

An Examination of Carpal Joint Function using Kinematic Analysis and Predictive
Modeling

By

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BS, Pennsylvania State University, The Behrend College, 2004

Thesis

Submitted in partial fulfillment of the requirements for the
degree of Doctor of Philosophy in Biomedical Engineering at Brown University

Providence, Rhode Island

May, 2012

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Acknowledgements

My experiences as a graduate student at Brown have been some of the most rewarding of my life. I have had the privilege of interacting with some of the most talented and gracious people in the field, and I have made many life-long friends inside and outside the lab. I would like to thank my colleagues and friends in the bioengineering lab. Despite different advisors and projects, my fellow lab mates were always willing to lend their expertise. I expect that I will continue to rely on their expertise throughout my career and their friendship throughout my life.

I am very grateful to my advisor, Dr. Trey Crisco. He was always there to provide feedback and guidance. Despite his busy schedule, the door to his office was always open. I also appreciate the freedom and independence he gave me to explore my own thoughts and ideas. It was under his guidance where I transitioned from being an engineer, interested in science, to a scientist with an engineering background.

I would also like to thank Dr. Robin Kamal, a resident in the Orthopaedic Department at Rhode Island Hospital, for the countless hours of phone and g-chat conversations debating and discussing wrist joint function. His input and feedback were crucial to every aspect of this thesis.

Doug Moore deserves thanks for his rigorous critiques of my work, and for great conversations both scientific and personal.

I would like to thank my committee members, who have provided guidance and feedback throughout this process. Dr. Braden Fleming gave me perspectives on modeling as a means

rather than an end, and a lot of great advice on grantsmanship. Dr. Thomas Roberts and the rest of the ecology and evolutionary biology group provided me with unique and creative perspectives on approaching biomechanical problems. Dr. David Laidlaw along with his former student, now professor, Dr. Liz Marai provided computational expertise that laid the foundation of this thesis. The studies presented herein would not be possible without the methodologies they have developed.

I will be forever grateful to Dr. Frank Buczek. I can honestly say that if it were not for Frank, I would not be where I am today. He introduced me to the biomechanics world and guided me toward becoming a better scientist and, more importantly, a better person. His words of encouragement have always come back to me whenever I have been frustrated or discouraged.

I am grateful to my family; Dad, Andrew and Megan, Steve, Caleb, and Claire, Tom, Karen, Grace, Rebecca, Sarah, Elizabeth, and Jake. My siblings are not just family, they are also my best friends. You have all been very supportive throughout this process and throughout my life.

Lastly, I would like to thank my fiancé, Dr. Roshni Patel. She has been extremely helpful and supportive throughout this entire process. She has done everything from patiently listening to me go on about the wrist, to offering me her scientific expertise, to keeping me company while I work. I'm very thankful to have her in my life.

Dedication

This work is dedicated to my mother, Christine Rainbow, whose passion for life continues to inspire me.

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Chapter 1

Introduction

Overview

As a versatile joint, the wrist remains stable through large motions under a variety of loading conditions. Wrist motion is achieved through the individual articulations of eight small carpal bones situated distal to the radius and ulna and proximal to the metacarpals. With minimal direct muscular attachments, the carpus is a passive joint. From an engineering perspective, it can be modeled as a mechanical linkage composed of multiple rigid bodies and a system of elastic elements. The carpal bones transmit force through tensile strain in the ligaments and through compressive strain in the articular cartilage. These forces facilitate the positioning of the metacarpals with respect to the forearm; the direct measurement of these cartilage and ligament forces would fully elucidate the function of the joint. However, the carpus is part of a living biological system and assembling a mechanical model is associated with a unique set of challenges.

The aim of this dissertation is to extend our understanding of carpal joint function with special emphasis on the carpal ligaments. The following Chapter provides background information on the carpus, focusing on functional anatomy, a brief history and discussion of related research, and specific aims designed to address unanswered questions about the function of the carpal joint.

1.1 Background

It is common to use the terms *wrist* and *carpus* interchangeably. Herein, *wrist* motion will specifically refer to motion of the metacarpals with respect to the forearm, while motion of the *carpus*, or carpal motion, will refer to the motion of the carpal bones.

1.1.1 Functional Anatomy of the Carpal Bones

The wrist joint facilitates motion and transmits force between the hand and the forearm through a complex set of interactions between the five metacarpals, eight carpal bones (the carpus), and the radius and ulna (Figure 1.1). The wrist is often simplified as a two-degree-of-freedom universal joint that moves in flexion, extension, ulnar deviation, or radial deviation (Figure 1.2). Traditionally, the carpal bones are organized into functional groups that attempt to explain the mechanics of these orthogonal motions.

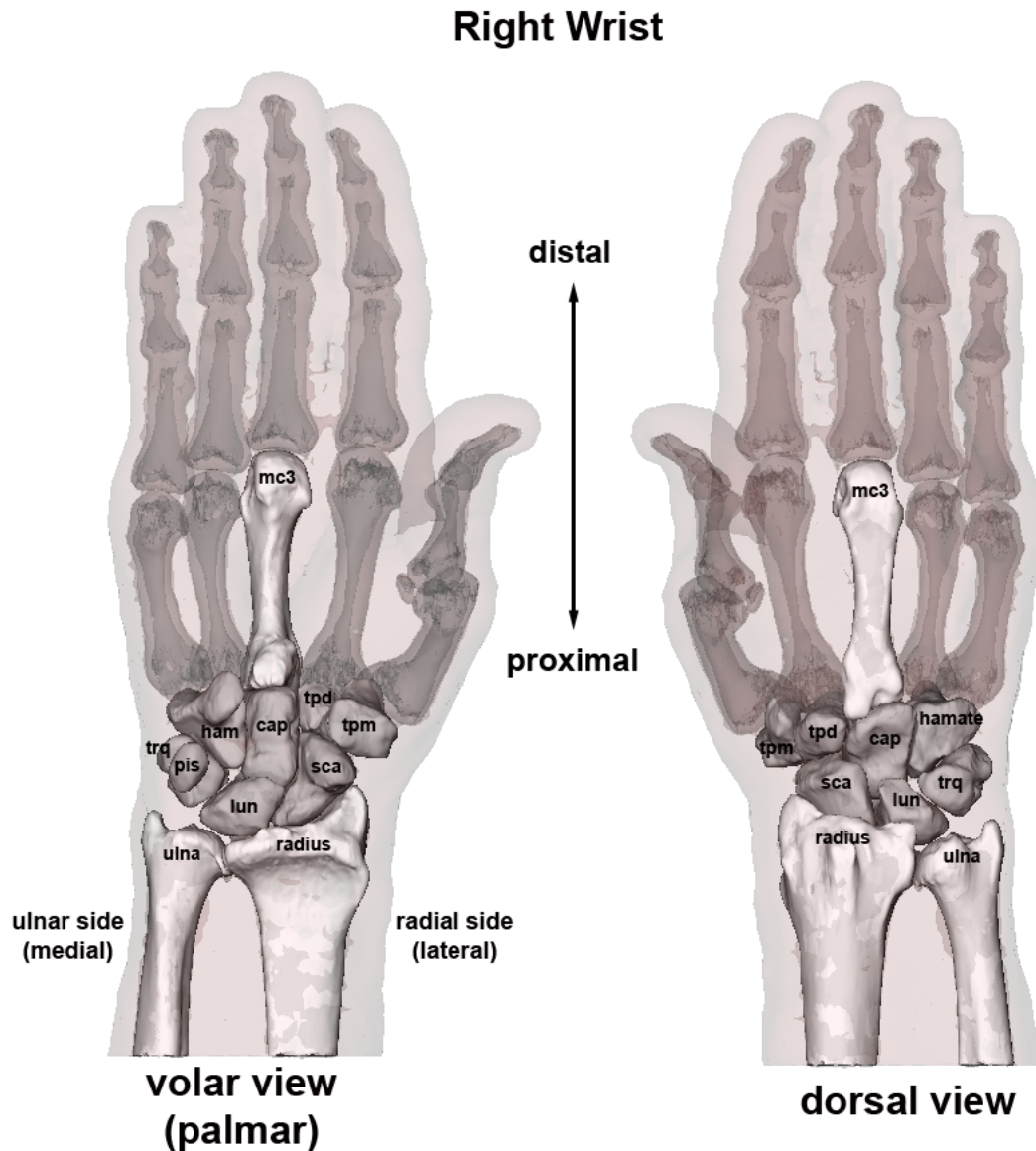


Figure 1.1: The wrist facilitates the positioning of the digits and load transfer between the metacarpals and the forearm. Wrist position is measured by the position of the third metacarpal (mc3) with respect to the radius. The wrist joint is composed of eight small carpal bones (carpus); the scaphoid (sca), lunate (lun), triquetrum (trq), trapezium (tpm), trapezoid (tpd), capitate (cap), and hamate (ham). The pisiform (pis) is a sesmoid bone for the flexor carpi ulnaris.

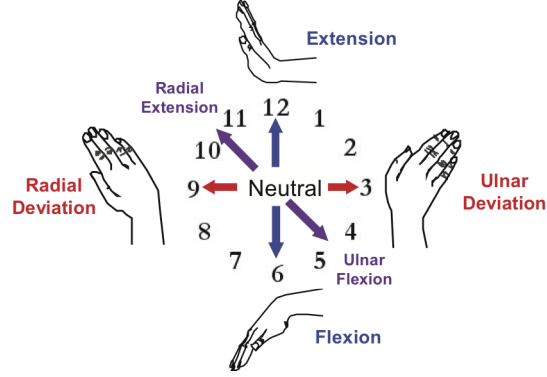


Figure 1.2: The wrist is approximated as a two-degree-of-freedom universal joint. Motion is defined in terms of two orthogonal anatomical axes; the flexion-extension axis and ulnar-radial deviation axis. The wrist more commonly moves along paths that are oblique to these orthogonal motions.

1.1.1.1 Row Theory

Originally, the carpus was organized into proximal and distal rows (Figure 1.3) [1]. The proximal row, referred to as the intercalated segment, consists of the scaphoid, lunate, and triquetrum. The distal row is very tightly bound to the metacarpals and consists of the trapezium, trapezoid, capitate, and hamate. It was noted that the scaphoid functions as the bridge between the rows because it sometimes moves with the distal row and other times with the proximal row [2]. Early studies of carpal function posited that wrist flexion and extension occurred between the two rows at the midcarpal joint, while ulnar and radial deviation occurred through the articulation of the scaphoid with the radius at the radiocarpal joint [1, 3].

1.1.1.2 Column Theory

Other studies suggested that the row theory was an oversimplification of the carpal joint mechanism. Instead, they proposed that the carpal bones should be conceptualized as radial, central, and ulnar columns (Figure 1.3). The radial column consists of the scaphoid, trapezium, and trapezoid, the central column consists of the capitate, hamate, and lunate, and the ulnar column originally consisted of the triquetrum and pisiform [4]. Later, the

pisiform was excluded from the ulnar column because it is a sesmoid bone [5]. The column theory states that wrist flexion and extension occurs through articulations in the central column and wrist radial and ulnar deviation occur through rotation of the scaphoid and triquetrum about the central column.

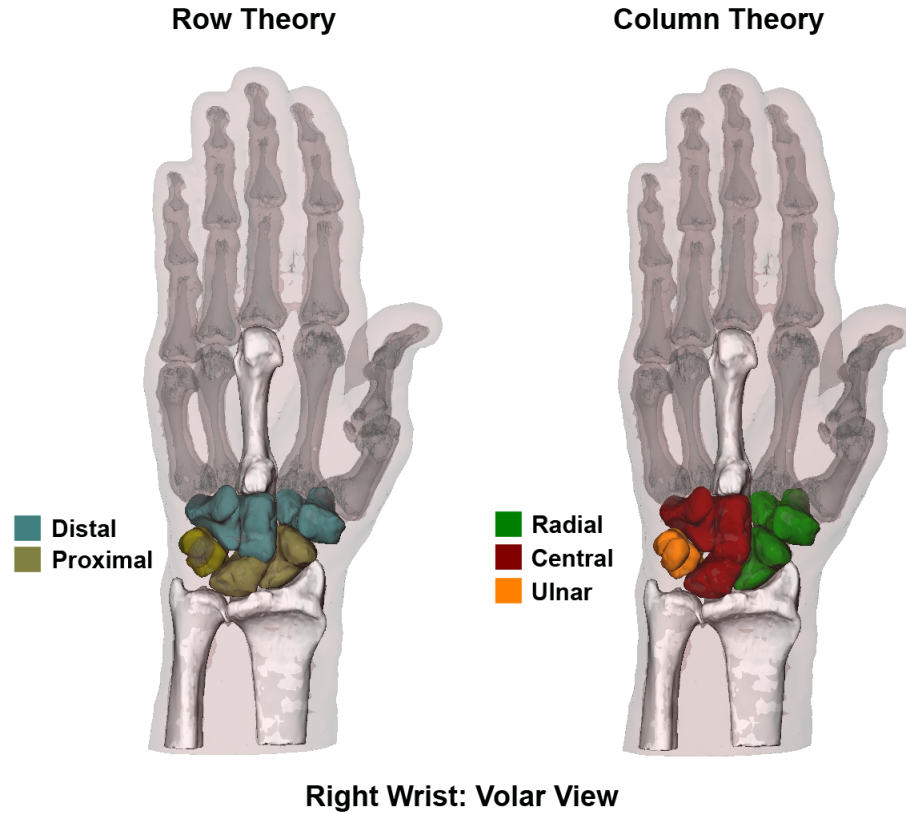


Figure 1.3: The row and column theory of carpal joint function. The row theory organizes the carpus into proximal and distal rows. It states that wrist flexion and extension occur between the two rows while wrist radial and ulnar deviation primarily occurs at the articulation of the scaphoid and radius. The column theory organizes the carpal bones into a radial, central, and ulnar column and states that wrist flexion and extension occur about the central column while wrist radial and ulnar deviation occurs by rotations of the radial and ulnar column about the central column.

1.1.1.3 Row or Column

The column theory of carpal joint function has been revisited and revised several times. Two-dimensional radiography studies postulated that there is a continuum of wrists from a row-type wrist to a column-type wrist [6, 7]. One study found a strong correlation between

clinically measured joint laxity and scaphoid extension with wrist ulnar deviation [7]. They concluded that during wrist ulnar deviation, a person with lax ligaments exhibits more scaphoid extension (column-type wrist), whereas a person with stiffer ligaments exhibits more scaphoid ulnar deviation (row-type wrist).

1.1.2 Three-Dimensional Carpal Bone Kinematics

The row and column theories were developed using observations from two-dimensional studies of carpal motion. Three-dimensional studies using either magnetic resonance imaging (MRI) [8] or computed tomography (CT) [9] are capable of assessing the six-degree-of-freedom kinematics of the carpal bones *in vivo*. The accuracy of these methods is on the order of 0.5 mm and 0.5° [10]. These technologies significantly advanced the understanding of carpal bone kinematics because they were not susceptible to artifacts due to out-of-plane rotations, as is the case with two-dimensional studies. Studies using these techniques have refined the row and column theory by providing detailed descriptions of carpal bone rotations that are either in or out of the plane of wrist rotation. Additionally, they have determined that the proximal carpal bones rarely move as a single rigid body, even when they are rotating in the same plane as the wrist. For example, as the wrist moves from neutral to 60° of flexion the scaphoid flexes approximately 70% the amount the capitate flexes, while the lunate flexes 45% of capitate flexion. From neutral to 60° of extension, the scaphoid extends with the capitate while the lunate extends approximately 65% of capitate extension [8, 11-15].

In vitro and *in vivo* studies reported carpal bone rotation while the wrist rotates about the traditional anatomical axes (flexion-extension and ulnar-radial deviation) [15, 16], and through oblique axes that combine these motions [14, 16, 17]. One path, referred to as the dart thrower's path, has received special attention in the literature [11], [17-20]. The dart thrower's motion (DTM) is defined as the path from wrist extension with radial deviation toward wrist flexion with ulnar deviation. The DTM defines the transition between scaphoid and lunate flexion with wrist flexion and scaphoid and lunate extension with wrist ulnar

deviation. During this transition the rotation of the scaphoid and lunate is minimized and the majority of wrist motion occurs at the midcarpal joint [17, 21].

1.1.2.1 Active and Passive Wrist Motions

The hand can function actively or passively. Most studies have analyzed carpal bone kinematics during activities, when the wrist is actively flexed or extended via its forearm musculature [8, 15, 19] Few *in vivo* studies have examined carpal bone posture where the hand is moved passively by an external force [22, [23]. In this situation, wrist position is defined by the position of the proximal segments and the externally applied load.

1.2 The Carpal Ligaments

The complex three-dimensional motions of the carpal bones depend on the morphology of the bones and an intricate web of ligaments. The exact functional role of the carpal ligaments, and their impact on maintaining normal kinematics in the wrist continues to be studied. Pathologic disruption of this complex system may result in predictable patterns of wrist instability that can lead to secondary degenerative joint disease [24]. While surgical procedures attempt to restore normal wrist kinematics, or rely on specific wrist ligaments for post-reconstruction wrist stability, a complete understanding of the kinematics and kinetics of the carpal bones and the intrinsic and extrinsic ligaments is needed to better understand wrist pathologies and improve potential surgical treatments.

1.2.1 Ligament Anatomy

The carpal ligaments are described as thickenings in the surrounding fascia (Figure 1.4a) [25]. Because the carpal ligaments are part of a confluence of tissues, the identification and isolation of a specific ligament is subject to large variability among specimens and observers. As a result there has been controversy in the identification and classification of the carpal

ligaments [26, 27]. Still, meticulous cadaver dissection studies have been performed to categorize, describe, and determine the function of these ligaments (Figure 1.4b) [25, 28-32]. Recently, methods of dissecting and reporting ligament structures have been extended to include three-dimensional databases of ligament insertion regions located on 3-D bone surface models [33-35].

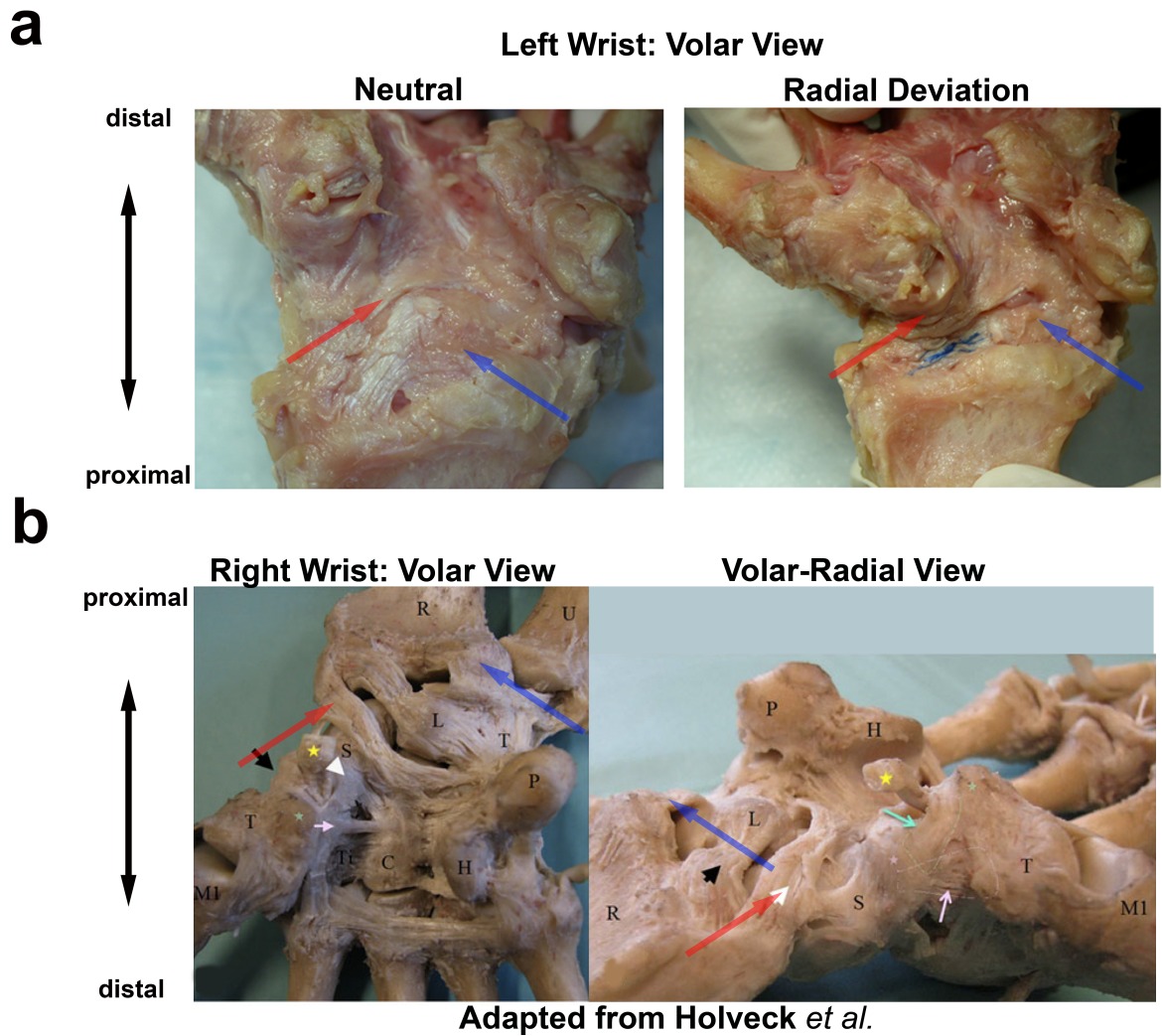


Figure 1.4: Wrist dissections illustrating (a) the confluence of tissues that make up the carpal ligaments, and (b) an osseoligamentous dissection of the carpus where the surrounding fascia has been removed [32]. The red arrow denotes the radioscaphocapitate ligament (RSC). Note that fibers of the RSC can be discerned from the surrounding tissue in both dissections. The blue arrow indicates the short radiolunate ligament (SRL). It is much more difficult to discern the SRL ligament from the surrounding fascia in (a). In radial deviation and slight flexion (top right), the RSC ligament is crimped, indicating that it is slack in this position.

The wrist ligaments are grossly categorized as either dorsal or volar. They are also named according to their depth and the nature of their attachments (Figure 1.5). The extrinsic carpal ligaments are within the most superficial layer and insert on the radius or ulna and one or more carpal bones. With the exception of the dorsal intercarpal ligament, which inserts on the scaphoid and triquetrum, the intrinsic ligaments are deep to the extrinsic ligaments and insert only on the carpal bones. The interosseous ligaments are within the deepest layer. The interosseous ligaments are described as stout ligaments connecting two adjacent bones. These ligaments are named by their location and nature of their attachments [25].

1.2.1.1 The Dorsal Ligaments of the Wrist

The dorsal ligaments consist of the dorsal intercarpal ligament (DIC) and the dorsal radiocarpal ligament (DRC) [36] which together form a lateral “V” shape, originating on the triquetrum and inserting on the scaphoid, and originating on the radius and inserting on the triquetrum respectively. Although these ligaments have been characterized as less substantive when compared to their volar counterparts, [28] recent studies have suggested that they play an important role in scapholunate stability and lunotriquetral stability [37, 38].

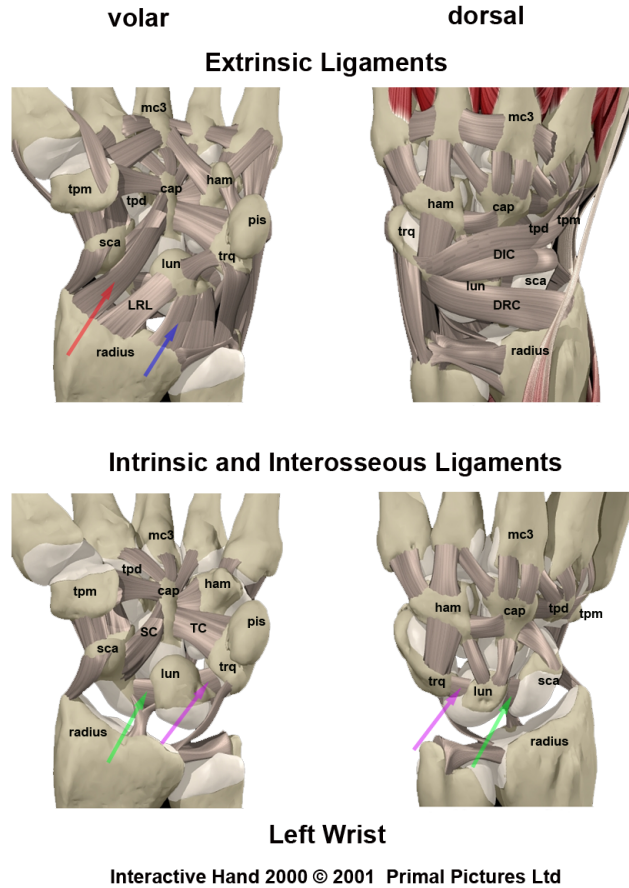


Figure 1.5: An artistic rendition of the carpal ligaments. The carpal ligaments are organized into layers. The extrinsic ligaments are within the most superficial layer (top). The intrinsic and interosseous ligaments (bottom) are deep to the extrinsic ligaments. The red and blue arrows denote the RSC and SRL ligaments defined in figure 4. The light green and purple arrows denote two very important interosseous ligaments, the scapholunate interosseous ligament (SLIL) and lunotriquetrum interosseous ligament (LTIO) respectively. Several other important ligaments are labeled (capitate letters). The long radiolunate ligament (LRL) (top left) is considered to be an important volar extrinsic ligament. The dorsal radiocarpal (DRC) and dorsal intercarpal (DIC) ligaments (top right) support the dorsal side of the wrist. The arcuate ligament (bottom left), made up of the scaphocapitate (SC), and triquetrocapitate (TC) ligaments, is thought to be an important stabilizer of the carpus.

1.2.1.2 Ligament Mechanical Properties

Reports of stiffness and strain vary among mechanical studies of the carpal ligaments [37, 39-43]. For example, one study measured the stiffness of the scapholunate interosseous ligament to be 66 N/mm [41], while another measured 250 N/mm [40]. Likewise failure strains of the extrinsic ligaments range from 20% to 50% between studies [26, 43]. Failure

strains of the interosseous ligaments are reported to be as high as 200% [43]. These values are much higher than other skeletal ligaments. For example, ligaments in the knee fail at 12-15% strain [44]. Aside from the inherent difficulty in identifying and isolating the carpal ligaments, differences in displacement rates, the definition of resting length (l_o), and testing protocol are sources for a lack of consensus. Additionally, anatomic studies have shown that there is variability amongst people in the insertion location of several ligaments, particularly the ulnocarpal and dorsal ligaments [33, 37].

Despite the wide variability of reported stiffness and failure strain values, one important pattern has emerged. The extrinsic carpal ligaments are consistently reported to be stiffer and weaker than the stout interosseous ligaments [26, 45, 46]. In addition to reported failure strain and load to failure values that are higher than the extrinsic carpal ligaments, one study reported that the interosseous ligaments are made up of 41% type-3 collagen compared to 19% in the extrinsic ligaments [47]. The authors indicated that type-3 collagen is found in other elastic structures such as blood vessels.

1.2.1.3 Ligament Elongation –Ligament Anatomy Combined with Three-Dimensional Carpal Kinematics

In the absence of the ability to directly measure ligament forces as a function of wrist motion, several studies have alternatively measured ligament length changes [48-53]. De Lange *et al* embedded radio-opaque markers directly into the ligament fibers and the carpal bones. They tracked the change in length of ligament fibers and the change in distance between ligament insertion sites using Roentgen stereophotogrammetry. They reported that tracking insertion site distances alone did not accurately reflect the true length changes of a ligament fiber and the path of the fiber should be considered [48]. For example, the radioscapophcapitate and long radiolunate ligaments underwent significant length changes due to deflection from the scaphoid, creating a situation where the insertion site distances decreased or remained constant while the ligament fiber increased in length. Marai *et al* developed an algorithm for computing ligament fiber length changes *in vivo*. The algorithm computed the shortest

path between two insertion sites with the constraint that bone surfaces were not penetrated [54]. Several studies have utilized these methods to measure length changes of the extrinsic carpal ligaments as the wrist moves along specific paths of motion [51, 52]. These methods are able to determine which wrist motions alter the fiber path such that the fiber elongates, shortens, or remains unchanged. However, they cannot determine whether a ligament is strained or slack because the resting length of the ligament cannot be directly measured with these methods.

1.2.1.4 Ligament Function

When a stiff ligament transitions from a slack state to a taut state, it provides passive restraint with minor changes in strain (on the order of 5%). Therefore, a slack ligament that becomes taut along a specific direction of motion will restrict further motion along this path. Once taut, a ligament will allow large rotations, while undergoing small length changes, if the bones pivot around its points of insertion.

Studies of ligament elongation have used this concept to classify function [55]. A ligament that undergoes a large length change along a specific path of motion is assumed to be slack during most of the motion, acting as an end-range limiter of that motion. In contrast a taut ligament that undergoes modest or no length changes over large rotations is assumed to be acting as guide along that motion path.

The extrinsic carpal ligaments have been proposed to function as limiters of end-range wrist motion [56]. However, this hypothesis has not been tested because the lengths of the carpal ligaments at the end range of motion have not been determined.

With multiple rotational and translational degrees of freedom, it is unlikely that a given carpal ligament functions exclusively as either an end-range limiter or as a guide through a large range of motion. It is more likely that the carpal ligaments function as both end-range limiters in one direction of wrist motion and guide motion through a large range of a different direction of wrist motion.

1.2.2 The Mechanism of Carpal Joint Motion

Several studies have considered all of the available data on carpal anatomy, kinematics, mechanical properties, and ligament elongation patterns in an attempt to identify a unifying theory of the carpal joint mechanism [5, 46, 56-58]. The main theories are summarized as follows.

1.2.2.1 The Carpus as a “Screw-Clamp” Mechanism

Macconail described the carpus as a “screw-vice” or clamp where the wrist bones of the proximal row are drawn together with wrist extension. He theorized that this mechanism occurs in two stages as the wrist extended [5]. First, the scaphoid and capitate are drawn together by taut ligaments causing them to move as a single unit. Second, the triquetrum and lunate “twist” along a male screw surface formed by the capitate and hamate, drawing them into the scaphoid. He also stated that as the wrist moves from radial to ulnar deviation, the forces from the forearm musculature are communicated differentially to the scaphoid, lunate, and triquetrum, in turn causing them to move differentially and become less closely packed.

1.2.2.2 The Triquetrum Hamate Articulation

Weber posited that the articulation between the triquetrum and hamate indirectly controlled the movement of the scaphoid through the lunate [57]. The triquetrum was described to move dorsal and volarly on the convex surface of the hamate as the wrist moved in flexion-extension or radial-ulnar deviation. This movement imparted a flexion or extension moment on the lunate through the lunotriquetral interosseous ligament. The flexion moment was amplified by the articulation of the lunate with the capitate and then transmitted to the scaphoid via the scapholunate interosseous ligament. When the wrist extended or ulnar deviated, the moment caused the scaphoid to extend. When the wrist flexed or radial

deviated, the moment caused the scaphoid to flex. The trapezium and trapezoid articulation was described as accommodating these movements.

1.2.2.3 The Longitudinal Chain Theory

The longitudinal chain theory shifts focus away from the triquetrum-hamate joint toward the radial and central columns [56]. The interosseous ligaments and articular contacts are considered drivers of carpal bone motion and the extrinsic ligaments are considered check-reins at the end of the range-of-motion. In this model, the scaphoid and lunate are described as having a wedge shape that tapers from volar to dorsal. Because of this wedge shape, proximally directed forces from the distal row create a tendency for the scaphoid and lunate to extend. The trapezium and trapezoid create a flexion moment on the scaphoid that stops it from extending. When the wrist moves in radial deviation or flexion, the scaphoid flexes due to contact from the trapezium and trapezoid. The scaphoid causes the lunate to flex in turn, despite its tendency to extend. When the wrist extends or ulnar deviates, the scaphoid rotates farther than the lunate by virtue of the higher curvature of the scaphoid fossa of the radius compared to the lunate.

1.2.2.4 Predictive Models of the Wrist

Even with guidance from quantitative studies of carpal kinematics, ligament elongation patterns, and mechanical properties, published descriptions on the carpal joint mechanism are not able to prove their assertions because the forces acting on the carpus have not been directly measured. Emerging predictive modeling approaches may be able to bridge the gap between quantitative studies that focus on one aspect of carpal joint function and unifying theories on the carpal joint mechanism [23], [59-62].

Predictive models are able to simulate the motion of a bone based on the forces and moments imposed on it by the surrounding soft tissues. These models could augment published descriptions of the carpal joint mechanism by quantifying the relative contributions of the cartilage-contact and ligament forces to carpal bone motion.

Marai *et al* developed predictive modeling methods [54, 60, 63] that are built on a foundation of *in vivo* carpal bone surfaces and kinematics [9, 10]. They estimated carpal cartilage and ligament forces, applied gravity, and showed that the scaphoid remained stable [60]. Fischli *et al* used a similar model to simulate wrist motion in a single cadaver specimen [62]. These studies have provided a foundation for predictive carpal modeling. However, *in vivo* motion has not yet been modeled in multiple subjects. The application of these models to a large database of *in vivo* motion may provide further insight into the carpal mechanism and provide a better understanding of the strengths and limitations of these models.

1.3 Significance

The wrist joint is subject to both acute and chronic trauma in patients of all ages, genders, and ethnicities and accounts for significant medical expenses, lost wages, and decreased quality of life each year. Activities of daily living require at least 105° of flexion-extension and 40° of ulnar-radial deviation [60]. Sufficient grip strength is also required to perform many activities of daily living [64]. Acute and chronic wrist injuries limit both range of motion and grip strength, impairing normal function. Acute injuries cause ligament tears, cartilage damage, and fracture. Even after treatment of these injuries, altered kinematics can lead to a limited range of motion, pain, and degenerative joint disease [7, 8].

Descriptive kinematic and ligament elongation studies provide insight into the mechanics of the normal and pathological wrist, however further understanding of the elongation of the carpal ligaments during active and passive motions are required to provide physicians with the best treatment options. Additionally, this work provides a foundation for the creation of a predictive carpal model (PCM) that is capable of accurately predicting the position and orientation of one or more carpal bones. This will provide new insight into the mechanics that cause the observed kinematics, and better inform surgical treatments focused on recreating normal mechanics. The model may also be used to examine injury modes of the carpus, such as by simulating ligament tears, and then calculating the resulting contact forces, ligament forces, and carpal kinematics.

1.4 Specific Aims and Chapter Summary

The following Chapters present kinematic analyses of the carpal bones and the elongation of specific ligaments. These analyses were performed using data collected during active and passive wrist positions. A predictive model was then constructed, tuned, and validated with the kinematic dataset. This model was used to determine the relative contribution of contact and ligament forces that cause the observed motion of the scaphoid. These goals were accomplished through the following specific aims.

Aim 1 - Describe ligament elongation of the dorsal carpal ligaments throughout the full range of active wrist motion

The objective of aim 1, as described in Chapter 2, was to compute ligament fiber lengths of the dorsal intercarpal and dorsal radiocarpal ligaments in 28 wrists. Each wrist was placed in 9 positions that sampled the range of active motion. A hierarchical linear model identified the paths of wrist motion that elongated the dorsal carpal ligaments over the range of motion.

Aim 2 - Describe carpal bone kinematics and ligament elongation patterns of the wrist through active and passive motions

The first objective of aim 2, as described in Chapter 3, was to evaluate the kinematic patterns of the capitate, scaphoid, and lunate at the extremes of passive wrist flexion and extension in the wrists of 12 subjects and compare those patterns to active motion. Kinematics from neutral-grip to extreme flexion and from neutral-grip to extreme extension were described using the helical axis of motion. Kinematic patterns were identified by modeling the flexion or extension rotation of the scaphoid and lunate as a function of capitate flexion or extension with linear regression. Novel joint space metrics evaluated the arthrokinematics of the radiocarpal and midcarpal joints in neutral-grip and at extreme flexion and

extension. The second objective of aim 2, as described in Chapter 4, was to compute ligament fiber recruitment over active and passive wrist motions. We determined differences between fibers within each ligament and evaluated whether a ligament was within 95% of its maximum length in each wrist position.

Aim 3 - Develop a subject-specific predictive carpal model that is capable of simulating scaphoid motion

The objective of aim 3, as described in Chapter 5, was to tune a predictive carpal model by optimizing ligament stiffness values through multiple simulations in the wrists of 12 subjects. During each simulation, rotational error was computed as the difference between simulated scaphoid kinematics and the experimental scaphoid kinematics that were analyzed in Aim 2. One hundred simulations were performed for each wrist while ligament stiffness values were varied. Optimal ligament stiffness values were those that minimized the error between simulated and experimental scaphoid kinematics. These optimal stiffness values were computed at the subject level. The model was validated by using these optimal stiffness values to evaluate the rotational error between simulated and experimental scaphoid kinematics in a wrist motion that differed from the motion used to tune the model.

Aim 4 - Estimate the contributions of cartilage and ligament forces associated with scaphoid motions that cause the scaphoid to extend or ulnar deviate when the wrist moves along the dart thrower's path

The objective of aim 4, as described in Chapter 6, was to identify the contributions from the cartilage and ligaments to the flexion and ulnar deviation moment of the scaphoid as the wrist moved through five positions along the dart thrower's path. The carpal bones that interacted with the scaphoid through cartilage contact or ligamentous attachment were driven through their experimental kinematics while the kinematics of the scaphoid were simulated according to estimated cartilage contact and ligament forces. The scaphoid was treated as a free body and its resultant moment was deconstructed into its component parts. The

component parts were the cartilage and ligament contributions to the net moment, individual cartilage contact contributions to the net moment from contact, and the individual ligament contributions to the net ligament moment. The moments were normalized by the absolute value of the maximum net, cartilage-contact, or ligament moment. The normalized moment contributions represent general trends; therefore, qualitative, rather than quantitative, descriptions were provided.

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Chapter 2

Elongation of Dorsal Carpal Ligaments: A Computational Study of *In Vivo* Carpal Kinematics

The following chapter is in revision for publication as a full manuscript by Rainbow MJ, Crisco JJ, Moore DC, Kamal RN, Laidlaw DH, Akelman E, and Wolfe SW to The Journal of Hand Surgery (February 2012).

Abstract

Purpose: The dorsal radial carpal (DRC) and dorsal intercarpal ligaments (DIC) play an important role in scapholunate and lunotriquetral stability. A crucial step towards determining *in situ* strain is the measurement of how ligaments elongate as a function of wrist position. The purpose of this study was to compute changes in ligament elongation as a function of wrist position for the DRC and the scaphoid and trapezoidal insertions of the DIC.

Methods: A computational model was developed that incorporated a digital dataset of ligament origin and insertions, bone surfaces models, and *in vivo* 3-D kinematics ($n = 28$ wrists), and an algorithm for computing ligament fiber path.

Results: The total changes in length of the DRC, DIC scaphoid component, and DIC trapezoidal component over the entire range of motion were 5.1 ± 1.5 mm, 2.7 ± 1.5 mm, and 5.9 ± 2.5 mm respectively. The DRC elongated as the wrist moved from ulnar-extension to radial-flexion and the DIC elongated as the wrist moved from radial deviation to ulnar deviation.

Conclusions: The DRC and DIC lengthened in opposing directions during wrist ulnar and radial deviation. Despite complex carpal bone anatomy and kinematics, computed fiber elongations were found to vary linearly with wrist position. Errors between computed values and model predictions were less than 2.0 mm across all subjects and positions. Clinical Relevance: The computed relationships between ligament elongation and wrist position should further our understanding of ligament function, provide insight into the potential effects of dorsal wrist incisions on specific wrist ranges of motion, and serve as a basis for modeling of the wrist.

2.1 Introduction

The wrist is a unique “joint” because the motions of its component carpal bones are determined almost exclusively by cartilage contact and ligament forces rather than by direct muscular attachments. Many surgical interventions attempt to restore ligament function [1-2], yet little is known about the *in vivo* mechanical behavior of the ligaments. One set of ligaments that has not been extensively studied *in vivo* are the extrinsic dorsal carpal ligaments. The dorsal ligaments consist of the dorsal intercarpal ligament (DIC) and the dorsal radiocarpal ligament (DRC) [3] which together form a lateral “V” shape, originating on the triquetrum and inserting on the scaphoid, and originating on the radius and inserting on the triquetrum respectively. Although these ligaments have been characterized as less

substantive when compared to their volar counterparts, [4] recent studies have suggested that they play an important role in scapholunate stability and lunotriquetral stability [5-7].

Determining how a ligament elongates as a function of wrist position is a crucial step in understanding ligament function. Much of the previous work on carpal ligaments has focused on mechanical testing and anatomical descriptions of the ligaments and their origin and insertion sites [3-4, 6, 8-9]. Measuring ligament length changes as a function of wrist position in living subjects is a challenge because current imaging modalities are limited in their ability to simultaneously resolve bone surfaces, kinematics, and soft tissue morphology. The current state of the art for *in vivo* imaging of the carpal ligaments is limited to identification of major ligaments and pathological conditions [10] [11].

Computational models can be a powerful tool for evaluating wrist kinematics and ligament function. These models have previously provided insights into pathology of the distal radioulnar joint [12]. Additionally, they have determined which wrist motions are likely to cause the carpal ligaments to become taut [13-14]. The elongation of the DRC and DIC have been previously studied along specific paths of ulnar-radial deviation [15] and a dart thrower's path [16] using this approach. The purpose of this study was to determine the relationship between fiber elongation and wrist position of these two important dorsal carpal ligaments through the full range of wrist motion.

2.2 Methods

2.2.1 Digital Database of Carpal Bone Surfaces and Kinematics

After IRB approval and informed consent, both wrists of 8 healthy males (mean age 24.4; range 22-26) and 6 healthy females (mean age 24.7; range 21-28) underwent computed tomography (CT) in nine wrist positions [17]. Both the right and left wrist were scanned simultaneously on a GE HiSpeed Advantage CT scanner (GE Healthcare Technologies, Waukesha, Wisconsin) with settings of 120kVP and 80mAs, in plane resolution of 0.9 X 0.9

mm² and slice thickness of 1.0 mm. The bone surfaces for each subject were obtained via segmentation of the CT volume (ANALYZE™, Mayo Clinic Foundation, Rochester, MN), and then tessellated using GEOMAGIC STUDIO (Raindrop Geomagic, Research Triangle Park, NC). Kinematic transformations from the neutral wrist position to the remaining positions were calculated using a previously reported markerless bone registration technique [18-19]. Briefly, the technique first identified tissue-classified voxels associated with the segmented point cloud of each carpal bone in the neutral position. Then, the algorithm optimized the fit between the neutral position voxels and voxels in each subsequent scan to compute the change in bone posture with sub-voxel accuracy [19]. Actual wrist position was defined by the position of the capitate with respect to the radius [20]. The targeted wrist positions of neutral, 40° flexion (F), 40° extension (E), 10° radial deviation (R), 30° ulnar deviation (U), and the combined motions of 40° E and 30° U, 40° F and 30° U, 40° E and 10° R, and 40° F and 10° R. resulted in a range of wrist motion for all subjects of 34.2° F to 62.2° E and 35.2° U to 26.5° R. The actual range of wrist motion per subject was $28.6^\circ \pm 6.4^\circ$ ulnar-radial deviation (UR) and $68.7^\circ \pm 8.1^\circ$ flexion-extension (FE).

2.2.2 Selection of Ligament Insertion Regions

The DRC and DIC insert on multiple bones and interdigitate with other ligaments. Only the largest portions of the DRC and DIC with insertion areas greater than 4 mm² were modeled in the present study [21]. These included the main band of the DRC, which originates on the dorsal medial ridge of the radius, passes over the lunate and inserts onto the dorsal pole of the triquetrum; and the main bundles of the DIC, which originate on the dorsal pole of the triquetrum and passes over the hamate, lunate, and capitate and inserts onto the dorsal ridge of the scaphoid (DIC_S) and trapezoid (DIC_T) [3]. These attachments of the DRC, DIC_S and DIC_T were reported to occur in 100%, 97%, and 42% of people respectively [6]. The regions of insertion were mapped to each subject’s bone surface model (BSM) using a template created in a previous study via careful digitization of insertion sites on cadaver bones which were then registered to the corresponding BSM [22]. Insertion site mapping was performed manually using GEOMAGIC with both models opened simultaneously for

visual comparison (Figure 2.1).

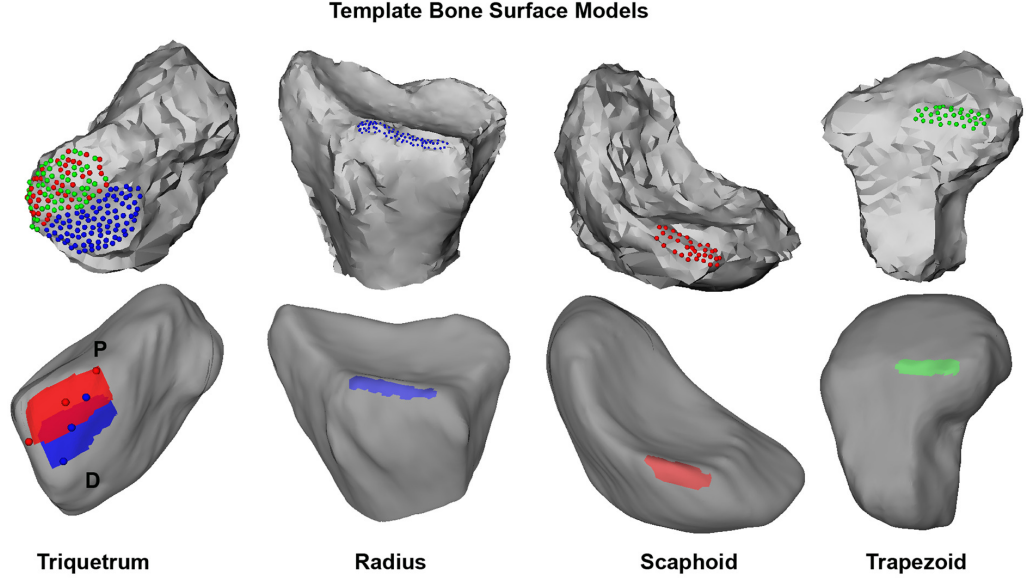


Figure 2.1: The template insertion BSM (top) with points representing the insertion regions for the DRC (blue), DIC_S (red), and DIC_T (green), and a representative subject's BSM with polygons selected to match the template insertion regions.. The points on the subject's triquetrum represent the three fibers analyzed; the most proximal point located on the perimeter, the most distal point, and a point in the center of the insertion region.

2.2.3 Ligament Fiber Generation

The length of a ligament fiber was defined as the distance along its path. The fiber path was computed as the shortest distance between insertions under the constraint that the BSM was not penetrated (Figure 2.2) [23]. A central fiber was defined between the central points of corresponding attachment regions. Two additional fibers were also defined using the most proximal and distal aspects of the triquetrum insertion region, and were roughly parallel to the central fiber based on previous qualitative studies [8, 24]. The path of each fiber was computed by creating 39 segments. Fiber length was defined as the sum of these 39 segments. The paths and lengths of the central, proximal, and distal fibers were computed for each subject in each of the 9 widely distributed wrist positions. The total excursion of each ligament was defined as the difference between its minimum and maximum length over all 9 wrist positions.

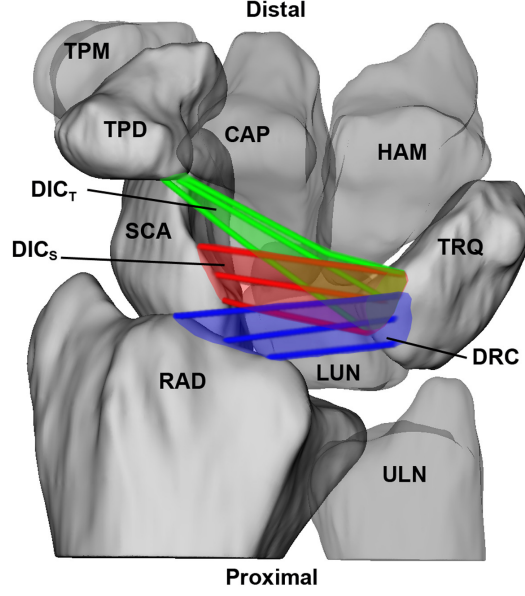


Figure 2.2: Right wrist, dorsal view. The proximal, central, and distal fibers of the dorsal radio-carpal ligament: DRC (blue), scaphoid attachment of the dorsal intercarpal ligament: DIC_S (red), and trapezoidal attachment of the dorsal intercarpal ligament: DIC_T (green) of a subject in a slightly extended wrist position.

2.2.4 Analysis

A hierarchical linear model (HLM) determined the relationship between fiber length and wrist position. The HLM can loosely be conceptualized as a set of subject-specific best-fit plane models with fiber length as the dependent variable and wrist position the independent variable. The neutral length of each fiber was the intercept of the model. The intercept was the fibers's length at 0° FE and UR. The coefficients of the best-fit model, termed elongation rates, quantified the change in fiber length per change in wrist position. The HLM estimated model parameters using a hierarchy consisting of three levels. The first level contained the central, proximal, and distal fibers within each ligament. The second level contained each ligament within a wrist, and the top level contained all wrists within the population. The HLM uses residual estimation of maximum likelihood to simultaneously estimate ligament lengths at each level of the hierarchy while accounting for interactions between levels and within each level [25].

The fidelity of this linear model was reported in terms of the root mean squared error

between the ligament lengths computed from the *in vivo* data set and the lengths estimated by the HLM. The HLM was used to determine the elongation rate along four paths of wrist motion: pure flexion-extension (with ulnar-radial deviation set to 0°), pure ulnar-radial deviation (with flexion-extension set to 0°), the dart thrower’s path (radial-extension to ulnar-flexion), and the anti-dart thrower’s path (ulnar-extension to radial-flexion). The dart thrower’s and anti-dart throwers paths were 45° oblique to the flexion-extension and ulnar-radial deviation paths and passed through 0° FE and UR. The HLM model tested which paths of wrist motion significantly ($p < 0.05$) lengthened each ligament. The HLM also determined if the fibers within a ligament had significantly ($p < 0.05$) different elongation rates.

2.3 Results

There were no statistically significant differences ($p < 0.001$) in elongation rates between the central and both the distal and proximal fibers within each ligament as the wrist moved through its range of motion. Therefore, the results presented herein are reported at the ligament level. The excursions, averaged across subjects, of computed ligament lengths throughout the range of wrist motion were 5.1 ± 1.5 mm, 2.7 ± 1.5 mm, and 5.9 ± 2.5 mm for the DRC, DIC_S, and DIC_T respectively. The average neutral length estimated by the HLM was 16.1 ± 3.7 mm, 21.8 ± 4.2 mm, and 26.0 ± 4.2 mm, for the DRC, DIC_S, and DIC_T respectively. Neutral fiber length of each ligament was found to significantly ($p < 0.05$) increase with an increase in volume of the subject’s capitate bone (Figure 2.3). This size effect was normalized by the HLM in the following analysis. Fiber length varied linearly with wrist position for the DRC, DIC_S, and DIC_T. This linear relationship was supported by low root mean square errors between computed fiber lengths and those estimated by the HLM of 1.0 mm, 1.8 mm, and 1.2 mm over all subjects for the DRC, DIC_S, and DIC_T, respectively.

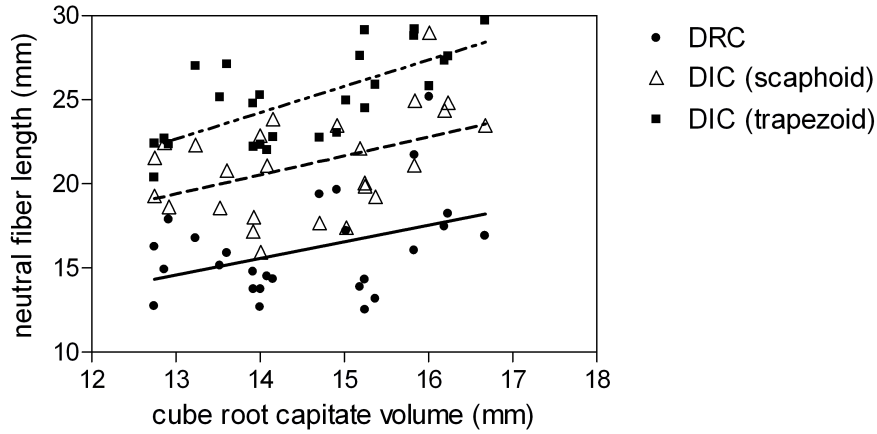


Figure 2.3: Neutral fiber length versus the cube root of capitate volume. Neutral position fiber length increased significantly with increasing wrist size as measured by capitate volume ($P < 0.05$). This size effect is accounted for by the hierarchical linear model (HLM), as it models length changes from neutral of each subject.

2.3.1 Flexion-Extension

The DRC lengthened significantly ($p < 0.0001$) with an elongation rate of 0.51 ± 0.03 mm (standard error of the mean) for every 10° of wrist flexion (Table 2.1). Neither wrist flexion nor extension lengthened the DIC_S ($p = 0.37$). The DIC_T lengthened ($p < 0.0001$) with an elongation rate of 0.23 ± 0.03 mm for every 10° of wrist flexion (Figure 2.4c-d).

2.3.2 Ulnar-Radial Deviation

Wrist radial deviation lengthened the DRC ($p < 0.0001$), while the DIC_S and DIC_T significantly lengthened with wrist ulnar deviation ($p < 0.0001$) (Figure 2.4a-b). The elongation rate of the DRC was 0.78 ± 0.09 mm for every 10° of wrist radial deviation, while the elongation rate of the DIC_S and DIC_T was 0.53 ± 0.09 mm and 1.47 ± 0.09 mm respectively for every 10° of wrist ulnar deviation.

2.3.3 Dart Thrower's Path (wrist radial-extension to ulnar-flexion)

The DRC lengthened with an elongation rate of 0.40 ± 0.08 mm for every 10° of wrist motion along a path from ulnar-flexion to radial-extension ($p < 0.0001$), which is the reverse of the

Table 2.1: Ligament elongations along specific paths of wrist motion predicted by the HLM.

Ligament	Neutral Length (mm (s.c))	mm/degree (standard error)	U (+) R (-)	UF (+) RE (-)	RF (+) UE (-)
		F (+) E (-)			
DRC	16.1 (0.7)	0.051 (0.003)	-0.078 (0.009)	-0.0395 (0.008)	-0.084 (0.004)
DIC _s	22.2 (0.8)	0.001 (0.003)	0.053 (0.009)	0.045 (0.008)	0.027 (0.004)
DIC _T	25.6 (0.8)	0.023 (0.003)	0.147 (0.009)	0.137 (0.008)	0.058 (0.004)

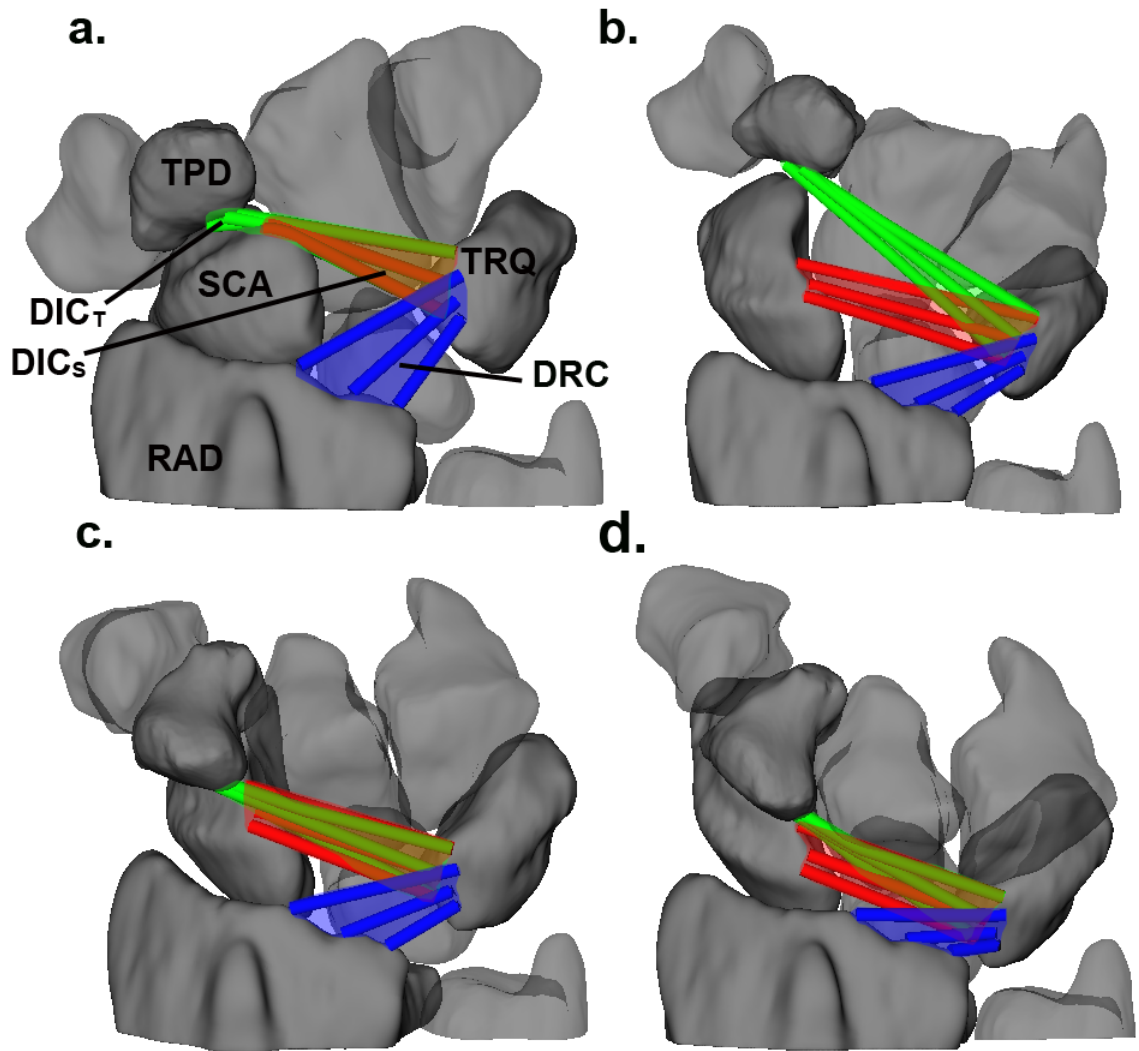


Figure 2.4: Ligament fiber path computed for a representative wrist in a. radial deviation b. ulnar-deviation c. flexion d. extension.

dart thrower's path. The DIC_S and DIC_T lengthened with an elongation rate of 0.45 ± 0.08 mm and 1.37 ± 0.08 mm respectively for every 10° of wrist motion along the dart thrower's path ($p < 0.0001$).

2.3.4 Anti-Dart Thrower's Path (ulnar-extension to radial-flexion)

The DRC lengthened with an elongation rate of 0.84 ± 0.04 mm for every 10° of wrist motion along the anti-dart thrower's path ($p < 0.0001$). Fibers of the DIC lengthened with an elongation rate of 0.27 ± 0.04 mm (DIC_S) and 0.58 ± 0.04 mm (DIC_T) for every 10° of wrist motion along a path from radial-flexion to ulnar-extension ($p < 0.0001$), which is the reverse of the anti-dart thrower's path.

2.4 Discussion

Ligaments may prevent extremes of motion, functioning as check-reins, or guide the carpal bones synchronously through a range of motion. It was proposed that a ligament that elongates through a range of motion functions as a check-rein while a ligament that does not lengthen functions as a guide [26]. Our results suggest that the dorsal carpal ligaments may function differentially as both check-reins and guides depending on the direction of wrist motion. In our model, the DIC elongates when the wrist moves toward ulnar deviation, while the DRC elongates when the wrist moves toward radial deviation. It follows that the DIC and DRC limit motion at the end range of ulnar and radial deviation respectively, acting as check-reins. In contrast, the elongation of the DRC was minimized along a dart thrower's path that had a coupling ratio of 1.5° of flexion-extension for every 1° of ulnoradial deviation. Elongation of the DIC_S was minimized in pure wrist flexion-extension, while elongation of the DIC_T was minimized along a path of flexion-extension and slight ulnar-radial deviation. These paths of minimal elongation would suggest that if the ligament fibers were strained, they could be functioning to guide the carpal bones through these ranges of motion.

When interpreting these data, it is important to note that the length at which a ligament is no longer slack and begins to generate tensile loading cannot be determined with these methods. A related limitation is that slack ligament lengths were presumably included in the HLM model. However, the inclusion of slack lengths in the model should not affect our elongation results because the carpal ligaments transition between slack and taut states in a linear fashion [27]. An additional implication of the linear model is that a ligament fiber that elongates in a specific direction will shorten by the same amount in the opposite direction. Finally, ligament fibers were modeled roughly parallel to one another. While there is qualitative evidence to support this assumption [8, 24], there has been no rigorous analysis of fiber orientation compared to the course of the carpal ligaments.

The elongation range, neutral length, and insertion site areas of the DIC and DRC compared well with previous *in vitro* and *in vivo* studies despite these limitations and large variability between studies [6, 8-9, 28]. The neutral length of the DRC and DIC fell within ranges of (14.7 mm to 17.7 mm) and (13.5 mm to 40.9 mm) reported by cadaver studies [6, 8, 21]. Neutral lengths and elongation of the DRC and DIC compared well with *in vivo* studies that used similar techniques to measure elongation in radial and ulnar deviation and the dart thrower's motion [15-16]. Neutral lengths of the DRC, DIC_S, and DIC_T differed between studies by 1 mm, 2.8 mm, and 1.4 mm respectively and were within the standard deviations of each other. Additionally, the mean insertion site areas were within one standard deviation of those reported in Nagao et al [21]. The average difference between insertion site areas reported in Nagao *et al* and those in the present study was 3 mm². In a cadaver study, Savelberg *et al* [27] injected radio-opaque beads into the carpal bones and selected ligaments and used Roentgen stereophotogrammetry. They found that the DRC lengthened $5.7 \pm 5.0\%$ when the wrist was flexed to $69.9 \pm 14.5^\circ$ but did not elongate in radial or ulnar deviation. Beginning in the clinical neutral position (0° FE, and 0° UR), the elongation of the DRC in our study increased to 22.2% (CI: 21.3% – 23.2%) in a position of 70° F; however, we found that the DRC also elongated more substantially in positions of radial deviation. Savelberg *et al* reported that the DIC elongated $6.4 \pm 4.3\%$ when the wrist was ulnarly deviated to $43.6 \pm 7.2^\circ$ but this value was not statistically significant. Our study found a greater

elongation of the DIC with values of 9.6% (CI: 9.4% – 9.8%) (DIC_S) and 22.6% (CI: 22.5% – 22.8%) (DIC_T) at 40° of ulnar deviation. The consistently higher elongation values found in this and other *in vivo* studies, compared to cadaver studies, highlights the assumption that neutral length occurs at the neutral wrist position (0° FE, and 0° UR).

It was hypothesized that the DRC and DIC work in tandem to support the scaphoid throughout the range of motion [6]. This is possible if the DRC is taut in the same positions that the DIC is taut. Our study suggests that the ligaments may be complementary; while one ligament is elongating, the other is shortening and possibly moving into a slack state. However, we did not model the lunate attachments of either ligament or their direct interactions with one another. Modeling these attachments may reveal an interaction between the DRC and DIC that supports the concept that they work in tandem to support the scaphoid. It is interesting to note that the DIC elongated in ulnar deviation but not in FE, consistent with Macconail’s description of the “screw-home” mechanism of the wrist and the importance of the dorsal intercarpal ligament to proximal row synchrony [29].

Ligament-sparing approaches to the wrist, such as fiber splitting dorsal capsulotomies [1], avoid transection of the dorsal ligaments and minimize interference with wrist motion. Our results concur, suggesting that even limited approaches to the wrist, including the dorsal approach to the scaphoid for nonunion, or “triangular fibrocartilage complex” approaches to the ulnocarpal joint, if associated with scarring of the dorsal extrinsic ligaments, would lead to deficits in wrist range of motion. A dorsal approach that leads to scarring of the DRC ligament, for example, could potentially lead to decreased wrist flexion, radial deviation, and ulnar-flexion, while an approach that leads to scarring of the DIC ligament would lead to decreased ulnar deviation and ulnar-flexion. We postulate that alternative approaches to the wrist, such as arthroscopy, that cause minimal disruption of these ligaments, along with early range of motion may decrease the risk of post-operative wrist stiffness.

We quantified the elongation of the fibers of the DRC and DIC over all anatomical and oblique planes of active motion. Previous studies have examined length changes in these ligaments along specific motion paths. This study generated a ligament elongation model

using multiple directions of wrist motion simultaneously. Computing ligament elongations as a function of wrist position is an important step to determining the function of these ligaments during motion. We identified differing behaviors of the DRC and the scaphoid and trapezoidal attachments of the DIC. An improved understanding of these dynamic relationships will better predict ligament response to injury and repair, and their effects on wrist range of motion.

2.5 References

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Chapter 3

In Vivo Kinematics of the Scaphoid, Lunate, Capitate, and Third Metacarpal in Extreme Wrist Flexion and Extension

The following chapter is in revision for publication as a full manuscript by Rainbow MJ, Kamal RN, Levnthal E, Akelman E, Moore DC, Wolfe SW, and Crisco JJ. to The Journal of Hand Surgery (February 2012).

Abstract

Purpose: Insights into the complexity of active *in vivo* carpal motion have been gained recently using 3D imaging studies; however kinematics during extremes of motion have not been fully elucidated. The purpose of this study was to determine motion of the carpus during extremes of wrist flexion and extension.

Methods: CT scans of 12 healthy wrists were obtained in neutral-grip, extreme loaded flexion, and extreme loaded extension. 3D bone surfaces and six-degree-of-freedom kinematics were obtained for the radius and carpal bones. The contribution of scaphoid and lunate rotation to capitate rotation was determined and the assumption that the capitate and third metacarpal function as a single rigid body was tested. Additionally, joint space metrics at the radiocarpal and midcarpal joints were used to describe the effect of kinematic changes at the joint surface.

Results: In extreme flexion, the scaphoid and lunate rotated 70% (CI: 66% – 73%) and 46% (CI: 43% – 49%) of capitate rotation, respectively. In extreme extension, the scaphoid rotated 74% (CI: 67% – 82%) and the lunate rotated 42% (CI: 33% – 50%). The MC3 rotated $3.9^{\circ} \pm 2.4^{\circ}$ ($P < 0.001$) farther than the capitate in extreme extension. The joint contact area decreased at the radiocarpal joint during extreme flexion. The radius-scaphoid contact center moved onto the radial styloid in extreme flexion from a more proximal and ulnar location.

Conclusions: The contributions of the scaphoid and lunate to capitate rotation are approximately 25% less in extreme extension compared to wrist motion through an active range of motion. More than half the motion of the carpus when the wrist is loaded in extension occurs at the midcarpal joint.

Clinical Relevance: These findings give insight into the possible etiologies of the predictable arthritic pattern seen with abnormal scaphoid flexion in SNAC and SLAC wrists.

3.1 Introduction

The hand can function actively or passively. Most studies have analyzed carpal bone kinematics during activities when the wrist is actively flexed or extended via its forearm musculature [1-5]. Few studies have examined carpal bone posture when the hand is moved passively by an external force [6-7]. In this situation, wrist position is defined by the position of the proximal segments and the externally applied load.

Extreme loaded extension of the wrist is a position ubiquitous to occupational settings (carpentry, plumbing), sports activities (push-up, yoga, Pilates, gymnastics), and activities of daily living (opening heavy door, supporting body weight while leaning) [8]. The wrist also achieves this posture during a fall onto an outstretched hand [6, 9-10]. While extreme loaded flexion is less frequently achieved than extreme extension during normal physiologic motion, certain occupational activities such as plumbing or playing string instruments with the left hand require the wrist to be maintained in an acutely flexed position. Loaded wrist flexion may also emulate post-traumatic wrist kinematics, in which one or more bones of the proximal carpal row may be pathologically flexed [11-13].

Previous studies of active wrist motion have shown that the carpal bones follow consistent kinematic patterns through 60° of wrist flexion and 60° of extension [2-3, 14-17]. These studies indicate that during wrist flexion the scaphoid flexes approximately 70%, while the lunate flexes 45% of capitate flexion. In extension, the scaphoid extends with the capitate while the lunate extends approximately 65% of capitate extension. Furthermore, it is well accepted that the third metacarpal and capitate are normally tightly bound throughout active wrist motion, behaving very nearly as a single rigid body [18-19]. In this work, we questioned whether these well established patterns of active carpal bone motion hold when the wrist is externally loaded at the extremes of motion. Our working hypothesis was that the contribution of the scaphoid and lunate motion to capitate rotation in extreme flexion and extension would differ from their contribution in the normal active range of motion. We also postulated that the capitate and third metacarpal would continue to behave as a rigid body at the extremes of motion.

The purpose of this study was to determine the kinematics and joint space changes of the scaphoid, lunate and capitate during wrist positions of extreme loaded flexion and extension. To explain our findings we also described kinematic changes at the interface of the radiocarpal and midcarpal joints using a joint space metric referred to as the joint contact center. The contact center elucidates the regions of the articulating surfaces that are likely in contact with one another.

3.2 Methods

3.2.1 Scanning Protocol

Following IRB approval and informed consent 6 male and 6 female, right-handed volunteers, mean age 24.8 ± 3.8 years were screened by a board certified orthopedic hand surgeon with PA and lateral radiographs and physical exam to confirm the lack of injury or diseases that could affect carpal motion. CT volume images of the wrist were obtained during three tasks: neutral-grip, extreme flexion and extreme extension. During scanning the volunteers were positioned prone on the CT table (chest supported with a pillow), with their dominant arms extended, in neutral shoulder abduction, parallel with the center axis of the CT gantry.

CT scanning was performed with a GE LightSpeed 16 scanner (General Electric, Milwaukee, WI) at tube settings of 80kVp and 80mA, slice thickness of 0.6 mm, and a field of view that yielded an in-plane resolution of 0.3mm x 0.3mm. During the neutral-grip scan, subjects were instructed to hold a plastic dowel (diameter 28 mm) with a relaxed grip with the forearm pronated and flat to the bed of the scanner. Extreme loaded extension was obtained by instructing the patients maintain a straight elbow and apply approximately 98 N by pushing with the palm of their hand against an analog weight scale, with a vertical orientation and fixed to the scanner table (Figure 3.1). Extreme loaded flexion was obtained by placing the dorsal aspect of the hand flat to the scale and instructing the subject to passively flex until slight discomfort was felt (estimated to be 50 N, Figure 3.1).

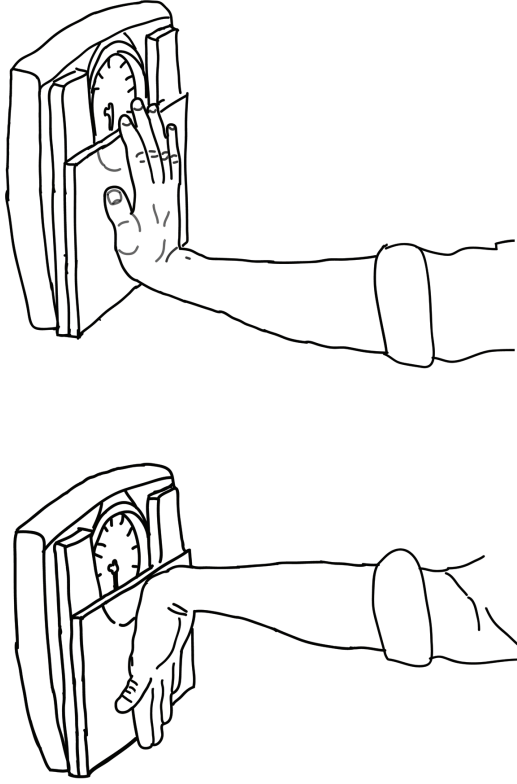


Figure 3.1: CT scanning protocol for (top) extreme loaded extension and (bottom) extreme loaded flexion.

3.2.2 Image Processing and Kinematic Analysis

Using an established markerless bone segmentation and registration method [20-21], polygonal bone surface models and six-degree-of-freedom kinematic transforms were generated for the MC3, capitate, scaphoid, lunate and radius. The bone surface models were segmented from the neutral-grip CT volume image using Mimics 9.11 (Materialize, Leuven, Belgium). Custom C++ (GNU gcc, Free Software Foundation, Boston, Massachusetts) and Matlab (The MathWorks, Natick, Massachusetts) code were then used to calculate the kinematic transformation of each bone independently from neutral-grip to extreme flexion and to extreme extension [21].

All kinematic variables were computed relative to a radius-based coordinate system (RCS). The RCS was generated from landmarks on the radius bone surface model such that the proximally directed X-axis defined wrist pronation-supination, the radially directed Y-axis defined the wrist flexion-extension axis, and the volarly directed Z-axis defined wrist ulnar-radial deviation [22-23]. Wrist position was reported in degrees of flexion-extension and ulnar-radial deviation by computing the orientation of the third metacarpal with respect to the RCS [22].

The rotations and translations of the individual carpal bones from neutral-grip to extreme flexion and extension were described using the helical-axis-of-motion variables. The helical axis of motion fully describes the kinematics of a rigid body by defining a unique rotation axis about which the body rotates and translates. The location of the rotation axis was defined by its point of intersection with the sagittal plane (X-Z) of the RCS. The X-coordinate of this point corresponded to the proximal-distal location and the Z-coordinate corresponded to the volar-dorsal location. The change in scapholunate angle was defined as the difference in flexion-extension rotation between the scaphoid and lunate from neutral-grip to extreme flexion and to extreme extension [17].

3.2.3 Joint Space Analysis

Joint space measurements were summarized by the joint contact center and contact area. The joint contact center was defined as the central point on the articular surface weighted by the average of the distances of each polygon on the articular surface to the closest point on the opposing joint surface. The polygons considered in the analysis were within a threshold distance of 2.5 mm or less from the opposing joint surface. This area, referred to as the contact area, was reported as a percent change from neutral-grip to extreme flexion and to extreme extension. The threshold value of 2.5 mm was chosen because it was found to capture the entire articular surface of the joint in a neutral posture [24].

The joint contact center and contact area was calculated between the radioscapoid, radiolunate, scaphocapitate, and capitollunate joints. Each joint has a pair of contact centers

associated with it at neutral-grip, extreme flexion, and extreme extension. For example, the radius-scapoid contact center is located on the surface of the radius due to its distance from the scaphoid while the scaphoid-radius refers to the contact center located on the surface of the scaphoid due to its distance from the radius.

3.2.4 Statistical Analysis

Linear regression was performed on scaphoid and lunate rotation as a function of capitate rotation independently for extension and for flexion. The contribution of scaphoid and lunate rotation to capitate rotation was compared to previous studies of active carpal bone motion [2, 5, 14, 17, 23]. To examine the rigid body assumption of the MC3-capitate joint, a paired Student's t-test was performed on the magnitude of flexion-extension rotation of each bone. Differences between the location of the rotation axis were evaluated with a two way repeated measures ANOVA followed by a Holm-Sidak pairwise comparison method. The ANOVA was performed for the proximal-distal, and dorsal-volar rotation axis location using bone and posture as factors. Between-subject variability in the rotation axis was calculated as the mean angle between each subject's rotation axis and the resultant rotation axis of all subjects. The joint contact centers and areas were used to qualitatively describe kinematic changes at the joint surface and no statistics were performed on these metrics.

3.3 Results

The total range of motion of the wrist from extreme flexion to extreme extension averaged (± 1 SD) $169.8^\circ \pm 10.9^\circ$ (Figure 3.2). In neutral-grip the wrist was extended $23.9^\circ \pm 12.5^\circ$. The wrist posture was in ulnar deviation in neutral-grip, extreme flexion, and extreme extension ($8.7^\circ \pm 6.0^\circ$, $8.7^\circ \pm 5.8^\circ$, and $11.3^\circ \pm 7.9^\circ$, respectively). The orientation of the rotation axes of the MC3, capitate, scaphoid and lunate were consistent across subjects and primarily aligned with the flexion-extension axis as defined by the RCS (Figure 3.3, Table 3.1).

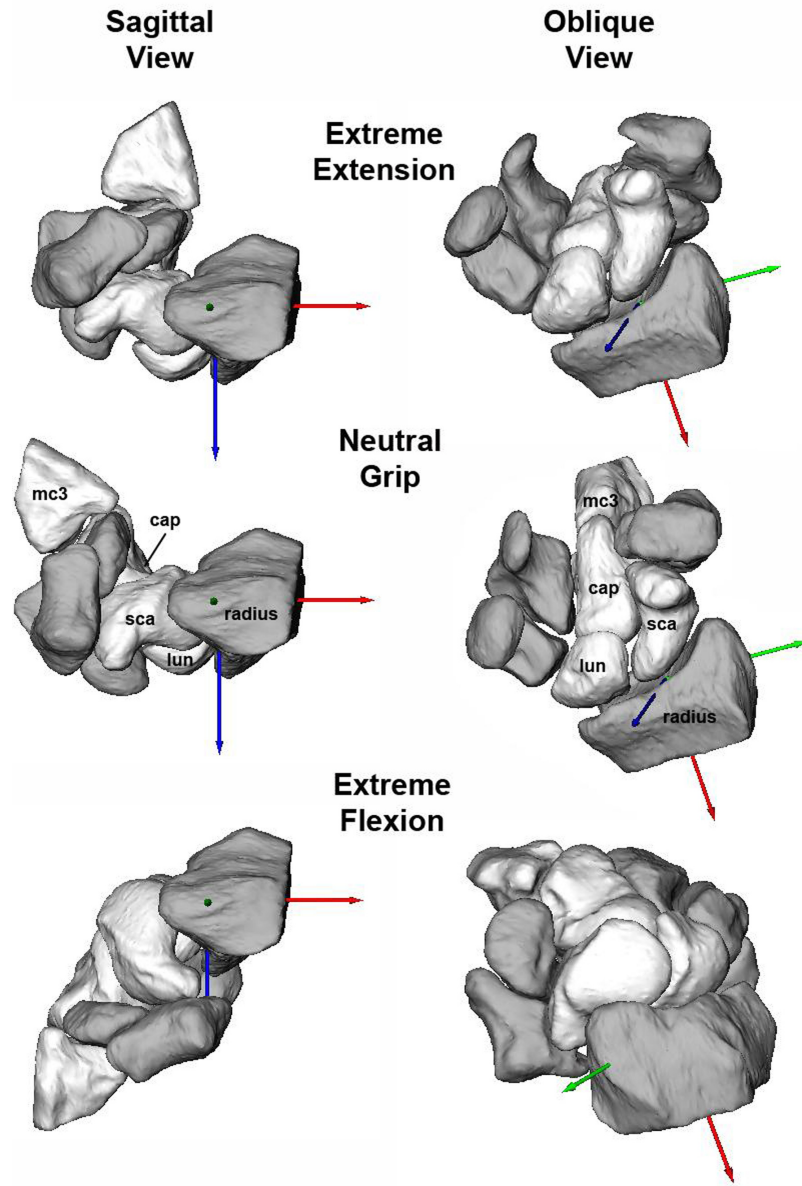


Figure 3.2: Scaphoid, lunate, third metacarpal, and capitate flexion-extension rotation vs. capitate rotation from neutral-grip to extreme flexion and extreme extension. Linear regression revealed that in extreme flexion, the scaphoid rotated 70% and the lunate rotated 46% of capitate rotation. In extreme extension, the scaphoid rotated 74% and the lunate rotated 42% of capitate rotation, resulting in a change in scapholunate angle of $16.2^\circ \pm 7.5^\circ$. The MC3 rotated $1.0^\circ \pm 1.2^\circ$ ($p < 0.05$) and $3.9^\circ \pm 2.4^\circ$ ($p < 0.001$) farther than the capitate in extreme flexion and extension respectively.

Table 3.1: Unit vector components of extreme flexion and extension helical axis vectors (radius-based coordinate system)

Bone	Motion	Axis			Variation With Helical Axis ($^{\circ}$)
		X	Y	Z	
MC3	Extension	0.001	-0.996	0.084	8.8
	Flexion	0.038	0.999	-0.025	5.5
Capitate	Extension	-0.003	-0.993	0.115	8.8
	Flexion	0.032	0.999	-0.029	5.3
Scaphoid	Extension	-0.13	-0.96	0.26	5.7
	Flexion	0.012	0.9997	-0.02	5.6
Lunate	Extension	-0.07	-0.92	0.376	7.8
	Flexion	0.002	0.986	0.168	6.6

3.3.1 Kinematic Patterns of the Scaphoid, Lunate, and MC3

In extreme flexion, the scaphoid rotated 70% (CI: 66% – 73%) and the lunate rotated 46% (CI: 43% – 49%) of capitate rotation (Figure 3.4). This resulted in an increase in the scapholunate angle of $27.2^{\circ} \pm 6.1^{\circ}$ as compared to the neutral position. In extreme extension, the scaphoid rotated 74% (CI: 67% – 82%) and the lunate rotated 42% (CI: 33% – 50%) of capitate rotation, resulting in an increase in scapholunate angle of $16.2^{\circ} \pm 7.5^{\circ}$. The MC3 rotated $1.0^{\circ} \pm 1.2^{\circ}$ ($p < 0.05$) and $3.9^{\circ} \pm 2.4^{\circ}$ ($p < 0.001$) farther than the capitate in extreme flexion and extension respectively (Table 3.2, Figure 3.2).

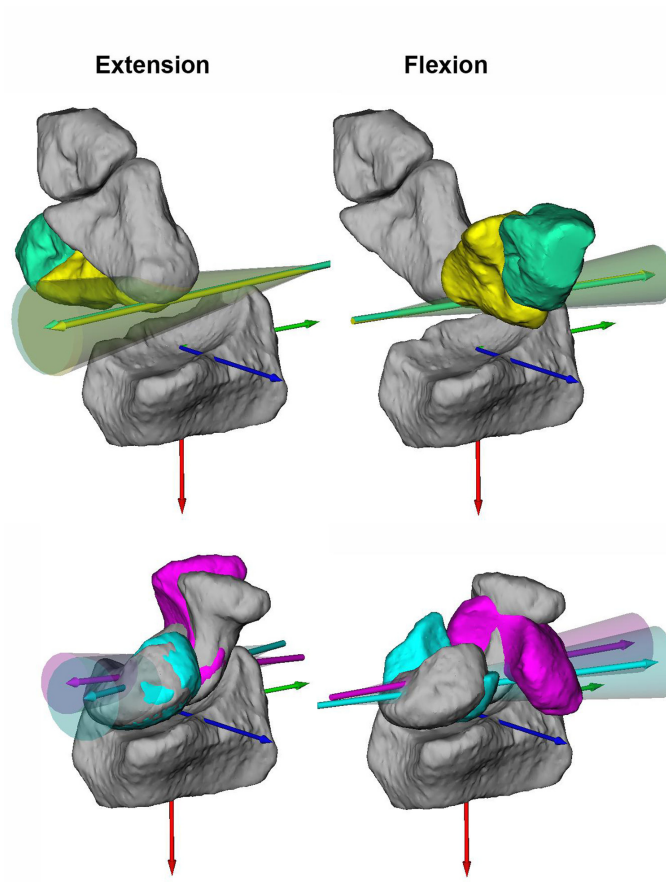


Figure 3.3: Carpal postures with respect to a fixed radial coordinate system of a representative subject in extreme extension, the neutral grip position, and extreme flexion. The capitate (cap), scaphoid (sca), lunate (lun), and third metacarpal (mc3) are highlighted in white.

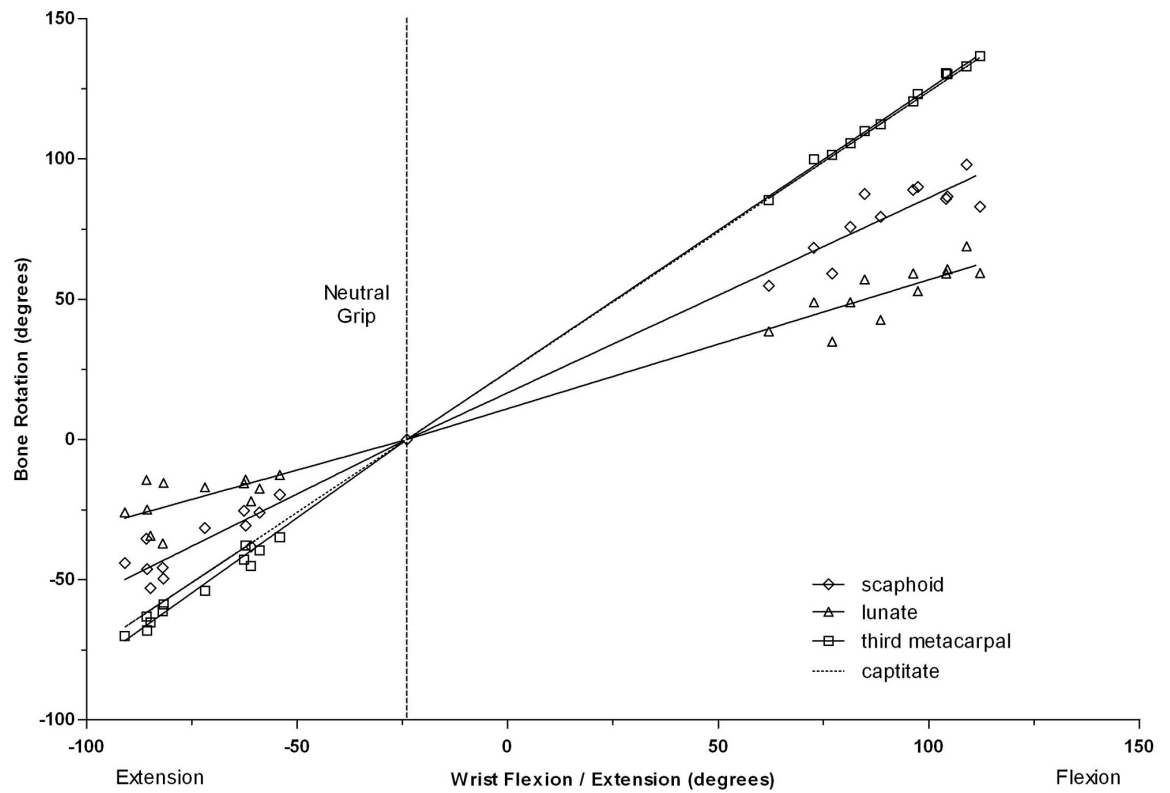


Figure 3.4: Helical rotation axes of the MC3 (green) and capitate (yellow) (top) and scaphoid (purple) and lunate (turquoise) for wrist motion from neutral-grip (gray) to extreme extension and extreme flexion. The translucent cones represent helical rotation axis variability across subjects (Table 3.2).

Table 3.2: Average (1 SD) rotation about and translation along the helical axis in the radius-based coordinate system

Bone	Motion	X		Y		Z	
		rot (deg)	t (mm)	rot (deg)	t (mm)	rot (deg)	t (mm)
MC3	Extension	0.05 (4.0)	0.04 (0.07)	-53.4 (12.7)	0.31 (0.86)	4.5 (8.6)	-0.15 (0.30)
	Flexion	4.7 (8.3)	0.07 (0.07)	115.7 (15.9)	1.2(1.0)	-2.6 (10.1)	-0.01 (0.11)
Capitate	Extension	-0.13 (3.5)	0.02 (0.07)	-49.5 (13.0)	0.61 (0.67)	5.7 (7.8)	-0.17 (0.27)
	Flexion	3.9 (7.4)	0.05 (0.08)	114.7 (15.7)	1.2 (0.9)	-3.1 (10.4)	-0.01 (0.13)
Scaphoid	Extension	-5.1 (2.5)	0.17 (0.08)	-37.1 (10.7)	1.2 (0.4)	10.1 (4.5)	-0.36 (0.20)
	Flexion	1.0 (5.5)	0.01 (0.08)	79.8 (13.0)	0.79 (0.67)	-1.5 (7.1)	-0.02 (0.08)
Lunate	Extension	-1.6 (1.7)	0.07 (0.09)	-21.0 (8.1)	1.2 (0.4)	8.5 (3.7)	-0.51 (0.25)
	Flexion	0.14 (4.8)	-0.01 (0.2)	52.6 (10.1)	1.9 (0.7)	8.9 (5.2)	0.36 (0.25)

3.3.2 Descriptive Kinematics

Although the primary motions of the scaphoid and lunate from neutral-grip to the extreme positions were in the flexion-extension plane, they also exhibited larger out-of-plane motions than either the MC3 or the capitate. The lunate ulnarly deviated $8.9^\circ \pm 5.2^\circ$ in extreme flexion and $8.5^\circ \pm 3.7^\circ$ in extreme extension while the scaphoid ulnar-deviated $10.1^\circ \pm 4.5^\circ$ in extreme extension only. The scaphoid supinated $5.1^\circ \pm 2.5^\circ$ in extreme extension, but demonstrated no supination in flexion. Both the scaphoid and lunate translated radially in extreme flexion and in extreme extension (Table 3.2). The rotation axes of the scaphoid and lunate were located at the proximal articular surface of the capitate. They were dorsal to the midpoint of the proximal pole of the capitate in extreme extension and volar in extreme flexion. There were no significant differences ($p = 0.46$) between the locations of the scaphoid and lunate rotation axes in extreme extension. However, in extreme flexion the lunate rotation axis was located an average of 1.4 mm more distal and an average of 4.6 mm more dorsal than the scaphoid helical axis ($p < 0.001$) (Figure 3.5). While no differences were detected in the volar-dorsal location of the MC3 and capitate rotation axes in extreme flexion, the MC3 rotation axis was located an average 0.9 mm more dorsal ($p < 0.05$) than the capitate in extreme extension (Figure 3.5).

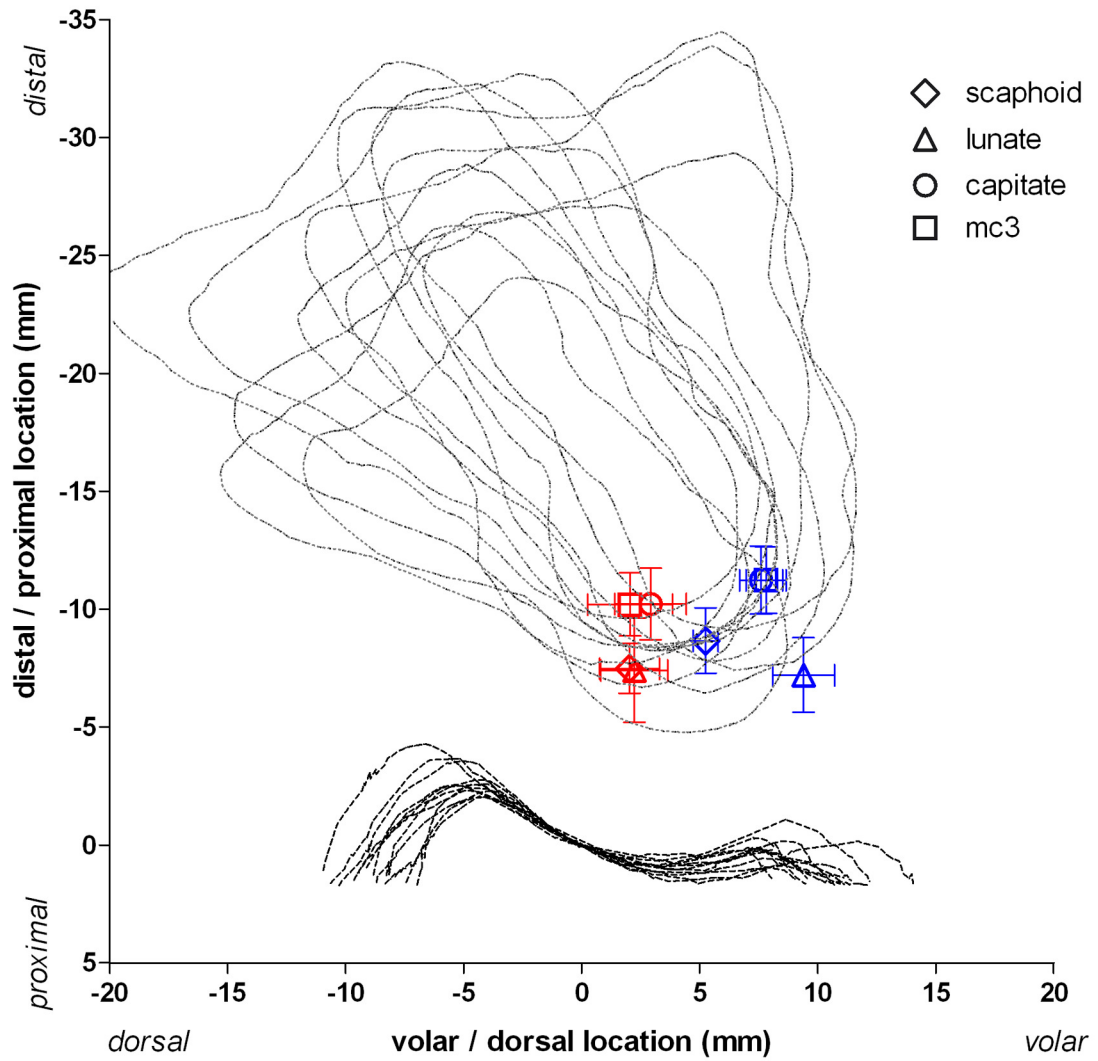


Figure 3.5: Helical rotation axis location in the sagittal plane plotted with sagittal outlines of the each subject's radius and capitate in neutral-grip. Axis location shifts dorsal to volar as the wrist moves from extreme extension (red) to extreme flexion (blue). The dorsal-volar distance between the MC3 and capitate increases in extreme extension.

3.3.3 Joint Space Measurements

The radius-lunate joint contact center was volar in all three poses (Figure 3.6). The radius-scaphoid contact center shifted toward the radial styloid in extreme flexion and dorsally in

extreme extension. When the lunate and scaphoid moved from neutral-grip to extreme flexion, their contact areas on the radius decreased $39\% \pm 22\%$ and $65.5\% \pm 12.6\%$ respectively (Table 3.3). In extreme extension the contact area increased $45.3\% \pm 21.6\%$ and $13.2\% \pm 16.1\%$ for the lunate and scaphoid respectively.

The capitate-scaphoid contact center was consistently located on the radial aspect of the proximal pole of the capitate. The scaphoid-capitate contact center shifted toward the distal tubercle of the scaphoid when the wrist was in extreme flexion. The contact centers of the capitate and lunate were located centrally in neutral-grip, volarly in extreme flexion, and dorsally in extreme extension. The contact area consistently increased from extreme flexion to extreme extension at each articulation of the midcarpal joint.

Table 3.3: Percent change of the 2.5 mm contact patch from neutral-grip at the radiocarpal and midcarpal joints.

Joint	Bone	% Change in 2.5 mm Threshold Surface Area	
		Neutral-grip to Flexion	Neutral-grip to Extension
Radiolunate	Radius	-40.0 (22.0)	45.3 (21.6)
	Lunate	-45.2 (19.0)	46.6 (26.2)
Radioscaphoid	Radius	-65.5 (12.6)	13.2 (16.1)
	Scaphoid	-62.4 (14.6)	23.3 (26.1)
Lunocapitate	Lunate	-9.1 (10.4)	7.2 (6.2)
	Capitate	-1.7 (13.8)	13.6 (10.1)
Scaphocapitate	Scaphoid	-22.9 (0.5)	9.4 (4.3)
	Capitate	-23.4 (10.3)	15.3 (6.7)

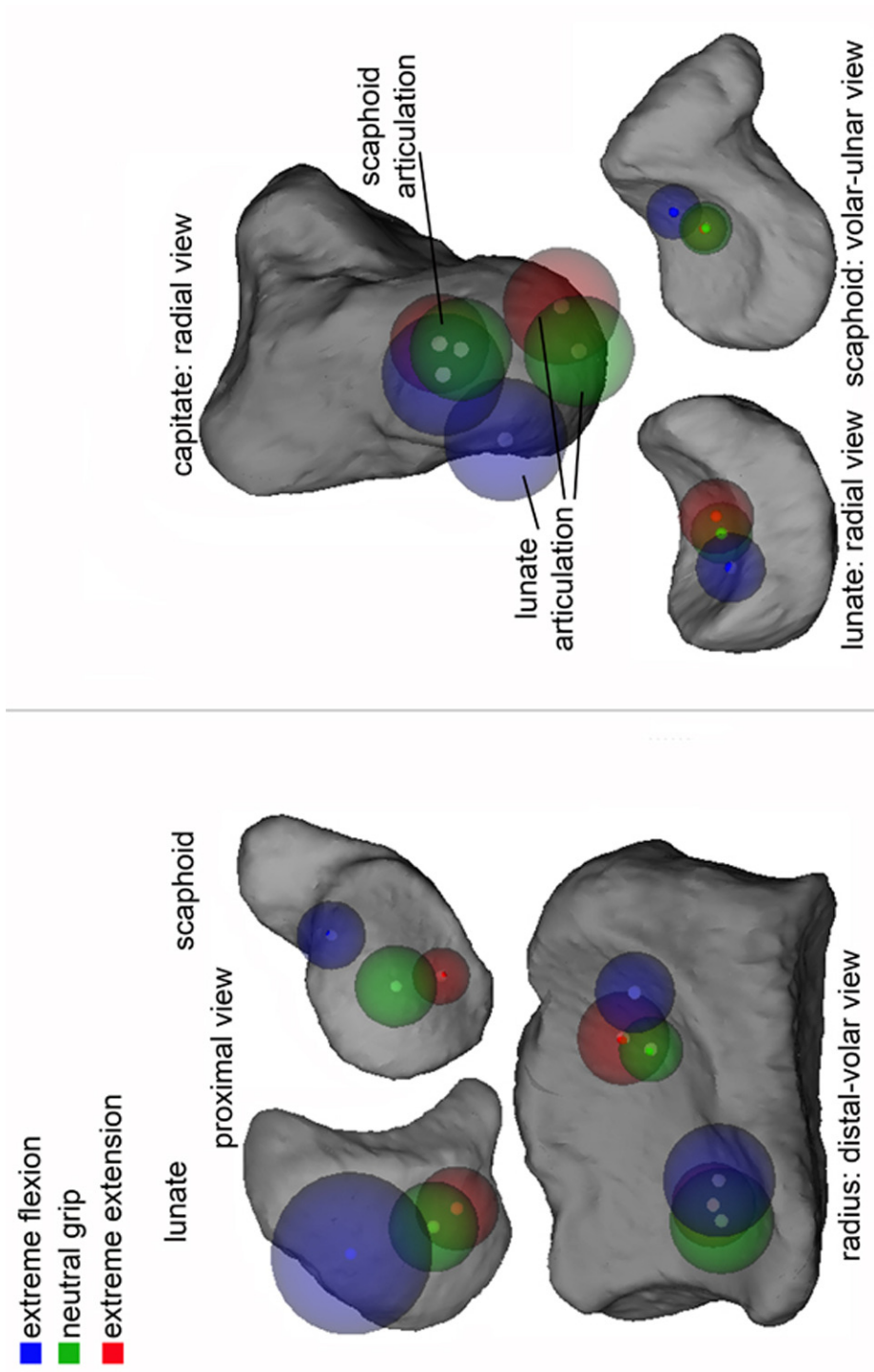


Figure 3.6: The mean (small central sphere) and standard deviation (large sphere) of the location of the joint contact center in all three wrist positions. At the radiocarpal joint (left), as the wrist moved from extreme extension to extreme flexion, the inter-bone distance increased as the contact area decreased. Note the volar location of the radius-lunate contact center in all three positions as well as the shift of the radius-scaphoid contact-center toward the radial styloid in extreme flexion. At the midcarpal joint (right), the capitate exhibited a rocking motion on the lunate as it moved from neutral-grip to extreme flexion or extreme extension. Note also, the consistency in the capitate-scaphoid contact center location in all three positions. However, the contact center shifted toward the distal tubercle of the scaphoid in extreme-flexion.

3.4 Discussion

This study described the three dimensional kinematics of the scaphoid, lunate, capitate, and third metacarpal at the physiological extremes of passive wrist flexion and extension. The contributions of the scaphoid and lunate to capitate rotation are approximately 25% less in extreme extension compared to previous *in vivo* studies of wrist extension through an active range of motion [2, 5, 14, 17, 23]. This finding indicates that while capitate motion is coupled with the scaphoid and lunate through the active range of wrist extension, when the wrist is loaded at extreme extension, the percentage of motion shifts toward the midcarpal joint from the radiocarpal joint. Whether the slightly increased MC3 rotation seen at the extremes of motion is a similar phenomenon or is secondary to the externally-applied force on the hand cannot be answered with these methods. In contrast, during the extremes of wrist flexion, the scaphoid and lunate rotate in a similar fashion as previously reported in active midrange flexion [2-3, 17]. The MC3 rotates farther than the capitate in extreme flexion but only by approximately 1° which is unlikely to be clinically significant.

Previous cadaveric kinematic analyses of the carpus during wrist flexion and extension have been unable to analyze terminal wrist motions, presumably due to external marker impingement and difficulty with motion tracking [25-27]. In our study, we identified a greater range of total motion (170°) than that identified in previous studies of active wrist motion that averaged 142° [28-34]. This is likely attributed to the application of force on the palmar and dorsal aspects of the hand forcing the wrist into terminal flexion and extension, simulating passive-assisted motion. Additionally, subjects in our study had an average of 8.7° of ulnar deviation in neutral, 8.7° in extreme flexion, and 11.3° in extreme extension. We postulate this may have been due to subjects attempting to keep their wrist in a stable position in neutral-grip [35], and due to the forearm posture required for central hand placement on the weight scale at the extremes of motion. Because this was passive wrist motion, the position of the forearm dictated wrist position, and may have caused ulnar deviation with both extreme flexion and extension. Alternatively, ulnar deviation may have been caused by the stress placed on the carpal ligaments at the extremes of motion.

Radiocarpal and midcarpal joint contact characteristics were quantified with subchondral bone surface measurements. The contact center does not necessarily represent physical contact between adjacent bones because cartilage thicknesses were not directly measured in this study. The stability of the contact center between neutral grip and extreme extension, combined with the similarly located helical axes and minimal rotation of these bones compared to the capitate, indicates that the scaphoid and lunate reach terminal extension somewhere between neutral-grip and extreme extension, shifting the remaining wrist extension to the midcarpal joint. In extreme extension, however, the scaphoid rotated into 16.2° more extension than the lunate. This difference in rotation between the scaphoid and lunate in extreme loaded extension may create a shear force across their interface that may help to explain the propensity towards scapho-lunate ligament interosseous rupture or scaphoid fracture when falling onto an outstretched hand.

In extreme flexion, the joint contact center of the scaphoid on the radius moved radial and distal, approaching the radial styloid. Additionally, the contact area decreased 65.5% from neutral-grip. This radial shift of the contact center is of clinical importance, as it is the location on the radius where osteoarthritis begins in stage 1 scapholunate advanced collapse (SLAC) as well as scaphoid non-union advanced collapse (SNAC) [11]. Both of these conditions are associated with dorsal intercalated segment instability (DISI), in which the lunate extends and the scaphoid flexes. Watson has previously established that scaphoid flexion and pronation leads to the predictable arthritic pattern seen in the SLAC wrist. He posited this was due to edge loading, comparing the scaphoid and radius to two incongruent spoons [11]. In this study, the contact center of the scaphoid moves radially toward the radial styloid in extreme flexion without excessive pronation. Therefore, it is possible that extreme flexion of the scaphoid underlies the predictable pattern of OA seen in the SLAC and SNAC wrist.

The midcarpal joint articulation between the scapholunate and the capitate also exhibited patterns that mimicked carpal instability. Both the capitate-lunate and lunate-capitate contact centers were volarly located in extreme flexion and dorsally located in extreme extension. Additionally, the helical axis of the capitate was more proximal in extreme

positions, compared to its position in the proximal pole during active wrist motion [18, 36-37]. These findings indicate that in extreme flexion and extension, the capitate begins to rock within the lunate fossa rather than rotating within it. This invites the potential for volar and dorsal lip impingement in extreme flexion and extension. This rocking phenomena may occur at lesser degrees of flexion-extension in SLAC or SNAC wrists, as the lunate is pathologically dorsally tilted. Interestingly, radiographic evidence of capitolunate OA is seen in stage 3 of the predictable arthritic pattern seen in SLAC/SNAC wrists [38] and could be secondary to incongruency in the capitolunate joint and abnormal patterns of midcarpal motion. On the radial side, the scaphoid appeared to contact the capitate in the same location throughout the range of flexion and extension, while the capitate shifted to articulate with the distal aspect of the scaphoid in extreme flexion.

The primary limitation to this study is that it infers carpal bone motion from three static positions. The helical axis calculation assumes the motions between the neutral-grip position and those at the end range of flexion-extension are at constant rate and in a single direction. The findings of this study demonstrate that the primary motion of the carpus when the wrist is loaded in extension is at the midcarpal joint and gives insight into the possible etiologies of the predictable arthritic pattern seen with abnormal scaphoid flexion in SNAC and SLAC wrists, and the vulnerability of the scaphoid and SLIL to injury in extreme wrist extension.

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Chapter 4

In Vivo Elongation of the Extrinsic Carpal Ligaments Computed at Active and Passive Wrist Positions

4.1 Introduction

The morphology of the carpal bones [1] and their complex system of ligaments [2] allow the wrist to maintain stability throughout a wide range of motion. The functional role of the intrinsic and extrinsic wrist ligaments, and their impact on maintaining normal kinematics in the wrist continues to be studied. For example, it is unknown whether the radiocarpal and dorsal intercarpal ligaments limit end-range wrist motion, serve as guides throughout mid-range wrist motion, or perform both functions. Several studies have reported that in some wrist positions, the carpal ligaments elongate up to 25% beyond their length at the neutral wrist position [3, 4]. With the exception of the yellow ligament in the spine, these values are well beyond the failure strains reported for other ligaments in the human body, such as the anterior cruciate ligament in the knee [5].

There are two possibilities for these apparently high elongation values; (1) the carpal liga-

ments are elastic and operate at high strains or (2) the carpal ligaments are stiff and were slack throughout much of the range of the wrist motions that were studied. As a logical extension of the second possibility, it has been proposed that these ligaments are only taut during the end ranges of wrist motion [6]. However, the carpal bones experience multiple rotational and translational degrees of freedom, and it is possible that a given carpal ligament functions as both an end-range limiter in one direction of wrist motion and as a guide through a large range in a different direction of wrist motion. Additionally, the lengths of the ligaments at the end-ranges of wrist motion have not been studied.

In this study, we evaluated whether the radiocarpal and dorsal intercarpal ligaments support the wrist through a wide range of wrist motion with the assumption that they are relatively inelastic; undergoing modest strain values of approximately 5% to 10%. Our approach differs from previous studies where the length at the neutral wrist position served as the reference length. Instead, we assume that the ligament reaches its maximal length in one of twelve wrist positions. These positions include active positions, and positions at the end range of wrist motion. We estimate strain throughout the remaining positions as a percentage of this maximum length and an assumption of the maximum strain.

4.2 Methods

4.2.1 Subjects, Imaging, and Carpal Bone Kinematics

After obtaining approval from our institutional review board and informed consent, we recruited six male and six female healthy volunteers (age: 24.8 ± 3.8 years). Their dominant (right) wrists were CT-scanned in twelve different positions. We organized these twelve positions into two groups; (1) a functional-task group (TASK) that contained discrete positions of neutral-grip (unloaded and loaded), extreme extension (unloaded and loaded), extreme flexion, carpal distraction (unloaded and loaded), and (2) five incremental positions along the dart thrower’s path (DTM). Details on these positions and the resulting

carpal kinematics were previously reported (Table 4.1) [7, 8]. Briefly, neutral-grip was acquired as volunteers lightly held a plastic dowel, and while a distally-directed load of 98 N was applied to the dowel. Two positions of extreme extension were acquired as the volunteers simulated a push-up position by placing the palm of their hand against a vertical plate and applying a light load and load of 98 N. Extreme loaded flexion required volunteers to place the dorsal aspect of their hand flat against a vertical plate until they reached maximum flexion. Unloaded distraction was acquired with finger-traps affixed to the volunteer’s fingers while they were lying prone, with the palm of their hand pressed flat to the scanner bed. Loaded distraction was the acquired by imposing a distally-directed tensile load of 98 N to the finger traps. Finally, five positions along a simulated hammering task were collected as the volunteers held a dowel against five equally spaced spokes that sampled a hammering motion. The resulting kinematics of this task were along a dart thrower’s path; a path from radial-extension to ulnar-flexion. Three-dimensional carpal bone surface models and six-degree-of-freedom kinematics of all eight carpal bones, radius and ulna, and all five metacarpals were obtained using a previously established markerless bone registration technique from the acquired CT volume images [9, 10].

4.2.2 Ligament Modeling

The volar and dorsal extrinsic radiocarpal ligaments and the dorsal intercarpal ligament were modeled. Boundaries of insertion of the radioscapophcapitate (RSC), long radiolunate (LRL), dorsal radiocarpal (DRC), dorsal intercarpal (DIC), and short radiolunate (SRL) ligaments were manually mapped onto each subject’s bone surface models in Geomagic (Raindrop Geomagic, Durham, NC). Published anatomical descriptions guided the mapping process [11–13]. The RSC, LRL, DRC, and DIC ligaments were modeled with two fibers placed at the most proximal and distal aspects of their respective regions of insertion. Due to its broad attachment, the SRL ligament was modeled with three fibers arranged from radial to ulnar (Figure 4.1). We determined fiber length l_p at each wrist position by computing the shortest path from the origin of the ligament to its insertion with the constraint that the bone surface models were not penetrated [14].

Table 4.1: Database of Functional Tasks and Dart Thrower’s Motion ($n = 12$ subjects) with CT-scanning parameters and wrist position measured by the orientation of the third metacarpal with respect to the radius.

Scan Name	Description	In-Plane Resolution (mm)	Slice Thickness (mm)	Flexion (+), Extension mean (SD)	Ulnar (+), Radial deviation mean (SD)	Group	Description of Carpal Bone Kinematics
Neutral-Grip	Forearm pronated, hand flat, gripping 28 mm diameter dowel. Light grip and with 98 N Distractive Load	0.3 x 0.3	0.625	-23.9 (12.5)	8.7 (6.0)	TASK	Ch. 3
Neutral-Grip Loaded				-22.0 (15.1)	5.9 (5.5)		Unpublished
Distraction Unloaded	Finger-traps, relaxed and loaded to 98N	0.60	1.9 (8.2)	8.5 (5.5)	[15]		
Distraction Loaded			-0.6 (3.8)	5.2 (4.4)			
Extreme Extension Lightly Loaded	Simulated Push-up, lightly loaded and 98 N load applied		-73.9 (5.5)	11.9 (7.8)	Unpublished		
Extreme Extension Loaded			-78.2 (5.4)	11.3 (7.9)	Ch. 3		
Extreme Flexion	Dorsum of hand flat, forced end-range flexion until slight discomfort	0.5 x 0.5	0.625	91.6 (8.2)	8.7 (5.8)	DTM	[16], Appendix 2
Hammer 1	Held wooden dowel against spokes placed at 5 positions that sampled hammering path from wind-up to strike			-59.6 (9.6)	-5.4 (8.9)		
Hammer 2				-48.7 (9.6)	1.5 (9.0)		
Hammer 3				-37.1 (7.9)	12.9 (9.3)		
Hammer 4				-24.4 (7.9)	26.1 (6.5)		
Hammer 5				-8.8 (10.7)	34.1 (5.8)		

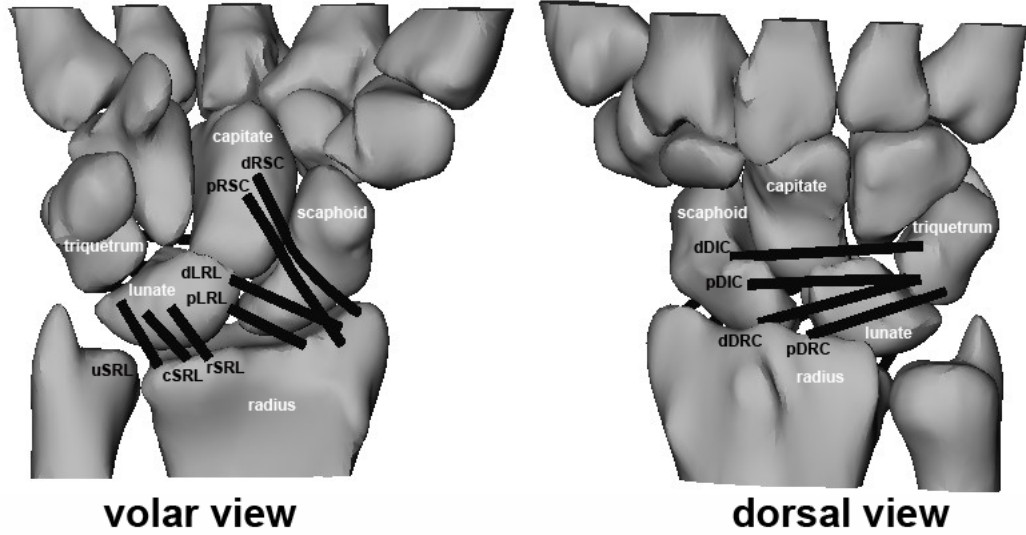


Figure 4.1: Ligament fibers modeled for a representative subject's right wrist. On the volar side (left panel), the radioscaphocapitate (RSC) was modeled with a distal (dRSC) and proximal (pRSC) fiber. The long radiolunate (LRL) was also modeled with a distal (dLRL) and proximal (pLRL) fiber. The short radiolunate (SRL) ligament was modeled with a radial fiber (rSRL), a central fiber (cSRL), and an ulnar fiber (uSRL). On the dorsal side of the wrist (right panel), the dorsal radiocarpal (DRC) ligament was modeled with a distal (dDRC) and proximal (pDRC) fiber and the dorsal intercarpal (DIC) ligament was modeled with a distal (dDIC) and proximal (pDIC) fiber.

4.2.3 Estimating Ligament Fiber Recruitment

Our analysis assumes that each fiber may achieve its maximum length l_{max} in one of the twelve wrist positions examined in this study. The estimated strain of the fiber throughout the remaining positions was then computed as a function its length l_p , the maximum length l_{max} , and an assumption of the maximum strain, ε_{max} . We note that ε_{max} occurs at l_{max} . We performed the analysis twice, assuming an ε_{max} of 0.1 and 0.05, based on the assumption that the carpal ligaments are stiff, and have similar mechanical properties to the anterior cruciate ligament [5].

The elongation E_p of each fiber at each position p was computed using the engineering strain equation.

$$E_p = \frac{(l_p - l_r)}{l_r}$$

where l_r is the reference length. Since we are unable to determine the resting length with these methods, we estimate it using an assumption of maximum strain and the maximum length [15].

$$l_r = \frac{l_{max}}{(1 + \varepsilon_{max})}$$

Because we are assuming the maximum strain occurs when the length of the ligament is at the maximum length in our dataset, we compute elongation, E_p as a percentage of ε_{max} .

$$R = 100 * (1 + E_p - \varepsilon_{max})$$

We termed R the recruitment value, and note that when R is equal to 100, the fiber's length is equal to l_{max} and the estimated strain is equal to ε_{max} .

4.2.4 Statistical Analysis

A paired Student's t-test determined whether setting $\epsilon_{\max}$ to 0.1 or 0.05 significantly changed the value of R . Paired Student t-tests also determined differences in R between fibers within each ligament with significance set to $p = 0.05$ for the RSC, LRL, DRC, and DIC. Because the SRL had three fibers, significance was set at 0.017 ($0.05 / 3$) using a Bonferroni-adjustment. In each TASK position, the mean R value and its standard deviation was reported for each fiber. To determine the value of R as the wrist moved through the DTM, linear regression was performed on R as a function of the angle along the dart thrower's path.

4.3 Results

Ligament fiber lengths ranged from 5.4 ± 0.8 mm (rSRL) to 27.9 ± 2.4 mm (pRSC) (Table 4.2). When $\epsilon_{\max}$ was set to 0.05 compared to 0.10, R significantly increased 0.38 ± 0.42 % ($p < 0.005$). However, the increase was negligible and we set $\epsilon_{\max}$ to 0.1 for the remainder of the analysis.

There was a significant difference in R ($p < 0.005$) between fibers within each ligament over all wrist postures, and the position where $l_{\max}$ occurred was variable between fibers, subjects, and positions (Table 4.3). However, R varied consistently within the TASK group (Figure 4.2) and the DTM group (Figure 4.3).

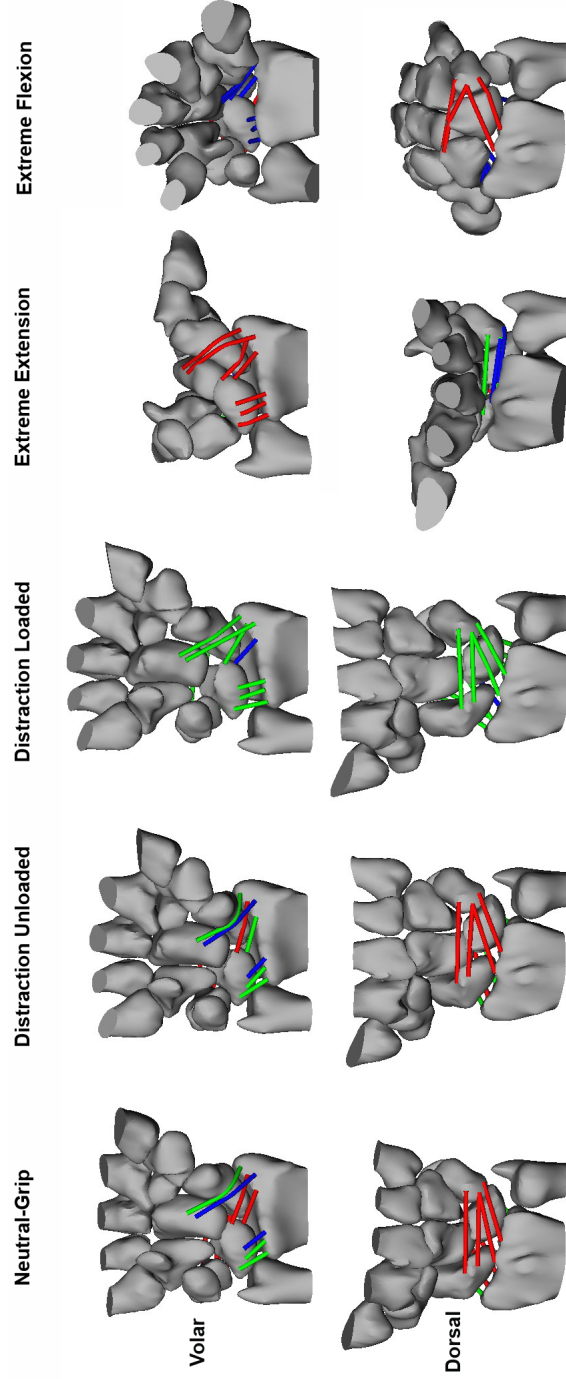


Figure 4.2: Ligament fibers of a representative subject's right wrist in each TASK position. The color of each ligament fiber correlates to its length as a function of the maximum length or R value, averaged across all subjects. Blue fibers represent R values below 90%, green fibers represent R values greater than or equal to 90% and less than 95%, and the R value of red fibers are 95% or higher. Loaded grip and unloaded extension are not shown because they were similar to unloaded grip and loaded extension respectively.

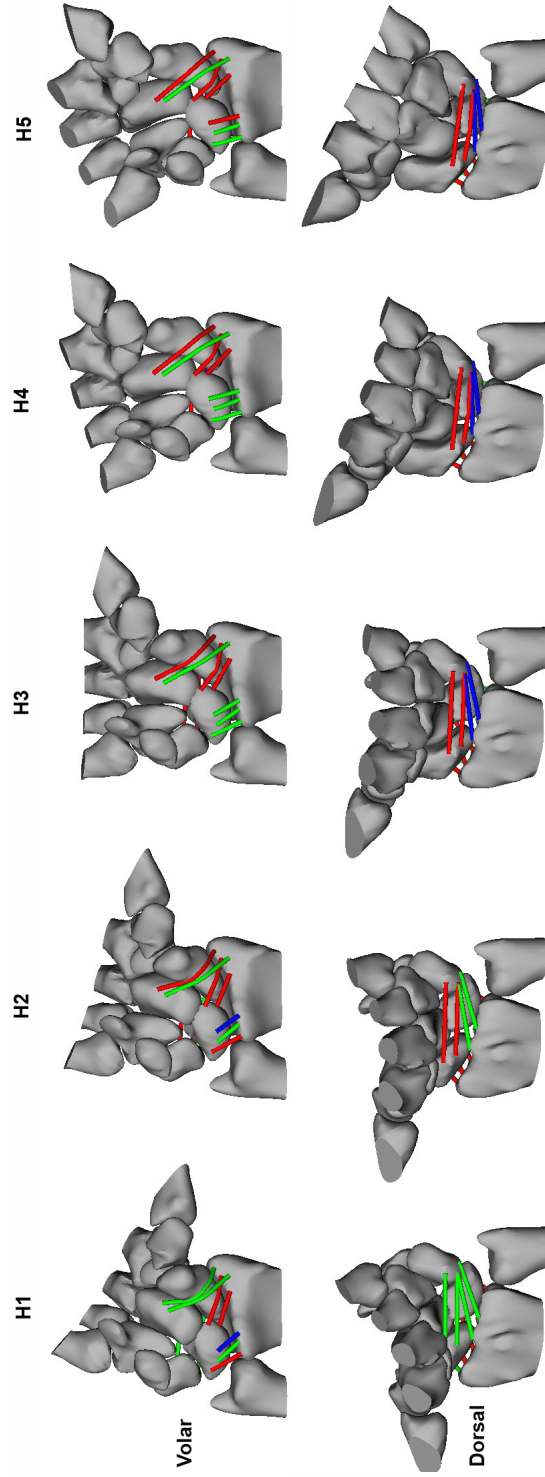


Figure 4.3: Ligament fibers of a representative subject's right wrist moving through the DTM. The color of each ligament fiber correlates to its length as a function of the maximum length or R value, averaged across all subjects. Blue fibers represent R values below 90%, green fibers represent R values greater than or equal to 90% and less than 95%, and the R value of red fibers are 95% or higher. From left to right the wrist moves from radial-extension in the first hammering position (H1) toward ulnar flexion in the final hammering position (H5).

Table 4.2: Mean (standard deviation) of Computed Fiber Lengths in Millimeters

	$l_r, \epsilon_{\max} = 0.05$	$l_r, \epsilon_{\max} = 0.1$	Neutral-Grip	Distraction Unloaded	Maximum Length	Minimum Length
dRSC	25.8 (2.3)	24.6 (2.2)	25.4 (2.4)	24.8 (2.5)	27.1 (2.4)	20.7 (2.0)
pRSC	26.6 (2.3)	25.4 (2.2)	24.9 (2.4)	23.6 (2.4)	27.9 (2.4)	18.7 (2.1)
dLRL	16.5 (1.6)	15.7 (1.6)	16.7 (1.6)	16.7 (1.7)	17.3 (1.7)	14.2 (1.7)
pLRL	12.0 (1.4)	11.5 (1.3)	11.9 (1.4)	11.7 (1.4)	12.6 (1.4)	7.9 (1.4)
rSRL	11.1 (1.5)	10.6 (1.4)	9.5 (1.4)	8.8 (1.4)	11.7 (1.6)	5.4 (0.8)
cSRL	10.1 (1.6)	9.7 (1.6)	9.1 (1.3)	8.5 (1.0)	10.6 (1.7)	6.3 (0.9)
uSRL	10.6 (1.4)	10.1 (1.3)	10.1 (1.4)	9.5 (1.1)	11.2 (1.5)	7.6 (0.9)
dDRC	22.4 (1.9)	21.4 (1.8)	22.5 (1.9)	22.9 (2.1)	23.6 (2.0)	18.9 (1.1)
pDRC	17.8 (2.5)	17.0 (2.4)	17.9 (2.6)	18.2 (2.7)	18.7 (2.7)	13.9 (1.5)
dDIC	19.9 (3.2)	19.0 (3.1)	20.3 (3.3)	20.3 (3.6)	20.9 (3.4)	18.7 (3.4)
pDIC	18.2 (2.1)	17.4 (2.0)	18.6 (2.2)	18.4 (2.3)	19.1 (2.2)	17.2 (2.2)

Table 4.3: The number of subjects whose fibers reached maximum lengths in each wrist position.

	Number of Subjects (out of 12) Whose Fibers Reached l_{max} in Each Wrist Position										
	dRSC	pRSC	dLRL	pLRL	rSRL	cSRL	uSRL	dDRC	pDRC	dDIC	pDIC
Neutral-Grip								1	1		
Neutral-Grip Loaded	3		1	3					4		3
Distraction Unloaded								2	2		
Distraction Loaded		1							1		
Extreme Extension Lightly Loaded	1	1		1	1	2	2				1
Extreme Extension Loaded		9		3	7	9	9				
Extreme Flexion								9	4		
Hammer 1	1		1	1		1	1				1
Hammer 2				1							
Hammer 3	1	1	1							1	1
Hammer 4	5		1		1					3	5
Hammer 5	1		8	3	3					8	1

Radioscaphocapitate (RSC)

Fibers of the RSC approached their maximum lengths in extreme loaded extension with R values for the dRSC and pRSC of $97.1 \pm 2.5\%$ and $98.4 \pm 2.0\%$, respectively (Figure 4.4a, b). During the DTM, the R values of the dRSC increased ($p < 0.001$) from 93.2 ± 2.3 to $98.5 \pm 1.2\%$ while the pRSC stayed constant ($p = 0.4$) at an R of $92.5 \pm 0.4\%$ (Figure 4.4c, d).

Long radiolunate (LRL)

In the TASK group, the R value of the dLRL was 95% or higher in all positions but loaded distraction ($89.2 \pm 8.3\%$) and extreme flexion ($82.3 \pm 5.7\%$). The pLRL had the highest R value during grip ($94.0 \pm 6.8\%$) and extreme extension ($98.2 \pm 2.5\%$). Both fibers of the LRL had the lowest R values in extreme flexion. Interestingly, loaded distraction also shortened the R value of the dLRL and pLRL to $89.2 \pm 8.3\%$ and $81.8 \pm 10.2\%$ respectively (Figure 4.5a, b). During the DTM the R value of the dLRL increased ($p < 0.001$) from $95.8 \pm 3.2\%$ to $99.4 \pm 1.0\%$ but the pLRL had a constant R value ($p = 0.4$) of $96.5 \pm 3.5\%$ (Figure 4.5c, d).

Short radiolunate (SRL)

Extreme extension elongated the rSRL, cSRL, and uSRL to l_{max} in at 8 out of 12 subjects for the rSRL and 11 out of 12 subjects for the cSRL and uSRL (Table 4.2, Figure 4.6a-c). In the DTM, the R value of the rSRL increased from $85.2 \pm 6.2\%$ to 96.1 ± 4.0 , the cSRL remained constant with an R value of $90.7 \pm 5.2\%$, and the R of the uSRL decreased from $95.4 \pm 3.2\%$ to $91.8 \pm 4.9\%$ (Figure 4.6d-f).

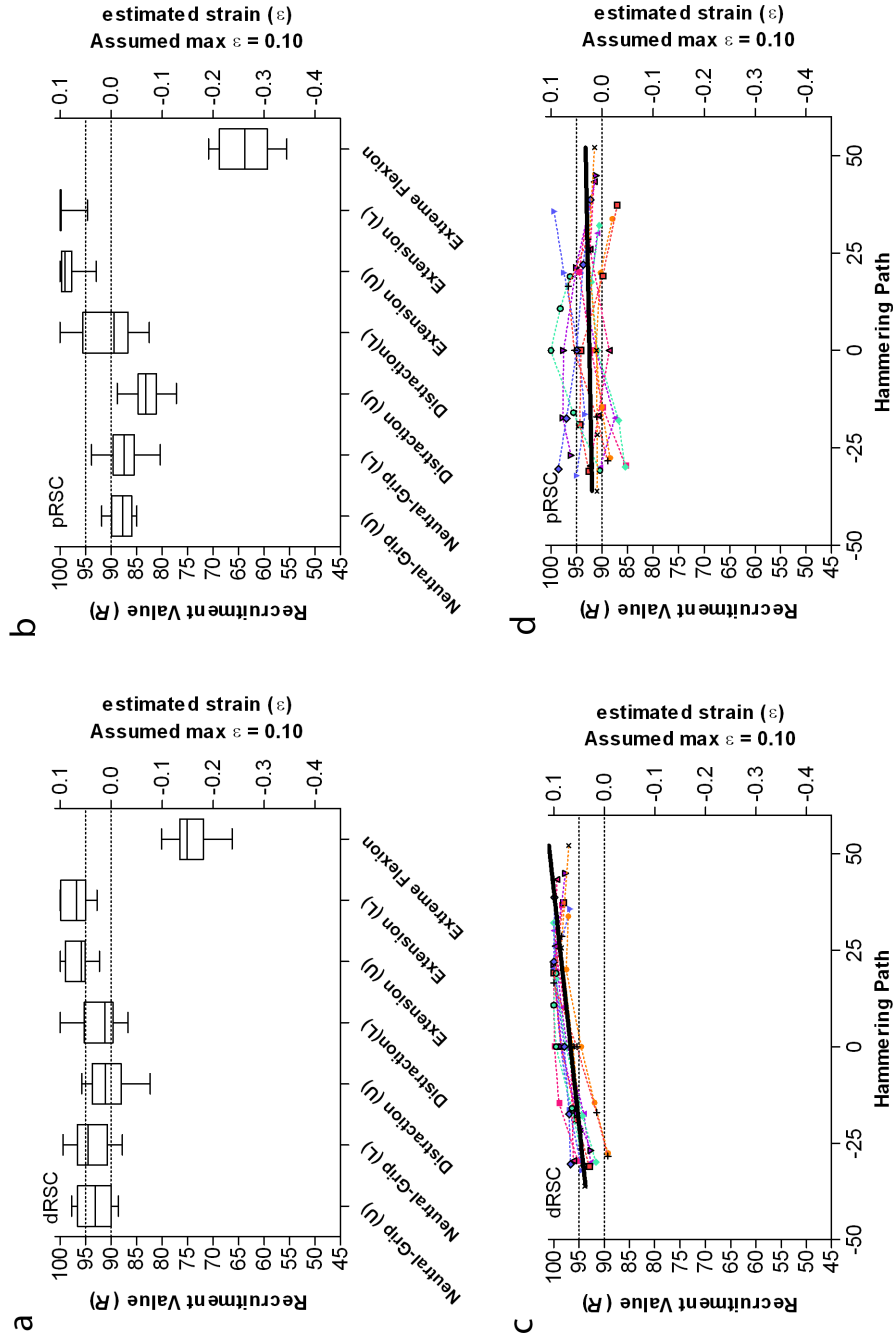


Figure 4.4: Elongation of the radioscaphocapitate ligament (RSC). The R value (left y-axis) is the percentage of ε_{max} (right y-axis). Top: R values of the distal (a) and proximal (b) fibers of the RSC for all subjects within the TASK group. These data are summarized with a box and whiskers plot. The box represents the 25th percentile, median, and 75th percentile. The whiskers represent the minimum and maximum values across subjects. Bottom: Elongation of the distal (c) and proximal (d) fibers of the RSC during the DTM. Each line represents a subjects elongation pattern. The RSC was elongated maximally during extreme extension. During the DTM, fibers of the dRSC also approached their maximum lengths as the wrist rotated from the neutral (0°) toward ulnar flexion.

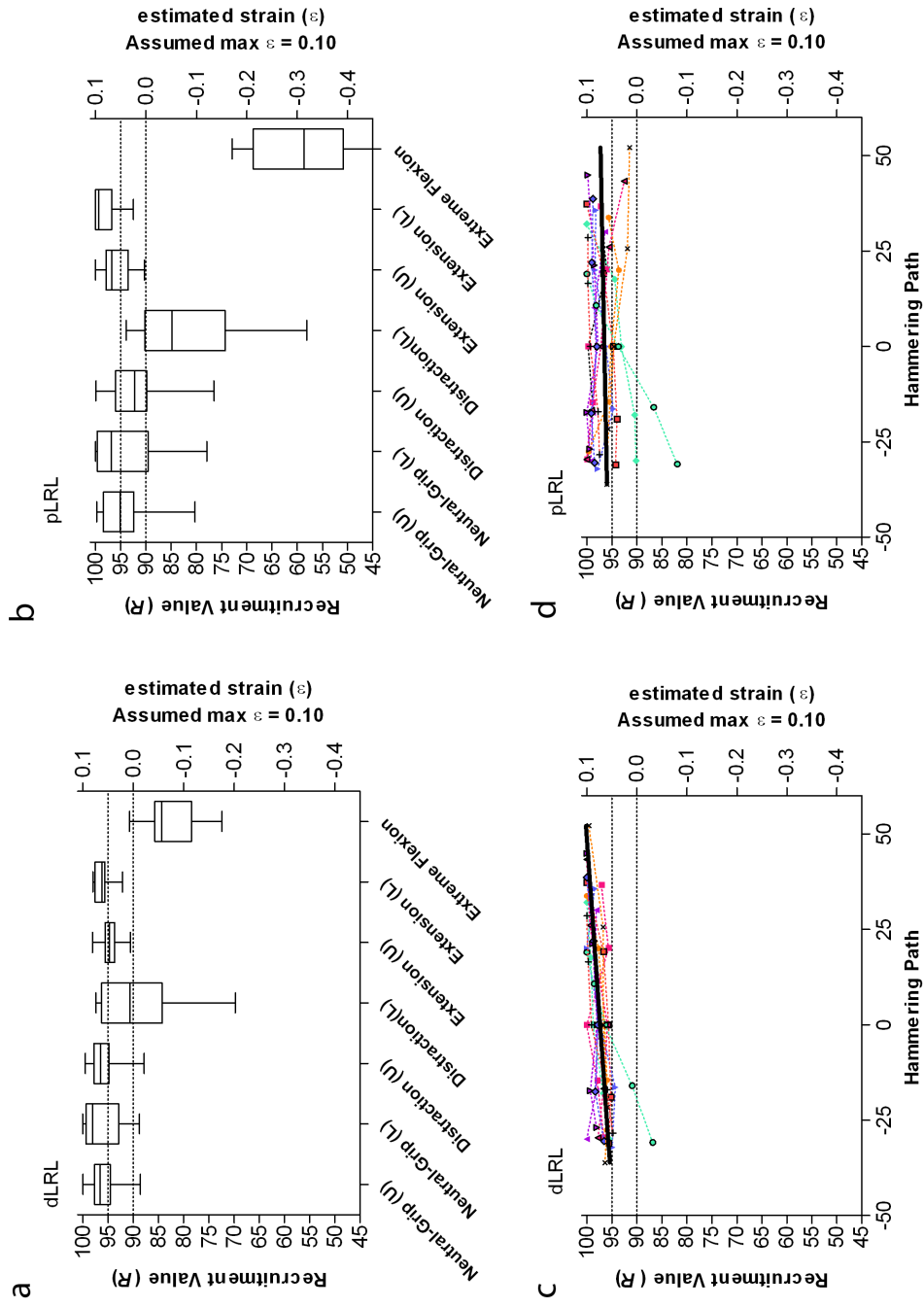


Figure 4.5: Elongation of the long radiolunate ligament (LRL). The R value (left y-axis) is the percentage of ε_{max} (right y-axis). Top: Elongation of the distal (a) and proximal (b) fibers of the LRL for all subjects within the TASK group. Bottom: Elongation of the distal (c) and proximal (d) fibers of the LRL during the DTM. With the exception of loaded distraction and extreme flexion, the LRL was elongated within 95% of its maximum length throughout the entire range of motion.

Dorsal radiocarpal (DRC)

The dDRC was elongated in extreme flexion ($99.8 \pm 0.5\%$) as evidenced by 9 out of 12 subjects reaching l_{max} in this position. The R value of the dDRC was greater than 95% in neutral-grip and unloaded distraction. The pDRC behaved similarly to the dDRC. However, the R value was greatest in unloaded distraction ($97.1 \pm 3.1\%$), neutral-grip ($97.1 \pm 3.8\%$), and extreme flexion ($96.4 \pm 3.7\%$) (Figure 4.7a, b). Both fibers of the DRC shortened ($p < 0.001$) as the wrist moved along the dart thrower's path (Figure 4.7c, d).

Dorsal Intercarpal (DIC)

The R value of the dDIC was greatest in neutral-grip ($97.3 \pm 1.9\%$) and unloaded distraction ($96.4 \pm 3.0\%$). The R value of the pDIC was between $94.2 \pm 3.3\%$ and $97.7 \pm 2.5\%$ for all positions within TASK, except loaded distraction where the R was $90.1 \pm 5.4\%$ (Figure 4.8a, b). The lengths of both fibers of the DIC increased as the wrist moved along the dart thrower's path, ranging from R values of $94.6 \pm 3.2\%$ in the first hammering position to R values of $99.4 \pm 0.7\%$ in the final position (Figure 4.8c, d).

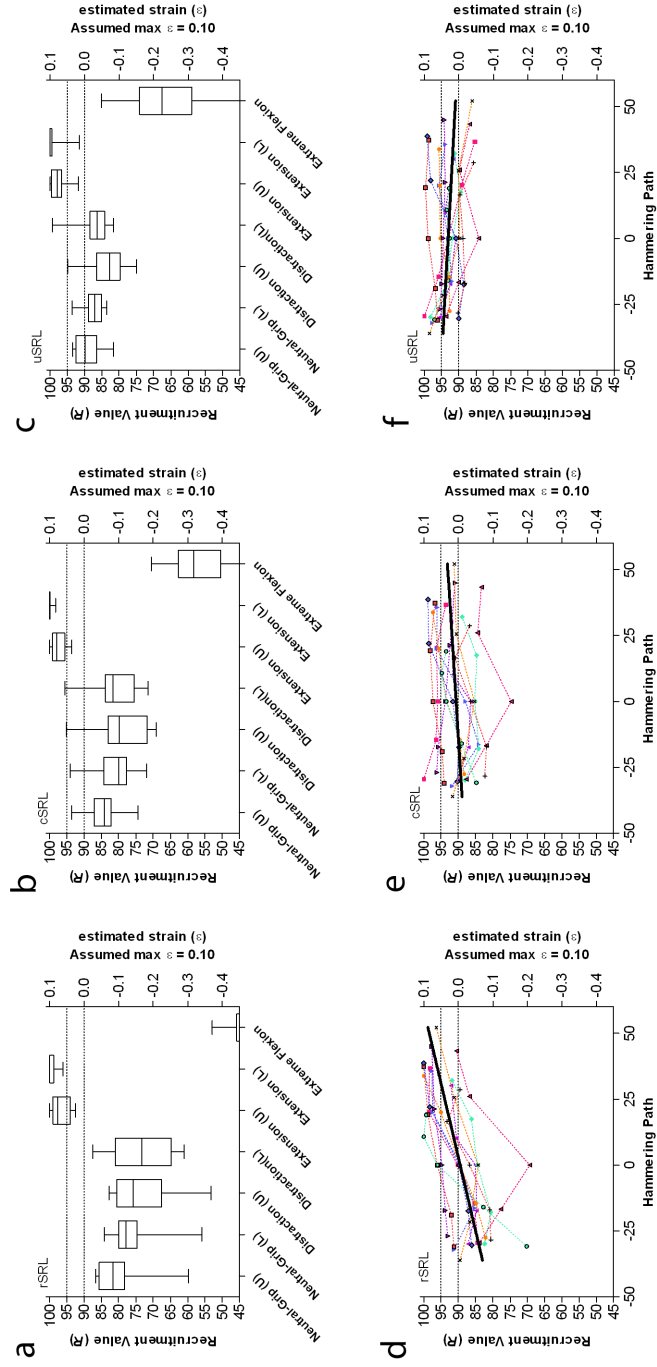


Figure 4.6: Elongation of the short radiolunate ligament (SRL). The R value (left y-axis) is the percentage of ϵ_{max} (right y-axis). Top: Elongation of the radial (a), central (b), and ulnar (c) fibers of the SRL for all subjects within the TASK group. Bottom: Elongation of the radial (d), central (e), and ulnar (f) fibers of the SRL during the DTM. With the exception of loaded distraction and extreme flexion, the LRL was elongated within 95% of its maximum length throughout the entire range of motion.

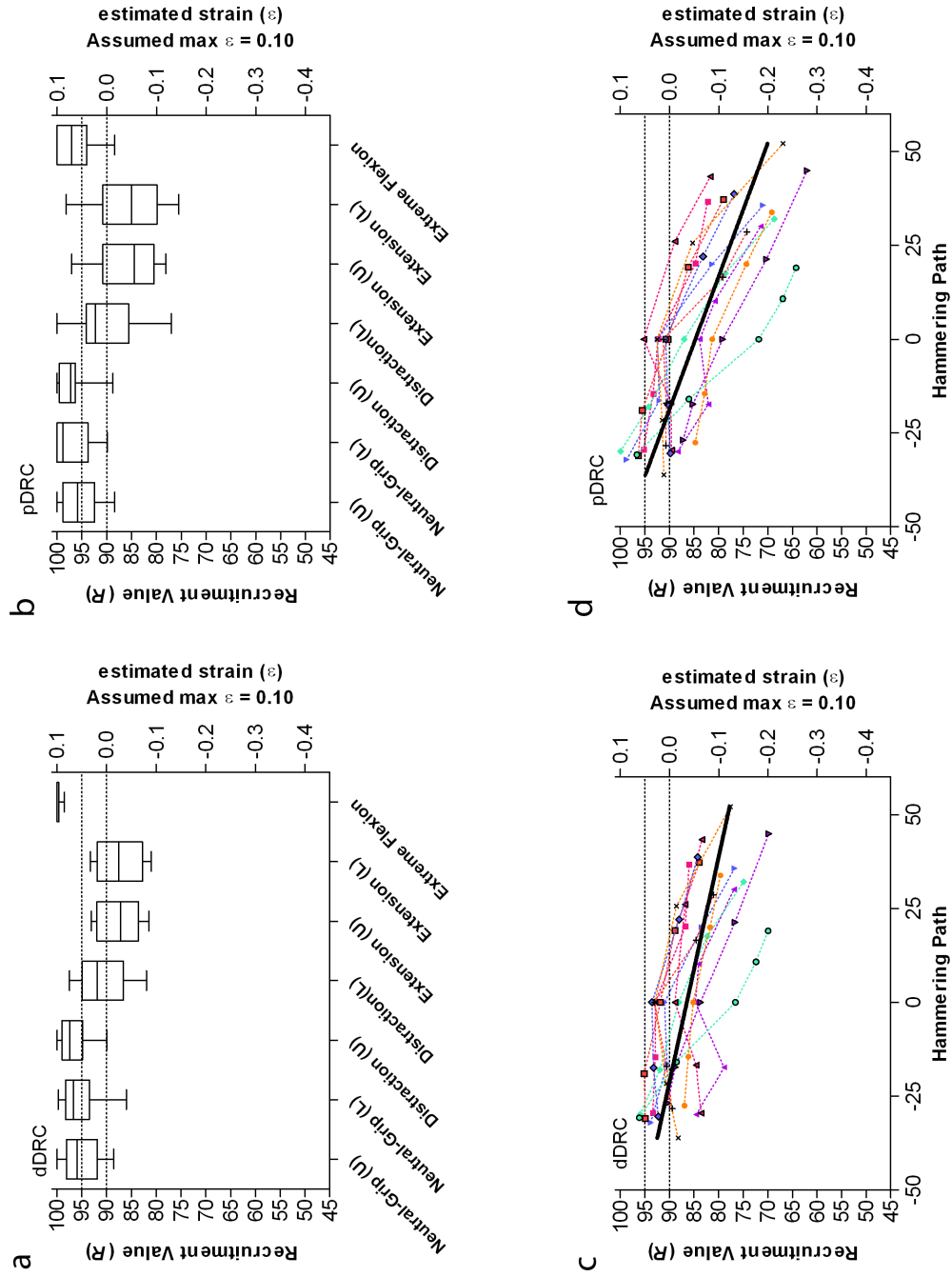


Figure 4.7: Elongation of the dorsal radiocarpal ligament (DRC). The R value (left y-axis) is the percentage of ε_{max} (right y-axis). Top: Elongation of the distal (a) and proximal (b) fibers of the DRC for all subjects within the TASK group. Bottom: Elongation of the distal (c) and proximal (d) fibers of the DRC during the DTM. The DRC was elongated within 95% of its maximum length during grip, extreme flexion, and unloaded distraction. The DRC shortened as the wrist moved along the DTM.

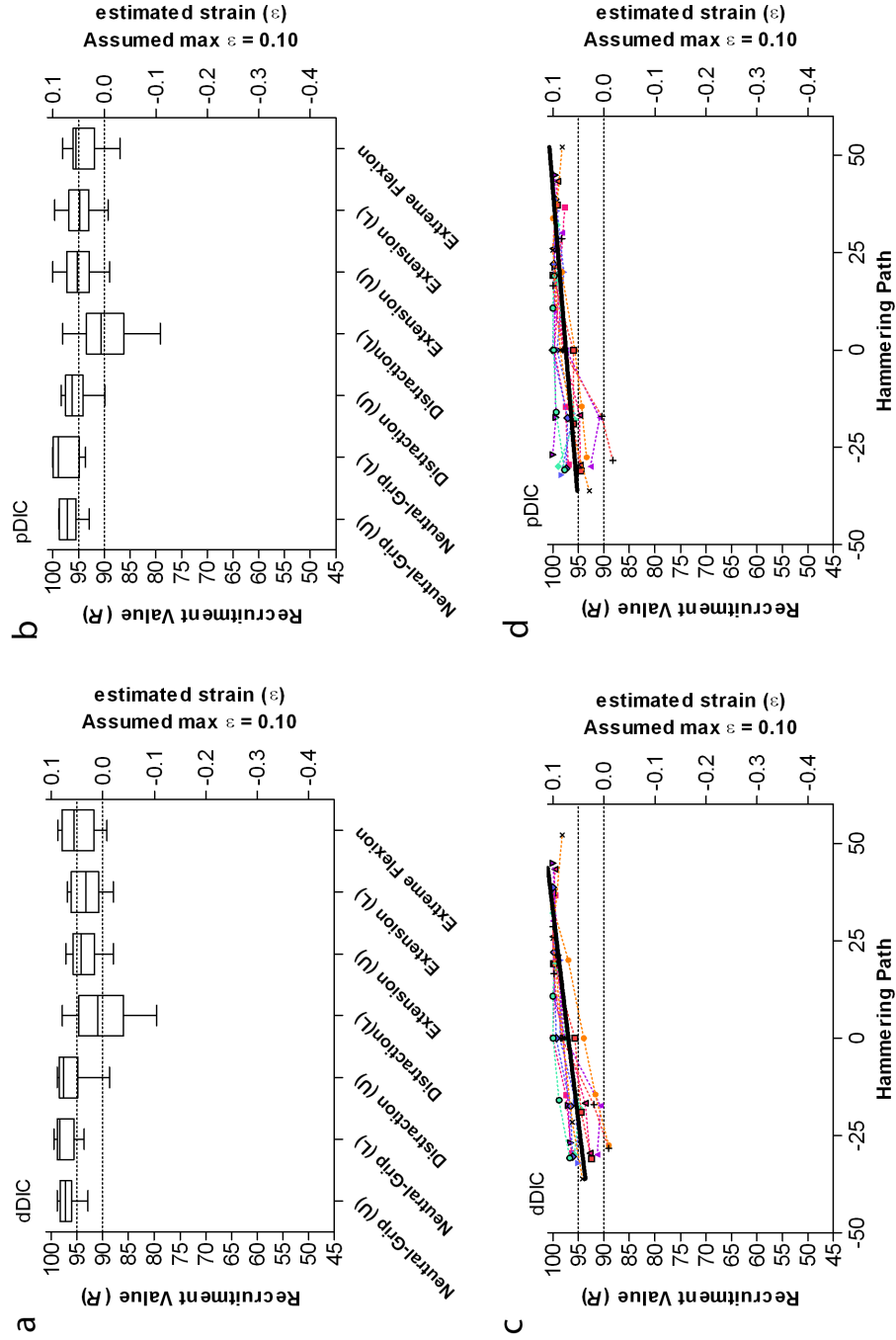


Figure 4.8: Elongation of the dorsal intercarpal ligament (DIC). The R value (left y-axis) is the percentage of ε_{max} (right y-axis). Top: Elongation of the distal (a) and proximal (b) fibers of the DIC for all subjects within the TASK group. Bottom: Elongation of the distal (c) and proximal (d) fibers of the DIC during the DTM. With the exception of unloaded distraction, the DIC was elongated within 94% throughout the range of motion.

4.4 Discussion

We studied the elongation of the dorsal and volar radiocarpal ligaments and the dorsal intercarpal ligament during twelve wrist positions that sampled active and passive wrist motions. Many ligaments are strained in these positions, particularly in extreme extension (volar ligaments) and in extreme flexion (dorsal ligaments). Considering that the normal operating strain in ligaments of other joints is approximately 5% [22], we propose that there is a high probability that ligaments within 95% of their maximum length are strained. As the ligament fiber lengths fall below 95% of their maximum values, the likelihood that they are strained falls as well [23]. Many of the volar and dorsal ligaments were within 5% of their maximum value throughout much of the range of wrist motion. Importantly, some ligaments were maximally elongated during an extreme position and within 95% throughout grip, and the entire dart thrower's path. These results indicate that the extrinsic carpal ligaments may serve as limiters to end-range wrist motion as well as guide the carpal bones throughout a large range of motion. We also found slightly different behavior within a ligament depending on fiber location. This finding highlights the concept that the carpal ligaments are a confluence of tissues and that different regions within a defined ligament are elongated differently depending on the direction of wrist motion. Previous studies of carpal ligament elongations [16] and studies of the anterior cruciate ligament [24] have come to similar conclusions regarding variable strains among fibers within a ligament.

Our approach of computing ligament fiber elongation as a function of the maximum fiber length differs from previous carpal ligament elongation studies in several important ways. First, the analysis focuses on feasible strain values. Engineering strain ε is defined as

$$\varepsilon = \frac{l_p - l_o}{l_o}$$

where l_o is the length at which a ligament fiber begins to undergo strain and produces a tensile force. Studies of ligament elongation are unable to determine the resting length and typically replace l_o by the length at the neutral wrist position. If the ligament is slack in the neutral position, the computed elongation of a strained fiber will be overestimated because the numerator of the strain equation increases as the denominator decreases. For example, if a ligament's fiber length is 5 mm in the neutral position and 8 mm in extension, the computed strain will be 60%. However, if the ligament is operating around 5% strain, the resting length should actually be 7.6mm. Here, the ligament is probably slack in the neutral position and using the neutral length as the resting length is misleading. Second, by computing the maximum fiber length and assuming that a stiff ligament will only elongate slightly once it is strained, our analysis avoids variability in targeting specific wrist positions to define l_o . However, a related limitation of our approach is the requirement that we estimate the maximum strain in order to compute l_o . We changed the maximum strain estimate from 0.1 to 0.05 and determined that the resulting choice minimally changed the resulting R values. Nevertheless, these methods are useful in determining the general behavior of the carpal ligaments because they utilize accurate *in vivo* kinematics and a robust fiber-wrapping algorithm that mimics the course of their physical counterparts.

The Radioscaphocapitate Ligament

The RSC ligament has been suggested to act as a radial collateral ligament to the wrist, a constraint against radiocarpal pronation or ulnocarpal dislocation, and as a stabilizer to the distal pole of the scaphoid [11]. Cadaveric and *in vivo* studies consistently report that the RSC elongates from the neutral position to ulnar deviation and from neutral to extension [4, 17-20]. There are divergent reports on the elongation of the RSC in radial deviation with some studies reporting elongation [19–21], some reporting shortening [3, 18], and others reporting minimal length changes [16]. Our results are in agreement that the RSC is maximally elongated in extreme extension. A recent study reported that the RSC shortened to 90% as the wrist moved from neutral into ulnar flexion along the dart thrower's path [4]. Our results suggest that the RSC maximally elongates through a large range of

the DTM and increases with increasing ulnar flexion. This could potentially serve as a stabilizer to the carpus as it moves into ulnar deviation or ulnar flexion [11]. Differences in elongation of the RSC in the present study compared to the previous study can be explained by different coupling ratios of flexion to ulnar deviation. The average coupling ratio of the present study was $1.14 \pm 0.43^\circ$ of flexion for every 1° of ulnar deviation, while the previous study had a coupling ratio of approximately 3° of flexion for every 1° of ulnar deviation. The RSC is an important stabilizer of the carpus after proximal row carpectomy, preventing volar and ulnar translocation of the distal carpal row [22]. Our results suggest that the RSC elongates during ulnar flexion and elongates during extreme wrist extension, potentially serving to prevent volar translation of the proximal capitate in the midcarpal joint.

The Long Radiolunate Ligament

The LRL ligament is separated from the RSC by the interligamentous sulcus, and has been suggested to prevent ulnar and distal translation of the lunate, while not interfering with flexion and extension [11]. However, studies consistently report elongation of the LRL with wrist extension [16, 18, 19, 21], suggesting that the LRL restrains lunate extension. Our results are in agreement with reports indicating that the LRL restrains the lunate with wrist extension. Previous *in vitro* studies using differential displacement transducers found that the LRL elongates with wrist radial deviation [19, 20]. In contrast, a cadaveric study that tracked implanted tantalum beads with stereoradiography found that while the proximal fiber remained unchanged, the distal fiber increased in length with ulnar deviation [16]. An *in vivo* study that used similar methods as the present study showed an increase of approximately 20% from neutral to ulnar deviation [3]. Our results suggest that fibers of the LRL elongate similarly to the RSC but are generally closer to their maximum fiber lengths. Similar to the findings of Savelberg *et al* [16], we found that in wrist ulnar deviation the proximal fiber remained constant throughout the DTM, while the distal fiber increased with increasing ulnar flexion. Additionally, our results suggest that the LRL restrains the lunate in neutral-grip. Rupture of both the LRL and RSC are associated with lunate injuries, and

both have been suggested to prevent ulnar translocation of the carpus after four-corner-fusion [31]. The functional roles of both the LRL and RSC were analyzed in a cadaveric model where Nakamura *et al* showed radialstyloidectomy can lead to release of the LRL and RSC and ulnar and palmar wrist instability. Our results make sense clinically, as both of these ligaments are oriented such that they may prevent ulnar translation of the lunate during ulnar deviation. They are also elongated (along with the SRL) during extreme wrist extension, potentially preventing volar wrist instability.

The Short Radiolunate Ligament

Berger and Landsmeer [23] first identified the short radiolunate (SRL) and proposed that it may restrain wrist flexion and extension, influence lunate kinematics, and prevent distal translation of the lunate. Berger later suggested that some part of the SRL ligament may be under tension throughout the range of wrist motion [11]. The present study provides the first description of the *in vivo* elongation of the SRL. Our results suggest that the SRL restricts the lunate in extension but not in flexion. The wide fan-shape of the SRL, consisting of the radial, central, and ulnar fibers, elongated differently during the dart thrower's path. The radial fiber potentially functions as a check-rein at the end of the DTM (ulnar deviation and slight extension) while the most ulnar fiber restricts the lunate in radial-extension. During carpal distraction the lunate translated distally by less than half the amount of the scaphoid [7], suggesting that the lunate was more restricted. However, it did not appear that the SRL was responsible for limiting translation of the lunate because it did not elongate from unloaded to loaded distraction. We propose that the translation of the lunate is not limited in distraction by the SRL because the lunate is not directly connected to the distal row. Therefore, the distractive forces are transferred obliquely to the direction of loading through the scaphoid and triquetrum interosseous ligaments. In addition to the LRL and RSC, the SRL is associated with lunate injuries. It may elongate during extreme wrist extension to prevent volar instability of the lunate. Ruch *et al* reported that those patients with residual impaction of the lunate facet of the distal radius, where the SRL originates, can present with loss of supination and wrist extension [24]. We postulate that in this malunion, the

SRL may be abnormally elongated and that by limiting lunate extension it may partially restrain wrist extension.

The Dorsal Radiocarpal Ligament

The DRC was originally described as thin and weak compared to its volar counterparts [2]. However, it was more recently proposed that it plays a stabilizing role in the neutral wrist position [21]. Cadaveric studies of DRC elongation patterns report that it shortens from neutral to wrist extension and remains constant or decreases from neutral to ulnar and radial deviation [16, 21]. A recent *in vivo* study reported elongation of the DRC in radial deviation. Our results confirm that the DRC is recruited during unloaded distraction (similar to clinical neutral) and neutral-grip, but also during extreme flexion. Additionally, the DRC is potentially recruited at the beginning of the DTM, when the wrist is in radial extension. Cadaveric analysis of the DRC ligament has shown that it reinforces the lunotriquetral interosseous ligament, and prevents static volar intercalary segmental instability in the setting of lunotriquetral ligament tears [25]. We postulate that similar to its volar counterparts, because the DRC is elongated in extreme flexion, it may prevent dorsal subluxation of the radiocarpal joint.

The Dorsal Intercarpal Ligament

Viegas previously emphasized the importance of the DIC ligament in maintaining normal carpal stability and normal carpal kinematics [26]. Berger suggested the DIC ligament may act as a labrum to the head of the capitate or the proximal pole of the hamate, or it may serve to constrain midcarpal motion [11]. An *in vivo* study of DIC elongation found that the scaphoid attachment of the DIC elongates up to 12% from neutral to ulnar deviation [3]. Our results show that the DIC ligament was elongated the most during neutral-grip and unloaded distraction, and through a large range of the DTM. Recently, Mitsuyasu *et al* showed that the DIC ligament serves to support the scaphoid and lunate articulation and prevents static scapholunate instability in the presence of a scapholunate interosseous

ligament tear [27]. While our results suggest that the DIC may function to stabilize the radiocarpal joint, they also suggest that the DIC ligament may function to stabilize the midcarpal joint during grip, distraction, and through the DTM. While both the DRC and DIC remained elongated during wrist flexion, which includes motion of both the radiocarpal and midcarpal joints, only the DIC remained elongated during the DTM, where motion predominantly occurs at the midcarpal joint. Interestingly, elongation of the DIC during the DTM mirrored elongation of the distal fibers of the LRL and RSC, while the DRC shortened.

This study describes computed *in vivo* carpal ligament elongations throughout a large range of wrist motion that includes positions at the end-range of motion. Our results were largely in agreement with previous ligament elongation studies. While our approach cannot definitely report strain values, it uniquely allows us to explore the potential that the carpal ligaments provide restraint at end-range as well as mid-range motion. A ligament acting as an end-range limiter is analogous to the rope attached to a boat drifting from a dock. Once its rope becomes taut, the boat ceases drifting and maintains a constant distance. Once the boat reaches its maximum distance, it may drift radially around the dock, at a constant distance, while the rope remains taut. The latter situation is analogous to a ligament acting as a guide. Our results suggest that, in addition to acting as end-range limiters in multiple directions of wrist motion, the extrinsic radiocarpal and dorsal intercarpal ligaments may serve as guides throughout much of the range of wrist motion. Studies of ligament elongation throughout the normal range of wrist motion highlight the importance of the extrinsic carpal ligaments to normal wrist function, and give insight into pathologic carpal kinematics and potential surgical interventions.

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Chapter 5

Development and Validation of a Predicative Model of Carpal Bone Kinematics

5.1 Introduction

Current CT and MRI-based registration techniques can measure the three-dimensional posture of the carpal bones with sub-millimeter and sub-degree accuracy [1]. These techniques facilitate a new understanding of normal and pathological carpal kinematics *in vivo*. Additionally, they have allowed researchers to formulate or refine hypotheses about the kinetic mechanisms that dictate carpal bone motion. However, these mechanisms cannot be determined without knowledge of the cartilage-contact and ligament forces that act on the carpal bones. Currently, the direct *in vivo* measurement of these forces is not possible and an indirect method is required.

A quasi-static predictive model is capable of indirectly estimating the forces in the carpal joint [2-4]. This model can simulate the motion of the carpal bones based on estimates of cartilage-contact forces and ligament forces using the equations of statics. Cartilage

contact forces are proportional to the magnitude of the penetration between two articulating cartilage surfaces. Ligaments are modeled as non-linear tension only springs.

Previous studies that have developed predictive carpal models have validated their approach in relatively few subjects, or in a single cadaver specimen. Additionally, they have estimated ligament stiffness values based on published data [3–5]. However, there is a high degree of variability in the reported mechanical properties of the carpal ligaments. For example, failure strains of the extrinsic ligaments range from 20% to 50% between studies[6, 7].

Here, we augment these models by presenting a method to develop a subject-specific model of the carpus using a database of twelve wrists with associated experimental kinematics, acquired with clinical CT. Our approach assumes that each wrist has different ligament stiffness parameters. Optimal parameters are determined by performing multiple simulations that vary the stiffness of the carpal ligaments. Model error is quantified by computing the differences between experimentally determined scaphoid kinematics and simulated kinematics. The optimal parameters are those that minimize the error. The experimental kinematics contain tasks that include passive, end-range flexion and extension and carpal distraction. The database also includes incremental positions that sample a dart thrower’s motion. These positions may provide a more robust estimate of ligament stiffness parameters. These wrist positions also provide a unique opportunity to decouple the model tuning process from the validation process, where tuning is the search for parameters that yield optimal simulation results.

5.2 Methods

5.2.1 Simulated Carpal Bone Kinematics

This model simulates the motion of the scaphoid. The bones in the model are placed in two categories; unconstrained bones, and kinematic-constrained bones. The unconstrained bones are free to rotate and translate according to the external forces that act on them. The

kinematic-constrained bones drive the unconstrained bones, whose motion is obtained from *in vivo* CT data, through a specified motion path (Figure 5.1). Here, the scaphoid is the unconstrained bone, the radius is fixed, and the lunate, triquetrum, capitate, trapezium, and trapezoid are the kinematic-constrained bones.

The scaphoid begins in static equilibrium. Then, the postures of the kinematic-constrained bones are slightly incremented along a motion path. This advancement in posture strains the ligaments and cartilage that act on the scaphoid. The resulting ligament and cartilage-contact forces are applied to the scaphoid and its posture is updated until it reaches a new equilibrium position. The posture is updated using an Euler integration scheme that iterates until a specified maximum number of iterations (convergence steps) are reached. As the scaphoid converges on a posture that places it in static equilibrium, the resultant forces and moments acting at its center of mass dissipate. In this model, the kinematics applied to the kinematic-constrained are derived from computed tomography scans (CT-based).

5.2.2 Subjects

Kinematics of the scaphoid were simulated for all twelve subjects in the functional tasks database (Table 4.1).

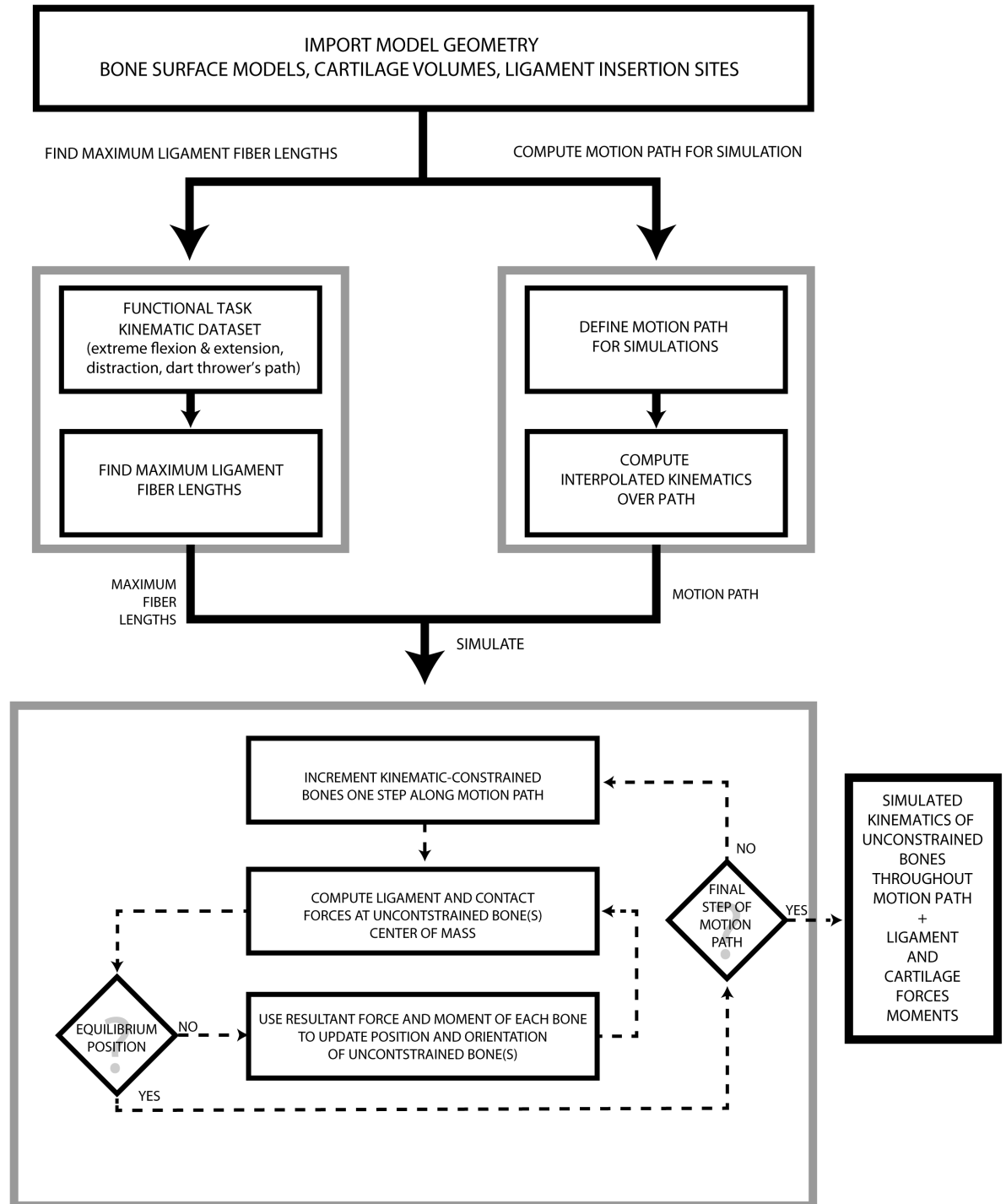


Figure 5.1: Overview of the predictive modeling algorithm.

5.2.3 Model Overview

Model geometry consists of the carpal bones, cartilage surfaces, and ligaments. *In vivo* kinematics were obtained from clinical CT-scans of the wrist in multiple static postures (Table 4.1). Cartilage contact forces were estimated using the method of deformable contact [2, 11]. Ligaments are modeled as non-linear tensile elements [2].

5.2.4 Cartilage Geometry

Cartilage surfaces were computed based on the joint space between two articulating bones over an area that includes closest point distances between the pair of bones, limited by a specified distance-threshold level [8]. Two surfaces, one extruded from each bone, were created by splitting the minimum inter-bone distance by half at each wrist position. To allow contact, each cartilage surface was extruded from its respective bone an additional 5% of this minimum distance in each position. The computed cartilage surfaces were validated with high resolution surfaces obtained from a cadaver model using μ CT (Appendix 3). We modeled cartilage contact between the scaphoid and the five adjacent bones (Figure 5.2, Table 5.1).

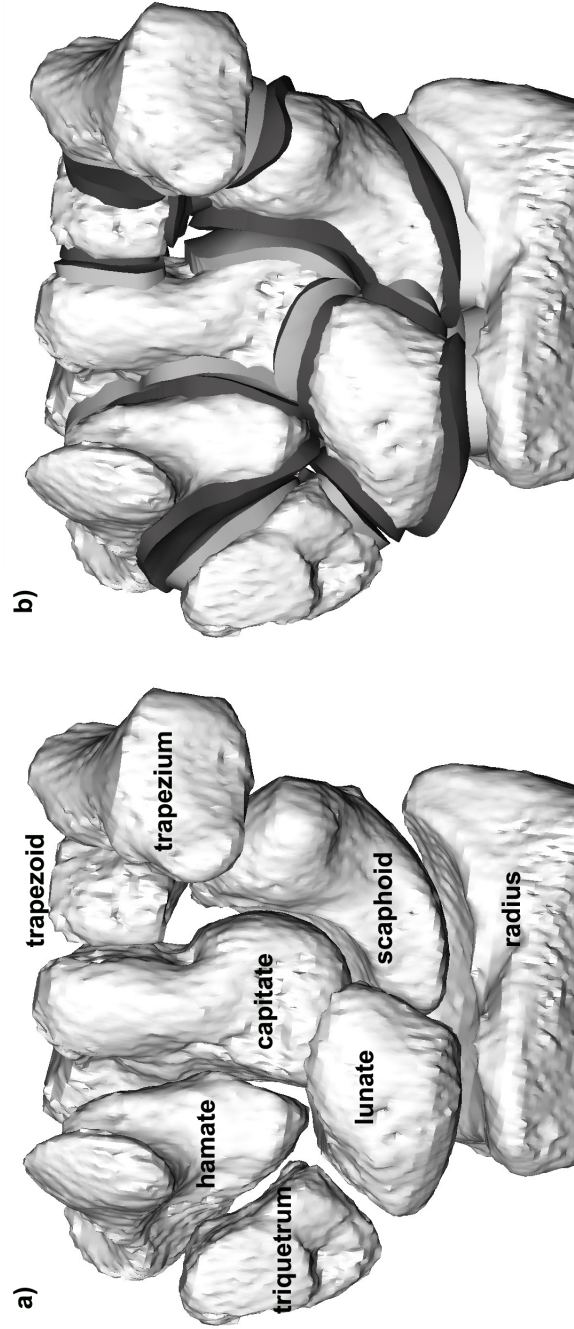


Figure 5.2: The radius and carpal bones in the neutral wrist position (right wrist, volar view). (a) Bone surface models derived from CT-scans. (b) Bone surface models with computed cartilage surfaces based on joint space calculations in multiple poses. Contact was modeled in the five bones that articulate with the scaphoid; the radius, lunate, capitate, trapezoid, and trapezium.

Table 5.1: The contacts modeled during the simulations of scaphoid motion (**bold**) and all possible contacts within the carpus. The carpometacarpal contacts are not shown.

Bone	radius	scaphoid	lunate	triquetrum	trapezium	trapezoid	capitate	hamate	Number of Contacts per Bone
radius	-	1	1	-	-	-	-	-	2
scaphoid	1	-	1	-	1	1	1	-	5
lunate	1	1	-	1	-	-	1	-	4
triquetrum	-	-	1	-	-	-	-	1	2
trapezium	-	1	-	-	-	1	-	1	3
trapezoid	-	1	-	-	1	-	1	-	3
capitate	-	1	1	-	-	1	-	1	4
hamate	-	-	-	1	-	-	1	-	2

5.2.5 Cartilage Contact Force Algorithm

Cartilage contact forces and moments were computed according to the method of deformable contact described by Blankevoort et al. Here, cartilage is modeled as a frictionless, isotropic, and linear elastic material. The contact algorithm was implemented in Matlab (Mathworks, Natick MA). Normal contact stress σ_n at each polygon is computed based on the amount of cartilage penetration

$$\sigma_n = S(u_n/b), \quad (5.1)$$

where S is the aggregate modulus, u_n is the penetration depth, and b is the combined thickness of both cartilage surfaces. The magnitude of the force f_n at each polygon node was computed as

$$f_n = \sigma_n A \quad (5.2)$$

where A is the area of the polygon.

Distance fields were computed around each cartilage volume to detect penetration. A master surface and slave surface were chosen for each contact. Penetration was computed by placing the polygon nodes of the master surface within the distance field of the slave surface. The direction of the force $\hat{\mathbf{f}}_n$ at the node was the negative of the polygon normal. Each $\vec{\mathbf{f}}_n$ was applied to the bone's center of mass and summed over m penetrating nodes.

$$\vec{\mathbf{F}}_C = \sum_{i=1}^m \vec{\mathbf{f}}_n \quad (5.3)$$

Newton's third law was enforced by mirroring the force vector $\vec{\mathbf{f}}_n$ associated with each penetrating node of the master surface and applying it to the center of mass of the bone associated with the slave surface. The resultant moment at the center of mass was

$$\vec{\mathbf{M}}_C = \sum_{i=1}^m \vec{\mathbf{r}} \times \vec{\mathbf{f}}_n \quad (5.4)$$

where $\vec{\mathbf{r}}$ is the moment arm directed from the polygon node to the respective bone's center of mass.

5.2.6 Ligament Fiber Model

Ligament fibers were defined according to methods described in Chapters 3 and 4. The ligament fibers that were modeled included the dorsal intercarpal, the scaphocapitate, the scaphotrapezium, and the volar, proximal, and dorsal fibers of the scapholunate interosseous ligament (Figure 5.3).

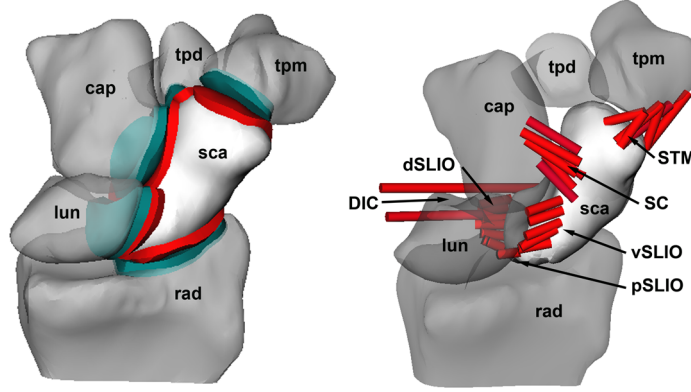


Figure 5.3: Model Definition. Contact forces were transmitted to the scaphoid (sca), from the fixed radius (rad), lunate (lun), capitate (cap), trapezium (tpm), and trapezoid (tpd). Modeled ligaments included the volar (vSLIO), proximal, (pSLIO) and dorsal (dSLIO) scapholunate interosseous ligament, the scaphotrapezium ligament (STM), the scapho-capitate ligament (SC), and the dorsal intercarpal ligament (DIC).

5.2.7 Ligament Model: Force Estimation

Ligament fibers were modeled as non-linear, tension-only springs. The non-linear behavior consisted of a small toe region when the fiber was initially strained. Following the toe region the force-length relationship of the ligament fiber was linear. A tuning constant c controlled the transition from the toe region to the linear region. The tuning constant c was set to 0.005 for the extrinsic-intrinsic ligaments and 0.05 for the interosseous ligaments. The force in the ligament fiber was zero below its computed resting length. The resting length l_o of the extrinsic and intrinsic (extrinsic-intrinsic) ligaments was computed based on the maximum fiber length, computed over the entire functional dataset using an assumption of the maximum possible strain ε_{max} (Chapter 4). Interosseous ligaments are reported to be more elastic and to operate under higher loads than the extrinsic-intrinsic ligaments

[7, 9, 10], therefore the resting length of the interosseous ligament fibers was set to the minimum length computed throughout all positions within the functional tasks dataset (Table 4.1). Ligament force was computed as a function of the stiffness constant k , the toe-region constant c , and the computed strain ϵ .

$$f(\epsilon) = kl_o(\epsilon + c(e^{-\frac{\epsilon}{c}} - 1)) \quad (5.5)$$

5.2.8 Simulated Motion Path Definition and Application of Kinematic Constraints

To perform a simulation, the relatively large rotations and translations between two CT-scanned positions were discretized into smaller increments according to the following kinematic interpolation method.

The change in position and orientation (pose) of a carpal bone from the neutral N scan to each subsequent scan p is represented by a rotation matrix $\mathbf{R}_{N \rightarrow p}$ and a position vector $\vec{\mathbf{v}}_{N \rightarrow p}$ referenced at the origin. For algebraic convenience the rotation matrix and position vector are written as a 4 x 4 transformation matrix,

$$\mathbf{T}_{N \rightarrow p} = \begin{bmatrix} & & & v_X \\ & \mathbf{R}_{N \rightarrow p} & & v_Y \\ & & & v_Z \\ 0 & 0 & 0 & 1 \end{bmatrix}. \quad (5.6)$$

The radius is mathematically fixed at each pose p ,

$$\mathbf{H}_{N \rightarrow p} = \mathbf{T}_{Radius_{N \rightarrow p}}^{-1} * \mathbf{T}_{Bone_{N \rightarrow p}}. \quad (5.7)$$

where $\mathbf{H}$ represents a transform with respect to the fixed radius. Let p_A represent the

initial posture of a given bone and p_B represent the final posture. First, the relative motion between p_A and p_B is computed,

$$\mathbf{H}_{A \rightarrow B} = \mathbf{H}_{N \rightarrow p_B} * \mathbf{H}_{N \rightarrow p_A}^{-1}, \quad (5.8)$$

and expressed as the helical axis of motion,

$$\mathbf{H}_{A \rightarrow B} \rightarrow \begin{pmatrix} \phi_{A \rightarrow B} \\ \hat{\mathbf{n}}_{A \rightarrow B} \\ t_{A \rightarrow B} \\ \mathbf{q}_{A \rightarrow B} \end{pmatrix}. \quad (5.9)$$

$\hat{\mathbf{n}}_{A \rightarrow B}$ is a unit vector defining the orientation of the rotation axis in space and $\mathbf{q}_{A \rightarrow B}$ is a point on the axis. The change in pose from p_A to p_B is achieved by a rotation of $\phi_{A \rightarrow B}$ about $\hat{\mathbf{n}}_{A \rightarrow B}$ and a translation of $t_{A \rightarrow B}$ along $\hat{\mathbf{n}}_{A \rightarrow B}$. Interpolated values ϕ_i and t_i between p_A and p_B are computed by holding $\hat{\mathbf{n}}_{A \rightarrow B}$ and $\mathbf{q}_{A \rightarrow B}$ fixed and interpolating linearly between $[0, \phi_{A \rightarrow B}]$, and $[0, t_{A \rightarrow B}]$ at $i = \{1, 2, 3 \dots n\}$ nodes. Here, n represents the number of interpolation steps. At each interpolated value of ϕ_i and t_i the helical axis parameters are converted back to a matrix that describes the transformation from neutral to the interpolated position p_i ,

$$\begin{pmatrix} \phi_i \\ \hat{\mathbf{n}}_{A \rightarrow B} \\ t_i \\ \mathbf{q}_{A \rightarrow B} \end{pmatrix} \rightarrow \mathbf{H}_{A \rightarrow p_i} \quad (5.10)$$

$$\mathbf{H}_{N \rightarrow p_i} = \mathbf{H}_{A \rightarrow p_i} * \mathbf{H}_{N \rightarrow p_A} \quad (5.11)$$

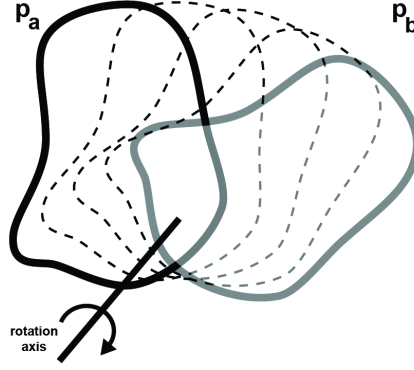


Figure 5.4: A representation of kinematic interpolation about the rotation axis. The rotation axis is computed between two poses p_a (solid black) and p_b (solid grey). Linear interpolation was performed over the rotational component ϕ and the translational component $\mathbf{t}$ of the helical axis of motion. This created smooth, incremental kinematics (dashed bones) between the two discrete postures that were scanned with clinical CT.

5.2.9 Model Tuning

The model was tuned by allowing ligament stiffness values to vary and by holding the remaining model parameters fixed. Since previous work shows model results to be relatively insensitive to cartilage stiffness values, ligament stiffness was chosen as the variable parameter [4]. Stiffness parameters were varied over multiple simulations of scaphoid kinematics as the wrist moved from unloaded distraction to loaded distraction (Table 4.1). In each simulation, two ligament stiffness values were given, one for the interosseous ligaments (vSLIO, pSLIO, and dSLIO), and one for the extrinsic and intrinsic ligaments (extrinsic-intrinsic) (DIC, SC, and STM). Interosseous ligament stiffness varied between 1 and 200 N/mm and the extrinsic-intrinsic ligament stiffness varied between 1 and 100 N/mm. Each simulation incorporated combinations of interosseous and extrinsic-intrinsic stiffness values by increments of 20 N/mm for the interosseous ligaments, and 10 N/mm for the extrinsic-intrinsic ligaments. We evaluated the magnitude of the rotational error between the simulated and kinematic-constrained scaphoid after each simulation. The optimal parameters were those that produced the smallest errors between experimental scaphoid rotation and simulated scaphoid rotation.

5.2.10 Model Tuning: Analysis

Contour plots of the rotational error as a function of the interosseous and extrinsic-intrinsic stiffness values were generated and examined qualitatively. Stiffness values that minimized the error were recorded for each subject. These minimum values were used in the tuned model. Additionally, the range of error was also reported.

5.2.11 Validation

The model was validated by comparing simulated scaphoid kinematics to experimental scaphoid kinematics as the wrist moved from neutral to extension. The initial position was unloaded distraction and the final position was neutral-grip (Table 4.1). These two positions effectively simulated wrist motion from clinical neutral (third metacarpal aligned with the forearm) to an extended gripping posture. Translational differences between the kinematic-constrained pose and the simulated pose were expressed as the change in distance between bone centroids. Rotational differences were expressed as Euler angles between the kinematic-constrained and the simulated orientation of the scaphoid. These translational and rotational differences were reported with respect to a radius-based coordinate system [11, 12] (Figure 5.5).

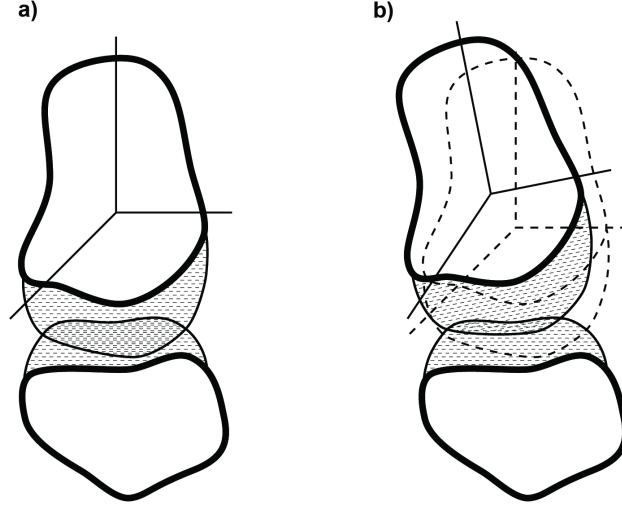


Figure 5.5: A representation of two bones (a) measured with CT (kinematic-constrained) and (b) adjusted to satisfy static equilibrium (unconstrained). Note in (b) there is a discrepancy in the coordinate systems of the kinematic-constrained and unconstrained bones. This coordinate system is the radius-based coordinate system moved to each bone's center of mass. Euler angles between the kinematic-constrained bone and unconstrained bone describe the rotational error. The distance between centroids, reported in the coordinate system of the kinematic-constrained bone, described the translational error.

5.2.12 Sensitivity Analysis

The sensitivity of the model to ϵ_{max} of the extrinsic-intrinsic ligaments, cartilage stiffness, kinematic interpolation steps, and convergence steps was determined by evaluating the error associated with varying these parameters one at a time. Each parameter was set at two different values above and below the default value (Table 5.2). We evaluated the sensitivity of the model to these parameters in one randomly selected subject by simulating scaphoid extension. Once the most sensitive parameter was identified, model error was visually evaluated by adjusting this parameter.

Table 5.2: Range of values tested for each parameter in the sensitivity analysis. When one parameter was varied, the remaining parameters were set to the default value.

	Default Value	Run 1	Run 2	Run 3	Run 4
Maximum Strain, ϵ_{max}	0.15	0.01	0.07	0.22	0.29
Cartilage Stiffness (MPa)	5	1	3	7	9
Kinematic Interpolation Steps	10	1	5	15	20
Convergence Steps	200	50	150	250	300

5.3 Results

5.3.1 Tuning: Carpal Distraction

Tuning the model by varying ligament stiffness as the scaphoid distracted resulted in minimum errors of $3.9 \pm 2.0^\circ$ (range: 1.8° - 8.6°), and maximum errors of $9.2 \pm 3.3^\circ$ (range: 5.3° - 15.0°) averaged across subjects. The minimum error corresponded to a wide range of stiffness values across subjects. Interosseous stiffness values averaged 70 ± 45 N/mm (range: $20 - 150$ N/mm) and extrinsic-intrinsic stiffness averaged 55 ± 32 N/mm (range: $20 - 100$ N/mm). Inspection of the contour plots of rotational error as a function of stiffness revealed wide ranges of minimized error within and amongst subjects (Figure 5.6).

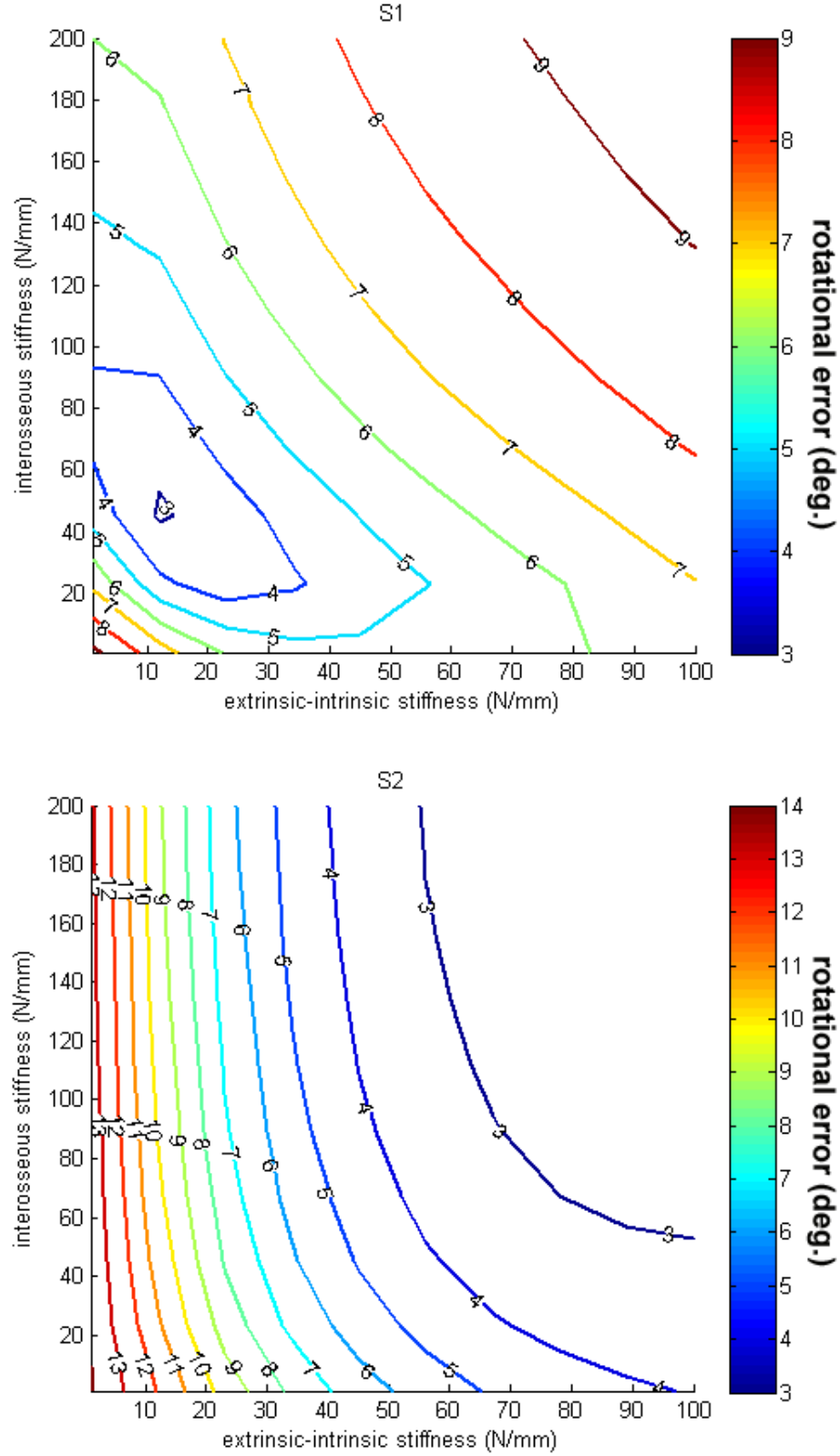


Figure 5.6: Contour plots of the magnitude of rotational error as a function of interosseous and extrinsic-intrinsic ligament stiffness of two subjects. The color bar and contour lines represent the magnitude of the rotational error in degrees. One hundred simulations were performed on the scaphoid as the wrist was loaded in distraction. These subjects were chosen to illustrate the differences in regions that minimized the error.

5.3.2 Validation: Wrist Extrension

Rotational and translational errors in wrist extension ranged from 0.5° to 8.5° and from 0.5 mm to 1.7 mm across subjects. (Table 5.3). These errors were of similar magnitude to simulated distraction using the same parameters.

Table 5.3: Rotational and translational errors between the tuned, simulated scaphoid and the CT-based scaphoid. The subject-specific, tuned parameters obtained during carpal distraction (Distraction) were applied to a simulation where the wrist extended $18.6 \pm 5.5^\circ$ (Extension).

		Distraction	Extension
		median (min, max)	median (min, max)
Rotation Error (degrees)	X	-2.4 (-7.4,-0.5)	-3.5 (-8.45, -2.1)
	Y	-2.8 (-5.2,0.8)	-0.9 (-3.7, 2.9)
	Z	-0.2 (-4.1,3.0)	0.5 (-3.0,2.0)
Translation Error (degrees)	X	-0.4 (-1.6,0.3)	-0.03 (-0.8,0.5)
	Y	0.2 (-2.1,1.4)	0.6 (-1.0,1.7)
	Z	-0.04 (-0.7,1.8)	-0.3 (-1.7,0.9)

5.3.3 Sensitivity of Simulated Scaphoid Extension to Model Parameters

The model was insensitive to the specified ranges of cartilage stiffness, convergence steps, and kinematic interpolation steps compared to maximum strain. In fact, setting ε_{max} to 0.07 from 0.15 reduced the rotational error of the scaphoid to 1.5° from 4.3° during simulated extension (Figure 5.7). Cartilage stiffness, convergence step, and kinematic interpolation step ranges of 9 MPa, 250 iterations, and 17 iterations respectively resulted ranges of error that were less than 1.5° . Setting ε_{max} to 0.5 resulted in errors in distraction and extension of 17.1° and 23.9° respectively (Figure 5.8).

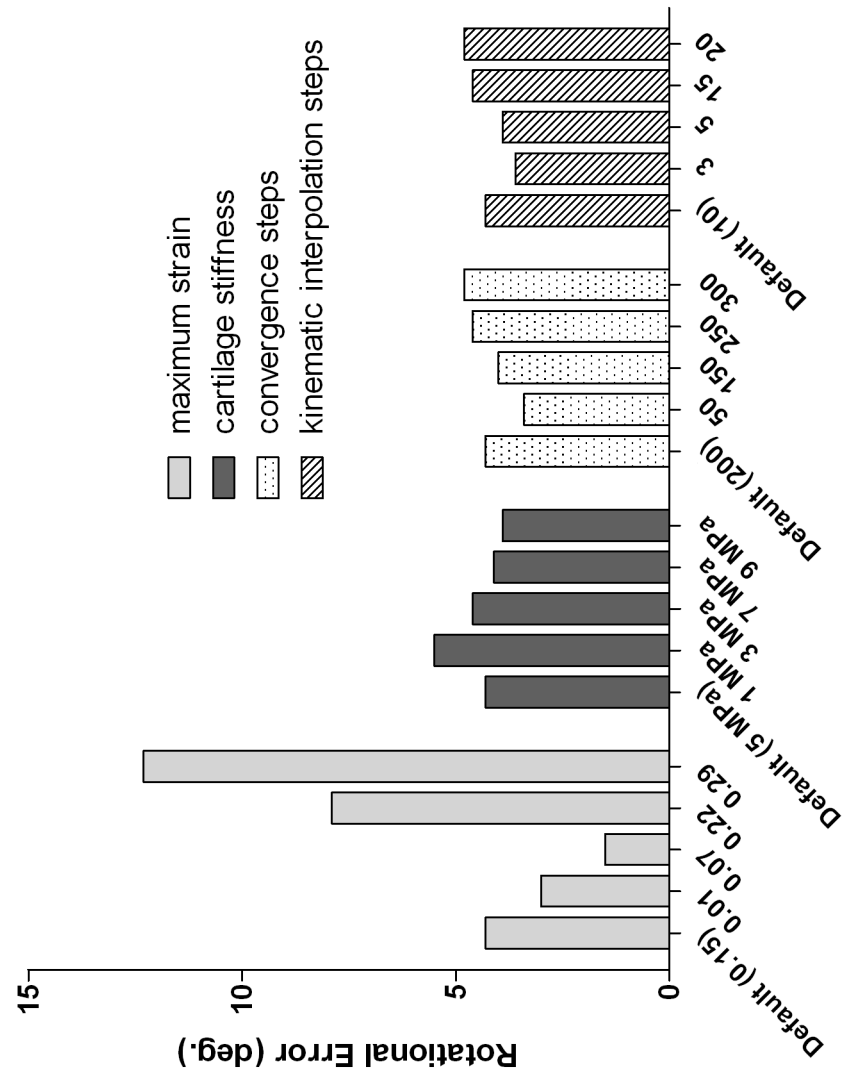


Figure 5.7: Sensitivity analysis of the simulated scaphoid during wrist extension. The model was relatively insensitive to cartilage stiffness, convergence steps, and kinematic interpolation steps compared to the choice of $\epsilon_{\max}$ within the specified ranges.

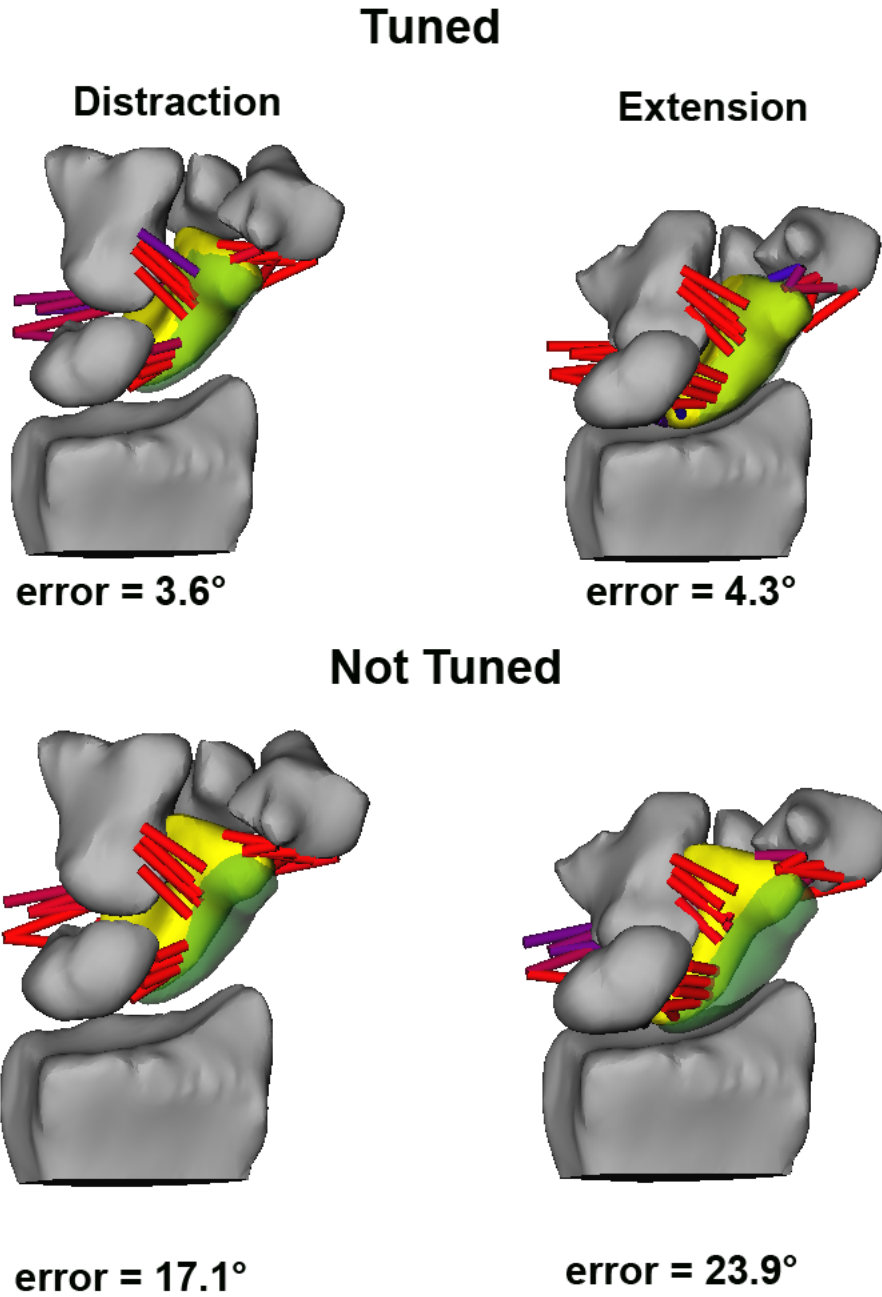


Figure 5.8: (top) Overlays of simulated and CT-based scaphoids using tuned ligament stiffness values and $\varepsilon_{max} = 0.15$. (bottom) Simulated and CT-based scaphoid with $\varepsilon_{max} = 0.5$

5.4 Discussion

This study constructed and validated a model that simulated scaphoid kinematics that were driven by the CT-based kinematics of neighboring bones. While the model was implemented on the scaphoid, it is capable of simulating any number of bones. The model was tuned for each subject by varying ligament stiffness until the simulated scaphoid kinematics during carpal distraction were optimized. We found that the set of stiffness values that optimizes scaphoid kinematics in distraction results in similar kinematic error during wrist extension. Additionally, the model remains stable given modest changes in cartilage stiffness, kinematic step size, and convergence step size parameters, but it is sensitive to maximum strain.

We found that the ligament stiffness values that minimized errors between simulated and CT-based scaphoid kinematics were variable amongst subjects. While identifying the origin of this variability is beyond the scope of this study, we postulate that it results from variability in joint laxity. It has been previously demonstrated that a population of healthy subjects have variable degrees of clinically measured joint laxity and that this laxity is highly correlated with scaphoid kinematics [13]. These results typically have implications in modeling studies which analyze few subjects and generalize their results to a population [4, 14]. This study provides evidence that multiple subjects are needed to evaluate characteristics of a model.

There are several important limitations and assumptions associated with this model. First, ligaments that interact with the scaphoid through contact were not modeled. These ligaments are thought to be important stabilizers of the wrist at the end range of motion, but they may also guide motion throughout a larger range (Chapter 4). However, errors between simulated and CT-based kinematics were low, indicating that these ligaments may not significantly affect the driving forces on the scaphoid. Similarly, the interaction between ligament fibers was not modeled. Because the carpal ligaments are a confluence of tissues, these interactions undoubtedly affect motion in the carpus. Additionally, cartilage-contact was modeled using a simple, penetration-based model. Cartilage is not a linear, isotropic material. However, cartilage is relatively stiff and we assert that local deforma-

tions do not need to be taken into account when considering the role of cartilage in carpal bone kinematics. Finally, the integration and convergence scheme was limited by a specified number of iterations. More sophisticated integration schemes are available and may improve model fidelity. The purpose of this model, however, was to determine the contributions of cartilage-contact and ligament forces to carpal bone motion, and this simple integration scheme seemed adequate for achieving that goal.

Previous studies have used predictive models to estimate joint contact forces in a single static position [3, 5, 15]. Emerging models are capable of simulating carpal bone motion [2, 5]. These models have simulated motion in a single subject. This study extends these models by evaluating the stability of the model to the input parameters in multiple subjects. Fischli et al. tuned a predictive model in ulnar-radial deviation and validated it in extension [2]. This method effectively decouples the tuning process from the validation process. We accomplished this by tuning the model in carpal distraction and validating it in extension, demonstrating that the model is robust. This tuning and validation method provides confidence that the model is simulating the physical system appropriately. The work described here represents a novel method for modeling carpal bone kinematics. The method drives a set of bones through kinematics measured with CT and allows other bones to move based on the external forces that act on them. This method is capable of determining the underlying forces that yield the complex kinematic patterns observed in the carpus.

5.5 References

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Chapter 6

Contributions of Cartilage and Ligaments to the Scaphoid Flexion and Ulnar Deviation Moment during the Dart Thrower's Motion

6.1 Introduction

The scaphoid is the most widely studied bone in the wrist because it transmits the majority of the load from the carpus to the forearm [1], bridges the distal and proximal carpal rows [2], and is susceptible to injury [3]. Fracture of the scaphoid or injury to its surrounding system of ligaments can lead to wrist instability and secondary degenerative joint diseases [4].

Numerous studies have quantified the three-dimensional kinematics of the scaphoid throughout the full range of wrist motion [5-8]. These studies have determined that the scaphoid flexes with wrist flexion or radial deviation and it extends with wrist extension or ulnar deviation. The dart thrower's motion (DTM) is defined as the path from wrist extension

with radial deviation toward wrist flexion with ulnar deviation [8-12]. The DTM defines the transition between scaphoid and lunate flexion with wrist flexion and scaphoid and lunate extension with wrist ulnar deviation [8, 13]

While studies of *in vivo* carpal kinematics have accurately described the complex motion of the scaphoid, the underlying mechanisms of scaphoid rotation have not been determined. Therefore, the purpose of this study was to compute the relative contribution of the ligament forces and cartilage-contact forces driving rotation of the scaphoid as the wrist moves along the dart thrower’s path. Our approach simulates scaphoid kinematics with a subject-specific predictive model, driven with *in vivo* kinematics [14].

6.2 Methods

6.2.1 Data Acquisition and *In Vivo* Kinematics

Following IRB approval and informed consent, 12 subjects (age: 24.8 ± 3.8 years) were computed tomography (CT)-scanned in 5 static wrist positions that sampled a path of wrist motion from radial-extension toward ulnar-flexion. Detailed descriptions of the protocol, segmentation and markerless registration process, and carpal bone kinematics were previously reported [15] (Table 6.1, Appendix 2, and Table 4.1). Relative motion between each of the 5 sequential positions was calculated for the radius, ulna, all eight carpal bones, and the metacarpals, resulting in four motion-steps. These motion-steps were expressed with the helical axis of motion variables and resolved into the radius-based coordinate system [16]. Scaphoid rotation was described in terms of flexion and extension, ulnar-radial deviation, and pronation-supination with respect to the radius-based coordinate system. To simplify the analysis, scaphoid pronation-supination was not considered.

Table 6.1: Wrist positions along the dart thrower’s motion, Average ($\pm$ SD). F = flexion, E = extension, U = ulnar deviation, and R = radial deviation.

Position	F(+), E	U(+), R
P1	-59.6 (9.6)	-5.4 (8.9)
P2	-48.7 (9.6)	1.5 (9.0)
P3	-37.1 (7.9)	12.9 (9.3)
P4	-24.4 (7.9)	26.1 (6.5)
P5	-8.8 (10.7)	34.1 (5.8)

6.2.2 Model Description

A detailed description of the predictive model and tuning process is available in Chapter 5, and is summarized as follows.

6.2.2.1 Cartilage-Contact

Cartilage surfaces were computed by tracking the minimum inter-bone joint space over multiple wrist positions and dividing the distance by half. To ensure contact, 5% of the minimum distance was added to the cartilage [17]. Cartilage forces were computed using the method of deformable contact [18].

6.2.2.2 Ligament Fiber Model

The manual mapping of ligament insertion sites onto each subject’s polygonal bone surface models were guided by previously published descriptions [19-21]. Six ligaments attached to the scaphoid were modeled as non-linear, tension-only elements (Figure 5.3). A ligament produced force only when its length was greater than its computed resting length. To compute resting length, the maximum strain of the extrinsic carpal ligaments was assumed to be less than 15%. The resting length of the interosseous ligaments was set to its minimum length across all wrist positions. This modeling choice was based on evidence that the interosseous ligaments are more elastic than the extrinsic ligaments, and that they operate under higher loads [22-24].

6.2.2.3 Quasi-static Simulation of Scaphoid Kinematics

All bones in contact with the scaphoid were driven through each motion-step in small increments, creating unbalanced forces on the scaphoid due to cartilage-contact and tension in the ligaments. The net force and moment were resolved at the scaphoid's center-of-mass and the position and orientation of the scaphoid was updated using simple forward Euler integration until the unbalanced force and moment dissipated (static equilibrium) [14].

6.2.3 Analysis

6.2.3.1 Model Fidelity

To validate the model, linear regression was performed on simulated and CT-based scaphoid flexion-extension and ulnar-radial deviation as a function of the magnitude of wrist flexion and ulnar deviation in a given motion-step. Additionally, RMS differences between simulated and CT-based scaphoid flexion-extension and ulnar-radial deviation were computed.

6.2.3.2 Scaphoid Flexion and Extension as a Function of Wrist Ulnar-Flexion

Scaphoid kinematics were pooled across all subjects and motion-steps, and then divided into two groups. The groups were defined by scaphoid flexion and extension. Motion-steps where the scaphoid flexed or extended less than 2.5° were excluded from the analysis. The net moment acting on the scaphoid was decomposed into flexion-extension, ulnar-radial deviation, and pronation-supination components. The net flexion-extension and ulnar-radial deviation moments were further decomposed into contributions from the ligament and cartilage-contact forces. Within each group, the moments were normalized by the absolute value of the maximum ligament, cartilage-contact, or net moment within each incremental motion-step. These normalized moments were then averaged across all motion-steps within the pooled groups of scaphoid flexion or scaphoid extension. The normalized moment contributions represent general trends; their quantities are only relevant within the context of each group. Therefore, qualitative, rather than quantitative, descriptions are provided.

6.3 Results

6.3.1 Model Fidelity

Visually, the simulated and CT-based scaphoids were aligned throughout the dart thrower's path (Figure 6.1). Simulated scaphoid flexion and extension consistently underestimated CT-based scaphoid flexion and extension; however, the trends of scaphoid flexion with wrist ulnar-flexion were similar (Table 6.2, Figure 6.2). There were no differences detected between simulated and CT-based scaphoid ulnar deviation with wrist ulnar-flexion. RMS differences between simulated and CT-based scaphoid flexion and ulnar deviation were 3.1° and 0.8° respectively.

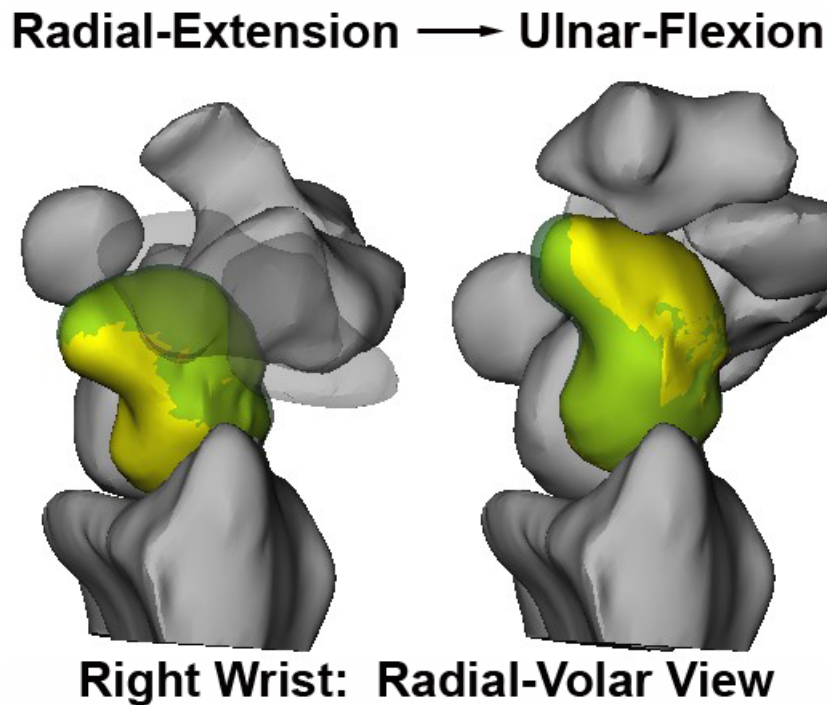


Figure 6.1: Simulated (yellow) versus CT-based (green) scaphoid of a representative subject (right wrist: radial view). Visually, there was good alignment between the simulated and CT-based scaphoid throughout the dart thrower's motion.

Table 6.2: Regression analysis between simulated and CT-based scaphoid kinematics as a function of the change in wrist posture. Slope $\pm$ SE. $P < 0.05^*$ denotes significant differences between simulated and CT-based kinematics. Units of slope: degrees of scaphoid rotation divided by degrees of change in wrist posture. NS = slope not significantly different than zero.

Change in Wrist Posture	Scaphoid F(+), E			Scaphoid U(+),R		
	CT	SIM	P	CT	SIM	P
ΔFE	0.9 ± 0.2	0.5 ± 0.1	0.13	-0.14 ± 0.1	-0.13 ± 0.1	0.96 NS
ΔUR	-1.2 ± 0.1	-0.8 ± 0.1	$< 0.05^*$	0.4 ± 0.1	0.4 ± 0.1	0.62

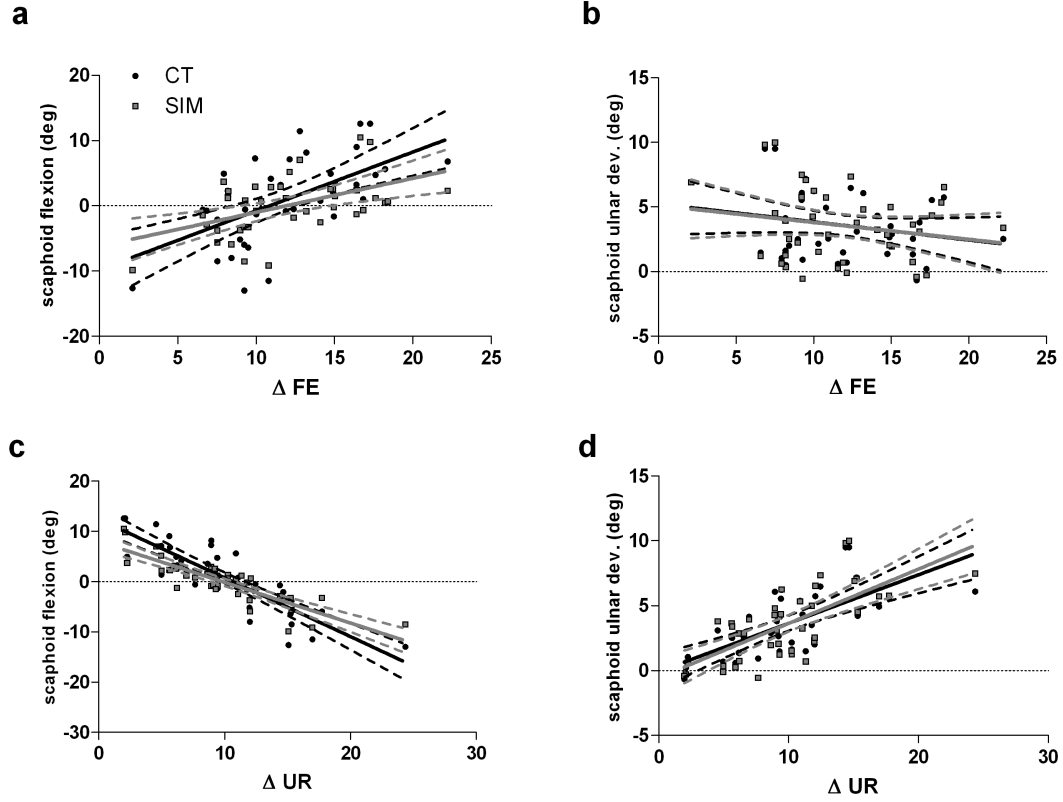


Figure 6.2: Scaphoid motion as a function of the increase wrist flexion (top) or ulnar deviation (bottom) in a given step. There was no difference in scaphoid flexion (a) or ulnar deviation (b) as a function of wrist flexion between simulated and CT-based kinematics. Although there was a significant difference between simulated and CT-based scaphoid flexion as a function of wrist ulnar deviation (c), the trends were similar. There was no difference in scaphoid ulnar deviation as a function of the wrist ulnar deviation (d).

6.3.2 Contributions of Cartilage-Contact and Ligaments to the Net Moment

6.3.2.1 Scaphoid Flexion and Extension

During scaphoid flexion, the net flexion moment was due to cartilage contact (Figure 6.3a). In contrast, during scaphoid extension, the primary extension moment was generated by the carpal ligaments, which opposed the flexion moment due to contact.

As the wrist flexed, scaphoid flexion was driven by cartilage contact from the lunate (Figure 6.3b). The net moment generated by the ligaments approached zero due to extension moments imposed by the DIC, SC, STM, and pSLIO balancing flexion moments imposed by the dSLIO and vSLIO.

When the wrist ulnar deviated more than it flexed, the scaphoid extended due to a reduced flexion moment from contact and an increased ligament extension moment (Figure 6.3a). In these subjects, the contact flexion moment was reduced due to a decreased flexion moment imposed by the trapezium and trapezoid, and an increased extension moment imposed by the capitate (Figure 6.3b). The ligament extension moment increased as the flexion moment imposed by the dSLIO and vSLIO decreased (Figure 6.3c).

6.3.2.2 Scaphoid Ulnar Deviation

As the wrist ulnar flexed, the scaphoid always ulnar deviated, regardless of whether it was flexing or extending (Figure 6.3d). The ulnar deviation moment due to contact was similar between scaphoid flexion and extension (Figure 6.3d, 6.3e). Increased ulnar deviation during scaphoid extension was caused by a reduced radial deviation moment from the carpal ligaments (Figure 6.3d, 6.3f).

Scaphoid ulnar deviation was driven by an ulnar deviation moment imposed by the lunate (Figure 6.3e). This moment was incompletely balanced by an opposed radial deviation

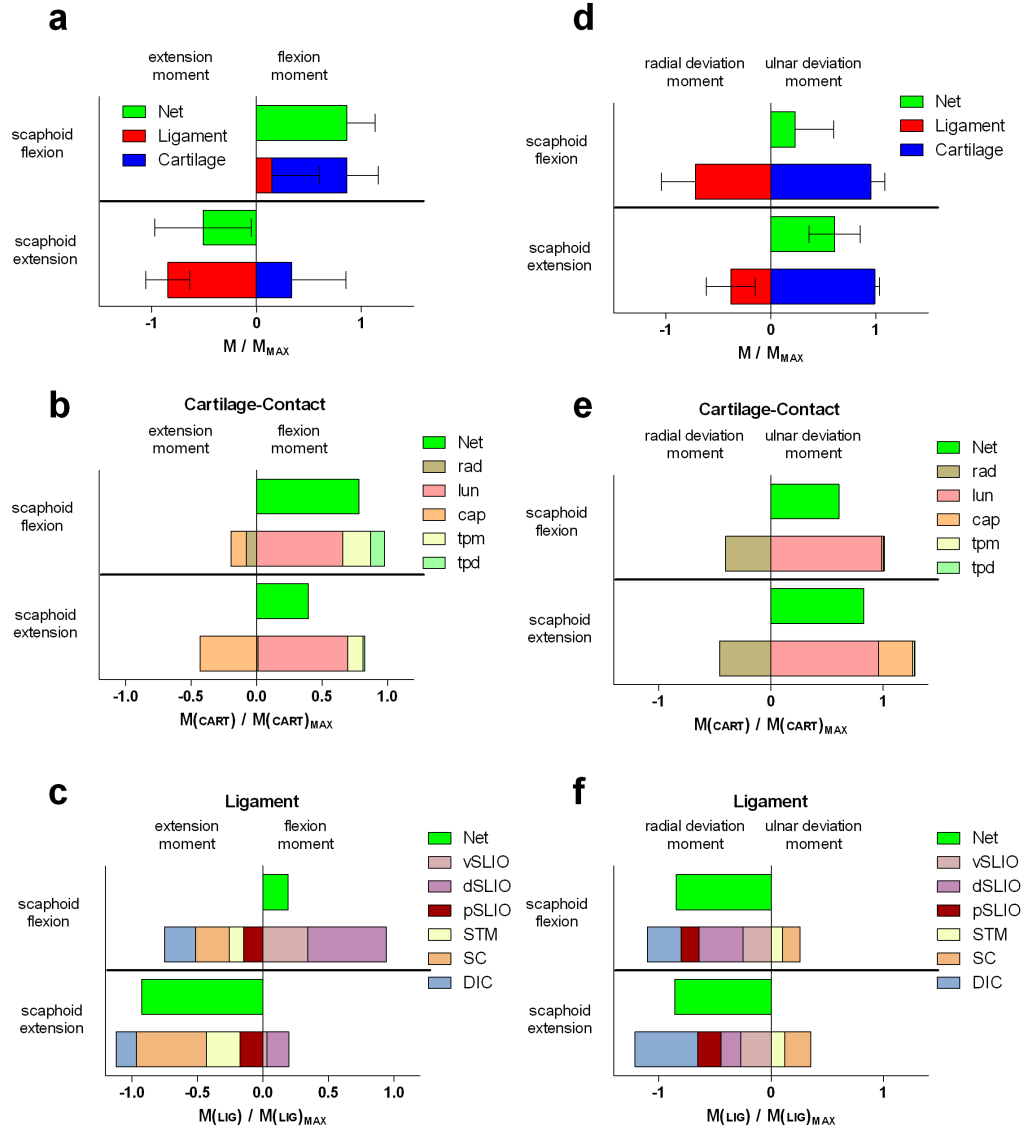


Figure 6.3: Contributions from the ligaments and cartilage-contact to the flexion-extension (left) and ulnar-radial deviation (right) moments acting on the scaphoid. A flexion moment due to the cartilage contacts caused scaphoid flexion during wrist flexion. An extension moment due to the carpal ligaments caused scaphoid extension during wrist ulnar deviation. Scaphoid ulnar deviation was primarily caused by an imposed ulnar deviation moment from contact with the lunate.

moment from the radius and the carpal ligaments. The carpal ligaments, with the exception of the SC and STM, supplied an opposing radial deviation (Figure 6.3f).

6.4 Discussion

A novel method to simulate and validate scaphoid kinematics using *in vivo* kinematics is presented, allowing us to determine the underlying mechanisms that drive scaphoid motion. We observed that as the wrist moved through the dart thrower’s path, scaphoid flexion and extension depended on the coupling ratio of flexion to ulnar deviation. When the wrist ulnar deviated more than it flexed, the scaphoid extended. Conversely, when the wrist flexed more than it ulnar deviated, the scaphoid flexed. We found that the ligaments controlled scaphoid extension when the wrist ulnar deviated and joint contact controlled scaphoid flexion when the wrist flexed.

Theories on the mechanism of the carpal joint describe the lunate as the controller of scaphoid flexion or extension [25, 26]. One theory describes scaphoid extension with wrist ulnar deviation occurring as a result of the triquetrum moving volarly on the hamate and transmitting an extension moment to the scaphoid through the lunate via the lunotriquetral and scapholunate interosseous ligament [25]. Kauer noted that the lunate and scaphoid taper from volar to dorsal and indicated that this slope would create a tendency for both bones to extend [27]. The theory stated that the trapezium-trapezoid joint prevent scaphoid extension during flexion and allows extension during ulnar deviation. We found that the lunate supplies the majority of the ulnar-flexion moment throughout the dart thrower’s path regardless of whether the scaphoid flexes or extends. Additionally, the dorsal and volar aspects of the scapholunate interosseous ligament always supplies a flexion moment to the scaphoid. These results indicate that scaphoid extension with wrist ulnar deviation is not caused by an extension moment imposed by the lunate. Instead, as the wrist ulnar deviated, scaphoid extension was explained by an increased extension moment from the scaphocapitate, scaphotrapezium, and dorsal intercarpal ligament, and an increased extension moment imposed on the scaphoid by the articular surface of the capitate.

Scaphoid flexion was primarily caused by a flexion moment imposed by the articular surfaces of the lunate, trapezoid, and trapezium. Additionally, the flexion-extension moment due to the ligaments was balanced by the scapholunate interosseous ligament.

When interpreting these results, there are several limitations to consider. However, we believe that these limitations did not significantly affect the accuracy of our results. First, motion was inferred from static wrist positions acquired from CT. However, static positions have been shown to approximate dynamic wrist motion [28], and the hysteresis effect on carpal kinematics varies as a function of path by less than 2° [29]. A related limitation is that the simulation depended on incremental wrist positions interpolated from larger motion-steps. However, the fidelity of the model demonstrates that interpolated kinematics over motion-steps on the order of 20° is a reasonable approach. Finally, the model parameters such as cartilage and ligament stiffness, ligament resting length, and convergence criteria were determined such that simulated kinematics most closely matched the kinematics measured with CT. However, this tuning process was performed on wrist motions that were not included in the analysis, demonstrating the robustness of the model. One final limitation is inherent to the modeled system. The carpus is a redundant system; several mechanisms and combinations of ligaments and cartilage-contact are capable of producing the same driving forces. It is possible that a different set of ligament and cartilage forces could create similar motions. However, the system was modeled using the best available information on the geometry and properties of the carpal joint. The model assumptions and tuning approach utilized here give us confidence that the contributions to the moments driving scaphoid flexion and extension are appropriate.

To our knowledge, this is the first description of the contributions from cartilage-contact and ligament forces to the moments that drive the scaphoid as the wrist moves through the dart thrower's path. We have identified a mechanism for scaphoid extension with wrist ulnar deviation that depends on the complex interplay between the cartilage-contact and ligaments. The importance of the scaphotrapezium, scaphocapitate, and scapholunate interosseous ligaments during load transfer has been previously identified [30-32]. Our findings suggest that they are also important structures to the kinematics of the scaphoid.

Understanding scaphoid kinematics will help elucidate the mechanism of carpal stability and the pathomechanics of trauma and of degenerative diseases, such as scaphoid nonunions and STT arthritis.

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Chapter 7

Conclusions

Overall, this dissertation provides new insights into carpal joint function with an emphasis on carpal ligament function. An *in vivo* database of carpal bone surfaces and kinematics was the foundation of these studies. This work identified previously unknown kinematic patterns during extreme passive motions, as well as ligament elongation patterns during both active and passive wrist positions. We also highlight the capability of a predictive carpal model to augment traditional kinematic analyses. Optimal model parameters were found with a novel tuning approach and scaphoid kinematics were simulated as the wrist moved through the dart thrower's motion. The predictive model identified the cartilage contact and ligament forces responsible for scaphoid flexion, extension, and ulnar deviation.

A primary focus of the presented work was to extend the current understanding of carpal ligament function. To this end, ligament fiber paths were computed in both active and passive wrist positions. We concluded that the dorsal radiocarpal and dorsal intercarpal ligaments limit motion at the end range of ulnar and radial deviation, respectively, and may potentially guide motion throughout the full range of flexion-extension and the dart thrower's path. Our results suggest that, even limited approaches to wrist surgery, if associated with scarring of the dorsal extrinsic ligaments, would lead to deficits in wrist range of motion. We postulate that alternative approaches to open wrist surgery, such as arthroscopy along with early range of motion, may decrease the risk of post-operative wrist stiffness.

Extreme loaded wrist motions, such as extreme loaded flexion and extreme loaded extension, are important because many of the extrinsic ligaments are likely to be taut in these positions. We quantified the kinematic patterns of the scaphoid and lunate from neutral-grip to these extreme positions and found that the scaphoid and the lunate contribute approximately 25% less in extreme loaded extension compared to active wrist extension, indicating that wrist extension shifts the motion of the carpus from the radiocarpal joint toward the midcarpal joint. These findings are not surprising considering the robust network of supporting extrinsic volar carpal ligaments as well as the interosseous ligaments. We postulate that as the wrist moves into extreme extension, scaphoid and lunate extension is restricted by the volar radiocarpal ligaments. Because the external load continues to apply an extension moment on the wrist, the capitate extends within the scapholunate fossa. This mechanism may explain the etiology of scapholunate ligament interosseous rupture or scaphoid fracture when falling onto an outstretched hand.

In contrast, the contribution of scaphoid and lunate flexion to capitate flexion during extreme passive flexion was similar to previous reports of active flexion. These findings are also not surprising given that there are minimal ligamentous attachments on the dorsal side of the wrist that directly connect the radius to the scaphoid or lunate. These bones are indirectly attached to the radius through the DRC and DIC ligaments, which both insert on the triquetrum. This ligament complex may allow more scapholunate mobility throughout a larger range of wrist flexion.

Computing ligament fiber paths in these extreme positions, carpal distraction, and throughout the dart thrower's path provided further insight into the behavior of the extrinsic carpal ligaments. Our approach assumed that a ligament reached its maximum value in one of twelve active and passive wrist positions and evaluated elongation with respect to this maximum value and an estimation of the maximum strain. This technique allowed us to question whether the carpal ligaments provide restraint in more than one direction of motion. We concluded that, in addition to restricting end-range wrist motion, the extrinsic radiocarpal ligaments might provide mechanical guidance throughout much of the range of wrist motion.

The latter half of this dissertation focused on the development of a predictive carpal model through tuning and validation. Our approach decoupled the tuning step from the validation step by using different wrist positions for each of these steps. We validated the model by demonstrating that the set of stiffness values that optimized simulated scaphoid kinematics in distraction results in similar kinematic error during wrist extension. We also found that the ligament stiffness values that optimized the model varied across subjects. While identifying the origin of this variability is beyond the scope of this study, we postulate that it results from variability in joint laxity and joint morphology. This inter-subject variability highlights the importance of modeling wrists and positions before drawing broad conclusions on carpal joint function.

Finally, we demonstrated the utility of the predictive model by presenting simulated scaphoid kinematics as the wrist moved through the dart thrower’s motion. Since our model provides the specific contributions of the contact forces and ligament forces to scaphoid motion, we were able to determine that the lunate supplies the majority of the ulnar-flexion moment throughout the dart thrower’s path regardless of whether the scaphoid flexes or extends. Additionally, the dorsal and volar aspects of the scapholunate interosseous ligament consistently supplied a flexion moment to the scaphoid. These results indicate that scaphoid extension with wrist ulnar deviation might not be caused by an extension moment imposed by the lunate. Instead, as the wrist ulnar deviates, scaphoid extension is explained by an increased extension moment from the scaphocapitate, scaphotrapezium, and dorsal intercarpal ligament, and an increased extension moment imposed on the scaphoid by the articular surface of the capitate. Scaphoid flexion is caused by a flexion moment imposed by the articular surfaces of the lunate, trapezoid, and trapezium. Additionally, we found that the flexion-extension moment due to the ligament is balanced by the scapholunate interosseous ligament.

Certainly, there are limitations to this work. Firstly, the ligament fiber algorithm does not model ligament-ligament interactions. The carpal ligaments are a complex web of tissue where fibers insert on bones and interdigitate with other ligaments. For example, the DIC interdigitates with the lunotriquetral interosseous ligament. The interaction of the

DIC with the lunotriquetral interosseous ligament may be important to carpal stability. Secondly, fibers within a ligament are also connected to one another. We treated ligament fibers independently in our model. It is unlikely that interactions within a ligament will affect elongation patterns, but they will affect the structural properties of the ligament in the predictive model. Finally, ligament wrapping over bones and other structures was not modeled. The fiber-wrapping algorithm was computationally expensive and increased simulation times significantly. Additionally, the fiber-wrapping algorithm was sensitive to the initial guess. When the motion of the scaphoid was simulated, an inappropriate initial guess created a fiber whose course did not mimic the physically relevant one and therefore created instabilities in the model. However, when fiber wrapping was successful, its contribution to simulated scaphoid kinematics was negligible.

This dissertation as a whole demonstrates the power of using a predictive carpal model in tandem with an *in vivo* database of carpal bone surfaces and kinematics to understand the function of the carpus. The model is meant to augment, not replace, kinematic analyses of the carpus. Kinematic analyses describe motion and predictive modeling explains the cause of motion. Both are essential to understand this complex “joint”. We stress that predictive models should not be used in isolation and are not a replacement for accurate experimental kinematic data. In fact, large experimentally acquired datasets are necessary to construct, tune, and validate these models. The assumptions of the model must be clearly stated and care must be taken that the assumptions or limitations of the model are not intertwined with the hypotheses. The goal of this work was to demonstrate how these tools can be used together to gain insights into carpal function.

To summarize, this work has four major findings. First, the kinematic patterns of the scaphoid and lunate are different in extreme extension compared to active extension. Second, the extrinsic carpal ligaments may not be simple restraints to end-range motion. They likely function as restraints at the end-range of one motion and serve as guides throughout a larger range of motion in a different direction. Third, inconsistency in the optimal ligament stiffness parameters across twelve subjects highlights the importance of using a large database to develop a predictive model. Finally, the contributions of ligaments and cartilage

to scaphoid rotation during the dart thrower's motion are highly complex and well balanced. This finding demonstrates the delicate balance and redundancy in the skeletoligamentous structures of wrist. While injury to a single ligament or loss of a contact surface may not greatly affect the kinematic patterns, it may overload the remaining ligaments or create points of high pressure on the remaining articular surfaces. The logical extension of this work is to apply these techniques to a larger set of experimentally measured positions and to add ligament wrapping and ligament-ligament interactions to the model. The techniques presented here have the potential to provide a full understanding of the carpal mechanism, and such understanding will better inform physicians on the treatment of the pathological wrist. Other potential applications of these techniques include use in simulated surgery, implant design, and the diagnosis and treatment planning of carpal injuries.

Appendix A

Gender Differences in Capitate Kinematics are Eliminated After Accounting for Variation in Carpal Size

The following appendix has been published as shown in the journal of Biomechanical Engineering.

Gender Differences in Capitate Kinematics are Eliminated After Accounting for Variation in Carpal Size

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Previous studies have found gender differences in carpal kinematics, and there are discrepancies in the literature on the location of the flexion/extension and radio-ulnar deviation rotation axes of the wrist. It has been postulated that these differences are due to carpal bone size differences rather than gender and that they may be resolved by normalizing the kinematics by carpal size. The purpose of this study was to determine if differences in radio-capitate kinematics are a function of size or gender. We also sought to determine if a best-fit pivot point (PvP) describes the radio-capitate joint as a ball-and-socket articulation. By using an in vivo markerless bone registration technique applied to computed tomography scans of 26 male and 28 female wrists, we applied scaling derived from capitate length to radio-capitate kinematics, characterized by a best-fit PvP. We determined if radio-capitate kinematics behave as a ball-and-socket articulation by examining the error in the best-fit PvP. Scaling PvP location completely removed gender differences ($P=0.3$). This verifies that differences in radio-capitate kinematics are due to size and not gender. The radio-capitate joint did not behave as a perfect ball and socket because helical axes representing anatomical motions such as flexion-extension, radio-ulnar deviation, dart throwers, and antidart throwers, were located at distances up to 4.5 mm from the PvP. Although the best-fit PvP did not yield a single center of rotation, it was still consistently found within the proximal pole of the capitate, and rms errors of the best-fit PvP calculation were on the order of 2 mm. Therefore, the ball-and-socket model of the wrist joint center using the best-fit PvP is appropriate when considering gross motion of the hand with respect to the forearm such as in optical motion capture models. However, the ball-and-socket model of the wrist is an insufficient description of the complex motion of the capitate with respect to the radius. These findings may aid in the design of wrist external fixation and orthotics. [DOI: 10.1115/1.2913332]

Keywords: carpal kinematics, pivot point, in vivo, capitate

Introduction

Due to the capitate's rigid attachment to the third metacarpal and central position in the carpus, it has been designated the keystone of the wrist and an indicator of overall wrist position [1]. Anatomically, the proximal pole of the capitate is nearly hemispherical, and it articulates in a hemispherically shaped fossa created by the scaphoid and lunate. The proximal convex surfaces of the scaphoid and lunate, in turn, articulate with the scaphoid and lunate fossa of the distal radius. Based on this anatomical arrangement, it seems reasonable to assume that the motion of the capitate with respect to the radius (radio-capitate kinematics) approximately behaves as a ball and socket. If it were an ideal ball-and-socket joint, there would exist a single fixed pivot point (PvP) through which all helical axes of motion (HAMs) intersect.

Previous studies of radio-capitate kinematics have shown that all HAMs nearly intersect one another within a tolerance of a few millimeters [2,3]. Although most studies have shown that the 2D axes of rotation or 3D PvP resides in the proximal pole of the capitate (e.g., Refs. [4–9]), there are discrepancies in the literature regarding the distance of flexion/extension axes from radio-ulnar deviation axes ranging between 2 mm and 7.5 mm [2,10]. Previous

studies have also found differences in the location of HAMs between men and women [11]. From these studies, it can be shown that the location of the radio-capitate HAM is closer to the capitate centroid in woman than in men.

It has been proposed, but not proven, that gender differences in HAM location are due to carpal size, not function. This is not unreasonable given that simple isometric scaling eliminates gender differences in carpal size and bone volume differences [11]. If carpal bone size is the largest factor contributing to male-female discrepancies in HAM location, then normalizing kinematics by bone size should eliminate these differences [11].

The differences in the HAM location with the direction of wrist motion is more difficult to reconcile. The variability in the distance of flexion/extension and radio-ulnar deviation HAMs across previous studies may be due to the fact that the previous calculations have been hampered by relatively small sample sizes and imprecise analytical methods. Also, to the best of our knowledge, no previous study has accounted for intersubject variability in HAM or PvP location by normalizing kinematics by carpal bone size.

The purpose of this study was to examine the separation of HAMs as a function of wrist position by using a large database of in vivo wrist kinematics, derived from markerless bone registration (MBR) of sequential computed tomography (CT) scans. In this dataset, we include the traditional anatomical motion paths of flexion/extension and radial/ulnar deviation, as well as motion paths that are considered to be important in activities of daily

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Contributed by the Bioengineering Division of ASME for publication in the JOURNAL OF BIOMECHANICAL ENGINEERING. Manuscript received April 20, 2007; final manuscript received November 12, 2007; published online May 23, 2008. Review conducted by Noshir Langrana. Paper presented at the ORS 2007.

living, such as the dart thrower's path [12]. We also examine whether normalizing by capitate size eliminates differences in radio-capitate PvP location (with respect to the capitate centroid) across subjects. Furthermore, we test the effect of kinematic scaling on gender differences.

We hypothesized that gender-related differences in best-fit PvP location would be eliminated when scaled by capitate size. To examine HAM separation as a function of direction of motion, we used the null hypothesis that a single, fixed PvP exists through which all anatomically and functionally defined radio-capitate HAMs pass.

Methods

A digital anatomic database containing CT volume images of both healthy wrists of 13 male (25.6 years old, range 22–34) and 14 female (23.6 years old, range 21–18) subjects in a total of 233 wrist positions, distributed throughout a complete range of wrist motion, was established after IRB approval and informed consent. For each subject, a high resolution neutral scan of both left and right wrists was segmented by using ANALYZE™ (Mayo Clinic Foundation, Rochester, MN) image processing software. The segmentation procedure yielded separate point clouds representing the outer cortical surface of each carpal bone. Closed surfaces for each carpal bone and distal radius were created by tessellating the point clouds using GEOMAGIC software (Raindrop Geomagic, Research Triangle Park, NC). To facilitate direct comparison of the kinematic data from the left and right wrists, all left wrists were mathematically converted to right wrists by inverting each individual CT slice at the image processing stage [13].

Kinematic transformations from the neutral wrist position to each subsequent position were calculated using a MBR algorithm [14] based on tissue classified distance fields [15]. For each wrist, the MBR algorithm registered the neutral position point clouds to all other CT volumes, creating six-degree-of-freedom global transformations from the neutral scan to each subsequent position. Radio-capitate transformations were generated by registering both the radius and the capitate in each position to the radius of the neutral scan, and then by computing the relative motion of the capitate with respect to the radius. Capitate volumes, centroids, and principal inertial axes were computed using previously established methods [16,17]. A capitate-based coordinate system was established using the capitate's inertial properties, with the origin at the capitate's centroid. The orientations of the inertial axes (I_1 , I_2 , and I_3) that defined the coordinate system were proximal-distal (I_1), dorsal-volar (I_2), and ulnar-radial (I_3) (Fig. 1). The capitate's shape and orientation are inherent such that its inertial axes are found to be in general alignment with the standard anatomical directions.

The best-fit radio-capitate PvP was generated using a least square spherical joint technique. A detailed derivation of the algorithm can be found in the Appendix of Piazza et al. [18]. Briefly, the best-fit PvP was determined for each wrist by using the radio-capitate kinematics to compute the location of two points, one fixed in the capitate coordinate system, the other fixed in the radial coordinate system, whose distance was minimized across the motions of the capitate [18]. If the articulation behaved like a perfect ball and socket and there was no measurement error, these two points would be coincident. Because we employ a radius-fixed coordinate system in our analysis, we report the best-fit PvP of the radius-fixed coordinate system. Each subject's kinematics (characterized by the PvP) was then reported with respect to the capitate-based coordinate system. The mean separation of the two best-fit PvPs throughout the range of motion was also calculated as a measure of the validity of the functional joint algorithm applied to radio-capitate kinematics. A second set of PvPs was computed for each wrist by isometrically scaling the original PvPs to the average subject by multiplying the I_1 , I_2 , and I_3 components of the PvP by the scaling factor described below.

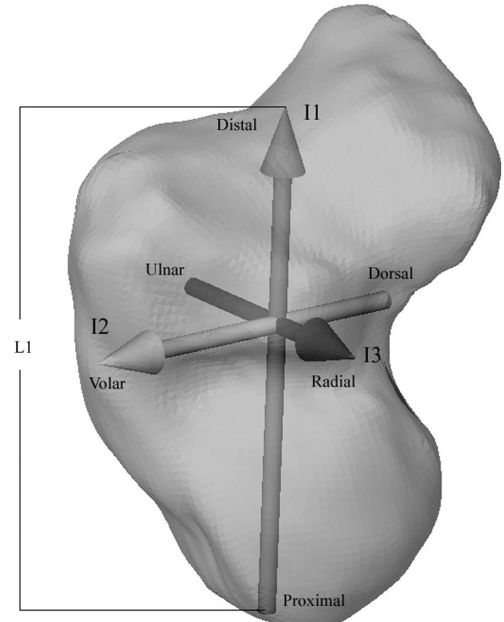


Fig. 1 The capitate coordinate system based on inertia: I_1 , I_2 , and I_3 =first, second, and third principal inertial axes. L_1 is the length of the capitate from the proximal surface to the distal surface along I_1 . L_2 and L_3 are similarly defined but corresponding to I_2 and I_3 , respectively.

To facilitate comparison of the HAM locations and PvPs across subjects, radio-capitate kinematics were scaled to the values of an average wrist. Scaling factors for capitate size were based on length measurements. For each capitate, the distance was calculated along I_1 , from the proximal to distal surfaces and defined as L_1 . Likewise, L_2 , and L_3 were found by computing the volar to dorsal and radial to ulnar lengths along I_2 and I_3 .

L_1 was chosen as the isometric scaling parameter and the measure of overall capitate size because it was found to correspond best to capitate volume (Fig. 2). To determine this, each inertial length was compared to the cube root of capitate volume. Linear regression analysis of this comparison yielded r -squared values for L_1 , L_2 , and L_3 of 0.86, 0.59, and 0.19, respectively. The average capitate volume and L_1 length were calculated across all

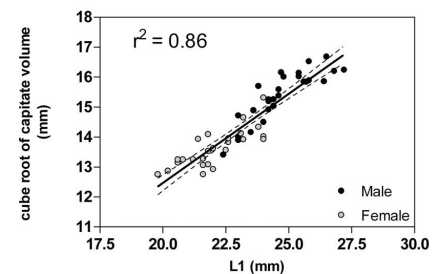


Fig. 2 There is a positive linear relationship between capitate long axis length and volume. Note that female capitates are smaller than males in both L_1 and volume.

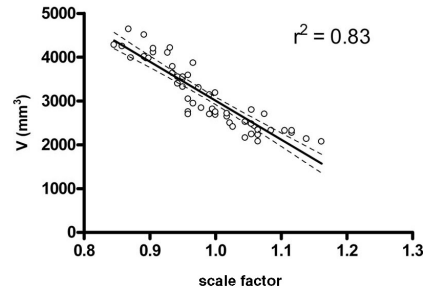


Fig. 3 The smallest capitate's PvP location was scaled up 16.2%, while the largest was scaled down 15.4%. Small changes in scaling factor correspond to large changes in volume.

wrists ($n=54$). A subject whose $L1$ value was closest to the group average ($n=54$), in this case less than 2% difference, was chosen as the average subject. The scaling factor for each subject was then determined by computing the ratio of the average subject's $L1$ values ($L1_{avg}$) to that of each subject.

The single, fixed PvP hypothesis was tested by determining if each HAM intersected the PvP within a tolerance of 0.5 mm. The distance (d) was calculated, for each subject, from the PvP to the closest point on each HAM. HAMs from all subjects were grouped into motion bins of flexion (F), extension (E), ulnar deviation (U), radial deviation (R), positive dart throwers (ulnar flexion, $+D$), negative dart throwers (radial extension $-D$), positive antidart throwers (radial flexion, $+AD$), and negative antidart throwers (ulnar extension, $-AD$) [13]. The HAMs were placed into a specific motion bin if they possessed less than 10 deg of any other motion. To find the representative HAM associated with each motion, HAM locations were scaled, and a quaternion average was performed on all of the HAMs of each bin [19]. The representative HAMs were useful in visualizing the relationship between the capitate and the PvP.

To examine the dependence of PvP location (PvP_0) on capitate size ($L1$), linear regression was performed on the $I1$, $I2$, and $I3$ components of PvP_0 ($PvP_{I1, I2}$, and $I3$) versus $L1$. This was done for the scaled and unscaled PvPs. Our hypothesis that gender differences are due to carpal size was tested by comparing male and female PvP_0 values for both the unscaled and scaled cases. For each case, a Student's t -test was performed to determine the significance of the mean differences between male and female PvP_0 . To test our hypothesis that radio-capitate kinematics behave as a ball and socket, HAMs were postulated to pass through the PvP if d was less than or equal to 0.5 mm. A one-population t -test with a hypothetical mean of 0.5 mm was performed on each motion bin.

Results

The mean capitate volumes for males were significantly greater than females at $3625 \pm 620 \text{ mm}^3$ and $2540 \pm 350 \text{ mm}^3$, respectively. These values were calculated in a previous study; however, the means for men differ from that study due to a slightly smaller sample size in the current study ($n=13$, instead of $n=14$) [11]. The mean $L1$ values for males and females were $24.7 \pm 1.3 \text{ mm}$ and $22.1 \pm 1.2 \text{ mm}$, respectively. The smallest capitate's PvP location, which belonged to a female, was scaled up 16.2%, while the largest, belonging to a male, was scaled down 15.4% (Fig. 3).

Regression analysis of the unscaled PvP_{I1} versus $L1$ revealed that the PvP moved further from the capitate's centroid as $L1$ increased ($P < 0.0001$) (Fig. 4(a)). Also, PvP locations of males and females had the same continuum. When isometric scaling was applied to PvP_0 , the dependence of PvP_{I1} on the capitate

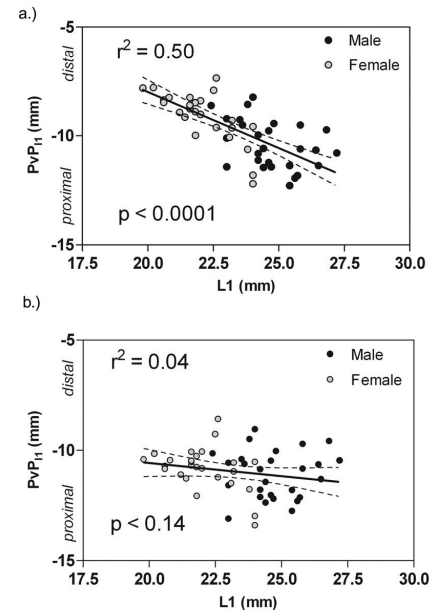


Fig. 4 Location of best-fit PvP from capitate centroid along $I1$. PvP_{I1} becomes more negative as it approaches the proximal pole of the capitate. (a) Although the R^2 value is low, a trend can be seen that reveals a relationship between capitate size and PvP location; as the capitates get larger, the distance from capitate centroid to the PvP increases. (b) Scaled PvP locations. The relationship between capitate size and PvP location vanishes.

length was completely removed ($P=0.14$) (Fig. 4(b)). No dependence was found of PvP_{I2} or PvP_{I3} on the capitate length in either the unscaled or the scaled case, with P values for scaled $PvP_{I2} = 0.90$ and $PvP_{I3} = 0.40$, and unscaled $PvP_{I2} = 0.48$ and $PvP_{I3} = 0.48$.

The unscaled PvP_0 was located ($P=0.0001$) significantly closer to the capitate centroid in females than in males. Scaling removed these gender differences, increasing P to a nonsignificant value of 0.3, supporting our hypothesis that simple isometric scaling would remove gender differences in PvP_0 . Linear regression analysis revealed that the dependence of PvP location on capitate size was removed with the scaling of the capitate (Fig. 5).

The single, fixed PvP model was not supported because significant differences ($P < 0.0001$) were found in the distance d for each motion bin (Table 1). Although the PvP did not serve as the intersection point of all HAMs, it was consistently found within the proximal pole of the capitate (Fig. 6). The average distance of a scaled PvP from the average of all PvPs was $1.9 \pm 1.1 \text{ mm}$. The average separation of capitate-fixed and radius-fixed PvPs was $1.7 \pm 0.6 \text{ mm}$. Qualitatively, the individual HAM axes for each motion bin were in the same general anatomical location with respect to the PvP. For example, flexion axes were located volar and distal to the PvP, extension axes tended to be located more proximal, ulnar deviation was distal, and radial deviation was distal and radial (Table 1 and Fig. 6).

Discussion

In vivo sequential CT scans were analyzed to examine gender differences in capitate kinematics and to explore the possibility that radio-capitate kinematics can be modeled as a ball-and-socket

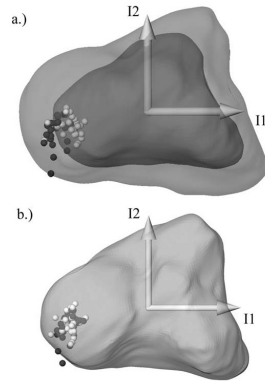


Fig. 5 Capitate PvP for each subject visualized in a lateral view of the capitate inertial coordinates, male (black), female (white). (a) Unscaled PvPs visualized within the smallest female (inner) and largest male (outer) capitates (registered to the inertial coordinate system). (b) PvPs scaled to $L1_{avg}$, visualized within the average subject's capitate. Both male and female PvPs tend toward the center of the proximal pole.

articulation. The unscaled PvPs were shown to have a near linear relationship with the proximal-distal length of the capitate. As capitate size increased across subjects, the PvP moved further from the capitate centroid. Once the PvPs were scaled to the average subject, the relationship between PvP location and capitate size was removed.

Gender differences in PvP location were also removed by using capitate length as a scaling factor, confirming that gender differences in capitate kinematics are simply a function of carpal size and not function. Our results are consistent with the findings of Neu et al. who also found that, in general, a female's radio-capitate HAM is located closer to the capitate centroid than a male's [2] (Fig. 5(a)).

Although the best-fit radio-capitate PvP was consistently found within the proximal pole of the capitate (Fig. 3(b)), with an average RMS error of 1.7 ± 0.6 mm, the distance of the closest point on the HAM from the PvP (d) varied with the direction of wrist motion (up to 4.5 mm). This verifies that the best-fit radio-capitate PvP should not be interpreted as an ideal ball-and-socket articulation. However, with errors on the order of 2 mm, the best-fit PvP may be used as a reasonable wrist joint center when studying segmental motion of the hand with respect to the forearm. The

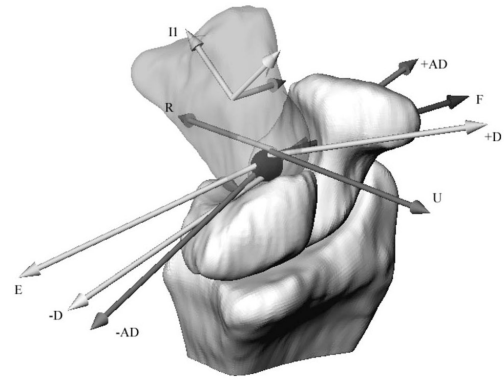


Fig. 6 The average of all scaled PvPs (black sphere). Representative HAMS for each motion bin. Refer to Table 1 for abbreviations.

variation in PvP_0 is on the same order of magnitude as the errors seen in biomechanical models of other joints created by using skin-mounted markers [20].

Numerous studies using a wide variety of two-dimensional (2D) and three-dimensional (3D) techniques have sought to quantify the rotational motion of the capitate. 2D studies have commonly used in vivo planar radiographic techniques, whereas previous 3D studies have typically used motion-tracking systems on cadaver wrists. Most studies have shown that the 2D axis of rotation or the 3D PvP resides in the proximal pole of the capitate [4,10,21,22]. However, discrepancies in literature exist in the distance of flexion/extension and radio-ulnar deviation axes from the PvP that range from 2 mm from each other to 7.5 mm [2,10]. Previous studies have methodological limitations, including the inability of a 2D study to measure out of plane motion [4], and the fact that previous 3D studies have been limited by the need for invasive procedures to attach markers to the carpal bones [10].

The current study addresses many of these limitations because of the increased accuracy of the MBR technique and its application to a relatively large number of healthy, normal subjects. While previous studies have revealed a great deal about the complexities of carpal motion, there have been relatively few studies that have examined radio-capitate kinematics using nonorthogonal, or combined, wrist motions, such as the dart thrower's motion [13,23,24]. Furthermore, to the best of our knowledge, no previous study has examined the effects of normalizing radio-capitate

Table 1 Pos= motion bin. d =the mean distance between HAM location and PvP for each motion bin. std=the standard deviation of d . n =number of positions in each motion bin. Average ROM=average total capitate rotation for each HAM. Direction with respect to PvP=relative position of the average helical axis to the average PvP. F=flexion, E=extension, U=ulnar deviation, R=radial deviation, +D=positive dart thrower path, -D=negative dart thrower path, +AD=positive anti dart thrower path, -AD=negative antidart thrower path.

Pos	d (mm) ^a	std (mm)	Distance of HAM axis to PvP		Direction with respect to PvP
			n	Average ROM (std)	
F	3.6	3.0	25	25 deg (19 deg)	Volar distal
E	3.4	3.0	63	36 deg (18 deg)	Proximal
U	4.5	2.0	23	26 deg (15 deg)	Distal
R	3.0	1.6	10	17 deg (12 deg)	Distal radial
+D	4.2	2.2	32	31 deg (14 deg)	Distal volar
-D	3.4	1.7	33	36 deg (16 deg)	Distal volar
+AD	4.1	2.3	31	36 deg (29 deg)	Volar
-AD	1.8	1.1	159	43 deg (19 deg)	Dorsal

^aAll distances of HAM location to the PvP were statistically different from a hypothetical mean of 0.5 mm.

kinematics by capitate size. The current study demonstrates that normalizing HAM or PVP location to the long axis length of the capitate facilitates the combination of male and female datasets of radio-capitate kinematics.

The most prominent limitation of this study is the small number of static positions used as input for the functional joint algorithm used to calculate the best-fit PVP. This algorithm is most often used with passive, high-speed optical motion capture systems. With capture rates of 60 Hz and up, hundreds of positions can be used as input into the least-squares, best-fit PVP algorithm. However, accurate and repeatable marker placement algorithms for high-speed optical motion analysis in the hand and wrist of living subjects have not yet been developed, and the MBR technique is considerably more accurate than skeletal models derived from skin surface markers [25].

Another potential limitation of this study is the relatively large number of negative antdard thrower's positions compared to other motions such as radial deviation. This may cause the PVP to be biased toward this position. However, the PVP was calculated on a subject by subject basis based on motions derived from targeted wrist positions in our CT protocol. Therefore, the PVP calculation represented the full range of wrist motion for each subject. One possible reason for the greater number of binned antdard thrower's positions is that both extension and ulnar deviation have the largest ranges of motion. Radial deviation has a limited range of motion, so it follows that there would be limited positions of pure radial deviation in similar ranges of motion. Because the PVP was calculated on a subject by subject basis, and each subject had CT images taken representing their entire range of wrist motion, the uneven number of positions in each bin does not negatively affect this analysis.

One potential application of the best-fit radio-capitate PVP is an estimation of the wrist joint center in an upper extremity model created by using optical motion capture and skin mounted surface markers. These models are becoming increasingly popular in the analysis of upper extremity function [26–28]. Another possible application of these data may be the creation of regression equations by which the best-fit PVP, and thus the proximal pole of the capitate, can be calculated by using its relationship to palpable bony landmarks. When considering carpal kinematics, the capitate does not rotate about a fixed center point for all motions (with respect to the radius). An improved understanding of the capitate center of motion, and its relationship to functional motion planes of the wrist, is important as a design consideration for articulated external fixation devices, wrist orthotic design, and prosthetic replacement of the wrist.

Acknowledgment

This study was supported by NIH AR44005 and NIH HD052127.

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Appendix B

In Vivo Triquetrum-Hamate Kinematics through a Simulated Hammering Wrist Motion

The following appendix was accepted for publication in the Journal of Bone and Joint Surgery. By Kamal RN, Rainbow MJ, Akelman E, and Crisco JJ.

ABSTRACT

Background: The shape and kinematics of the triquetrum-hamate joint have been the subject of continued research as its articulation provides for wrist stability and range of motion. The purpose of this study was to measure the in vivo articulation of the TqH joint as the wrist moved along an important functional wrist motion, the dart thrower's path.

Methods: The right wrist of 6 male and 6 female, (24.8 ± 3.8 years) volunteers were imaged with CT in 5 positions along a simulated hammering path. 3-D kinematics of third metacarpal, triquetrum, hamate, and radius were analyzed using the helical axis of motion and the path of contact areas.

Results: As the wrist ulnar-flexed with respect to the radius, the triquetrum translated 3.7 ± 1.7 mm distally on the hamate. Approximately midway through this distal course, when

the triquetrum appeared to engage the distal ridge of the hamate, the triquetrum began translating volarly. Total volar translation was $2.6 \pm 1.1\text{mm}$. As the wrist ulnar-flexed, there was also a decrease in the distance and variability in the location of the triquetrum-hamate rotation axis from the hamate-centroid from $11.7 \pm 4.1\text{ mm}$ to $3.3 \pm 1.4\text{ mm}$ ($P < 0.0001$).

Conclusion: Our findings support the concept that the triquetrum rotates on the convex ellipsoid surface of the hamate, and that the helicoidal description of the triquetrum's motion on the hamate may be an oversimplification.

Clinical Relevance: Our results suggest that the TqH joint is less constrained in radial extension than it is in ulnar flexion. The concave distal ridge of the hamate may guide the triquetrum towards the hook until it is fully engaged, which could block further wrist ulnar deviation. This may provide for carpal stability while also serving as a rationale for triquetrum excision to increase wrist range of motion.

B.1 Introduction

The ulnar sided triquetrum-hamate (TqH) joint has received less attention in the literature than its radial counterparts, however its unique articulation serves a vital role in carpal stability [1]. For normal carpal stability, a balance exists between the flexion moment of the scaphotrapezotrapezoid joint, and the extension moment of the triquetrum-hamate joint. These moments have been described as a spring with two distally directed arms where any alteration in the moment arm of the TqH joint disrupts the equilibrium, and results in non-dissociative volar intercalated segmental instability (CIND-VISI) [2-3]. Although the role of the TqH joint in carpal stability has been well-accepted, the exact kinematics and kinetics by which the TqH joint imparts an extension moment on the lunate are unknown. While the TqH joint has been shown to allow for static wrist stability in grip, its role in dynamic stability and wrist range of motion has received less attention. Clinically, however, Bain et. al have shown that triquetrum excision leads to 15-27 degrees of increased ulnar deviation of the wrist in the setting of radioscapulunate fusion and distal scaphoidectomy for

radiocarpal arthritis [4]. This finding agreed with prior cadaveric studies that showed that triquetrum excision lead to increased ulnar deviation of the wrist after radioscapulunate fusion with distal scaphoidectomy or 4-corner fusion with distal scaphoidectomy[5-6]. There is no biomechanical evaluation of the role of the triquetrum-hamate joint in wrist range of motion or an explanation for the increase in ulnar deviation achieved by triquetrum excision.

The purpose of this study is to measure the in vivo articulation of the TqH joint as the wrist moved along the dart thrower’s path (DTM). We chose the dart thrower’s path because it provides a stable base for common grip activities, and has been suggested as being the most important functional motion of the wrist [7-8]. Furthermore, the articulation of the TqH joint has yet to be studied as the wrist moves through the dart thrower’s motion. We study TqH motion using an accurate markerless bone registration technique applied to serial CT scans of the wrist as it moved along a simulated hammering path.

B.2 Methods

B.2.1 Data Collection

Following IRB approval and informed consent, 6 male and 6 female, right-handed volunteers, average age 24.8 ± 3.8 years were screened with radiographs and physical exam to confirm the lack of any injury or diseases that would affect carpal motion by a board certified orthopaedic hand surgeon (EA). Subjects held a dowel (simulated hammer handle) and placed their right wrist in a custom jig designed to hold the wrist steady at 5 evenly-spaced wrist positions along a simulated hammering path[9]. CT scans (GE LightSpeed 16, General Electric, Milwaukee, WI) of the wrist were obtained beginning in radial extension and moving toward ulnar flexion through each of 5 hammering positions: -40° (windup, radial-extension), -20° , 0° (neutral, hammer handle perpendicular to forearm), 20° , and 40° (impact). Imaging parameters were set at 80kVp and 80mA, with 0.5mm x 0.5mm in-plane resolution and slice thickness of 0.625mm. To provide better quality 3-D bone

surface models and aid the registration algorithm, a sixth, higher resolution neutral scan (0.3 X 0.3 X 0.625 mm) was also acquired.

B.2.2 Data Analysis

Relative motion between each sequential hammering position was calculated for the third metacarpal, hamate, and triquetrum with respect to the radius, and also for the triquetrum with respect to the hamate using an established markerless bone registration method[10] [11]. This resulted in four hammering steps (-40° to -20° (motion H1), -20° to 0° (motion H2), 0° to 20° (motion H3), and 20° to 40° (motion H4)). Each step was described using the helical axis of motion variables (HAM) that completely describe 3-D motion as a single rotation and translation about a unique axis, referred to as the rotation axis. The orientation of the rotation axes of each bone with respect to the radius was reported as a function of hammering position in a radius-based coordinate system[12]. The components of the TqH rotations (triquetrum with respect to the hamate) in each anatomical direction were reported as the percent of total rotation about the axes of a coordinate system that was aligned with hamate anatomy. The X-axis was directed from the proximal pole of the hamate toward the distal end. The Y-axis was directed from dorsal to volar such that it was parallel with the hook of the hamate, and the Z-axis was directed from the ulnar to the radial aspect of the hamate. The variability in rotation axis orientation across subjects was reported as the mean angle between each subject's rotation axis and the resultant rotation axis vector computed using all subjects. The location of the TqH rotation axis throughout the path was reported as the distance between the closest point on the rotation axis to the centroid of the hamate. Differences between rotation as well as rotation axis orientation between the triquetrum and hamate and hammering motion were evaluated using a two-way repeated measures ANOVA and a Holm-Sidak comparison procedure.

The translation of the triquetrum on the hamate was also described by the path of the centroid of the contact area of the TqH joint. Because the cartilage was not imaged directly the contact area was estimated using the distances between the subchondral bone of the tri-

quetrum and hamate. Iso-contour distances were computed according to previous methods [13] and the region on the subchondral bone surface of the hamate that was within 1.8 mm of the triquetrum was calculated. A value of 1.8 mm was chosen based on the estimated thickness of the articular cartilage on each facet as approximately 0.9 mm. The centroid of the 1.8 mm contour region was then computed for each of the five hammering positions. Centroid location in both the dorsal-volar and proximal-distal directions was plotted as a function of hammering path. To determine if there was a relationship between centroid location and wrist position, linear regression was performed at the subject level for each direction and statistical significance was tested to determine if the slope was non-zero using an F-test. Correlation coefficients (R^2) were also computed to examine the fidelity of the linear model.

B.3 Results

B.3.1 Kinematics of the wrist (MC3), triquetrum, and hamate with respect to the radius

Throughout the hammering pathway, the wrist ulnar-flexed through a range of motion of $71.1^\circ \pm 8.1$ (Figure B.1, Table C.1). The coupling ratio of flexion-extension to ulnar-radial deviation was linear ($P < 0.01$) with a mean of $1.14^\circ \pm 0.43$. This is equivalent to 1.14° of flexion-extension for every 1° of ulnar-radial deviation. The motion was biased towards wrist extension with the amount of extension changing from $59^\circ \pm 9.6$ during windup to $8^\circ \pm 10$ during impact. The orientation of the rotation axis of the wrist was approximately midway between the flexion-axis and the ulnar deviation-axis as the wrist moved through the first four hammering positions (-40° to 20°). As the wrist moved from the forth (20°) to the last (40° , impact) hammering position, it primarily flexed, shown by a significant shift ($P < 0.001$) in the angle the rotation axis made with the flexion-axis from $43.8^\circ \pm 15.6$ during windup to $27.2^\circ \pm 10.3$ during impact (Figure B.1, H4). The hamate rotated about an axis that was oriented nearly parallel to the MC3 axis ($P = 0.7$) but it rotated by

approximately 2.0° less during each progressive hammering step ($P < 0.001$). There were no statistically significant differences in the orientation of the triquetrum axis with either the MC3 or hamate axis, however the orientation of the triquetrum-axis was less consistent than the MC3 and hamate axes throughout the hammering motion. This was shown by larger variability in the triquetrum rotation axis orientation of 19.8° compared to 9.0° and 10.5° for the MC3 and hamate, respectively. The triquetrum rotated approximately 10° less ($P < 0.001$) than both the MC3 and hamate at each hammering step which resulted in a range of motion that was $44\% \pm 11$ the range of motion of the hamate. Although there were no statistical differences in orientation of the triquetrum rotation axis, likely due to the increased variability, the triquetrum ulnar deviated more than it flexed and in some cases extended while ulnar deviating during the middle hammer steps (20° to 0° and 0° to 20°) (Figure B.1, H3).

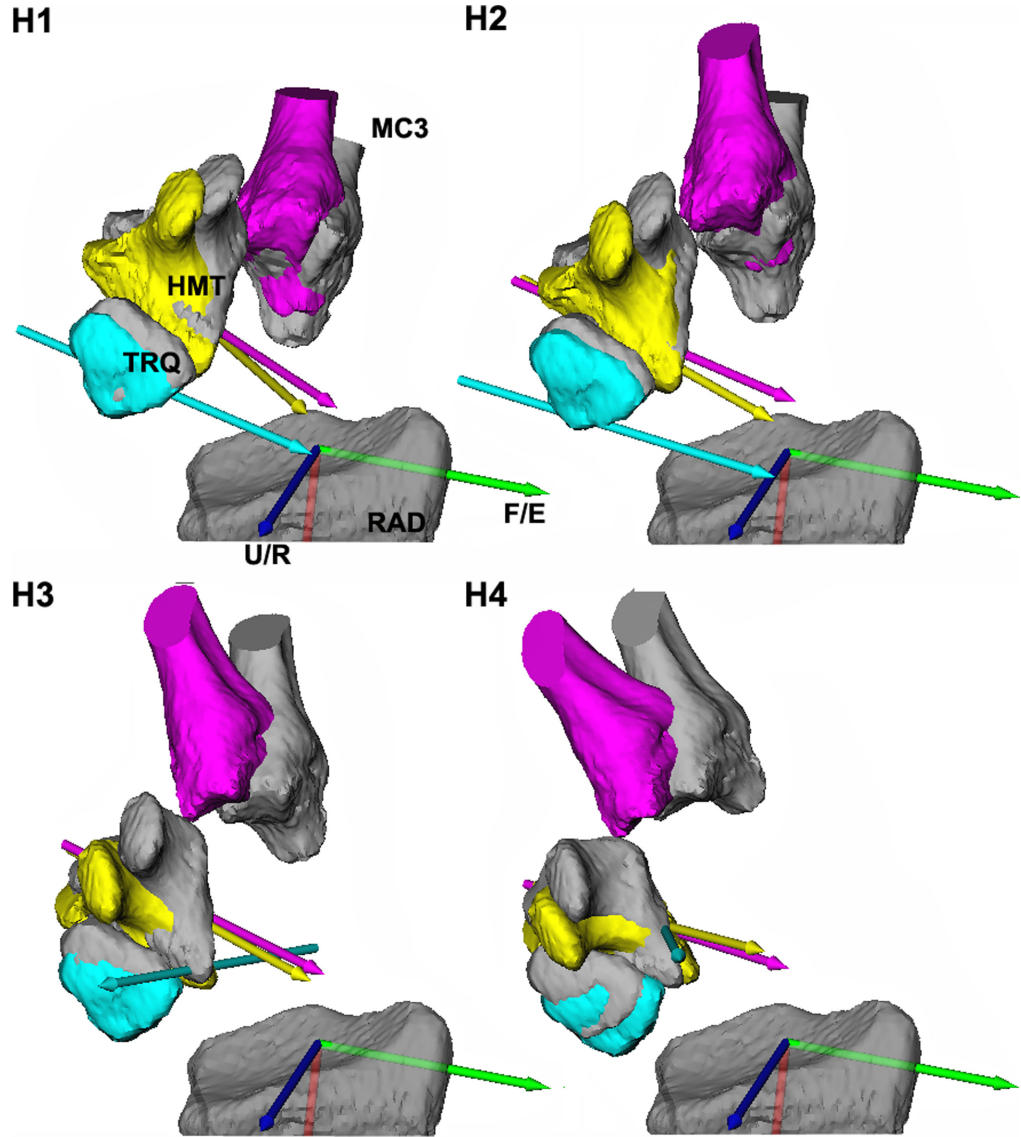


Figure B.1: Volar view of a right wrist. Motion of the MC3, hamate, and triquetrum with respect to the radius. The radius-based coordinate system is imbedded in the radius with the flexion-extension axis (F/E) shown in green and the ulnar-radial deviation axis (U/R) shown in blue. The grey bone represents the first position and the colored bone along represents the second position as the wrist moved from -40° to -20° (motion H1), -20° to 0° (motion H2), 0° to 20° (motion H3), and 20° to 40° (motion H4). The color of the rotation axis of each bone coincides with bone color. The MC3 and hamate rotate similarly while the triquetrum has a range of motion less than half that of the mc3 and hamate.

Table B.1: Rotation and orientation of the third metacarpal, hamate, and triquetrum with respect to the radius throughout the hammering motion. Rotation (degrees) is described about the helical axis of motion. Orientation is the angle formed between the rotation axis and the wrist flexion axis, where $0^\circ = \text{pure flexion}$, $90^\circ = \text{ulnar deviation}$, $45^\circ = \text{equal ulnar deviation and flexion}$, and $180^\circ = \text{pure extension}$.

Motion	Wrist (MC3)			Hamate		Triquetrum	
	Rotation (SD)	Orientation (SD)	Rotation (SD)	Rotation (SD)	Orientation (SD)	Rotation (SD)	Orientation (SD)
Total Motion H1-H4	71.1 (8.1) $^\circ$	41.0 (9.0) $^\circ$	62.0 (6.7) $^\circ$		47.1 (10.5) $^\circ$	27.5 (5.9) $^\circ$	76.1 (19.8) $^\circ$
Motion H1 (-40 $^\circ$ to 20 $^\circ$)	17.6 (4.0) $^\circ$	46.3 (18.0) $^\circ$	14.2 (4.7) $^\circ$		55.4 (23.1) $^\circ$	7.8 (3.0) $^\circ$	51.9 (25.6) $^\circ$
Motion H2 (-20 $^\circ$ to 0 $^\circ$)	19.4 (2.3) $^\circ$	49.5 (17.7) $^\circ$	17.7 (2.2) $^\circ$		57.2 (19.4) $^\circ$	8.4 (2.9) $^\circ$	88.1 (35.8) $^\circ$
Motion H3 (0 $^\circ$ to 20 $^\circ$)	20.3 (4.9) $^\circ$	43.8 (15.6) $^\circ$	18.5 (4.4) $^\circ$		49.6 (17.4) $^\circ$	8.9 (2.5) $^\circ$	101.0 (29.1) $^\circ$
Motion H4 (20 $^\circ$ to 40 $^\circ$)	16.9 (4.5) $^\circ$	27.2 (10.3) $^\circ$	15.9 (4.5) $^\circ$		28.5 (10.7) $^\circ$	7.7 (3.0) $^\circ$	63.4 (25.0) $^\circ$

B.3.2 Rotation of the triquetrum with respect to the hamate

Qualitatively, we observed that as the wrist moved into ulnar flexion, the triquetrum translated along the hamate distally and volarly in a path oblique to the sagittal plane along the ulnar surface of the hamate (Figure B.2). As the triquetrum approached the distal aspect of the ulnar surface (at a prominent ridge), it shifted from an oblique course to a volar course towards the convex surface of the base of the hook. This shift in motion corresponded to a change in the location of the rotation axis from outside of the hamate in a variable orientation to within the hamate in a predominantly radial/ulnar orientation (rotation axis variability 15.2°) (Table B.2, Figure B.2). The distance and variability in the location of the TqH helical axis of motion from the hamate-centroid decreased from 11.7 ± 4.1 mm, to 3.3 ± 1.4 mm ($P < 0.0001$) from the first step (wind up to 20°) to the last step (-20° to Strike). The variability in orientation of the rotation axis also decreased from the first step to the last step. This was expressed as the angle between the resultant helical axis (across subjects) and each subject's rotation axis decreasing from 33.7° to 15.2° .

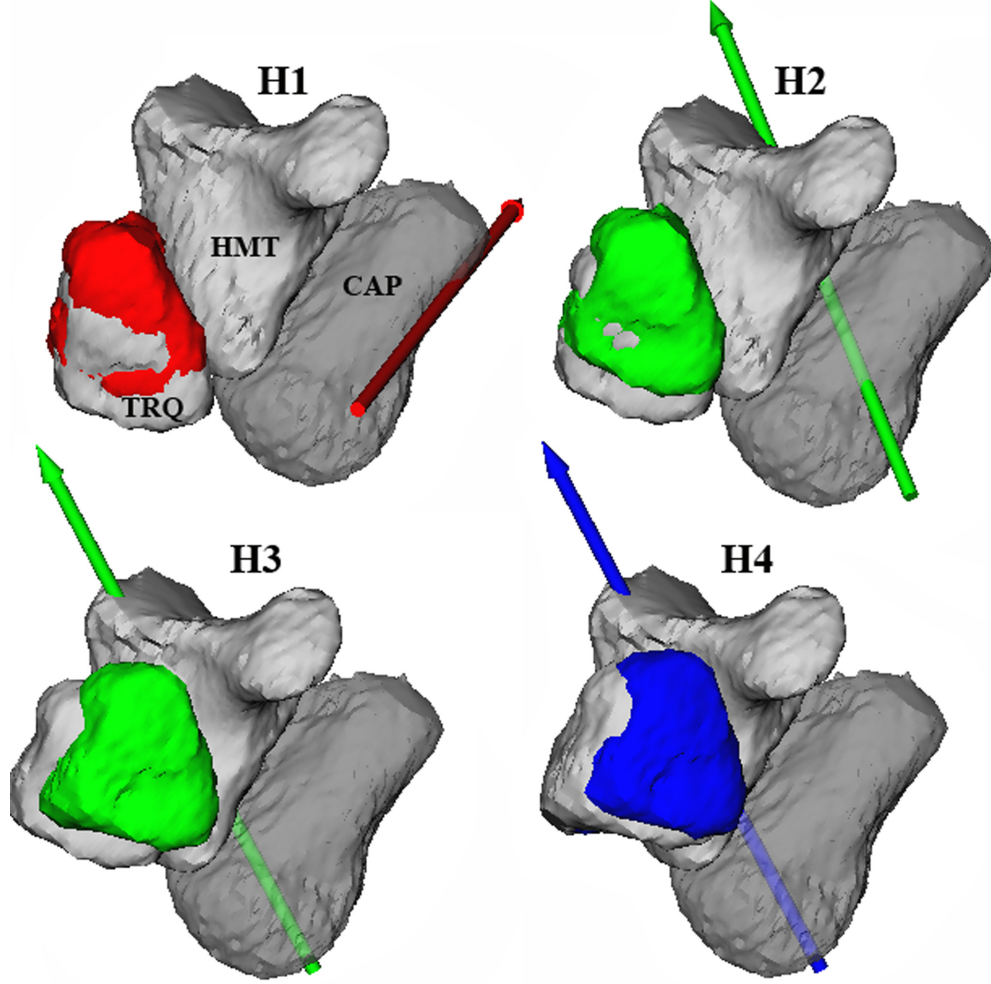


Figure B.2: Oblique (volar-ulnar) view of a right wrist. The rotation axis of the triquetrum with respect to the hamate moves from a variable location outside the hamate to a radial-ulnar orientation within the hamate as the wrist moves through the hammering path. The gray triquetrum represents the first position and the colored triquetrum represents the second position as the wrist moved from 40° to -20° (motion H1), -20° to 0° (motion H2), 0° to 20° (motion H3), and 20° to 40° (motion H4).

B.3.3 Translation of the triquetrum on the hamate

The path of the centroid of the contact area mirrored triquetrum motion and confirmed our visual findings (Figure B.3a,b). There was a positive linear ($R^2 = 0.93 \pm 0.06$) relationship

Table B.2: Total rotation of the triquetrum relative to the hamate and its components with respect to hamate anatomy. Rotation axis variability is computed as the average angle between each subject's individual helical axis and the average helical axis of all subjects.

Motion	Total HAM Rotation	Radial (+) / Ulnar (-)	Volar (+) / Dorsal (-)	Distal (+) / Proximal (-)	Helical Axis Variability
Motion H1 (-40° to 20°)	7.9° (3.4)	12% (25)	-35% (25)	40% (23)	33.7°
Motion H2 (-20° to 0°)	12.2° (3.6)	-18% (15)	-35% (11)	42% (12)	16.0°
Motion H3 (0° to 20°)	14.6° (5.3)	-30% (12)	-33% (10)	36% (9)	14.6°
Motion H4 (20° to 40°)	10.8° (5.6)	-45% (11)	-24% (14)	28% (9)	15.2°

($P < 0.05$) between the proximal/distal movement of the contour-centroid and hammering path (Figure B.3c). The total excursion of the contour-centroid along the proximal-distal path was 3.7 ± 1.7 mm. As the triquetrum moved distally it began moving in a volar direction towards the base of the hook of the hamate (Figure B.3d). In this plane, the triquetrum translated 2.6 ± 1.1 mm, and had a positive linear relationship with the dart thrower's path in 8 out of 12 subjects ($P < 0.05$). Interestingly, dorsal-volar movement of contour-centroid location along the hammering path was better modeled by a quadratic function, with R^2 values increasing from 0.74 ± 0.22 in a linear model to 0.96 ± 0.03 in a quadratic model.

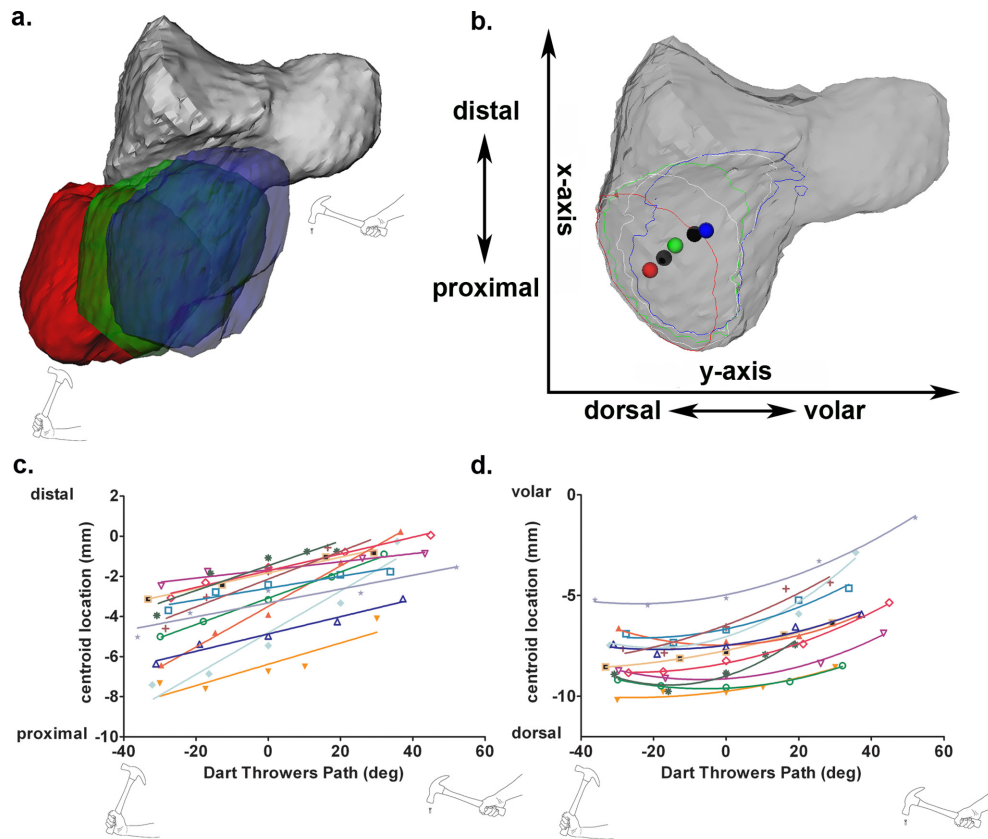


Figure B.3: a.) Translation of the triquetrum on the hamate. The position of the triquetrum (colored) on the hamate (grey) of a subject for the windup (red; wrist at -40°), neutral hammer (green; wrist at 0°), and impact (blue; wrist at 40°). b.) The 1.8 mm contour centroid (CONTOUR-CENTROID) and corresponding area for the same positions in a. The black spheres represent the intermediate hammering positions. c.) proximal-distal contour-centroid location vs. dart throwers path for each subject and a linear function fit to each subject. d.) dorsal-volar contour-centroid location vs. dart throwers path and a quadratic function fit to each subject.

B.4 Discussion

This study describes the in vivo kinematics of the TqH joint during combined motions of ulnar-radial deviation and flexion-extension, specifically motion along a dart thrower's path. The shape and articulation of the TqH joint has been the subject of some controversy. It was originally described as a saddle shaped joint because the surfaces of the hamate and triquetrum have both concave and convex parts [14]. Further studies extended the saddle joint concept to describe the concave and convex parts of the hamate as forming a spiral groove that caused a helicoidal or screw type motion of the triquetrum as it articulates with the hamate [15] [16]. Recently, Moritomo et al. measured the in vivo kinematics of the TqH joint throughout the orthogonal motions of ulnar-radial deviation and flexion-extension and found that the triquetrum articulates primarily with the convex surface of the hamate making the joint ellipsoidal rather than saddle shaped or helicoidal [17]. We found that as the triquetrum moved distally on the hamate, there was little dorsal / volar translation early in radial-extension and then increasing volar translation as the wrist moved into ulnar flexion. These findings suggest that the helicoidal description of the triquetrum's motion on the hamate is an over simplification [15]. True helicoidal or screw motion would imply a constant ratio of proximal-distal translation to dorsal-volar translation of the triquetrum as it spirals along the hamate groove. Rather, these findings suggest that the motion of the triquetrum is more complex as the amount of dorsal-volar translation increases with increasing distal translation. Our findings support the concept that the motion of the triquetrum can be explained by rotation on the convex ellipsoidal surface of the hamate as suggested by Moritomo et al. Our results suggest that the concave distal ridge of the hamate may guide the triquetrum as it articulates with the convex part of the hamate until its translation ends at the hook. The triquetrum first rotates about a minor axis (larger radius of curvature) of the ellipsoidal surface of the hamate, and finally rotates about the long axis (smaller radius of curvature) of the ellipse as evidenced by the helical axis moving from outside the body of the hamate to within the hamate in a radial-ulnar orientation.

The dart thrower's motion has been described as the most commonly used plane of wrist

range of motion for activities of daily living, and has been implicated in the use of a mallet, tools of percussion, such as for carpenters, stone masons, motor mechanics, fisherman, tennis players, and orthopaedic surgeons, and conducting an orchestra or throwing a dart. The DTM, described as a plane of motion from radial extension to ulnar flexion, is used for most activities of daily living and is oriented from 30° to 45° to the sagittal plane. In our experiment, subjects held a dowel rod with their hand and wrist in a custom-made jig along 5 evenly-spaced positions along a simulated hammering path from radial extension towards ulnar flexion. Interestingly, analysis of the hammering path in our experiment was oriented in an axis oblique to the sagittal plane of the wrist, from radial extension to ulnar extension. This path is similar to that described previously by Palmer et al. where the DTM was described as the wrist moving from an extended and radially deviated position to a less extended and ulnarly deviated position. Because the position of the wrist for power grip is ulnar deviation and extension, it is possible that because subjects were told to squeeze a dowel rod during the hammering pathway, they preferentially kept their wrist in extension while moving towards ulnar deviation to maintain power grip. We are unsure, however, if subjects in this experiment followed a path towards ulnar deviation and less extension (while remaining extended) because of the jig we used during experimentation instead.

Carpal kinematics are dependent upon the geometry of each carpal bone [18] and the constraints of the extrinsic and intrinsic carpal ligaments [1]. Carpal stability under load, is maintained by a balance between the flexion moment of the scaphoid and the STT joint and the extension moment of the TqH joint [1-2]. The scaphoid, by virtue of its obliquity, tends to rotate into flexion during radial deviation, creating a flexion moment on the lunate [1-2]. Conversely, the flexion moment of the scaphoid is balanced by an extension moment from the TqH joint during ulnar deviation. The current understanding of the function of the TqH joint is based on its roughly helicoidal shape creating an extension moment on the triquetrum, with an additional extension moment on the triquetrum from the flexor carpi ulnaris during ulnar deviation, however this continues to be studied [19]. The dynamic relationship between these two balancing forces is highlighted by the creation of a VISI

deformity of the carpus after a lunotriquetral ligament disruption, and a DISI deformity of the carpus after a scapholunate ligament disruption or STT degeneration [2]. Because our study is limited to a kinematic analysis, without regard to the forces that generate the motions, we cannot make definitive statements on the role of the TqH in joint stability. However, the decrease in variability in both location and orientation of the helical axis of the TqH joint as the wrist moves along the dart thrower's path implies that the joint is less constrained in radial extension than it is in ulnar flexion. The lack of constraint in radial-extension may imply that the joint has more mechanical slop in this position or it may be due to variations in the shape of bone surfaces between subjects. We are unsure if an extension moment is created in the TqH joint as the wrist goes into ulnar deviation through the DTM, or if stability is imparted simply by the articulation of the triquetrum with the distal ridge and base of the hook of the hamate. Future studies are needed to better understand the kinetic role of the TqH joint and its bony-articulation and ligamentous constraints.

Recently, Pervaiz et al. and Scobercea et al. described triquetrum excision leading to increased flexion/extension and radial/ulnar deviation of the wrist after radioscapholunate fusion[5] and midcarpal arthrodesis [6]. While triquetrum excision has shown clinical success in allowing for increased range of motion, no biomechanical study has explained why this procedure is effective. Our study suggests that the distal ridge of the hamate may guide the triquetrum towards the hook until it is fully engaged and fully constrained at the end of the DTM (flexion and ulnar deviation). This constrained position of the triquetrum at the hook of the hamate may serve as a stop to further wrist ulnar deviation. Based on this understanding, surgical procedures that excise the triquetrum would allow for increased ulnar deviation. From a kinetic standpoint, excising the triquetrum would lead to a VISI deformity of the wrist due to loss of the extension moment of the TqH joint on the lunate. In the setting of midcarpal or proximal row limited fusions, however, the carpus can be stabilized in neutral while triquetrum excision would allow for increased wrist range of motion.

To our knowledge, this is the first description of in vivo TqH kinematics through a dart

thrower's motion using a 3-D markerless bone registration technique. A complete understanding of the kinetics and kinematics of the TqH joint is important in understanding wrist stability through various ranges of motion, including the DTM and may aid in the treatment of wrist disorders including clunking wrists and dissociative carpal instabilities. Our results provide a more complete understanding of TqH joint kinematics through the DTM, and also provide a biomechanical basis for future research into carpal stability and triquetrum excision with limited wrist fusion for carpal instabilities.

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Appendix C

Carpal Cartilage Validation

C.1 Introduction

Numerous cadaveric and in vivo studies have provided detailed descriptions of the complex kinematics of the carpal bones [1, 2]; however, the fundamental interactions between ligament and cartilage contact forces have yet to be determined. Biomechanical models that include an accurate three-dimensional depiction of the cartilage surfaces are critical in computing these interactions.

However, obtaining accurate, high-resolution cartilage surfaces in vivo are not possible using current imaging techniques. An alternate approach has focused on developing an algorithm to compute cartilage surfaces based on polygonal bone surface models and carpal bone kinematics [3]. The purpose of this study is to validate this algorithm. The algorithm uses the joint space between articulating bones at multiple static positions imaged with a clinical CT scanner to create three-dimensional cartilage surfaces. We examine the validity of the algorithm by comparing computed cartilage surfaces to accurate three-dimensional cartilage surfaces generated with μ CT.

C.2 Methods

C.2.1 Imaging Procedures and 6-dof Kinematics

Six intact fresh-frozen cadaver wrists (4M and 2F, age 70 ± 6 years) were scanned in a clinical CT scanner (resolution: 0.3mm X 0.3mm X 1mm) in 9 wrist positions including neutral and 8 positions that sampled the range of orthogonal and oblique motions [4]. The targeted range of motion was 80° of flexion-extension and 40° of ulnar-radial deviation. Six-degree-of-freedom transforms from the neutral position to the 8 subsequent positions for the radius and all eight carpal bones were generated using an established markerless bone registration technique [5, 6].

Each cadaver was then carefully dissected, and selected bones including the cartilage surfaces were scanned in air at a resolution of 60 μm (isometric voxels) using a desktop micro-computed tomography system (μCT 40, Scanco Medical, CH).

Tessellated bone and surrounding soft tissue models were created by manual segmentation in Mimics 13 (Materialise, Leuven Belgium). Three-dimensional cartilage surfaces were isolated from the surrounding soft tissue using a previously established method [7, 8]. Briefly, the cartilage surfaces were manually cropped from the surrounding fascia using Geomagic (Raindrop, Geomagic, Durham, NC). A cartilage surface was easily identifiable from the surrounding tissue because it was denoted by a uniform, smooth region in the polygonal mesh compared to the non-uniform and rough surface features of the surrounding fascia.

These μCT -derived cartilage surfaces ($C_{\mu\text{CT}}$) were used as the gold standard compared to cartilage surfaces computed from joint space measurements and carpal kinematics.

C.2.2 Computed Cartilage Surfaces

The μCT -derived bone models were registered to the clinical CT-derived models using a closest-points algorithm in GEOMAGIC (Raindrop Geomagic, Durham, NC). Cartilage

surfaces were computed by calculating the joint space between the μ CT-derived bones over an area that includes closest point distances between the pair of bones, limited by a specified distance-threshold level [3]. Two surfaces, one extruded from each bone, were created by splitting the minimum inter-bone distance by half at each wrist position. Each cartilage surface was extruded from its respective bone an additional 5% of this minimum distance to allow contact.

Four separate cartilage surfaces were generated for selected radiocarpal and midcarpal articular surfaces. The first two were created by repeating the algorithm described above using two different distance-thresholds set at 2 mm (C_2) and 3 mm (C_3). The remaining two cartilage surfaces (C_{2C} and C_{3C}) were generated by trimming the 2 mm and 3 mm surfaces using bony landmarks (Figure 1a).

C.2.3 Outcome Measures and Statistical Analysis

Cartilage thickness was defined as the closest point of the bone to the cartilage surface averaged over the entire surface. Surface area was computed by numerical integration over the polygonal surface. Cartilage surface perimeter distance was defined as the average distance between the outer edges of the μ CT-derived cartilage surface and each computed surface (C_2 , C_3 , C_{2C} , C_{3C}).

The μ CT-derived facets were used as the gold standard to evaluate opposing facets of the radiocarpal (n=6), scaphocapitate (n=6), lunocapitate (n=6), and triquetrohamate (n=3) joints.

For each facet and outcome measure (thickness, surface area, and perimeter location), a one way repeated measures ANOVA using a Holm-Sidak comparison procedure was performed to identify differences between the μ CT cartilage surface and each computed cartilage threshold condition (C_2 , C_3 , C_{2C} , C_{3C}).

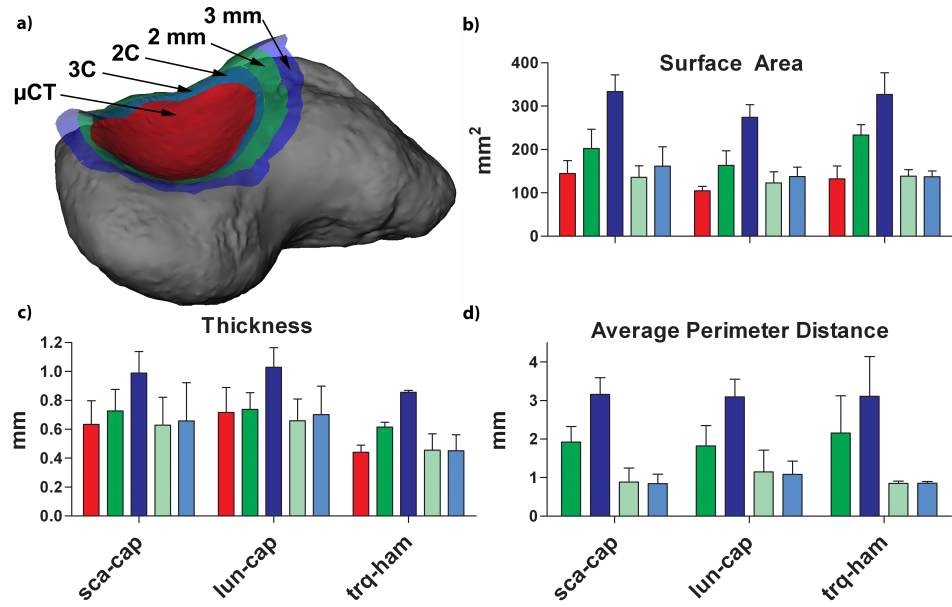


Figure C.1: (A) Scaphoid cartilage surface that articulates with the capitate (sca-cap). The μ CT cartilage surface ($C_{\mu\text{CT}}$) (red) cut 2 mm ($C_{2\text{C}}$) and cut 3 mm ($C_{3\text{C}}$) surfaces are overlaid in turquoise. The outer surfaces are computed using the C_2 (green) and C_3 (blue) threshold levels. (B) Mean cartilage thickness for representative facets for each condition; μ CT (red), 2 mm (green), 3 mm (blue), 2C (pale green), 3C (pale blue). (C) Cartilage surface area. (D) Mean distance from the perimeter (PD) of the μ CT cartilage surface for selected computed surfaces.

C.3 Results

Cartilage, computed at the C_3 threshold level significantly overestimated $C_{\mu CT}$ thickness and surface area ($P < 0.001$) (Figure C.1B, C). The C_3 cartilage surfaces had the largest perimeter differences, averaging 2.2 ± 0.3 mm over all specimens and joints (Figure C.1 D).

The C_2 , C_{2C} , and C_{3C} conditions approximated cartilage thickness and surface area with greater accuracy (Figure C.1 A, B). This was demonstrated by the number significant differences for surface area and thickness decreasing from 10 and 9 facets using the 3 mm threshold to as few as 1 and 4 facets respectively using the C_{2C} , or C_{3C} method (Table C.1).

The C_{2C} , and C_{3C} surfaces approximated $C_{\mu CT}$ perimeter with the greatest accuracy with P decreasing significantly ($P < 0.001$) across all facets (Figure C.2). For example, the min and max P for C_{2C} decreased to 0.8 ± 0.2 mm (sca-cap) and 1.9 ± 0.7 mm (rad-lun).

Table C.1: Thickness, Surface Area, and Average Perimeter Distance for $C_{\mu CT}$ and 3 mm threshold cartilage trimmed using bony landmarks (C_{3C}). * $P < 0.001$

Facet	Thickness (mm)		Surface Area (mm ²)		Perimeter Distance (mm)
	$C_{\mu CT}$	C_{3C}	$C_{\mu CT}$	C_{3C}	$C_{\mu CT}$ vs. C_{3C} *
rad-sca	0.88 (0.21)	0.85(0.17)	159(38)	201(42)*	1.5(0.6)
sca-rad	0.77(0.14)	0.84(0.17)	304(102)	264(90)	1.7(0.7)
rad-lun	0.95(0.19)	0.84(0.19)*	157(42)	193(44)	1.9(0.7)
lun-rad	1.17(0.27)	0.91(0.19)*	229(43)	214(40)	1.3(0.5)
sca-cap	0.64(0.16)	0.66(0.26)	144(30)	162(45)	0.8(0.2)
cap-sca	0.55(0.12)	0.71(0.23)*	187(42)	200(71)	1.1(0.3)
lun-cap	0.72(0.17)	0.70(0.20)	105(10)	138(22)*	1.1(0.3)
cap-lun	0.65(0.10)	0.71(0.16)	213(29)	168(13)*	1.0(0.3)
trq-ham	0.44(0.05)	0.45(0.11)	132(30)	137(13)	0.9 (0.05)
ham-trq	0.36(0.06)	0.52(0.11)	191(56)	188(17)	1.2(0.5)

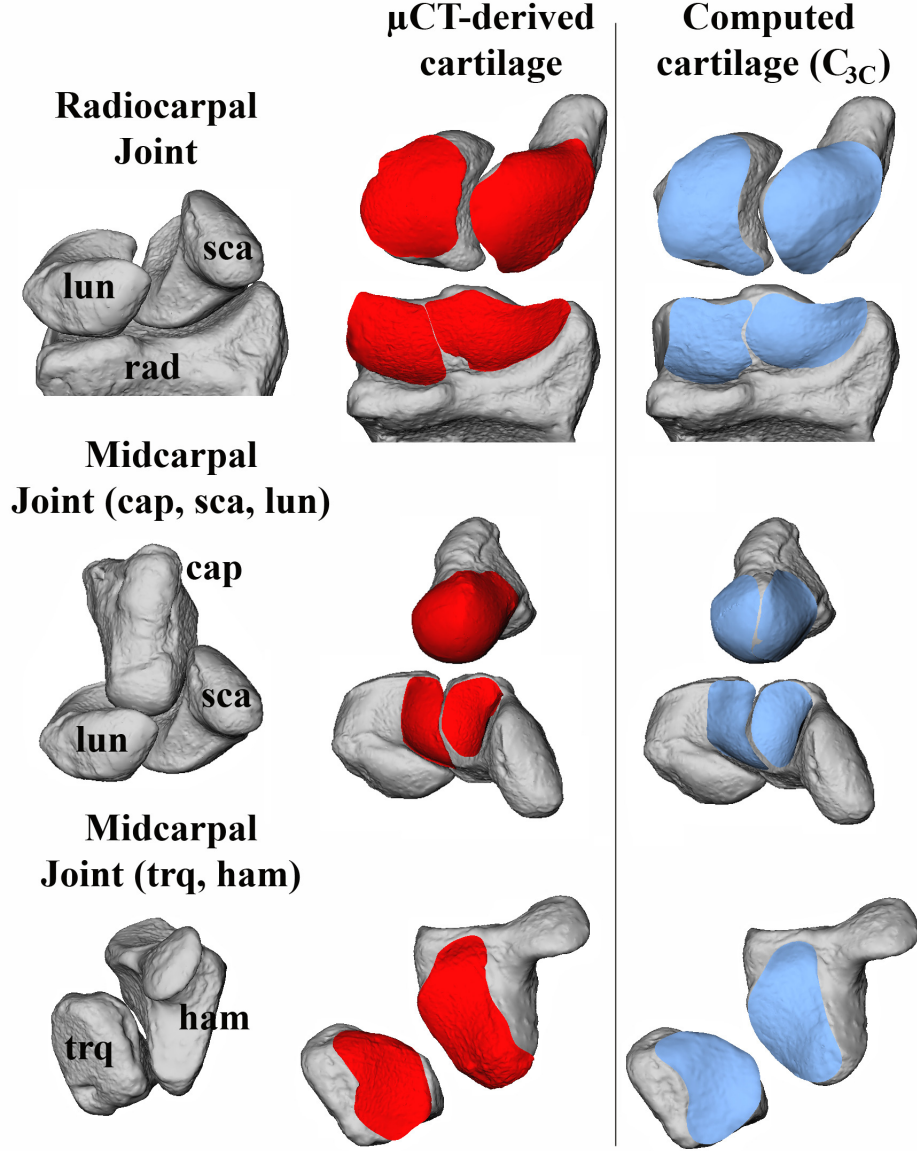


Figure C.2: Volar view of the radiocarpal and midcarpal joints of a representative right wrist. μ CT-derived cartilage surfaces ($C_{\mu CT}$) are displayed in red and cartilage computed using a 3 mm threshold and trimmed using bony landmarks (C_{3C}) are displayed in pale blue.

C.4 Discussion

The computed cartilage algorithm generated surfaces that closely mimic the underlying anatomy. Cartilage thickness was estimated within tolerances that are below the in-plane resolution of the clinical CT scanner. While cartilage surfaces computed using the 2 and 3

mm threshold values were located similarly to physiological cartilage surfaces, surface area and perimeter location were often either underestimated or overestimated. A reasonable approach to correct these discrepancies is to overestimate the cartilage surface area using the 3 mm threshold value, and then use anatomical landmarks to trim the excess cartilage. The validity of this technique (C_{3C}) is demonstrated by trimmed cartilage thickness, location, and surface area values approaching that of the μ CT-derived cartilage surfaces.

When interpreting our results there are several important limitations to consider. First, early stage osteoarthritis may have affected the geometry of the carpal bone cartilage we studied. The results of our direct comparison are unaffected by these changes, however, we caution against drawing broad conclusions about cartilage geometry in a younger population using this dataset. Finally, our computed cartilage was generated using μ CT-derived bone surface models that were driven with kinematics obtained from clinical CT. The use of bone surface models derived from clinical CT would likely result in smaller thickness values because the bone volumes would be larger due to partial volume effects. The use of CT-derived bones instead of μ CT-derived bones to generate cartilage would not likely affect cartilage perimeter location or surface area because the current resolution of clinical CT sufficiently captures the landmarks used to isolate the cartilage surfaces. In vivo imaging of cartilage using CT or MRI remains a challenge. Cartilage thickness ranges between 0.36 mm and 1.17 mm and accuracy of these modalities is of the similar magnitude. Second, isolation of an individual cartilage surface is difficult due to lack of contrast with the opposing facet and the surrounding soft tissue. We have presented a method to compute cartilage surfaces that are geometrically accurate compared to high-resolution cartilage surfaces obtained from cadavers using μ CT. These surfaces can be used as input into a model to study articular contact in the carpus.

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