring of the volunteers with the Polaris IR marker tracking system was performed at a 12-Hz sampling frequency, based on camera specifications. The individual’s head motion, through the surrogate IR markers, was subsequently transmitted to an in-house developed LabVIEW (2012, National Instruments) code which controls the robotic device shown in the volunteer setup in Figure 7.3. The control code communicated between the IR camera system and the robotic device simultaneously, transmitting the six linear actuator trajectories using an 8-axis stepper motor controller (PCI-7358, National Instruments). [70]

After 15 minutes of head tracking, the percent-of-time within the clinical threshold of 0.5 mm and 0.5 degrees was calculated. This analysis was performed on both the robot-corrected case and the uncorrected case. The two cases were then quantitatively compared in order to evaluate the system’s effectiveness at maintaining a stable head position for humans, both translationally and rotationally.

7.4.3 Results of 6DOF head stabilization

The motion compensation system performance was evaluated on six volunteers. Comparing the uncorrected case, in which the 6DOF cranial motion was monitored but not corrected for, and the corrected case, in which the robotic system performed a compensation trajectory to return the volunteer’s head to the initial position, improvements in 6DOF immobilization

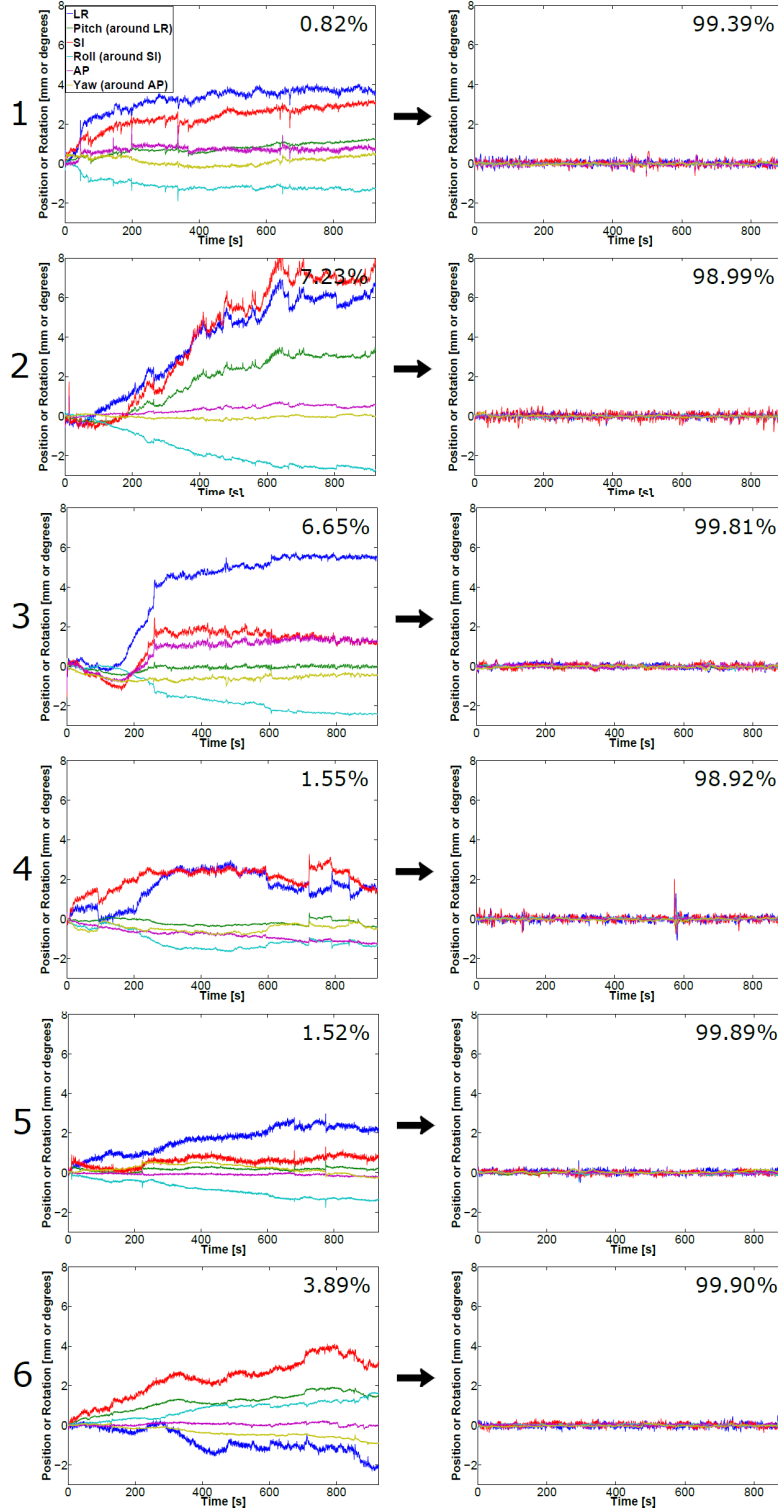


Figure 7.4: Example motions for all six volunteers, comparing uncorrected (left) to corrected (right) motion. For each volunteer's case presented, the percent of time within the clinical threshold of 0.5 mm and 0.5 degrees is also displayed in the each sector. Direct comparison shows that the corrected cases have improved the 6DOF positioning errors of the simulated tumor position by several factors.

Volunteer	Translational Percentage	Rotational Percentage	6DOF Percentage	Improvement Ratio
1	99.4	99.9	99.4	121
	1.0	3.0	0.8	
2	98.0	99.9	98.0	14
	7.2	14.6	7.2	
3	99.3	100.0	99.3	7
	13.9	23.1	13.8	
4	99.9	99.9	98.9	80
	1.2	9.1	1.2	
5	99.8	99.9	99.7	40
	2.5	27.1	2.5	
6	99.9	100.0	99.9	25
	4.0	10.8	4.0	
Overall	99.0	99.9	99.0	21
	4.8	9.5	4.7	

Table 7.1: Numerical results for motion compensation on 6 volunteers with 15 minutes of tracking in each session. The shaded rows for each volunteer show the uncorrected percent of time within the clinical threshold of 0.5 mm and 0.5 degrees, while the unshaded rows show the percent of time the volunteer’s motion was within the same clinical threshold when motion correction was enabled.

were observed. Figure 7.4 displays example motions for each volunteer, with an uncorrected case shown to the left of a corrected case. For the 15-minute tracking periods, an improvement factor of 21 was observed in this frameless maskless simulated SRS regime, comparing the uncorrected and corrected cases. When the robotic system was used to respond to volunteer head motion, the 6DOF localization of the virtual point, which was located at a simulated treatment isocenter within the brain, was within the clinical threshold of 0.5 mm and 0.5 degrees more than 99% of the time, compared to less than 5% when no correction was used, for all volunteers aggregated.

The volunteer-specific results and overall quantitative improvements are presented in Table 7.1. In this Table, if the six volunteers are considered collectively, a factor of 21 improvement between the uncorrected and corrected cases was achieved using the 6DOF motion compensation system. For the motion corrected cases, volunteer head motion remained within the 0.5 mm and 0.5 degree threshold 99.0% of the time, compared to 4.7% when no correction is used. For a 15-minute simulated treatment period, this corresponded to only 9 seconds outside the threshold when correction was used, compared to 14 minutes

and 17 seconds when no 6DOF robotic correction was used. Finally, a cumulative distribution plot was constructed to compare the uncorrected and corrected cases for the set of all volunteers (Figure 7.5). Examining the threshold of 0.5 mm / 0.5 degrees, it appears that the robotic motion compensation system ensures the volunteer’s head motion rarely exceeds this cutoff, compared to the uncorrected cranial motion.

To address issues with current SRS techniques, a novel robotic SRS system demonstrated high cranial motion control is achievable without the use of a metal frame of thermoplastic mask. The system was tested on six volunteers to evaluate performance in maintaining head stabilization throughout mock SRS deliveries. Comparing corrected and uncorrected head motion, the 6DOF parallel kinematics robotic system demonstrated an overall improvement factor of 21 with regards to maintaining cranial localization within the 0.5 mm / 0.5 degree threshold. Although variations in effectiveness between cases were observed, for the aggregated volunteer data, the target position remained within the clinical tolerances 99.0% of the time when the robotic motion compensation system was used, compared to only 4.7% of the time when the system was disabled. This technique therefore has been shown to be effective with volunteers, and has the potential to improve upon current frameless SRS treatment approaches.

The use of a 6DOF robotic real-time compensation platform may improve target coverage even more when rotations have a larger impact, such as the treatment of non-spherical lesions, or in the single-isocenter multi-target regime. In these cases, when rotations occur, a larger volume of the target lesion or lesions will deviate from the intended position, further increasing the deleterious effects of poor immobilization. [90] For such cases, the improvements demonstrated here can have a more profound effect, and future studies will consider the dosimetric difference between corrected and uncorrected motion in these cases. However, the current study demonstrated the feasibility of integrating 6DOF robotic systems into intracranial frameless maskless SRS treatments in order to improve the intrafractional immobilization by responding to measured head motion in real time.

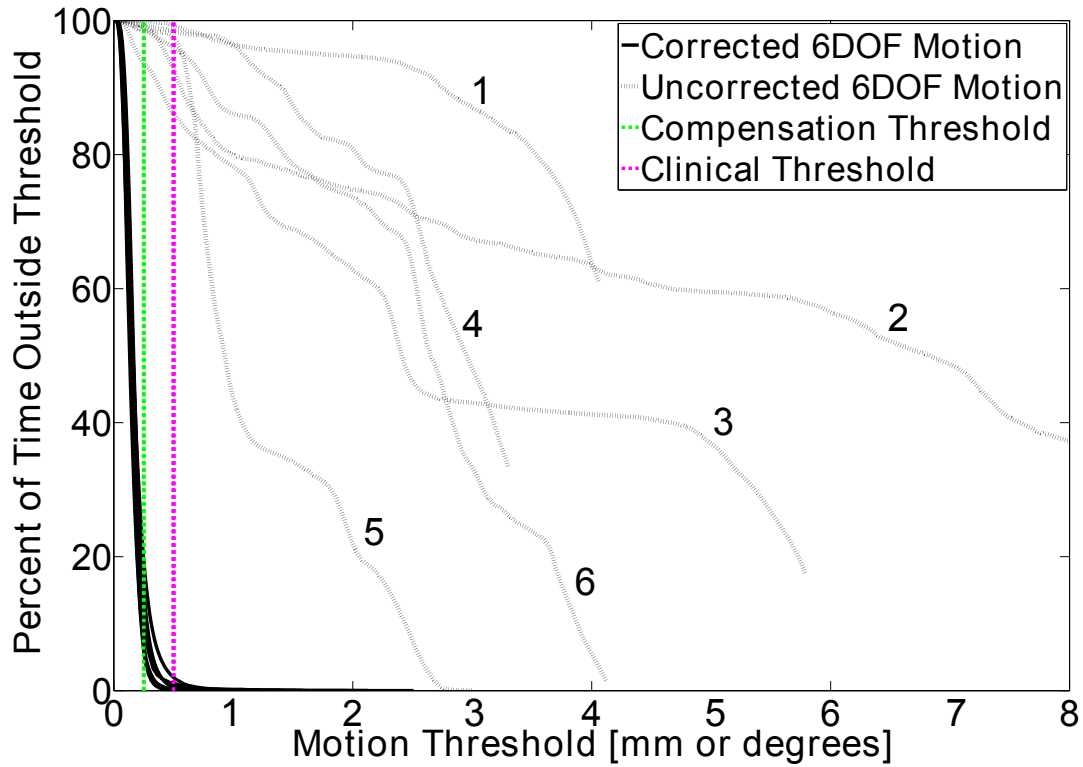


Figure 7.5: Cumulative distribution for all six volunteers, comparing uncorrected to corrected motion. The compensation threshold of 0.25 mm / 0.25 degrees was used in the control algorithm to determine when to engage the robotic system for correction; the clinical threshold of 0.5 mm / 0.5 degrees was used as a measure of system performance, chosen to reflect the target localization goal in treatment. The plot compares the corrected motion (solid lines) with the uncorrected motion (dashed lines) for all volunteers, with their anonymized labels, in terms of the percent of time they remained within the motion thresholds defined on the horizontal axis.

The platform on which the volunteers rested their heads consisted of four posts on a swivel, and since contact with these points is critical for correction, an improved setup can increase the percent of time within clinical threshold by enhancing volunteer comfort and better ensuring constant contact. Additionally, this study placed the robotic system directly beneath the volunteer’s head (Figure 7.3), but such a configuration may not be ideal for some clinical cases since posterior beams cannot be utilized in the treatment delivery. The proposed solution is to place the robotic system beneath the patient’s neck, away from the treatment beam, and control their cranial motion from that position. Since the calibration technique is a general approach to associate the coordinate frames considered for motion correction, no fundamental changes to the overall control algorithm are expected. [91]

The robotic SRS system was evaluated in this study using a stereoscopic IR camera, but it should be noted that its functionality is not restricted to any specific tracking system. The only requirement is that the tracking system is able to provide real-time 6DOF positional data. The robot uses this measurement data in the closed loop feedback configuration shown in Figure 6.4, where it attempts to zero out detected camera deviations, after appropriate transformations have been applied. Given that the robot could perform such corrections at the sub-millimeter level, overall accuracy was primarily limited by the 6DOF tracking accuracy of the head tracking system. [70] This can be observed in Figure 7.5, where positional control was limited by the 0.25 mm accuracy of the IR camera used for this pre-clinical evaluation. In this study, the real-time 6DOF motion was measured by tracking IR markers fixed to the volunteer’s forehead using tape and a contoured metal plate with the stereoscopic IR camera. Such a configuration is sufficient to demonstrate the goal of the study, where, given real-time 6DOF input, the robotic SRS system is able to stabilize motion to a sub-millimeter and sub-degree level. In actual clinical scenarios, it is necessary that the camera tracking point is accurate in terms of absolute target position as well, and therefore calibrated to the treatment isocenter. [68] To this end, the tracked surrogates, such as the reflective IR markers used in this study, must be rigidly connected to the internal tracking

target. Unfortunately, examination of the accuracies of various 6DOF tracking systems is beyond the goals of this study, which focuses on the robot and the iterative compensation algorithm, however some clinically suitable head tracking methods include the use of IR markers attached to a patient dental block, or a 3D surface map tracking of the rigid facial structures. [91, 76, 47, 48]

For clinical implementation, it will be necessary to incorporate beam gating into the treatment technique. The robotic system described above is able to respond to 6DOF cranial motion in real-time using the iterative algorithm, however, for large discrete head movements which exceed the preset clinical threshold, gating the beam would help minimize any risk of patient mistreatment. However, because the robot responds quickly, the beam can be automatically turned back on once the target returns to the treatment position. In this way, clinician intervention would be reduced, and the treatment workflow would improve upon current frameless techniques, where it is necessary to interrupt the treatment and enter the room in order to re-position the patient.

In this section, we described the application of the SRS robotic system for use in real-time translational and rotational cranial motion stabilization. The system performance was evaluated on six healthy volunteers, and was shown to dramatically improve target localization within a 0.5 mm and 0.5 degree threshold over 15-minute tracking periods. Using the novel iterative closest-point compensation algorithm described in Chapter 6, the robot was able to effectively cancel measured 6DOF cranial motions without the use of a frame or mask. The system’s success with volunteers established its potential for clinical implementation in a frameless maskless SRS approach with improved workflow, and treatment accuracies as good or better than traditional frame-based intracranial SRS.

7.5 Development of a cranial motion control evaluation system

A systematic approach to evaluating the 6DOF robotic head motion compensation system is important for long-term use and performance assessment under changing conditions. Im-

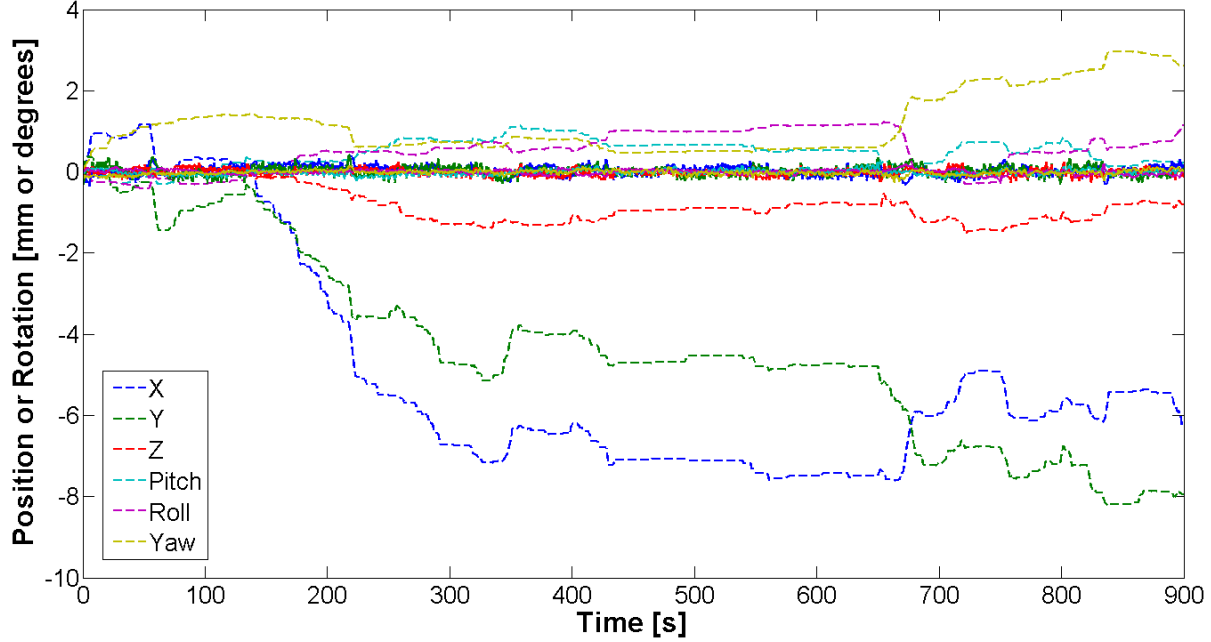


Figure 7.6: 6DOF robot-stabilized head motion of one volunteer (solid lines), with the robot’s top platform correction trajectory used to perform the real-time head motion compensation (dashed lines). Forward kinematics was used to convert the six linear actuator trajectories used to correct for the volunteer’s motion into motion of the top platform over the 15-minute study period.

plementing head stabilization with a 6DOF parallel kinematics robot with volunteers was described in the previous section, and system efficacy was determined using a percent-of-time within a threshold of 0.5 mm and 0.5 degrees metric. Although this approach is useful to demonstrate the system’s capability for maintaining stable head positions translationally and rotationally, it is highly dependent upon the volunteer’s unique motion and conditions specific to each trial. Shown in Figure 7.4 (left), when the robot is not enabled, cranial motion can follow unexpected trajectories over the 15-minute tracking period. In this section, we discuss alternative metrics for evaluating performance of the 6DOF robot head stabilization system under changing conditions in algorithm parameters and setup configurations.

The head motion observed in the volunteer study was unique to the individual subject, and instantaneous motion is highly unpredictable. This condition required the development of a novel iterative compensation algorithm, described in Chapter 6. The degree to which the robot was required to compensate for measured motion was not only dependent upon the

particular volunteer, but also the head support, which connected the subject's head to the robot. Because the clinical goal is a frameless and maskless system, the coupling between the head and the robot is critical for proper immobilization. Under the loose assumption used in the 6DOF control algorithm that the subject's head is rigidly connected to the robot when the compensation is performed, it is important that when the robot's top platform attempts to correct for head positioning error, the subject responds to this input. The degree to which this coupling holds is an important factor to consider when evaluating the whole 6DOF head stabilization system, which includes the head support configuration, the robot control accuracy, and the compensation algorithm parameters.

The weak coupling issue of frameless and maskless SRS was discussed in the context of developing the iterative motion compensation algorithm in Chapter 6; slippage between the patient's head and the robot system is important to consider in the context of evaluating the system's performance as well. Figure 7.6 presents an overlay of the parallel kinematics robot's top platform movements and the corresponding stabilized motion for one example volunteer. The full 6DOF trajectories of both the robot $\mathbf{H}$ and virtual point $\mathbf{M}$ coordinate frames demonstrate the degree to which the robot is actively compensating for recorded motion, both in terms of frequency of intervention and overall magnitude of robotic stabilization response. In the case shown in Figure 7.6, the robot needed to move over 10 mm and 3 degrees to effectively maintain the volunteer's head motion within the preset thresholds.

Conducting an analysis which compares the top platform motion and the stabilized head motion is useful for evaluating the system in several ways. First, it provides general information about how well the robot was able to compensate for measured head motion, and whether the top platform stabilization trajectory was near to the robot's workspace limitations. Second, we can examine instantaneous changes in the response of the 6DOF robot to the individual's head motion to evaluate slippage. For example, if the measured motion was recorded as 1 mm, and to bring the virtual point back to its initial position the robot was ultimately required to move 1.2 mm, or 1.8 mm under different conditions, we can extract

some information about the degree to which the individual responds to the robot's compensation. Although many factors can influence this over-correction, including the control algorithm parameters, the head support, the individual's response to robotic head motion stabilization, and instantaneous camera noise, this analysis was useful in determining the system's response effectiveness. And third, one can examine instantaneous response timing of the system by observing when the individual's head motion exceeded the 0.25 mm and 0.25 degree threshold, and how many frames before a correction was observed.

7.6 Dosimetric impact of 6DOF compensation: a phantom study

Removing the rigid metal immobilization system in frame-based SRS, as well as the thermoplastic mask in frameless SRS has the potential to not only degrade the 6DOF positioning of the patient's head, but also the dose coverage. By eliminating highly restrictive immobilization devices when treating lesions in the brain, in favor of approaches which improve clinical workflow and patient comfort, there is the potential for degrading dose delivery accuracy. 6DOF head motion stabilization using highly precise robotics was demonstrated on volunteers earlier in this chapter, but ultimately the objective of altering standard-of-care with the robot is to provide the benefits of a frameless maskless SRS system while maintaining dose coverage with the same or better accuracy as a frame-based system. In this section, we investigate the potential improvements in target dose coverage using the 6DOF robot in a frameless maskless SRS setting with a phantom.

To evaluate the head motion compensation system's efficacy in the dose accuracy setting, an anthropomorphic head phantom from CIRS (Computerized Imaging Reference Systems Inc., Norfolk, Virginia) was selected as a surrogate for a human head. The phantom was scanned using a CT simulator, under frameless SRS guidelines. For the purposes of this study, a thermoplastic mask was used to properly position the CIRS phantom in the CT scanner such that an internal cavity was aligned to the axes of the scanner. When the 6DOF robotic head stabilization system will be used in a clinical setting, the robot may also be

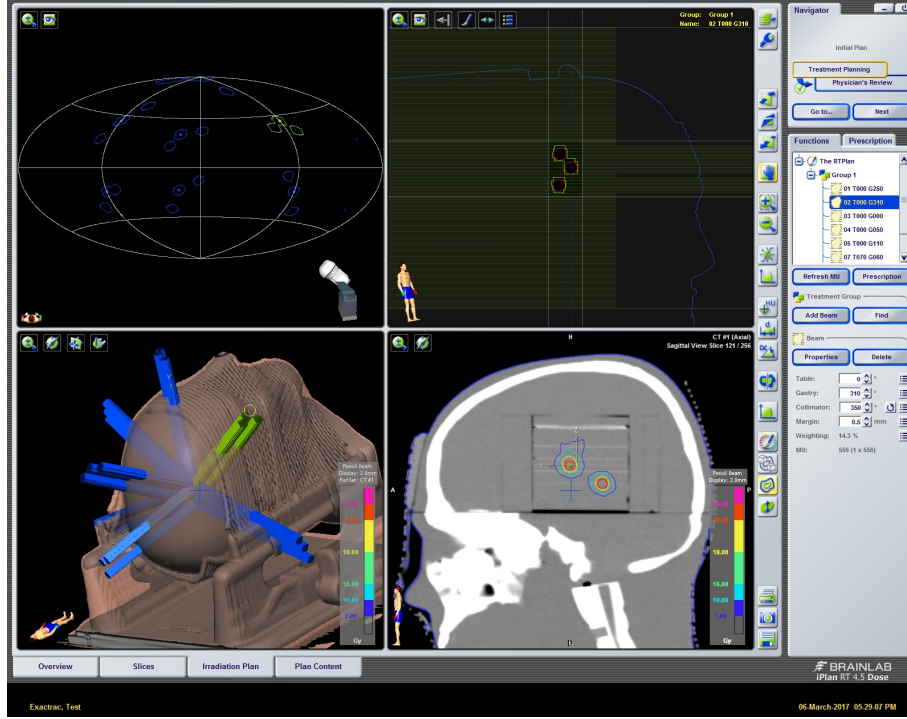


Figure 7.7: Treatment plan for a single-isocenter multi-target SRS treatment.

used for immobilization and artifact prevention when the treatment planning CT scan is performed, further reproducing the frameless maskless setting during treatment.

From the CT scan, a treatment plan was generated using BrainLab iPlan (Brainlab AG, Munich, Germany). In frameless and maskless SRS, both translational and rotational errors may increase dose coverage uncertainty due to a reduction in rigid immobilization, as described previously. This can be particularly problematic in single-isocenter multi-target SRS, where head rotations may further degrade treatment efficacy due to a leveraging effect. Consequently, we developed a treatment plan with three spherical targets and a single isocenter in an SRS protocol. Figure 7.7 depicts development of the plan with the iPlan software. The three lesions were placed within an internal cavity of the CIRS phantom, with the two distal lesions located approximately 1.5 cm away from the central lesion. The central lesion is placed at isocenter in this single-isocenter multi-target plan. The internal cavity of the head phantom has an insert with several tissue-mimicking slabs with 4 mm thickness; this setup allows users to place radiographic film between the slabs inside the

phantom for dosimetric analysis. The geometry of this phantom insert can be seen in Figure 7.7. The orientation of the head phantom, and subsequently the internal cavity, was chosen for ease-of-use, with the slab axis positioned to be in-line with the scanning axis.

The treatment plan was exported to a Varian TrueBeam (Varian, Inc., Palo Alto, CA) LINAC, and the head phantom was positioned on the treatment couch, as shown in Figure 7.8. The 6DOF parallel kinematics robot was placed beneath the head phantom; the treatment plan was developed to prevent incoming radiation beams from interacting with the robot’s aluminum top platform, reducing potential dose blurring from scatter. A thermoplastic mask was used to secure the head phantom to the robot’s mobile top platform, with a Brainlab SRS head support serving as the connector between the mask’s support and the robot. This setup helped ensure similarity to the CT scan geometry and sufficient rigidity between the robot and the CIRS phantom. Although frameless and maskless SRS inherently doesn’t use a thermoplastic mask, its use here was helpful to ensure the head phantom followed the motion of the robot faithfully in 6DOF.

After the treatment plan was generated and the experimental setup established, the treatment was delivered under three conditions, based upon the trajectories followed by the robot. Volunteer head motion was previously tracked and stabilized in real-time using the 6DOF robot, as described in Chapter 7.3. In that study, volunteer motion was also monitored translationally and rotationally in the simulated frameless and maskless setup without robotic head stabilization. In this way, head motion files were acquired for both the robot-corrected and uncorrected cases. These motion files served as inputs to the 6DOF robot for the purposes of this study into target dose coverage during 6DOF robotic frameless maskless SRS. The 6DOF head motion files were converted to six linear actuator trajectories using the formalism described in Chapter 2, allowing the robot to reproduce the previously recorded motion. Although these motions are relative to the virtual point of the particular volunteer, the setup geometry of the volunteer study is similar to that shown in Figure 7.8, with the robot beneath the head.



Figure 7.8: Experimental setup of the dosimetric evaluation study, featuring the 6DOF robot, an anatomical CIRS phantom with Gafchromic film placed internally, and a Varian TrueBeam LINAC. Although a thermoplastic mask is shown here, the motion reproduced by the robot follows head motion tracked under frameless maskless conditions, and the mask is used to ensure the phantom remains fixed to the top platform and consequently mimics this motion faithfully.

The single-isocenter multi-target treatment was carried out three times on the CIRS phantom: when the robot was reproducing unstabilized 6DOF frameless maskless head motion from the volunteer study, when the robot was reproducing 6DOF-stabilized head motion, and when the head phantom was stationary. For dose coverage analysis, 5 pieces of Gafchromic EBT3 film (Ashland Inc., Lexington, Kentucky) were placed within the phantom's internal cavity for each of the three treatment conditions, for a total of 15 films. Film was located at the center of each of the three spherical targets, as well as one placed 4 mm superior to the center of the middle lesion and one 4 mm inferior to the central lesion. The film was replaced for each run of the SRS treatment.

After the three trials were performed, the 5 films from each treatment were scanned using an Epson 11000XL scanner (Epson America Inc., Long Beach, California) with 600 dpi. Films from the two motion cases - uncorrected head motion and robot-stabilized motion - were compared to the no-motion case using a Gamma analysis procedure. Analysis software from RIT (Radiological Imaging Technology Inc., Colorado Springs, Colorado) was used for evaluating the target dose coverage. For the analysis, films from the no-motion case and either the robot-stabilized or the uncorrected motion cases were imported, once for each of the five films. The images were registered using markers on the film; the markers were based on the positioning of the block insert within the CIRS phantom, and was therefore unaffected by slight film alignment variations within the phantom. Because target coverage and positional accuracy were the analysis metrics for this study and not absolute dose, a dose calibration was not used. The images were then normalized, and an ROI around the target was selected. The Gamma analysis for the target region was then performed, independently comparing the no-motion case to both motion cases of the robot.

For film-based analysis on SRS treatments, a 95% pass rate for the target region on a 1% / 1 mm metric is necessary for plan approval. Comparing the uncorrected motion case to the no-motion case, only the slice through the center of the middle target passed this condition, with a 98.4% pixel pass rate. For the comparison between the robot-corrected motion case

and the no-motion case, all slices excluding the film located at a plane through the center of most inferior target passed at the 95% level. Of the films that passed in the robot-stabilized case, an average pixel pass rate of 99.1% was observed. Comparing all five analyses for both motion cases, the stabilized case improved Gamma analysis agreement by 6.8% compared to the uncorrected motion case on average.

In this study, we examined the dose coverage improvements when a 6DOF parallel kinematics robot was used for head motion stabilization in frameless and maskless SRS with a single isocenter and multiple targets. Reproducing both uncorrected head motion and robot-corrected motion on an anthropomorphic cranial phantom, target dose coverage agreement improved when the robot was used to automatically stabilize the head. When the robot was reproducing the 6DOF motion of an uncorrected case, only the central slice of the middle target passed the Gamma analysis; this slice was the only plane intersecting treatment isocenter, and as a result the leverage effect due to rotations in the single-isocenter multi-target regime appeared to have a greater impact on the more distal targets. Overall, introduction of motion degrades target localization, but the use of simulated robotic head motion stabilization showed good agreement with the no-motion case, demonstrating the potential for real-time 6DOF robotic head stabilization to improve frameless maskless SRS treatment accuracy.

CHAPTER 8

FUTURE WORK AND CONCLUSION

8.1 Future work and clinical implementation

8.1.1 Development of a second generation robot for clinical implementation

A first generation parallel kinematics robot with full 6DOF control was used for the majority of our research into real-time robotic head motion stabilization for frameless and maskless SRS. Throughout the course of evaluating the system performance, and examining the limitations of the robot shown in Figure 3.2, potential improvements were considered on a robot before clinical implementation. Based on the extensive evaluation of the robot used throughout this research project, a second generation parallel kinematics robot was designed for use with patients in a frameless and maskless SRS setting.

This new design features stronger motors capable of controlling cranial motion more effectively under different clinical setup geometries. NEMA 8 stepper motors used in the first generation design were replaced with NEMA 11 motors, which feature substantially greater torque. Earlier assumptions regarding patient setup during treatment were altered such that most of the weight of the neck and head would be potentially supported by two motors alone. As the ultimate application of the robotic system will involve positioning the device under the patient's neck, rather than the head directly, in order to avoid interaction with the treatment and imaging beams, a new system must be developed with stronger linear actuators. Additionally, the new linear actuators feature an updated design, wherein the motors are not in-line with the drive shaft, but rotated 180 degrees and placed adjacent to the lead screw. This will reduce the weight load directly on the stepper motor shaft, and is expected to improve motor longevity and performance.

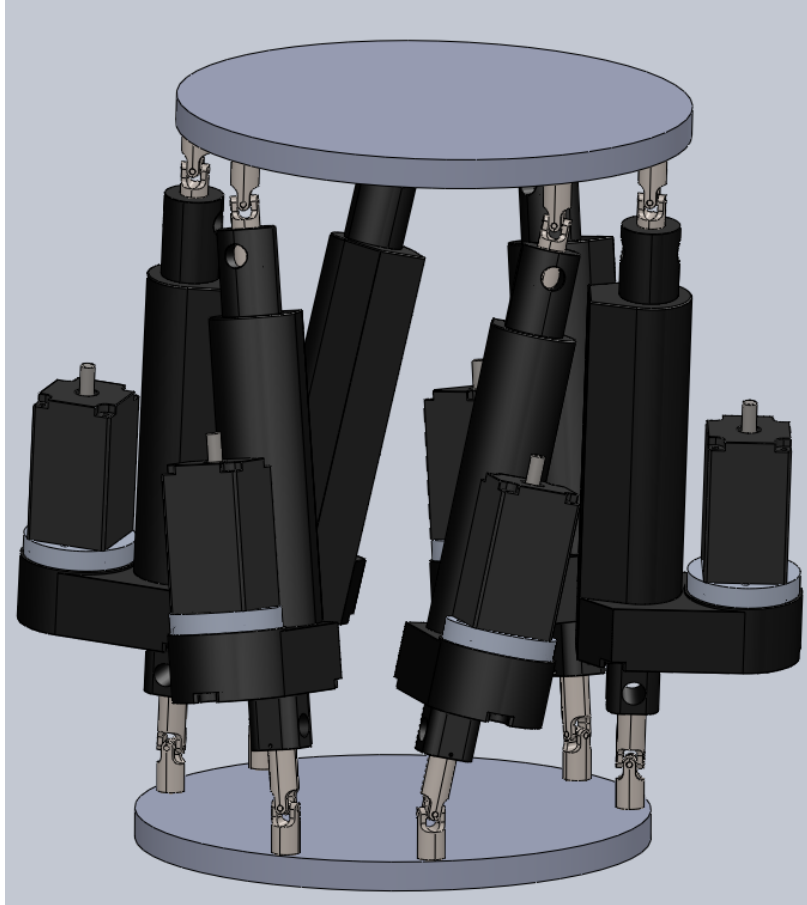


Figure 8.1: Proposed second generation parallel kinematics robot. The stepper motors used in the earlier design are replaced by larger NEMA 11 motors, which improve torque. Additionally, motors are rotated 180 degrees relative to the linear actuator travel direction to prevent high loads in the motor axis of rotation. A coupling between the lead screw and the motor was developed, which converts rotations of the motor to rotations of the lead screw.

8.1.2 *Proposed clinical setup*

The experimental setup for evaluating the motion compensation system’s performance used in the study presented in Chapter 7 was chosen to mimic a clinical setup. However, due to experimental limitations, the robot was placed beneath the subject’s head and the head support connection between the robot and the volunteer did not perfectly match a clinical setup. In this section, we describe the changes that would be made when transitioning from the laboratory to the treatment room.

Figure 8.2 shows the proposed clinical setup with the second generation parallel kinematics robot (Figure 8.1) placed beneath the patient’s shoulder blades. In this configuration, a couch attachment connects the bottom platform of the robot to the couch, and the top platform to the head support. This proposed attachment would include a coupling to standard treatment couches, such that therapists would simply connect the robot and head support system to the end of the couch. Such an approach further reduces treatment setup time, keeps therapist intervention at a minimum, and ensures proper positioning through ease-of-use. Furthermore, by using a standard attachment, the position of the 6DOF robot’s stationary bottom platform would be consistent and known, based upon the couch position. In this way, the setup remains consistent between treatments.

Finally, the clinical setup will include an updated head support. Previous chapters describe the importance of the head support in performing effective head motion stabilization during frameless and maskless treatments; patient discomfort, significant degradation of the rigidity assumption during control, and increased slippage can occur when the head support is not carefully designed. Additionally, to maintain ease-of-use and minimal setup time, the head rest should be universal and not dependent upon patient-specific anatomy. The proposed solution is to use a contouring foam with a high-friction surface. The pliable foam helps with the universality requirement and improves conformality between the head and the support structure. By having a surface with good traction, intrafraction motion can be reduced, and when robotic compensation is required, the patient is more likely to respond

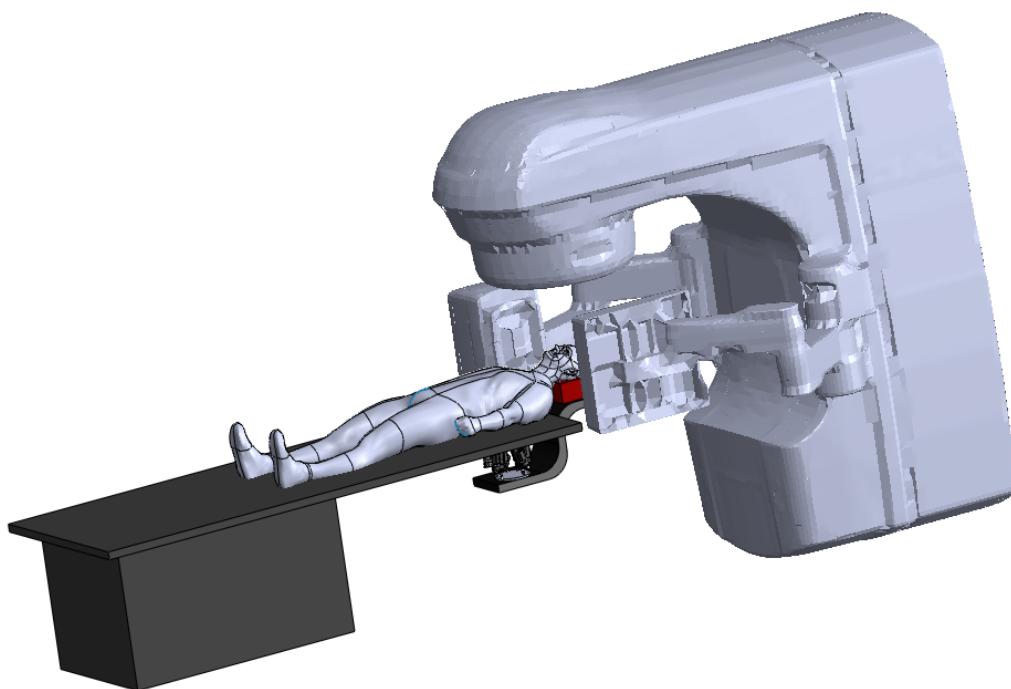


Figure 8.2: Proposed setup for clinical implementation in a frameless maskless SRS treatment regime. A second generation parallel kinematics robot is shown, positioned beneath patient's neck and shoulder area. Connecting the couch to the robot, and the robot to the head support (shown in red) is a couch attachment.

to the input stabilization. Additional work is required to develop this head support, and to determine how it will integrate into the CT scanning setup when the patient is receiving the treatment planning scan.

8.1.3 Proposed clinical procedure

The robotic SRS treatment strategy relies on the combination of real-time 6DOF surface tracking provided by AlignRT (AlignRT, VisionRT, UK) and occasional snapshots of patient head motion using the ExacTrac X-ray system (ExacTrac, BrainLab, Germany), which provides patient position verification in initial setup and at key points during treatment. This system, which is also frequently utilized in our clinic for cranial radiotherapy treatments, uses two non-coplanar kV x-ray units to provide accurate head positioning information based on internal anatomical structures. 6DOF fusion software compares the intrafraction position of the patient's bony skull anatomy to the CT-based digitally reconstructed radiographs (DRRs), and automatically determines the 6DOF deviations during treatment. The ExacTrac frameless radiosurgery system has sub-millimeter and sub-degree tracking accuracy, operates non-invasively, and will integrate easily with our proposed robotic SRS approach. [8]

Patients undergoing SRS treatments with the robotic motion compensation system will first receive a CT scan for treatment planning purposes; the robot with a head support will be placed under the patient's neck during the scan to match treatment conditions. In the same way that the robot's position relative to the patient's head will not interfere with the treatment radiation, CT imaging will not be disrupted by using the robot. Digitally reconstructed radiographs (DRRs) and cranial surface contours will be generated from the CT scan. After treatment planning, patients will be brought to the treatment room and positioned on the couch with the same technique as in current frameless SRS approaches, but with the robot attachment placed beneath their neck and the thermoplastic mask removed. The patient lies down on the couch with their head positioned on the head support, which

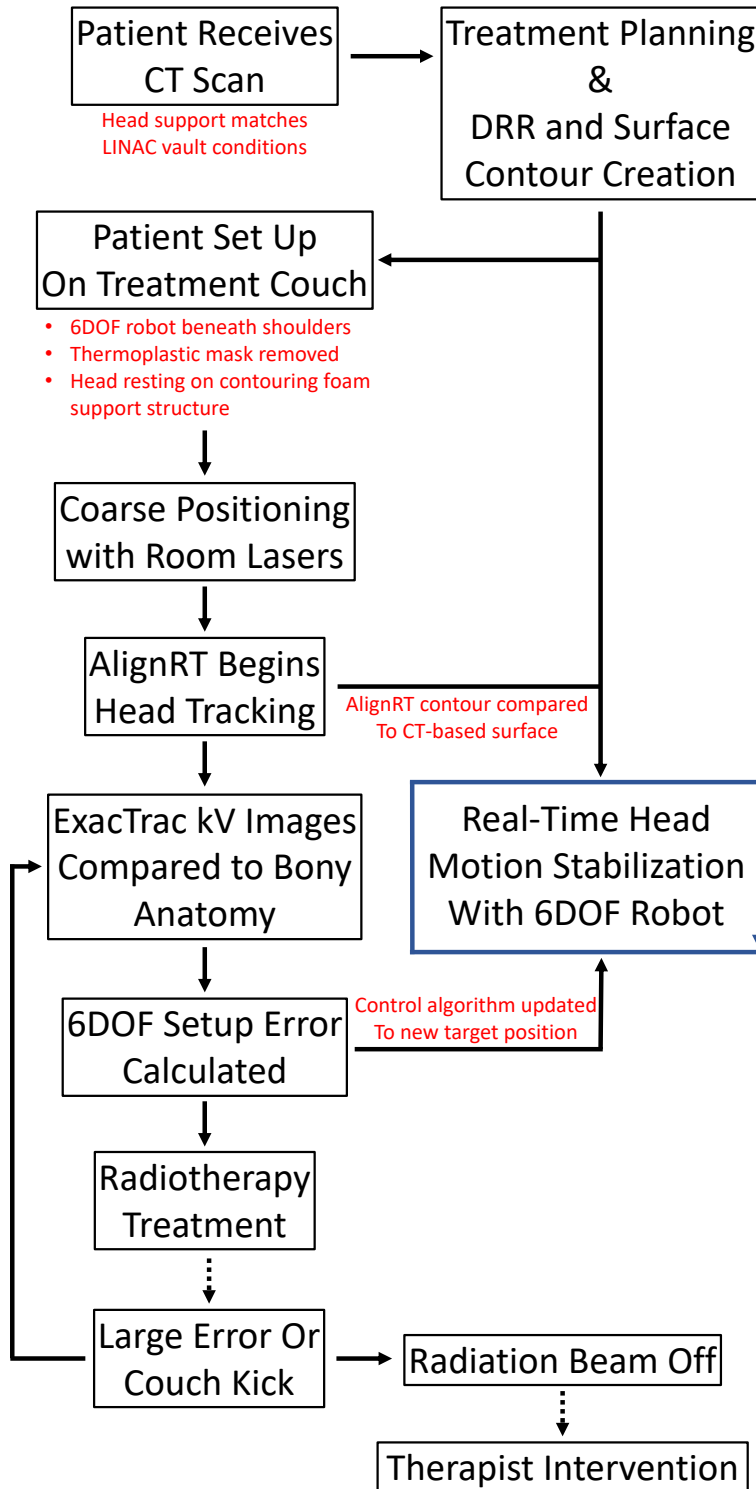


Figure 8.3: Proposed clinical workflow diagram when using the 6DOF robot for real-time head motion stabilization during radiotherapy procedures. Solid lines indicate a procedure which occurs for every treatment, and dashed lines indicate a step which may not always occur in all cases.

is rigidly connected to the parallel kinematic robot (Figure 8.2) and made of a contouring foam material to increase traction between the head and the head support. The same head rest will also be used during the CT scan procedure. Therapists will then perform coarse patient positioning using the room lasers. The AlignRT system begins tracking the facial contour and compares positioning to the CT-produced surface contour, and the 6DOF robot automatically begins stabilizing the patient’s head to the treatment position.

Images of the internal bony anatomy from the ExacTrac system will be compared to the DRRs, and a 6DOF deviation will be computed with BrainLab software. The initial setup corrections based on this calculation will then be transmitted to AlignRT and the robot, which will update the head stabilization control to begin automatically maintaining the patient’s head position to this updated location. Another comparison between the ExacTrac kV images and the setup DRRs will be made for final verification of initial patient setup. Therapists will then begin the treatment, and real-time 6DOF surface tracking from the AlignRT system will be simultaneously used to determine intrafraction head motion for gating purposes, and sent to the robot for real-time head stabilization. The robot will compensate for the intrafraction cranial motion based on the real-time AlignRT 6DOF measurements until a couch kick is used during treatment, thereby significantly altering the treatment setup, and halting the robot’s stabilization. At each updated couch position, ExacTrac kV images will be generated and compared to the DRRs for verification. Following the same procedure as before, 6DOF head position deviations at the new couch position will be calculated with ExacTrac software, and the robot’s control algorithm is updated to begin stabilizing the patient to the new treatment position. Treatment resumes at the new couch position, and the parallel kinematics robot continues to compensate for measured head motion in real-time. The procedure is repeated for every couch position, or if other substantial changes to the setup geometry are made during the course of treatment.

8.2 Conclusion

This dissertation presents the development of a robotic system for the purpose of providing real-time 6DOF head motion stabilization during frameless and maskless stereotactic radiosurgery. The motivations for this work were presented in Chapter 1, and were primarily aimed at improving upon the current approaches for treating brain disorders with high-dose low-fraction radiotherapy. Chapter 2 presented the control theory behind parallel kinematics robots. Both inverse kinematics and forward kinematics formalisms were discussed; the former was more prominently used for producing 6DOF trajectories for the robot, while the latter was useful in validating the system performance. In Chapter 3, we describe the process for designing, fabricating, and evaluating the capabilities of a parallel kinematics robot. Specifically, the design iterations are described and motivated, and the hardware used for the device’s construction are explicitly discussed. This chapter also used the inverse kinematics approach to simulate movements of the robot under several conditions to inform the system’s design. Chapter 3 finishes by examining the limitations of the robot in terms of 6DOF workspace and velocity, as well as its ability to faithfully reproduce several 6DOF trajectories, including previously recorded cranial motion. In Chapter 4, we develop a comprehensive and generalized calibration procedure for the robot. This procedure uses a well-defined 6DOF trajectory of the robot to connect the coordinate frames of the robotic system, the clinical tracking camera, and the treatment position, both translationally and rotationally. Proper calibration of the system was essential to understand how to stabilize head motion during SRS, as well interpret real-time cranial motion outputs from tracking systems. In Chapter 5, we built upon the developments of Chapters 2-4 and used the 6DOF robot as a quality assurance phantom to evaluate the tracking accuracy of several external patient tracking systems. Comparing the intended input motion of the 6DOF robot to the tracking system-measured motion, we determined the capabilities of three camera systems, translationally and rotationally, and at several positions, including points located away from treatment isocenter. This chapter also described a clinical investigation into patient head

motion, tracking using an optical surface mapping system during whole-brain radiotherapy procedures.

Chapter 6 returns to the use of the robot for 6DOF head motion stabilization, and presents a motion compensation algorithm. The chapter first describes an earlier generation system with 4DOF (3 translational + 1 rotational degree of freedom) control, and the development of a novel feed-forward control approach. After this architecture was described, an iterative closest-point 6DOF algorithm was presented, and an example for how the algorithm functions was provided. The dissertation culminates in Chapter 7, which brings together the developments of previous chapters to present the evaluation of the 6DOF robotic system for use in frameless and maskless SRS. The chapter first presents an initial investigation which uses the robot for real-time motion compensation in 6DOF, before proceeding to evaluating the system with volunteers. Chapter 7 concluded by presenting methods to evaluate the capabilities of the overall head stabilization system, as well as the results of a study which examined the dose coverage improvements of using 6DOF robotic stabilization in a single-isocenter multi-target SRS regime.

Taken together, the chapters of this dissertation presented the development process for a novel solution to improving upon frame-based and frameless SRS using robotics. Incorporating a highly precise 6DOF parallel kinematics robot into frameless and maskless SRS has the potential to improve target coverage, reduce therapist intervention, improve the clinical workflow, and minimize patient discomfort.

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