March 2018

The Kinematic and Biomechanical Effects of Bracing on the Rehabilitation of the LCL Injured Elbow

Sara M. Banayan
The University of Western Ontario

Supervisor
Johnson, James A.
The University of Western Ontario

Co-Supervisor
King, Graham J.W.
The University of Western Ontario

Graduate Program in Biomedical Engineering

A thesis submitted in partial fulfillment of the requirements for the degree in Master of Engineering Science

© Sara M. Banayan 2018

Abstract

Lateral collateral ligament (LCL) injuries are often treated non-operatively or with surgical repair. If instability persists, hinged elbow orthoses (HEOs) are often recommended. However, these orthoses are designed as a straight hinge, which does not account for the native carrying angle of the elbow. A custom HEO was designed to adjust the orthosis valgus angulation to measure *in vitro* elbow kinematics and biomechanics. An *in vitro* study investigated the effect of HEO valgus angulation during simulated active and passive flexion, in the vertical dependent and varus positions, with the forearm pronated and supinated. In the vertical dependent position, the orthosis did not produce instability and in the varus position, greater HEO angles trended towards improving elbow stability. Passive flexion was not found to worsen instability. In a subsequent study, a novel LCL tensioning mechanism is introduced to examine the effects of orthosis valgus angulation on LCL loads. No significant differences were found, as the tension did not change much throughout the range of motion. Future work is proposed to further improve the understanding of elbow kinematics and biomechanics to optimize rehabilitation techniques, surgical protocols and orthosis designs.

Keywords

Elbow joint, lateral collateral ligament, hinged elbow orthosis, brace, instability, rehabilitation, biomechanics, kinematics

Co-Authorship Statement

Chapter 1   **Sole Author** – Sara Banayan

Chapter 2   **Study Design** – Sara Banayan, Armin Badre, David Axford, Jim Johnson, Graham King

Data Collection – Sara Banayan, Armin Badre, David Axford, Martine McGregor

Data Analysis – Sara Banayan, Dan Langhor

Statistical Analysis – Sara Banayan

Manuscript Preparations – Sara Banayan

Manuscript Revisions –Jim Johnson, Graham King

Chapter 3   **Study Design** – Sara Banayan, Armin Badre, David Axford, Jim Johnson, Graham King

Data Collection – Sara Banayan, Armin Badre, David Axford, Martine McGregor

Data Analysis – Sara Banayan, Dan Langhor

Statistical Analysis – Sara Banayan

Manuscript Preparations – Sara Banayan

Manuscript Revisions –Jim Johnson, Graham King

Chapter 4   **Study Design** – Sara Banayan, Armin Badre, David Axford, Jim Johnson, Graham King

Data Collection – Sara Banayan, Armin Badre, David Axford, Martine McGregor

Data Analysis – Sara Banayan, Dan Langhor

Statistical Analysis – Sara Banayan

Manuscript Preparations – Sara Banayan

Manuscript Revisions –Jim Johnson, Graham King

Chapter 5   **Sole Author** – Sara Banayan

Appendix A   **Sole Author** – Sara Banayan

Appendix B   **Sole Author** – Sara Banayan

Appendix C   **Sole Author** – Sara Banayan

Acknowledgments

First and foremost, I would like to thank my supervisors and mentors, Dr. Johnson and Dr. King, without whom this work would not have been possible. Dr. Johnson, thank you for your continuous guidance and expertise. Your enthusiasm, patience and encouragement has made these past two years an amazing experience. I consider myself incredibly lucky to have had the opportunity to work with such an accomplished researcher and engineer. Dr. King, thank you for your never-ending encouragement, and medical expertise, particularly during our many testing day crises. Your passion for elbow research has greatly inspired me. I am very grateful to have had the honour to work with you.

I would like to thank Team Elbow for their continued dedication to this project. Armin, David and Martine, I appreciate all the long hours and weekends spent in the lab and could not have completed this work without you. Armin, thank you for being my research partner and supporting me throughout this project. I am in awe of your knowledge and technical abilities. David, thank you becoming our expert on the elbow simulator, and taking a genuine interest in this project. Your knowledge of software and hardware played a vital role in this project and I am very grateful. Martine, thank you for commuting to London on those early mornings to accommodate everyone's busy schedules. Testing days would not have gone as smoothly without your help, positivity, and music playlists.

Thank you to all my colleagues at HULC for their support and friendship. Dan, thank you for taking the time to answer all my software and statistical questions. Data analysis and statistical analysis would not have been possible without you. Steph, thank you for being the most encouraging roommate, colleague and friend.

I would also like to thank my amazing parents, brother, family and friends. Mom and dad, I would not have been able to do this without your endless support as well as the values and work ethic you have instilled in me. Jake, thank you for continuously motivating me and keeping me sane. I appreciate your unwavering support.

Table of Contents

List of Tables

List of Figures

List of Appendices

Chapter 1

1   Background and Literature Review

Overview

The purpose of this thesis is to investigate the effect of hinged elbow orthoses (HEOs) on kinematics and biomechanics of the lateral collateral ligament (LCL) deficient elbow. This chapter reviews the relevant anatomy, biomechanics and kinematics. A summary of injury pathology and management, as well as joint motion simulation is also described. Additionally, the rationale, objectives and hypothesis pertinent to this thesis are summarized.

1.1   Elbow Anatomy

The anatomy of the elbow joint can be divided into the bones, muscles and ligaments that comprise it. These anatomical categories are necessary to understand the motion and mechanics of the elbow joint.

1.1.1 Osteology

The elbow joint is made up of three bones - the humerus, radius and ulna (Figure 1-1). The humerus is situated in the upper forearm, and the radius and ulna are located laterally and medially in the forearm, respectively. The distal end of the humerus articulates with the proximal ends of the radius and ulna (Ramponi & Kaufmann, 2012). The orientation of these bones results in three articulations at the joint: the ulnohumeral joint, the radiocapitellar joint and the proximal radioulnar joint (Ramponi & Kaufmann, 2012).
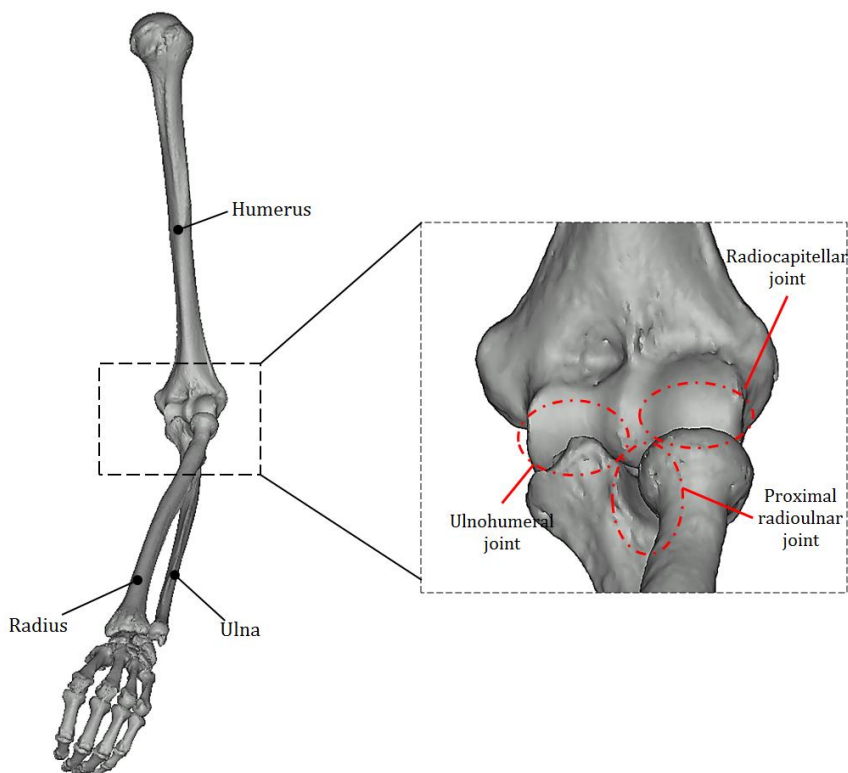
Figure 1-1: Elbow Joint Orientation. *The orientation of the humerus, ulna and radius bones that comprise the elbow joint. The alignment of these bones results in three joints – the radiocapitellar joint, the ulnohumeral joint, and the proximal radioulnar joint.*

Figure 1-2 shows the structure of the humerus. At the anterior, distal end, there are two condyles: medially, the trochlea, which articulates with the ulna and laterally, the capitellum which articulates with the radial head (Morrey & Sanchez-Sotelo, 2009). On the medial- and lateral-most ends of the humerus there are the medial and lateral epicondyles, respectively, which serve as ligament attachment points (Fornalski, Gupta, & Lee, 2003; Bernard F. Morrey & Sanchez-Sotelo, 2009). The radial and coronoid fossae, located anteriorly, house the radial head of the radius and coronoid process of the ulna, respectively. The olecranon fossa, located on the posterior, distal end of the humerus, accommodates the olecranon of the ulna (Morrey & Sanchez-Sotelo, 2009).
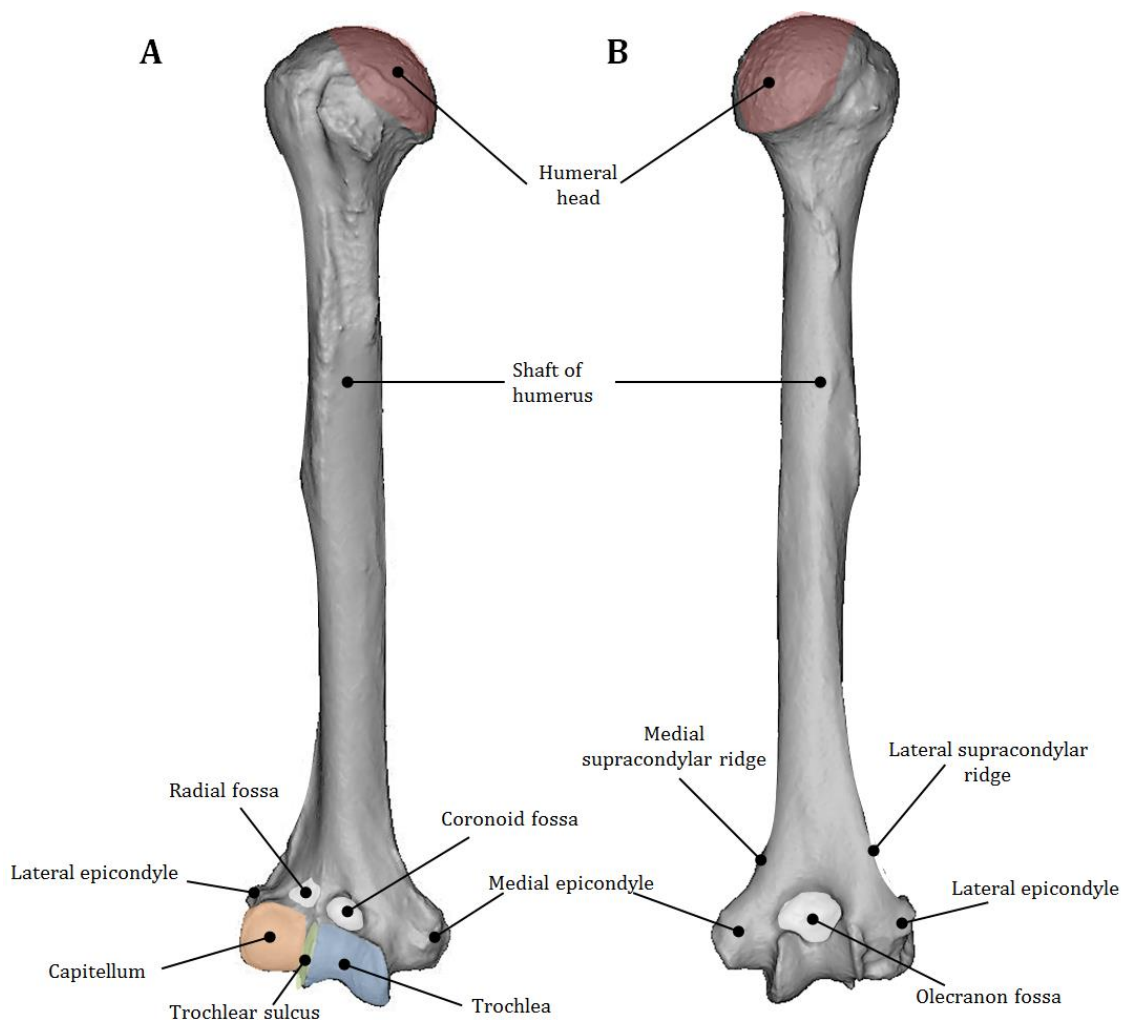
Figure 1-2: Bony Anatomy of the Humerus. *Bony anatomy of the right humerus from the (A) anterior view and (B) posterior view.*

Figure 1-3 depicts the anatomy of the ulna. The proximal end of the ulna contains the coronoid and olecranon processes (Fornalski et al., 2003). The arc that joins these articular surfaces is called the sigmoid notch and creates approximately a 30° angle with the ulnar shaft (Fornalski et al., 2003; Pederzini, Eygendaal, & Denti, 2016). The greater sigmoid notch articulates with the trochlea of the humerus, and contributes greatly to the inherent stability of the elbow (Fornalski et al., 2003; Morrey & Sanchez-Sotelo, 2009). The lesser sigmoid notch articulates with the radial head (Fornalski et al., 2003).
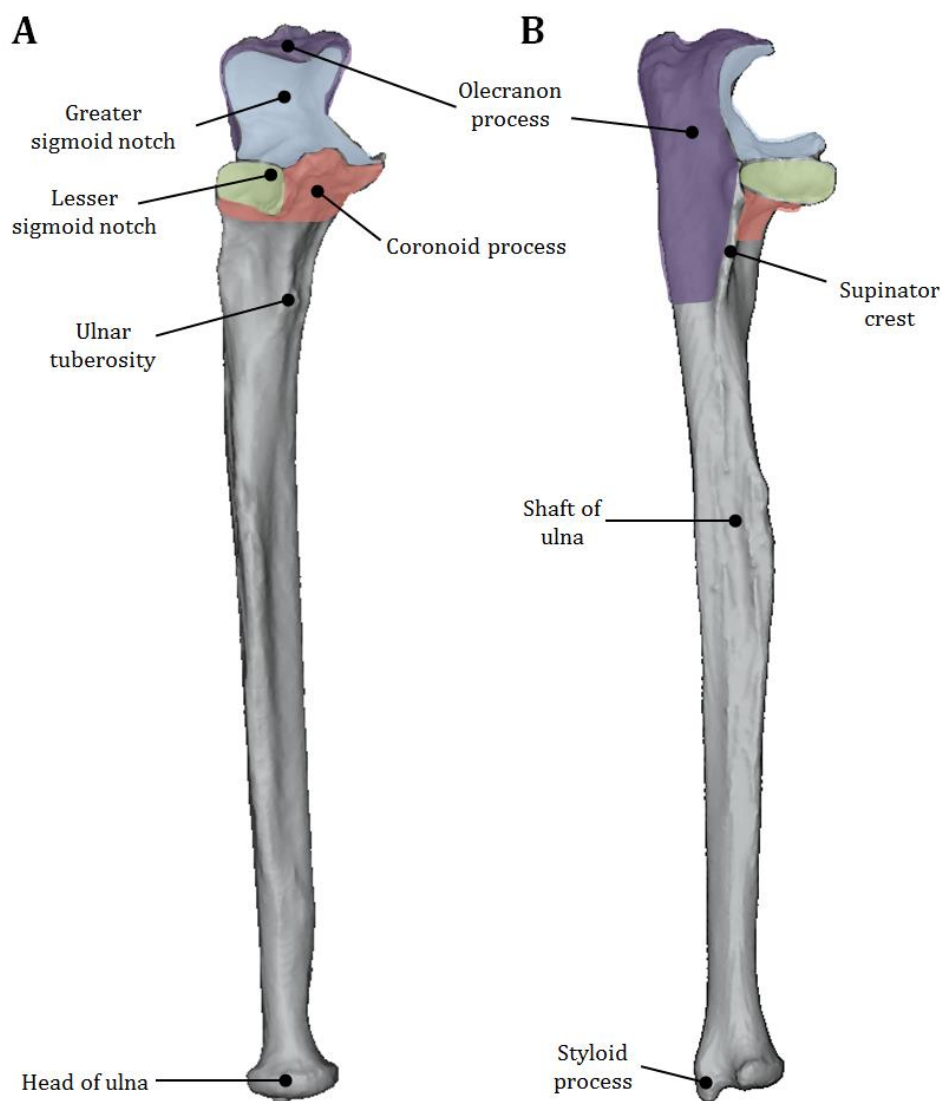
Figure 1-3: Bony Anatomy of the Ulna. *Bony anatomy of the important structures of the right ulna from the (A) anterior view and (B) lateral view.*

Figure 1-4 illustrates the structure of the radius. The radial head, located proximally, has a cylindrical shape and forms a 15° angle with the shaft of the radius (Pederzini et al., 2016). The head also has a concave dish, covered with hyaline cartilage, that articulates with the capitellum of the humerus (Fornalski et al., 2003; O'Driscoll, Jupiter, King, Hotchkiss, & Morrey, 2000). Figure 1-5 depicts the articulation between the radius and the ulna. Proximally, the posteromedial section of the radius articulates with the radial notch on the ulna and distally, the ulnar notch of the radius articulates with the head of the ulna (Fornalski et al., 2003; Morrey, 2000; Morrey & Sanchez-Sotelo, 2009; O'Driscoll et al., 2000).
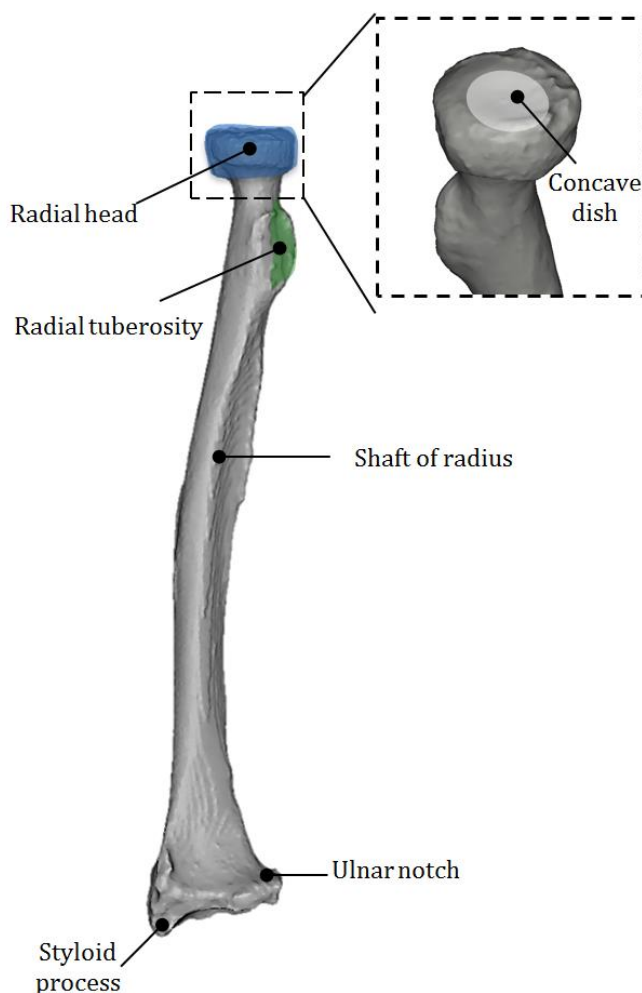
Figure 1-4: Bony Anatomy of the Radius. *Anterior view of the bony anatomy of a right radius with important structures highlighted.*
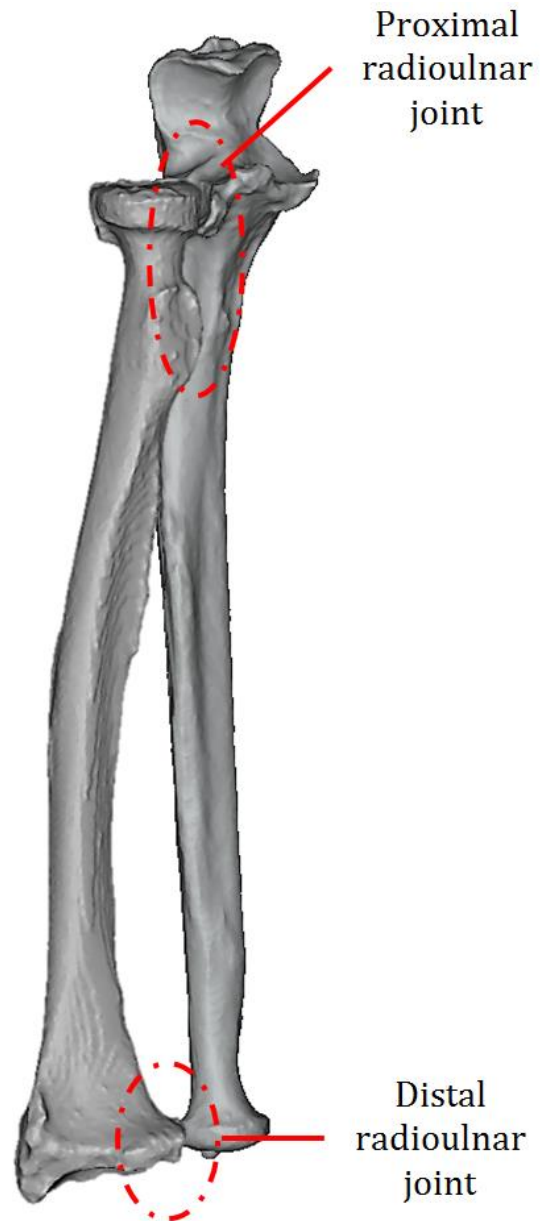
Figure 1-5: Articulation Between the Radius and Ulna. *Proximal and distal articulation of the radius and ulna resulting in the proximal radioulnar joint (PRUJ) and distal radioulnar joint (DRUJ). Right radius and ulna shown.*

1.1.1.1   Joint Capsule

The elbow joint is surrounded by a joint capsule, innervated by various nerve branches (Figure 1-6) (Morrey & Sanchez-Sotelo, 2009). The anterior portion of the capsule attaches above the coronoid and radial fossae (Morrey & Sanchez-Sotelo, 2009). The posterior capsule attaches above the olecranon fossa (Morrey & Sanchez-Sotelo, 2009). There is greater laxity on the anterior and posterior sides to permit flexion and extension, respectively (Pederzini et al., 2016).
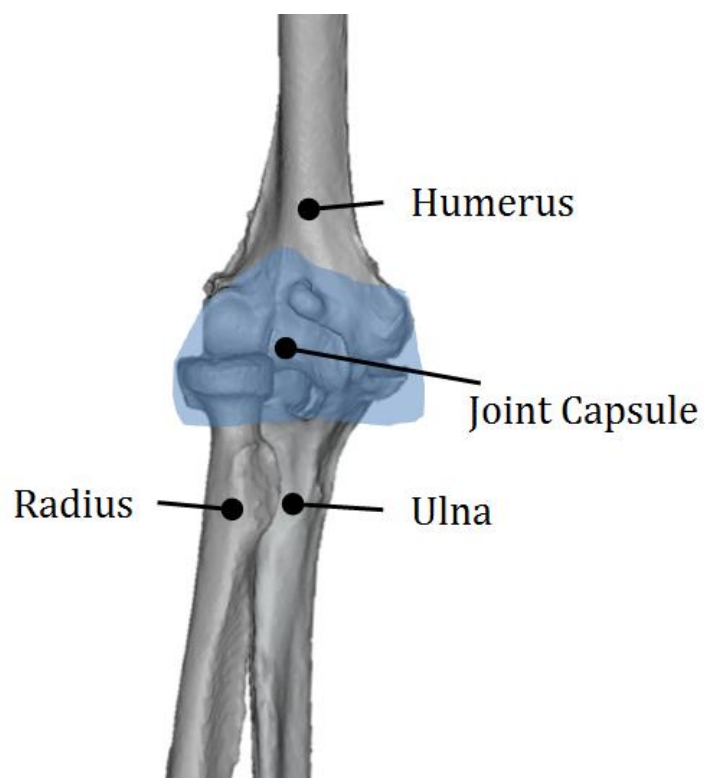
Figure 1-6: Elbow Capsule. *Anterior view of the capsule surrounding the elbow joint. Right elbow shown.*

1.1.2 Myology

There are 4 groups of muscles surrounding the elbow: the flexor/extensor muscles of the elbow and flexors/extensors of the wrist (Figure 1-7) (Pederzini et al., 2016). Important extensor muscles of the elbow are the triceps brachii and anconeus. The triceps has 3 origins (the long head on the scapula, the lateral head on the lateral humerus, and the medial head on the medial humerus) and inserts on the olecranon tip (Jacobson, 2007; Pederzini et al., 2016). The anconeus is a triangular extensor muscle that originates on the lateral epicondyle of the humerus and inserts on the lateral side of the proximal ulna (Pederzini et al., 2016).

The main flexors of the elbow, located proximally in the upper arm, are: the biceps, the brachialis and the brachioradialis. The biceps inserts on the radial tuberosity, and the brachialis inserts mainly on the coronoid process of the proximal ulna (Jacobson, 2007; Morrey & Sanchez-Sotelo, 2009). The brachioradialis crosses over the anterior side of the elbow joint. It originates on the lateral side of humerus and inserts on the distal end of the radius at the styloid process (Morrey & Sanchez-Sotelo, 2009). Weaker flexors of the forearm include the pronator teres and supinator muscles. The pronator teres originates from the medial epicondyle as well as the coronoid process of the ulna and inserts between the proximal and middle radius. The supinator, a rhomboid shaped, flat muscle, originates on the lateral anterior side of the lateral epicondyle, the lateral collateral ligament and proximal anterior crest of the ulna. It inserts on the proximal radius and pronator teres (Morrey & Sanchez-Sotelo, 2009).

There are also flexors and extensors of the wrist that are relevant to the elbow joint. On the medial side of the joint, the common flexor tendon originates on the medial epicondyle of the humerus (Jacobson, 2007). It consists of the flexor carpi radialis, palmaris longus, flexor carpi ulnaris and flexor digitorum superficialis (Jacobson, 2007). The common extensor tendon originates laterally, at the lateral epicondyle of the humerus and is composed of the extensor carpi radialis brevis, extensor digitorum, extensor digiti minimi and extensor carpi ulnaris (Jacobson, 2007). There is also the extensor carpi radialis longus, which originates near the lateral epicondyle of the humerus (Jacobson, 2007).
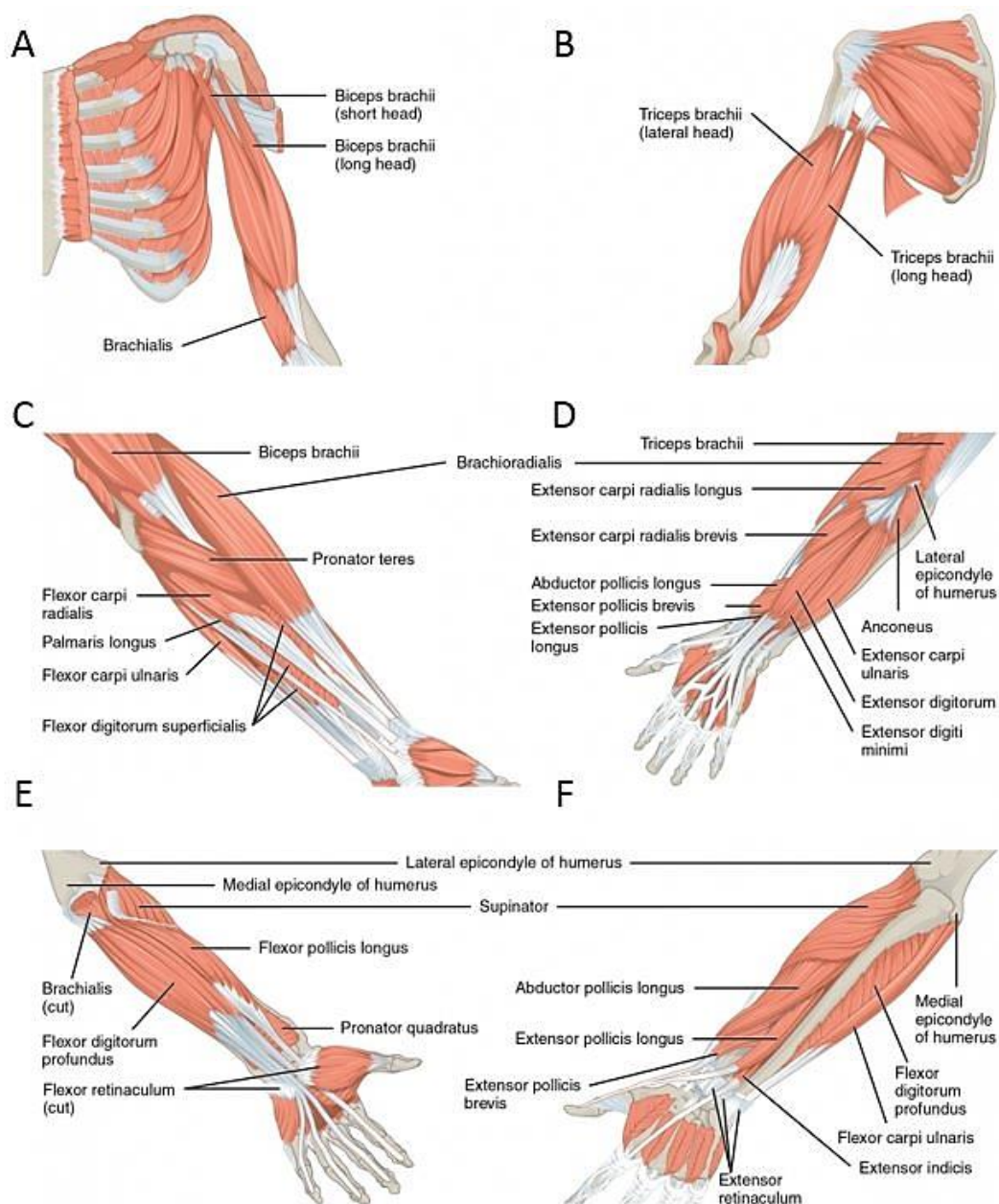
Figure 1-7: Elbow Myology. *Important extensors and flexors of the elbow joint. (A) Upper arm muscles, anterior lateral view, (B) upper arm muscles, posterior view, (C) forearm superficial muscles, palmar view, (D) forearm superficial muscles, dorsal view, (E) forearm deep muscles, palmar view, and (F) forearm deep muscles, dorsal view ("Upper extremity muscle anatomy muscles of the pectoral girdle and upper limbs anatomy and - upper extremity muscle anatomy," 2016).*

1.1.3 Desmology

There are two major ligament complexes associated with the elbow: the medial collateral ligament (MCL) and the lateral collateral ligament (LCL) (Figure 1-8) (Morrey & Sanchez-Sotelo, 2009). The MCL supports the ulnohumeral joint and provides stability under valgus loading (Cohen & Bruno, 2001). It can be classified into anterior, posterior, and transverse bundles. The anterior bundle is divided into anterior (AMCL), posterior (PMCL) and deep medial segments (Morrey & Sanchez-Sotelo, 2009). The anterior bundle originates on the inferior portion of the medial epicondyle and inserts on the sublime tubercle of the proximal ulna (Dines & Altchek, 2015). The posterior bundle also originates on the medial epicondyle, but inserts on the olecranon process (Dines & Altchek, 2015). The transverse segment originates on the coronoid of the ulna and inserts onto the olecranon tip (Morrey & An, 1985).

The LCL is divided into the radial collateral ligament (RCL), the annular ligament (AL), the accessory lateral collateral ligament (ALCL) and the lateral ulnar collateral ligament (LUCL) (Morrey & Sanchez-Sotelo, 2009). The AL originates on the lesser sigmoid notch and inserts on the supinator crest of the ulna to hold to the radial head against the ulna (Mirzayan & Itamura, 2004). The RCL originates on the lateral epicondyle of the distal humerus and runs distally into the AL (Cohen & Bruno, 2001). The LUCL originates at the lateral epicondyle and merges into the AL before it inserts on the supinator crest (Canale & Beaty, 2012; Mirzayan & Itamura, 2004). The ALCL is attached to both the AL and the supinator crest of the ulna and provides support to the AL and RCL during varus stress (Sivananthan, Sherry, Warnke, & Miller, 2012; Starkey & Brown, 2015). Regan et al. (2001) investigated the biomechanical behaviour of the AMCL, PMCL and RCL. They found that the AMCL is the strongest ligament with a failure load of 260.9 N, followed by the RCL, then the PMCL (Regan, Korinek, Morrey, & An, 1991).
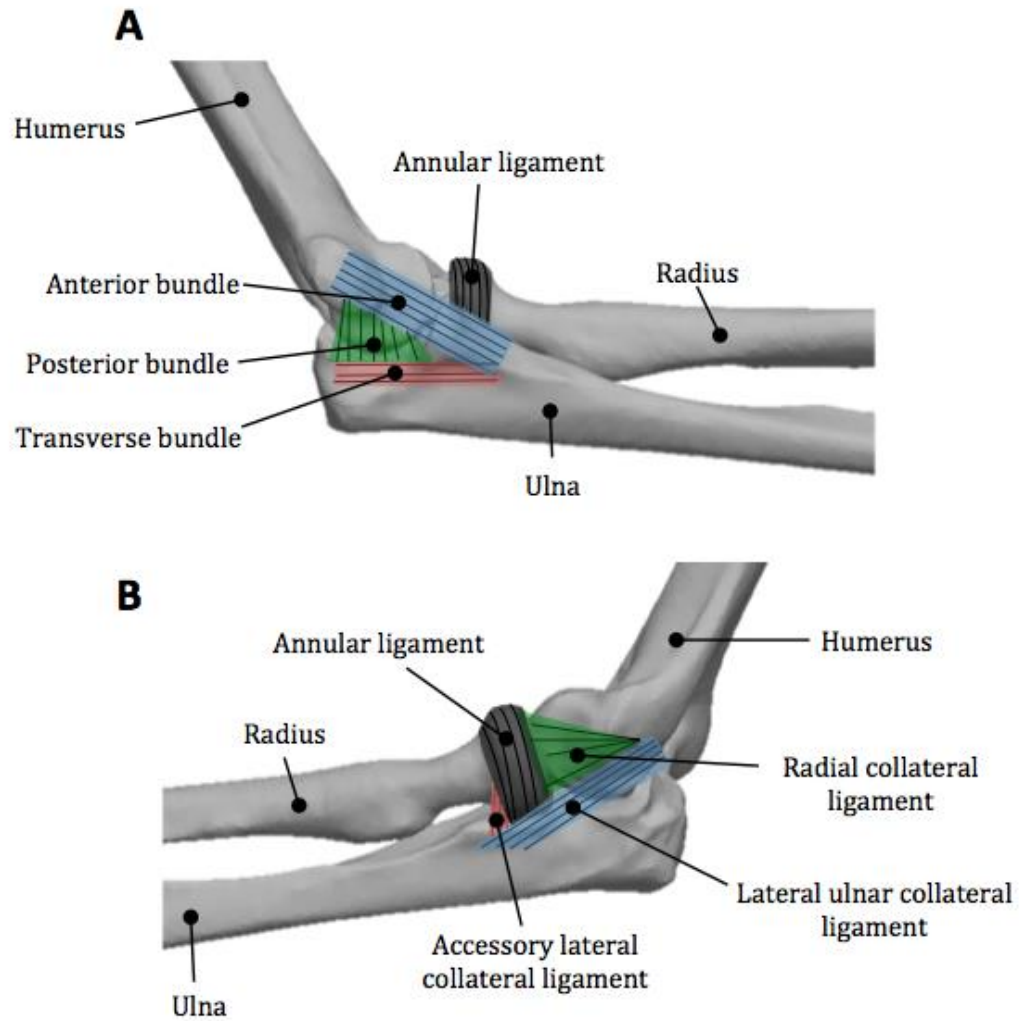
Figure 1-8: Ligaments of the Elbow. *(A) Medial view of a left elbow showing the medial collateral ligament (MCL), and (B) lateral view of a left elbow depicting the lateral collateral ligament (LCL).*

1.2 Biomechanics and Kinematics of the Elbow Joint

1.2.1 Range of Motion

The elbow joint is described as a trochoginglymoid joint because it permits 2 degrees of freedom: the ulnohumeral joint permits flexion and extension (hinge joint) and the radiocapitellar and radioulnar joints facilitate axial rotation (pivoting motion) (Morrey & Sanchez-Sotelo, 2009; Schenck, 1999).

1.2.1.1   Flexion/Extension

The elbow flexion-extension range of motion is between 0° and 150° for extension and flexion, respectively, however Morrey et al. (1981) determined that most essential daily activities require between 30° and 130° of flexion (Figure 1-9) (Fornalski et al., 2003; Morrey & Sanchez-Sotelo, 2009).
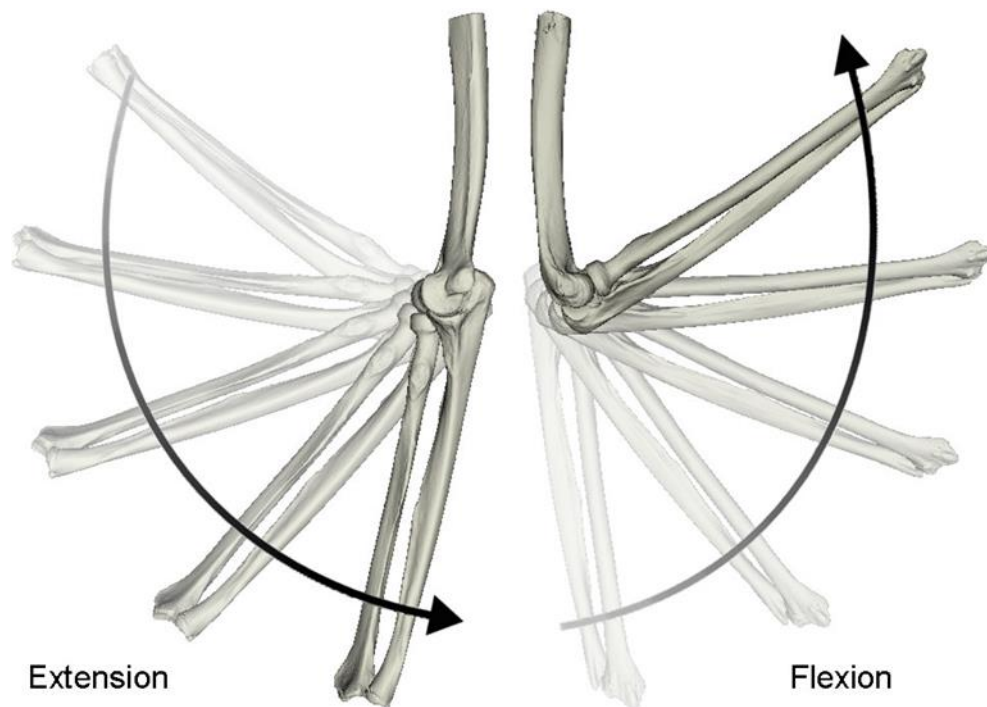
Figure 1-9: Flexion-Extension of the Elbow. *Medial (left) and lateral (right) views of a right elbow from extension to flexion. The elbow is capable of approximately 0° of extension to 150° of flexion (Ferreira, 2011).*

The axis of rotation for flexion-extension passes through the center of the capitellum and trochlear sulcus (Figure 1-10) (Deland, Garg, & Walker, 1987; London, 1981). It is oriented approximately 4-8° in the valgus direction relative to the long axis of the humerus and 3-8° internally rotated with respect to the humeral epicondyles (Ferreira, 2011; O'Driscoll et al., 2000). The distal humeral articular surface is also oriented 30° anteriorly (Fornalski et al., 2003; Morrey & Sanchez-Sotelo, 2009; Tanaka, An, & Morrey, 1998).
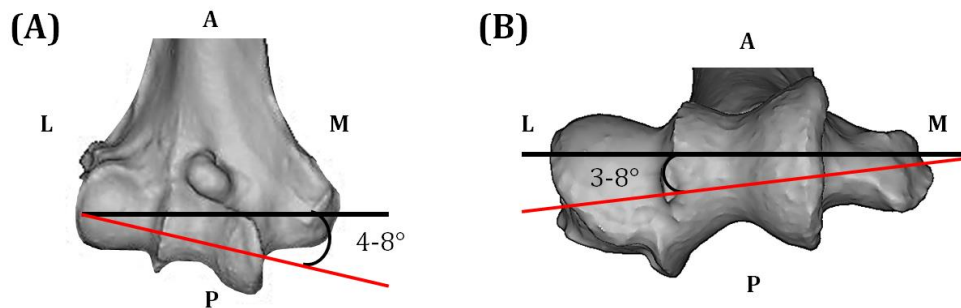
Figure 1-10: Flexion-Extension Axis of the Elbow Joint. *The flexion-extension axis of the elbow is oriented (A) 4-8° in the valgus direction, and (B) 3-8° internally rotated. M = medial view, L = lateral view, A = anterior view and P = posterior. Right humerus shown.*

Throughout elbow motion, the tautness of the ligaments varies. One experiment concluded that the AMCL and RCL are taut throughout most of the arc of flexion, whereas the PMCL is only taut with the arm in full flexion (Fuss, 1991; Regan et al., 1991). Unlike the MCL, the LCL is isometric through the entire ROM (Mirzayan & Itamura, 2004; Morrey & An, 1985; Schwab et al., 1980). The MCL originates posterior to the axis of rotation of the elbow, therefore each segment of the MCL tightens at a different angle of flexion-extension. During full extension, the AMCL is taut. At 60° of flexion the PMCL begins to tauten (Mirzayan & Itamura, 2004). Between 60° and 90° the posterior bundle tightens as well (Mirzayan & Itamura, 2004; Morrey & An, 1985). At 120°, the posterior bundle and PMCL are the major stabilizers as the AMCL begins to loosen (Mirzayan & Itamura, 2004). An *in vitro* study found that load of the AMCL was higher in the valgus arm position compared to the dependent position and in flexion relative to extension (Fay, Lalone, Ferreira, Johnson, & King, 2010).

1.2.1.2   Pronation/Supination

The forearm may be rotated 75° so that the palm of the hand is facing down (pronation) or rotated 85° the other way with the palm facing up (supination) however most activities of daily living fall within 50° forearm pronation to 50° of supination (Figure 1-11) (Fornalski et al., 2003; Morrey & Sanchez-Sotelo, 2009). Forearm pronation is largely attributed to the pronator muscle, whereas the biceps brachii is largely responsible for supination. The supinator muscle also aids in supination, however it is considered a weaker supinator (Morrey & Sanchez-Sotelo, 2009). During rotation from pronation to supination the radius rotates around the ulna and up to 6° of ulnar rotation occurs (Morrey & Sanchez-Sotelo, 2009; Olsen et al., 1998; Pederzini et al., 2016). During flexion and pronation, there is high contact between the radial head and the capitellum (Pederzini et al., 2016). However, during extension and supination, there is minimal and no contact, respectively (Pederzini et al., 2016).
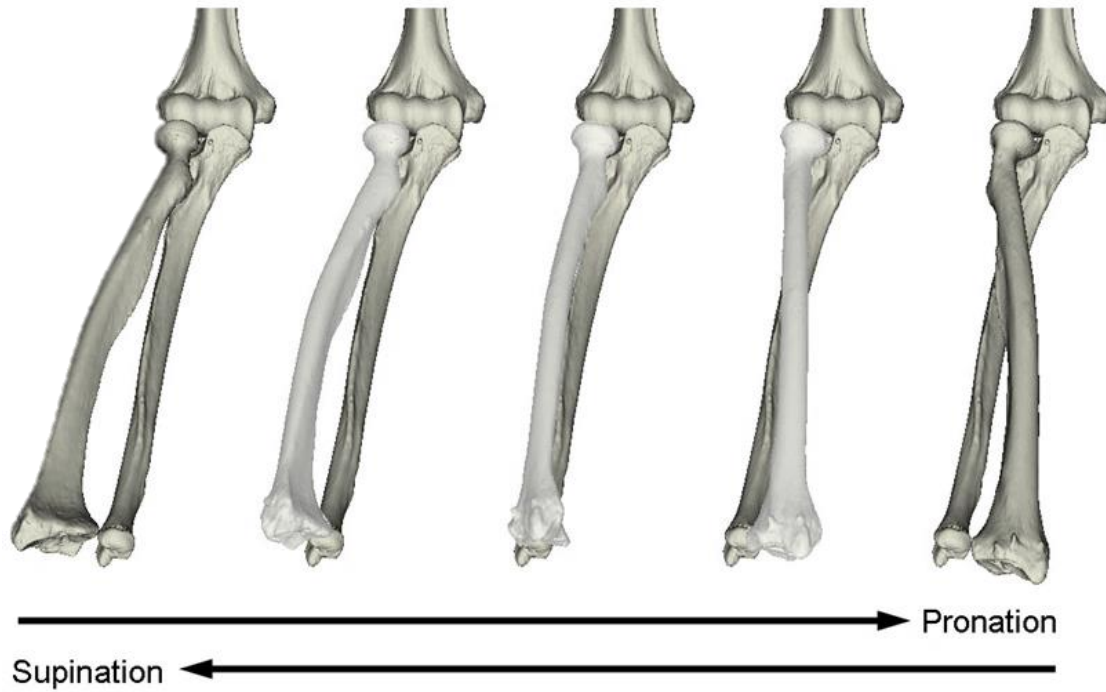
Figure 1-11: Pronation-Supination of the Elbow. *Anterior view of a right elbow from supination to pronation. The elbow can rotate from approximately 75° in pronation to 85° in supination (Ferreira, 2011).*

1.2.2 Carrying Angle

The angle formed by the long axis of the humerus and the long axis of the forearm in full extension is called the carrying angle (Figure 1-12) (Morrey & Sanchez-Sotelo, 2009). It is inversely proportional to the length of the bones in the forearm, where shorter individuals have larger carrying angles and vice versa (Khare, Goel, Saraf, Singh, & Mohanty, 1999). Most studies agree that the mean carrying angle is approximately 11-14° in men, and 13-16° in women (Atkinson & Elftman, 1945; Morrey & Sanchez-Sotelo, 2009; Paraskevas et al., 2004; Terra et al., 2011; van Roy, Baeyens, Fauvart, Lanssiers, & Clarijs, 2005). Additionally, it is approximately 1 degree more on the dominant side, compared to the non-dominant (Morrey & Sanchez-Sotelo, 2009; Yilmaz et al., 2005). The carrying angle is in a valgus orientation when the arm is extended, and in a varus orientation when the arm is flexed (Morrey et al., 1981). As the arm is flexed, the carrying angle decreases to a varus angle of $1.8° \pm 2.9$ in men and $1.6° \pm 2.3$ in women (van Roy et al., 2005).
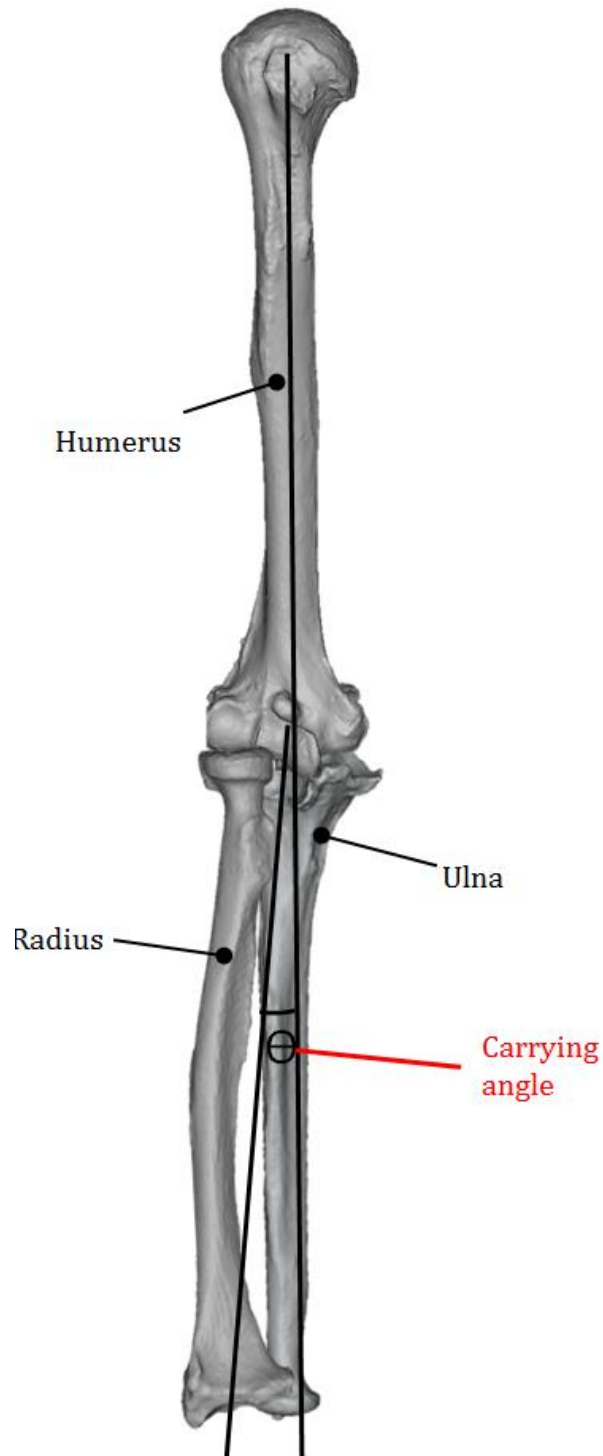
Figure 1-12: Carrying Angle. *The carrying angle, Ɵ, is generally between 11-14° in men, and 13-16° in women. Right arm shown.*

1.2.3 Stability

The elbow is an inherently stable joint primarily due to articular geometry and ligamentous structures and secondarily due to soft tissue stabilizers (Cohen & Bruno, 2001; Morrey & An, 1983). During flexion, the coronoid and radial fossae provide stability by protecting against anterior impingement Similarly, during extension, the olecranon fossa inhibits posterior impingement (Pederzini et al., 2016).

Soft tissue stabilizers can be classified as static or dynamic (O'Driscoll et al., 2000). Dynamic stability is determined by neuromuscular factors, whereas static stability is governed by the osteoarticular structure as well as the capsule and ligamentous structures (de Haan et al., 2011). The dynamic stabilizers include the aforementioned muscles associated with the elbow joint that provide stability by compressing the joint (de Haan et al., 2011; Pederzini et al., 2016). The important dynamic stabilizers that produce compressive forces on the elbow joint include the anconeus, the triceps and the brachialis (O'Driscoll et al., 2000).

The static stabilizers can be divided into primary and secondary constraints. The primary static stabilizers cause laxity upon release and include the ulnohumeral articulation, the AMCL, and the LCL (de Haan et al., 2011; O'Driscoll et al., 2000; Pederzini et al., 2016). It has been found that AMCL provides greater joint stability than the PMCL (Schwab et al., 1980). A secondary stabilizer, defined by the insufficiency of release alone to cause laxity, includes the radial head and common flexor and extensor origins (de Haan et al., 2011; O'Driscoll et al., 2000).

Morrey and An (1983) conducted an investigation on the primary stabilizers of the elbow and found that with the arm fully extended, valgus stability is accredited to the MCL, anterior capsule and bone articulation equally, whereas varus stability is divided between soft tissue constraints and joint articulation (Morrey & An, 1983). Conversely, another study reported that the elbow capsule has no role in stability when the ligaments are intact (Nielsen & Olsen, 1999). At 90° of flexion, majority of valgus stability is contributed by the MCL (55%) and varus stability is mainly attributed to joint articulation (75%) (Morrey & An, 1983).

The LCL resists varus instability, with the LUCL as the main lateral stabilizer (Canale & Beaty, 2012; Morrey & An, 1985; Sanderhoff OIsen et al., 1996). However, the LCL only provides 14% of the elbow's stability during complete extension and only 9% in 90° of flexion (Canale & Beaty, 2012). The bulk of the stability is attributed to the bony articular surfaces as well as the anterior capsule (Canale & Beaty, 2012).

1.2.4 Static Analyses

The elbow joint is often analyzed as a static, two-dimensional system. The elbow is assumed to be a hinge joint with the net forces and moments acting on it equal to zero (Morrey & Sanchez-Sotelo, 2009; Nordin & Frankel, 2001). Figure 1-13 shows a free body diagram of the forces that are often included in the analysis. These include the total weight of the forearm acting at the center of gravity, the brachialis force exerted on the ulna and the forces applied on the radius by the biceps and brachioradialis (Nordin & Frankel, 2001). A weight being held in the hand may also be included in the analysis. Anthropometric data may be used to approximate body segment lengths and weights as well as the location of center of mass (Winter, 1990).

When all three muscle forces are included, it becomes an indeterminate problem that can be solved by either reduction or optimization methods. The reduction method involves grouping forces together or increasing the number of equations using knowledge of the cross-sectional area and electromyography of the muscle. The optimization method assumes that the muscular force distribution optimizes some physical parameter (Prendergast, van der Helm, & Duda, 2005). The elbow joint force is at a maximum in full extension, reduces until approximately 100°, and then increases until full flexion (Ilbeigi & Ramezani, 2014). Typically, the magnitude of elbow joint forces ranges from 0.3 – 0.5 times body weight using these approaches (An, Kwak, Chao, & Morrey, 1984).
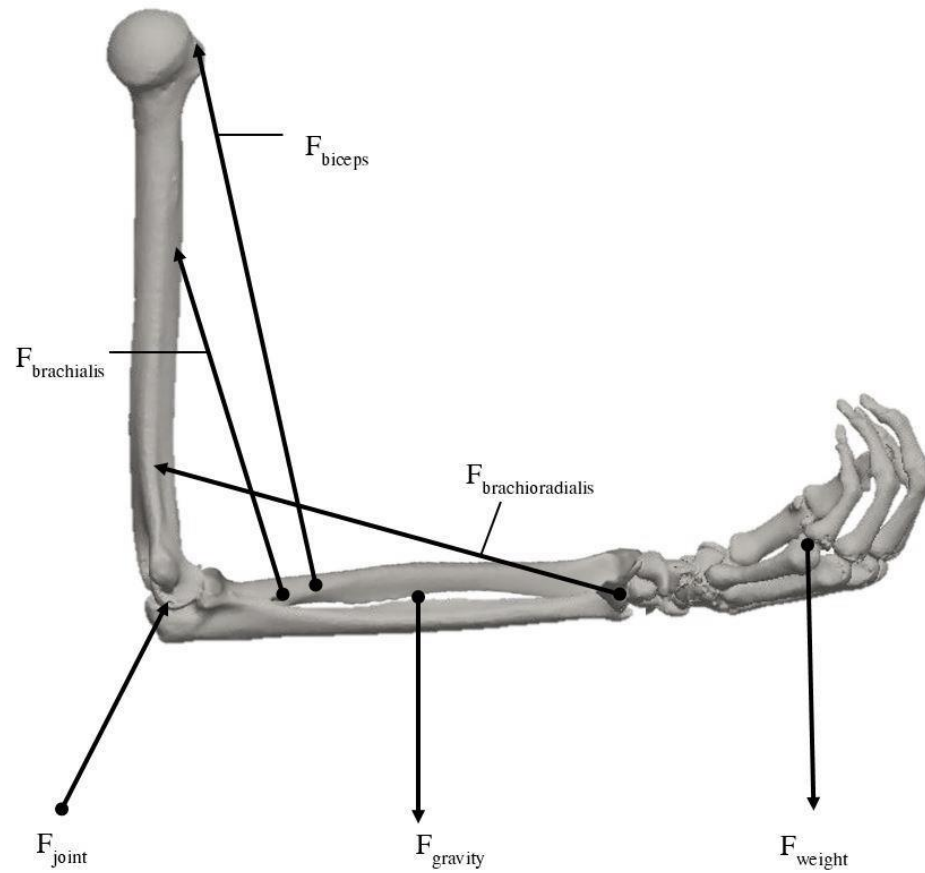
Figure 1-13: Free Body Diagram of the Elbow. *Free body diagram of the elbow at 90° including important forces often included in a static analysis. These analyses are often used to determine the joint reaction force (F$_{joint}$) Left arm shown.*

1.2.5 Bone and Joint Loading

It has been suggested that the joint forces are greater in extension and pronation compared to flexion and supination (Azar, Canale, & Beaty, 2016). During axial loading with the elbow extended, the radiohumeral joint withstands approximately 60% of the axial load, with the remaining 40% being transferred to the ulnohumeral joint (Azar et al., 2016; O'Driscoll et al., 2000). However, load transmission varies based on arm and forearm position. In the valgus position, there is contact between the radial head and capitellum, and load transmission occurs via direct axial loading of the radius (Markholf, Lamey, Yang, Meals, & Hotchkiss, 1998). In the varus position, contact between the radial head and capitellum is absent, and the load is transferred from the radius to the ulna through the interosseous membrane (Markholf et al., 1998).

Markholf et al. (1998) determined the mean forces with the forearm in neutral rotation. They concluded that with the elbow in the valgus position, the mean force in the proximal end of the ulna was approximately 11.8% of the load applied at the wrist. In the varus position, the force averaged 93% (Markholf et al., 1998). These findings are important to consider when designing a rehabilitation regimen.

1.3   Lateral Collateral Ligament Injuries

The LCL is a critical constraint against external and varus rotation, thus disruption of this ligament will often lead to instability (Olsen, Michael, Søjbjerg, Helmig, & Sneppen, 1966). LCL injuries most commonly occur as the result of an elbow dislocation and secondarily can result from iatrogenic injury (Babhulkar, 2015; Osbourne & Cotterill, 1966; Reichel, Milam, Sitton, Curry, & Mehlhoff, 2013).

1.3.1 Pathology of Elbow Dislocations

The elbow joint is the second most frequently dislocated major joint in the adult population and occurs in approximately 5.21 per 100 000 people, annually (Kuhn & Ross, 2008; Stoneback et al., 2012). Most commonly, these injuries are the result of a fall on an outstretched arm (Mehlhoff, Noble, Bennett, & Tullos, 1988; Osbourne & Cotterill, 1966). Elbow dislocations may be simple or complex, characterized by the absence or presence of fractures, respectively (Hildebrand, Patterson, & King, 1999). One study found that 58% of simple dislocations occur on the non-dominant side (Josefsson, Johnell, & Gentz, 1984). Simple dislocations are generally described according to the direction that the ulna moves in relation to the proximal humerus with approximately 90% dislocating posteriorly or posterolaterally (Cohen & Hastings II, 1998; Morrey & Sanchez-Sotelo, 2009; Shillingford & Levine, 2015). Simple dislocations will disrupt both the LCL and MCL (Armstrong, 2016; Bell, 2008; Josefsson et al., 1984).

O'Driscoll et al. (2000) described three stages of acute elbow instability. Stage 1 is characterized by a partial or complete disruption of the LUCL, causing a posterolateral rotary subluxation. This stage causes posterolateral rotatory instability (PLRI) (O'Driscoll et al., 2000). In stage 2, further anterior and posterior disruption results in an incomplete posterolateral dislocation. Stage 3 may be further divided into three sections. In stage 3A, the posterior section of the MCL is disrupted and only the AMCL is left intact. In this situation, the elbow is stable to valgus stress and pronation of the forearm will maintain stability (O'Driscoll et al., 2000). In stage 3B, the AMCL is also disrupted, causing instability in varus, valgus and posterolateral rotations (O'Driscoll et al., 2000). In stage 3C, all of the soft tissues of the distal humerus are compromised causing severe

instability (O'Driscoll et al., 2000). More recently, Rhyou & Kim (2012) proposed a different mechanism of elbow dislocation that contrasts O'Driscoll. They suggest that simple posterior lateral elbow dislocations begin on the medial side (Rhyou & Kim, 2012).

Posterolateral rotatory instability (PLRI) is the most common type of recurrent instability after a simple elbow dislocation (Reichel et al., 2013). PLRI occurs after external rotation of the radius and ulna relative to the distal humerus, which causes the radial head to shift posteriorly relative to the capitellum (Charalambous & Stanley, 2008). Patients often experience symptoms including elbow pain, locking, clicking, snapping or recurrent dislocations during supination, extension and/or valgus force (Clitherow, McGuire, & Bain, 2014; Reichel et al., 2013).

There are several methods used to diagnose PLRI including the "drawer sign", the "chair sign", and the "push up sign" (Arvind & Hargreaves, 2006; Regan & Lapner, 2006; Reichel et al., 2013), however the most common is the lateral pivot shift test (or PLRI test) (O'Driscoll, Bell, & Morrey, 1991). Often under general anesthesia, the patient is placed on their back with the arm in the overhead position and the forearm supinated. The patient's arm is placed in full extension and then supination, and valgus and axial compression forces are applied. The elbow is then flexed and reduction of the elbow subluxation occurs at about 40° of flexion and may be accompanied by a large clunk (Morrey & Sanchez-Sotelo, 2009; O'Driscoll et al., 1991).

1.3.2 Management

Simple dislocations are usually managed non-operatively with a closed reduction under general anesthesia (Clitherow et al., 2014; Josefsson, Johnell, & Wendeberg, 1987b; Mica, Caekebeke, & Riet, 2016; O'Driscoll, Morrey, Korinek, & An, 1992; Ross, McDevitt, Chronister, & Ove, 1999; Shillingford & Levine, 2015; Szekeres, Chinchalkar, & King, 2008). Following reduction, there are several treatment options based on the stability of the elbow: immobilization in a static plaster, surgical repair of the MCL and/or LCL, or the most common, early motion with or without a sling, hinged brace or functional plaster (de Haan, Schep, Tuinebreijer, Patka, & den Hartog, 2010). A

rehabilitation regimen is also initiated to minimize the effects of immobilization, promote healing, recover ROM and muscle flexibility, and co-ordinate with the phases of histological healing (Fusaro, Orsini, Szorza, Rotini, & Benedetti, 2014; Wilk, Arrigo, & Andrews, 1993).

1.3.2.1   Non-Operative Management

Following a closed reduction, the elbow is immobilized in a splint at 90° for 5 to 10 days, then ROM exercises are initiated (Mehlhoff et al., 1988; Shillingford & Levine, 2015). LCL injuries are immobilized in pronation whereas MCL injuries are placed in supination (Fusaro et al., 2014). If there is injury to the MCL and LCL , the forearm is immobilized in the neutral position (Fusaro et al., 2014). Immobilization periods greater than 2 weeks are not recommended and have been shown to correlate with greater limitation of motion (Mehlhoff et al., 1988; Schippinger, Seibert, Steinböck, & Kucharczyk, 1999). Early motion helps to prevent elbow stiffness (de Haan et al., 2010; Maripuri, Debnath, Rao, & Mohanty, 2007; Mehlhoff et al., 1988; Szekeres et al., 2008). Ross et al. (1999) examined patients treated with immediate active range of motion and found that they achieved a final range of motion of -4° to 139° within an average of 19 days following reduction. Passive motion is avoided early on to reduce the risk of swelling, pain inhibition and ectopic ossification and is usually introduced at week 6 (Cohen & Hastings II, 1998; Dunning, Zarzour, Patterson, Johnson, & King, 2001c; Szekeres et al., 2008).

If instability is present when the elbow is extended, the forearm is placed in pronation and extension greater than 60° is avoided but flexion is permitted (Clitherow et al., 2014; Hunt, 2016; O'Driscoll et al., 2000; Szekeres et al., 2008). Studies have found that active motion and supination provide greater stability for the MCL-deficient elbow and active motion and pronation stabilize the LCL-deficient elbow (Alolabi et al., 2012; Armstrong et al., 2000; Dunning et al., 2001c; Fraser et al., 2008; Pichora et al., 2007). After a simple dislocation, injury to the lateral side of the elbow is typically worse than the medial side because the common extensor origin is usually disrupted whereas the common flexor origin is left intact (Alolabi et al., 2012; McKee, Schemitsch, Sala, & O'Driscoll, 2003; O'Driscoll et al., 1992). Thus, pronation is more common to maintain stability (Alolabi et al., 2012). If pronation restores stability a hinged elbow orthosis

(HEO) is recommended by some to be worn for 6 weeks with the arm in pronation (Hunt, 2016; O'Driscoll et al., 2000).

The humerus can be positioned in the gravity overhead, vertical dependent, horizontal, varus or valgus positions during rehabilitation (Figure 1-14). When LCL instability is present, ROM exercises in the overhead position are often recommended as it is thought to minimize the effects of gravity (Wolff & Hotchkiss, 2006). The vertical dependent arm position is also suggested, whereas the varus position should be avoided (Dunning et al., 2001c; Manocha, 2016). When both the LCL and MCL are compromised, active motion with the arm in the horizontal and vertical dependent positions provides the most stability, whereas the varus and valgus positions are avoided (Alolabi et al., 2012).

Although surgical repair is often not needed for a simple elbow dislocation, Mehlhoff et al. (1988) found that 60 percent of patients treated with only a closed reduction reported long-term symptoms. The patients had a mean loss of extension and flexion of 12.3° and 6.3°, respectively. Additionally, 15% of patients reported more than 30° of flexion contracture, 45% described residual pain, and 35% experienced pain following valgus stress. Josefsson et al. (1987a) found that there was no significant difference between non-operative management and surgical repair after a simple elbow dislocation suggesting that regardless of the treatment approach, residual symptoms are common.
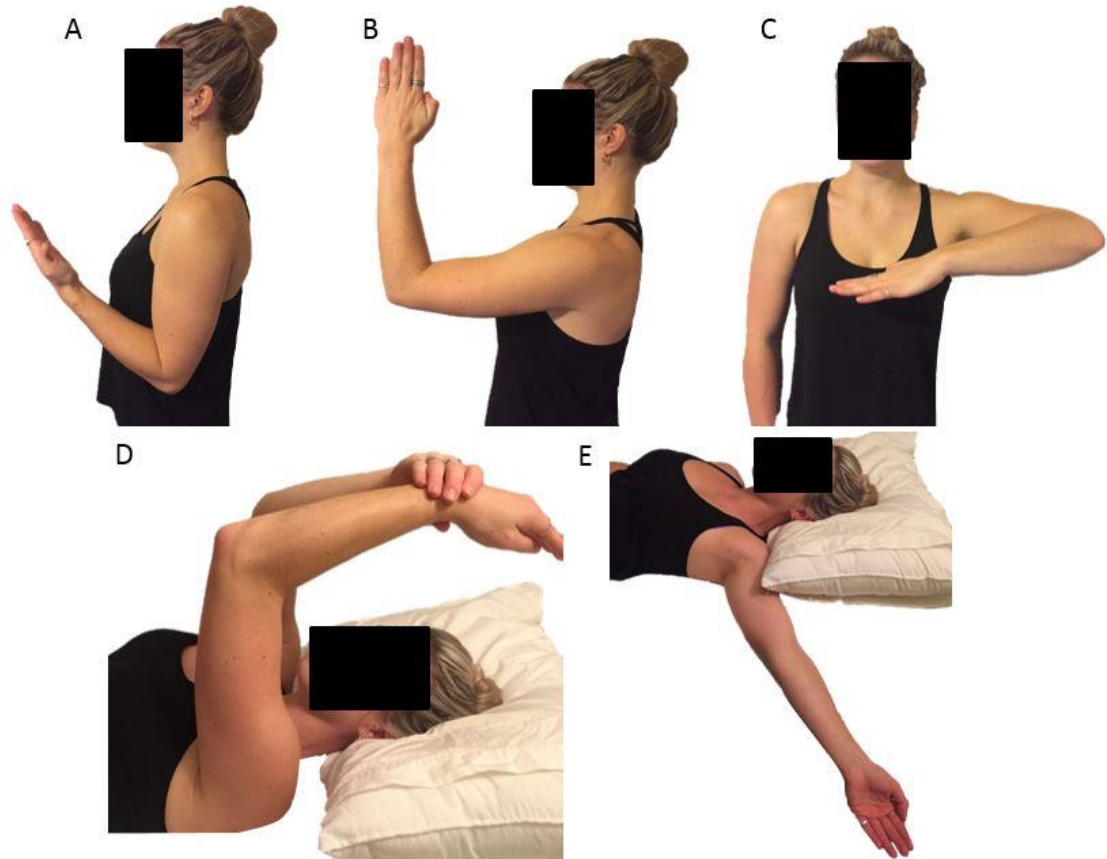
Figure 1-14: Humeral Rehabilitation Positions. *Following elbow injury, the arm may be positioned in the following gravity-loaded positions: (A) vertical dependent, (B) horizontal, (C) varus, (D) vertical overhead, and (E) valgus.*

1.3.2.2 Operative Management

Patients affected by recurrent dislocations, or who have persistent instability following a closed reduction may have surgery to repair the LCL and/or MCL (Charalambous & Stanley, 2008; Mica et al., 2016; Osbourne & Cotterill, 1966; Sheps, Hildebrand, & Boorman, 2004). If the elbow has remained dislocated for more than 2 weeks after the time of injury, an open reduction with ligamentous repair or reconstruction may be required (Sheps et al., 2004).

The LCL may be repaired or reconstructed, however in acute cases, a repair is often sufficient as long as the ligament is of sufficient quality (Morrey & Sanchez-Sotelo, 2009; Sheps et al., 2004; Smith, Savoie, & Field, 2001). If a repair is not possible, surgical reconstruction is performed with an autogenous or allograft tendon graft (Sheps et al., 2004). Sanchez-Sotelo et al. (2005) found that LCL repair or reconstruction improved or eliminated stability in 89% of their patients. If instability persists after LCL repair, the MCL may be repaired as well (Armstrong, 2016). Heo et al. (2015) concluded that after a simple elbow dislocation, an LCL repair alone is often sufficient to obtain stability. This may be because while damage occurs to the ligaments on both sides, the LCL and common extensor origin often incurs more damage than the medial structures (Osbourne & Cotterill, 1966).

Many authors have reported LCL reconstruction techniques (Nestor, O'Driscoll, & Morrey, 1992; Olsen & Søjbjerg, 2003; Sanchez-Sotelo, Morrey, & O'Driscoll, 2005; Tawari, Lawrence, & Stanley, 2013) however only repair techniques will be discussed herein. If an LCL repair is possible, the Kocher approach is usually utilized. This lateral approach uses the interval between the anconeus and extensor carpi ulnaris, which is directly over the LUCL (Antuña & Barco, 2014; Armstrong, 2016; Hunt, 2016). Bone tunnels or suture anchors are then used to repair the LCL (Wiesel, 2015). Using a humeral bone tunnel, non-absorbable sutures are used to attach the LCL to the isometric point of insertion on the humerus, where avulsion is most common (Armstrong, 2016; G. Fraser et al., 2008; Hunt, 2016; Josefsson, et al., 1987a; McKee et al., 2003; Mimic, Kim, Park, Kim, & Jeon, 2009; Nestor et al., 1992; Sheps et al., 2004; Wiesel, 2015). This point is located in line with the axis of rotation of the elbow, at the lateral epicondyle

(Armstrong, 2016). In a transosseous repair method described by Fraser et al. (2008), the suture is passed through the tunnel at the center of curvature of the capitellum to the lateral supracondylar ridge of the humerus. They investigated the effects of repair tensions of 20, 40 and 60 N and recommended an LCL repair tension of 20 N or less to reestablish the normal kinematics of the elbow. Suture anchors can also be used but carefully controlled ligament tensioning should be performed and elbow stability checked before locking the anchor (Lee, Eng, Keogn, McLean, & Bain, 2012).

If the elbow remains unstable following ligament repair or reconstruction, a hinged external fixator with pins inserted into the humerus and ulna may be applied to maintain reduction while allowing joint motion (Cohen & Hastings II, 1998; Hildebrand et al., 1999; Sheps et al., 2004). If the elbow is stable after the repair, a HEO with or without an extension block is recommended by some for 6 weeks (Sheps et al., 2004).

1.3.3 Orthoses

Elbow orthoses or braces may be ready-made or individually customized to the patient (Beam, 2011; Fusaro et al., 2014). Customized braces are often made of thermoplastic materials with thicknesses ranging from 2 to 5 mm and are often more expensive (Beam, 2011; Fusaro et al., 2014). Low-temperature thermoplastic braces are advantageous because they are lightweight and can be fabricated soon after injury, however they often cannot provide adequate stiffness (Lunsford & DiBello, 2008). Ready-made braces are often designed with an aluminum frame, and silicone or neoprene padding for patient comfort (Fusaro et al., 2014). While stainless steel is extremely strong, it is better suited for lower limb orthosis due to its heavy weight. Aluminum is less strong and stiff than stainless steel but is also approximately one-third of the weight (May & Lockard, 2011). Hook and loop fastener straps or press-studs are used to secure the joint and minimize movement, (Fusaro et al., 2014).

Based on their intent, orthoses can be classified into immobilization (or supportive), mobilization (or corrective) or restriction (or protective) braces (Jacobs, 2003; Lunsford & DiBello, 2008). In the initial period post-injury or post-surgery of the LCL the patient is usually placed in an immobilization brace at 90° to 120° of flexion for 3 to 5 days with

the forearm pronated (Fusaro et al., 2014; Jacobs, 2003; Rotini & Marinelli, 2008; Szekeres et al., 2008). The goal of this brace is to reduce edema and pain, and promote healing (Fusaro et al., 2014).

After 1 to 7 days, restrictive braces may be used to limit motion if residual instability is present following LCL injury or repair (Cohen & Hastings II, 1998; Jacobs, 2003; Szekeres et al., 2008). These braces are designed with adjustable ROM in flexion/extension and/or pronation/supination (Fusaro et al., 2014; Jacobs, 2003). Generally, extension is limited after a posterior elbow dislocation by a hinged elbow orthosis (HEO) (Jacobs, 2003). The hinge can be locked to limit extension but permit full flexion while the extension limit is gradually reduced over a period of 4 to 6 weeks (Jacobs, 2003; Wolff & Hotchkiss, 2006).

1.3.3.1   Hinged Elbow Orthoses (HEOs)

There are many components that must be considered in the design of a HEO. They may have a hinge on the medial side alone, or on both sides of the elbow. The hinges may also be exposed or protected in a sleeve (Griffin, Kercher, & Shoop, 2008). Most HEOs use a four-point pressure system with two forces provided by the two straps on the upper arm and two as a result of the straps on the forearm (Lunsford & Contoyannis, 2008). The orthosis operates as a three-point lever system with one middle force acting in the opposite direction to two other forces, proximal and distal to the middle force (Edelstein & Bruckner, 2002; Fess, 1995). The upper arm and forearm act as the proximal and distal lever arms, with the hinge providing the opposing force and acting as the fulcrum (Fess, 1995; Lunsford & Contoyannis, 2008).

Patient comfort must also be taken into account when designing an orthosis (Fess, 1995). Placement of the elbow cuffs closer to the hinge can maximize the lever system and increase skin comfort for the patient (Griffin et al., 2008). The orthosis may also be designed with longer longitudinal arms to increase the surface area of force applied to the arm, which can decrease the pressure at either end, thus increasing patient comfort (Edelstein & Bruckner, 2002; Fess, 1995).

HEO's often preferred by surgeons are the Bledsoe Brace (Bledsoe Brace Systems, Grand Prairie, TX) or the Mayo Clinic Universal Elbow Brace (Aircast, Summit, NJ) (Wolff & Hotchkiss, 2006). The Bledsoe Hinged Elbow Brace can be used following a dislocation to prevent hyperextension. It features cross straps to control elbow extension as well as dual removable hinges (Figure 1-15). Bledsoe also offers hinged elbow braces for use specifically following injury or surgery. The three post-op braces, the T-Chek Brace, Telescoping Elbow Brace, and Extender Arm Brace, feature a single hinge to limit range of motion as well as large cuffs to improve patient comfort. The Mayo Clinic Universal Elbow brace is another HEO that features a mechanical hinge and a control knob to control range of motion using an allen key (Figure 1-16). The brace has four longitudinal metal arms connected by four arm straps (2 on the forearm and 2 on the upper arm).

There is limited research on the efficacy of elbow braces following ligamentous injury. One study using the Mayo Clinic Universal Elbow Brace found that following injury to the LCL, an HEO did not significantly affect elbow stability during active motion and interestingly, was more harmful during passive motion (Manocha, 2016). Another study utilized the Bledsoe Telescoping Elbow Brace to quantify the efficacy of HEO's following LCL injury. During passive motion in the dependent orientation, it was found that the brace caused nearly double the ulnohumeral distraction compared to the intact elbow, however it was not statistically significant (Lee et al., 2013).

Figure 1-15: Bledsoe Hinged Elbow Orthoses. *Bledsoe's hinged elbow orthoses include (A) Bledsoe Hinged Elbow Brace, (B) Bledsoe T-Chek Post-Op Elbow Brace (C) Bledsoe Telescoping Elbow Brace and (D) Bledsoe Extender Arm Post-Op Elbow Brace (Bledsoe Brace Systems, Grand Prairie, TX).*

Figure 1-16: Mayo Clinic Universal Elbow Brace. *Mayo Clinic's hinged elbow orthosis (HEO) is often used following elbow trauma (Aircast, Summit, NJ) (DJO Canada, 2016).*

1.4 *In Vitro* Assessment of the Elbow

Modeling *in vivo* elbow motion is difficult to achieve but it is an important tool to gain a better understanding of elbow physiology, disorders and injuries. Elbow simulation *in vitro* is often achieved with the arm in one of four positions: vertical, horizontal, varus and valgus positions (Ferreira, Johnson, & King, 2010).

1.4.1 Motion Simulators

In vitro elbow simulation models motion pathways and can provide a better understanding of elbow kinematics and joint stability following injury (Dunning, Gordon, King, & Johnson, 2003). In the intact state, *in vitro* active simulation has been shown to more accurately represent *in vivo* elbow motion and provide more stability than passive motion (Dunning, Duck, King, & Johnson, 2001a; Johnson, Rath, Dunning, Roth, & King, 2000). Additionally, as discussed in Section 1.3.3.1, elbow dislocations are frequently rehabilitated actively for the first six weeks post-injury, while passive motion is avoided (Alolabi et al., 2012; Armstrong et al., 2000; Cohen & Hastings II, 1998; Dunning et al., 2001c). Thus, active motion would better model the early stages of rehabilitation and provide a greater understanding of the *in vivo* state.

Active elbow motion simulators will often control muscles using pneumatic, hydraulic or electromechanical actuators, or by motor control (Ferreira, 2011; Magnusen, 2004; Schimoler, 2008; Tanaka et al., 1998). The five main muscles of elbow motion most often controlled during simulation are the biceps, triceps, brachialis, brachioradialis and pronator teres (Lazar et al., 2015).

At the Roth McFarlane Hand and Upper Limb Centre (HULC) in London Ontario an *in vitro* active elbow motion simulator was developed that will be used for this thesis (Figure 1-17). The simulator was first described in 1997 and featured load-controlled motion by connecting the five main elbow muscles to pneumatic actuators via stainless steel cables (Rath, 1997). The muscle loads were proportionally calculated using electromyography (EMG) data and cross sectional areas (CSA) (Amis, Dowson, & Wright, 1979; Funk, An, Morrey, & Daube, 1987). Repeatable active motion in the vertical position was achieved and passive motion could also be evaluated in the varus

and valgus positions (Dunning et al., 2001a). The simulator was later modified to incorporate displacement-controlled motion into the load-controlled system (Dunning et al., 2003). The brachialis, designated as the prime mover, was position-controlled by moving the piston of the actuator at a constant velocity. The remaining muscles were load-controlled based on the muscle-loading ratio obtained from EMG and CSA data (Amis et al., 1979; Dunning et al., 2003; Funk et al., 1987). This version provided highly repeatable results in the vertical, varus and valgus orientations, however varus and valgus positions were achieved with greater difficulty (Dunning et al., 2003). The simulator was again modified to simulate flexion/extension in the vertical, varus, valgus and horizontal positions (Ferreira, 2011). Servo-motors were incorporated to control the biceps, brachialis and triceps and strain-gauge instrumented motor mounts provided load feedback. For each arm position and forearm position (supination, neutral and pronation) a muscle was designated as the prime mover. The remaining muscles were connected to pneumatic actuators. To negate the effects of gravity, wrist extensors (extensor carpi ulnaris and radialis longus) and flexors (flexor carpi ulnaris and radialis) were also actuator-controlled (Ferreira et al., 2010). Most recently, the simulator was adapted to include motion in the vertical overhead position (Kusins, 2015).
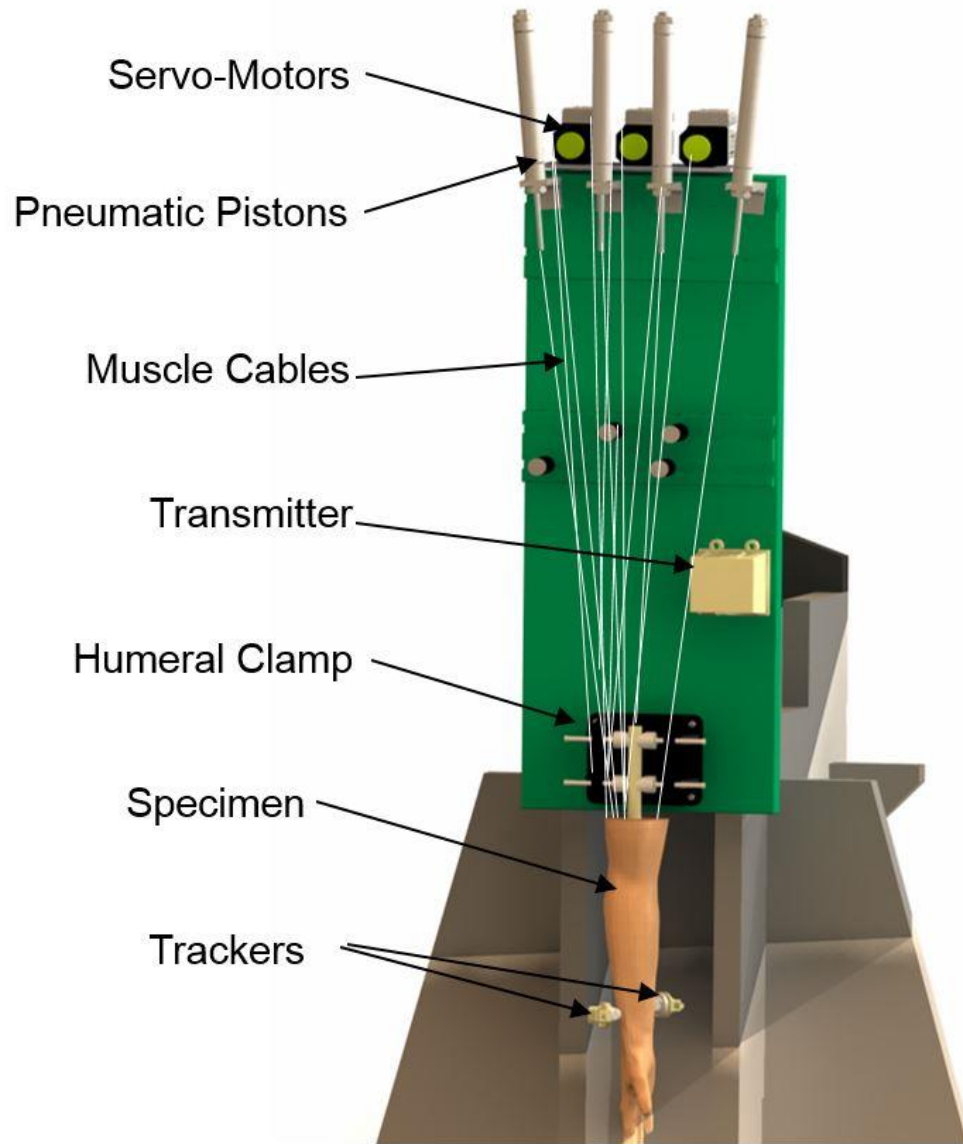
Figure 1-17: Active Elbow Motion Simulator. *Elbow motion simulator designed at the HULC to simulate active elbow flexion-extension using a combination of computer controlled servomotors and pneumatic actuators. Right arm shown in the vertical dependent arm position (Manocha, 2016).*

1.4.2 Motion Tracking

Real-time kinematic modeling can be accomplished using tracking systems to determine both the position and orientation of an object. The two main systems used for kinematic assessment are optical and electromagnetic tracking (Eckhouse, Penny, & Maulucci, 1996). While optical trackers are very precise, a direct line of sight is required from the camera to the tracker. Electromagnetic systems are also very precise however interference may be caused by ferromagnetic or electrical devices nearby (Craig, 2013; Poulin & Amoit, 2002). For the purpose of this thesis, the rotation the simulator into different arm position and the HEO mounted onto the cadaveric arm were expected to obstruct the line of sight, thus electromagnetic tracking was chosen for this application.

Electromagnetic tracking systems feature a transmitter with three orthogonal antennas located in a fixed position that generates a magnetic field by alternating current (AC) or directed current (DC) (Birkfellner, Hummel, Wilson, & Cleary, 2008; Craig, 2013). The signal acquired by one or more receiving units (sensors), also with three orthogonal antennas, is analyzed to determine the sensor's location and orientation (Craig, 2013). These systems can track motion in six degrees of freedom (Craig, 2013).

A popular electromagnetic tracking system used to model *in vitro* motion of the elbow following ligamentous injury is the Flock of Birds® (Ascension Technology Corporation, Burlington, VT) technology, which will be referred to and used later on in this thesis (Figure 1-18) (Armstrong et al., 2000; Dunning, Zarzour, Patterson, Johnson, & King, 2001b; Dunning et al., 2001c; Fraser et al., 2008; Johnson et al., 2000; Pichora et al., 2007). This system uses pulsed DC and can track up to four sensors. DC-based systems are better suited than an AC-based system when conductive metals such as stainless steel and aluminum are present. A DC-based system can minimize distortions caused by metals by turning the magnetic field on and off at certain frequencies, whereas AC-based systems emit a continuous magnetic field (Birkfellner et al., 2008). The manufacturer for the Flock of Birds® tracking system specifies static position and angular accuracy of 1.8 mm root mean square (RMS) and 0.5 RMS, respectively (Ascension, 2000).
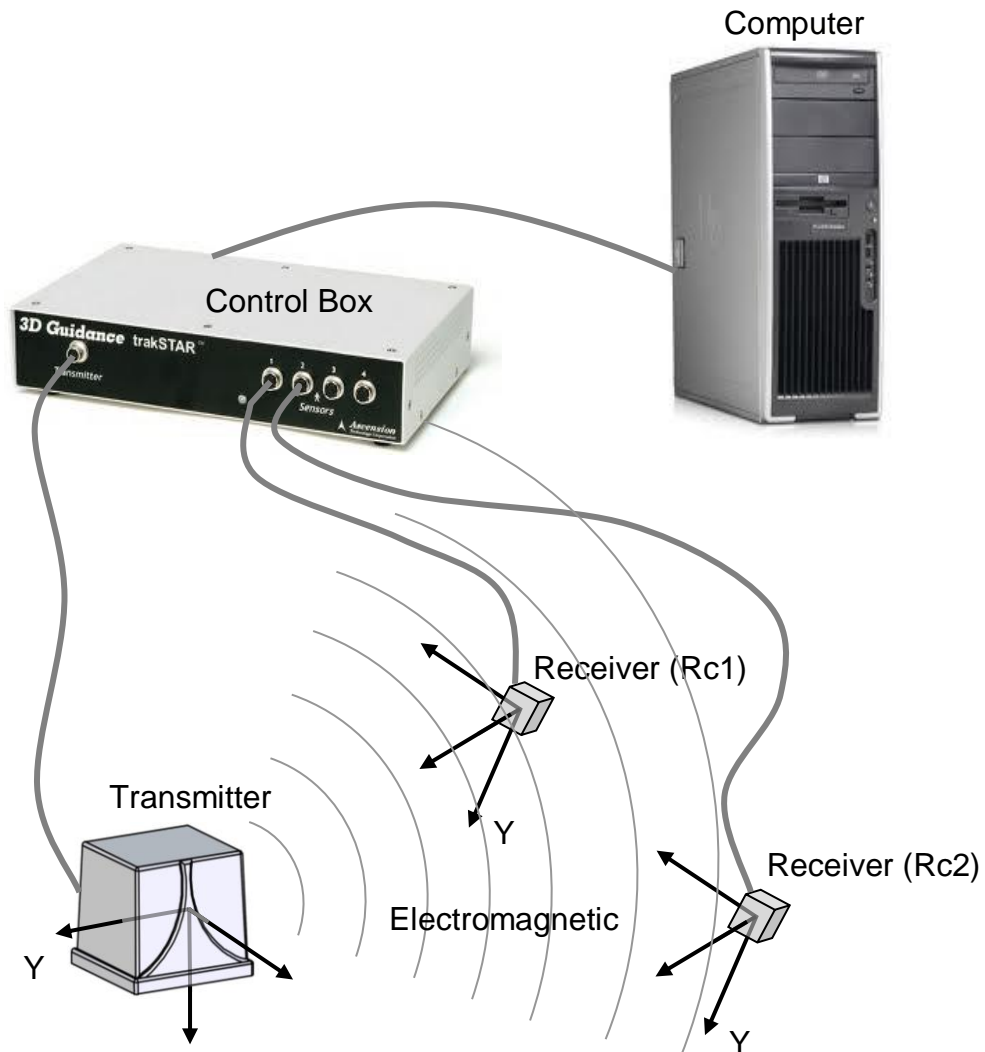
Figure 1-18: Flock of Birds® Electromagnetic Tracking System. *The electromagnetic tracking system designed by Ascension Technologies Inc., (Burlington, VT) features a fixed transmiitter (Tr) that emits an electromagnetic field from three orthogonal coils. The field induces currents in the receiver's antennae (RC1 and RC2) . The control box interprets the signals and outputs the magnitudes as positions and rotations of each receiver relative to the transmitter (Ferreira, 2011).*

Studies using the simulator designed at the HULC have often used the Flock of Birds® tracking system to investigate elbow joint kinematics (Armstrong et al., 2000; Dunning et al., 2001a; Dunning et al., 2001c; Johnson et al., 2000) The transmitter is fixed on the simulator with respect to the humerus and a receiver is fixed to the ulna. Following ligamentous injury, the kinematic data can be used to compare the varus-valgus angulation (VVA) and internal-external rotation (IER) of the intact elbow to its injured state, to quantify stability (Figure 1-19) (Alolabi et al., 2012; Dunning et al., 2001a; Dunning et al., 2001c; Fraser et al., 2008; Johnson et al., 2000).
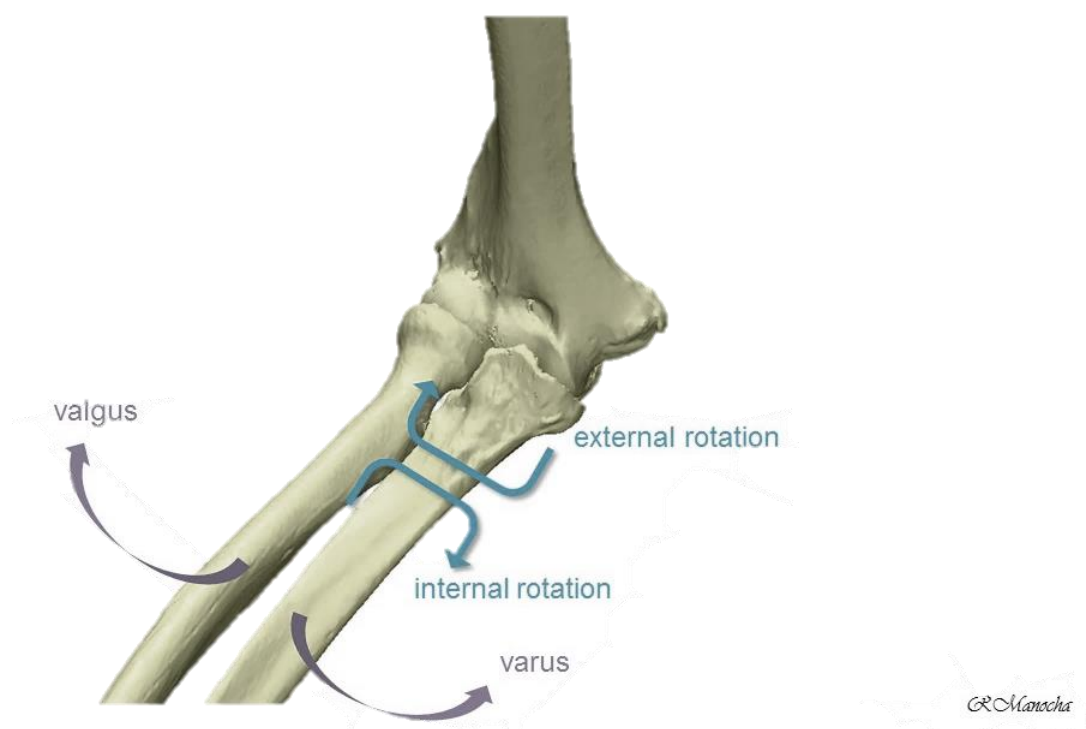
Figure 1-19: Kinematic References of the Elbow. *Kinematics of the elbow are often used to measure elbow stability. These include varus and valgus motion of the ulna with respect to humerus in the coronal plane as well as internal and external rotation of the ulna about its long axis (Manocha, 2016).*

1.4.3 Measurement of Ligament Tension

Knowledge of ligament loading following ligamentous injury and repair is imperative to optimize treatment. Many techniques have been developed for load measurement in the lower extremity, however there is minimal research reporting ligament tension in the elbow. There are a variety of devices for direct ligament measurement including buckle transducers and load cells.

Buckle transducers are commonly used *in vivo* to determine ligament load through deformation and therefore strain. Under axial loading, an increase in ligament tension causes deformation in a transverse beam instrumented with strain gauges (An, 2003). A previously mentioned study inserted a buckle load transducer into the AMCL (Fay, Lalone, Ferreira, Johnson, & King, 2010).

Load cells are electrical transducers that are a feasible load measurement option *in vitro*. An electrical signal is generated in response to an applied force. Strain gauge load cells are the most common. The typical configuration includes a Wheatstone bridge, which has four strain gauges that elastically deform when subjected to a force (Elbestawi, 2014). The Subminiature Model 11 load cell (Honeywell, Golden Valley, MN, USA), with a 100 lb capacity, capable of measuring forces in tension and compression will be utilized for this thesis (Figure 1-20).
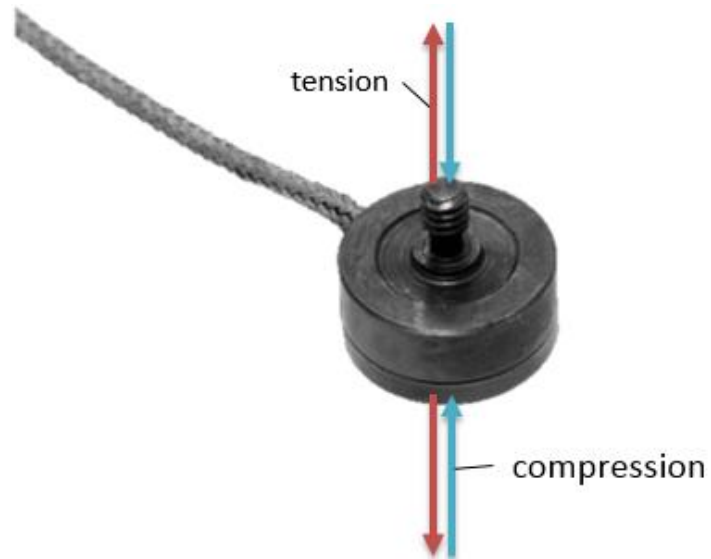
Figure 1-20: Subminiature Model 11 Load Cell (Honeywell, Golden Valley, MN, USA). *A one degree of freedom (DOF) load cell capable of measuring loads in compression (blue) and tension (red).*

1.5 Thesis Rationale

The elbow is the second most commonly dislocated joint among adults and frequently results in injury to the LCL (Babhulkar, 2015; Osbourne & Cotterill, 1966; Reichel, Milam, Sitton, Curry, & Mehlhoff, 2013; Kuhn & Ross, 2008). Orthoses are regularly prescribed to patients following LCL injury or repair, however the effects on elbow stability have not been sufficiently investigated. Interestingly, one study found that hinged elbow orthoses did not significantly improve stability in most situations, and worsened instability in certain arm positions (Manocha, 2016). In another study, only passive motion in the vertical dependent position was investigated and found no significant differences (Lee et al., 2013). It is important for clinicians to understand if HEOs are, in fact, improving stability or if they are increasing stress and preventing optimal healing.

The elbow has a native carrying angle of approximately 11-14° in men, and 13-16° in women (Atkinson & Elftman, 1945; Morrey & Sanchez-Sotelo, 2009; Paraskevas et al., 2004; Terra et al., 2011; van Roy, Baeyens, Fauvart, Lanssiers, & Clarijs, 2005), however, the orthoses that currently exist feature a straight hinge and do not account for this valgus angulation. It is important to investigate the impacts of changing this angle through the design of an adjustable hinged elbow orthosis. Understanding the optimal varus-valgus angle for an elbow orthosis may influence future designs and manufacturing as well as clinical practices.

Tension in the MCL has been investigated, however, due to its complex anatomical structure, LCL loads are more difficult to examine *in vitro*. The biomechanical implications of the load changes and patterns of the LCL, with and without an orthosis, could be applied to clinical interventions. In combination with the kinematic investigation presented in this thesis, the load data could improve rehabilitation regimens and provide useful information when developing better LCL repair techniques and reconstructions.

1.6   Objectives & Hypotheses

The specific objectives of this these are as follows:

1. To design and construct an adjustable hinged elbow orthosis for examining *in vitro* kinematics and biomechanics of the elbow following LCL injury;

2. To determine the effects changing the valgus angulation of the hinged elbow orthosis on elbow kinematics during simulated LCL injury;

3. To quantify the LCL load throughout simulated active and passive elbow flexion-extension with the arm in dependent and varus orientations and with the forearm pronated and supinated.

The specific hypothesis of this work are as follows:

1. Active motion and pronation will provide more elbow stability than passive motion and supination in the vertical dependent and varus arm positions.

2. The orthosis will provide more stability following an LCL injury or repair in varus and dependent positions.

3. Increasing the valgus angle of the orthosis will provide more stability in elbow extension, however at higher valgus angles it will cause greater instability with the elbow in flexion.

4. The orthosis will decrease the load through the LCL and will further decrease the load as the valgus angulation of the orthosis is increased.

1.7 Thesis Overview

Chapter 2 describes the design and development of a variable hinged elbow orthosis. The orthosis features an adjustable length and width as well as varus and valgus angle.

Chapter 3 presents an *in vitro* kinematic study to determine the effectiveness of a hinged elbow orthosis following LCL injury. LCL injury is simulated and the orthosis is tested at 0°, 5°, 10°, 15° and 20° of valgus angulation. The arm is placed in the dependent and varus positions, with the forearm pronated and supinated. In each position, simulated active and passive flexion is conducted.

Chapter 4 investigates the effects of an orthosis on the *in vitro* LCL loads during flexion. An LCL repair at 20 N is simulated. The loads are evaluated during simulated active and passive motion with the arm in dependent and varus positions and with the forearm both pronated and supinated.

Chapter 5 provides an overview and discussion of the work presented. The potential impacts for rehabilitation following LCL injury and future directions are also described.

1.8 References

Alolabi, B., Gray, A., Ferreira, L. M., Johnson, J. A., Athwal, G. S., & King, G. J. W. (2012). Rehabilitation of the medial- and lateral collateral ligament-deficient elbow: An in vitro biomechanical study. *Journal of Hand Therapy*, *25*(4), 363–373. https://doi.org/10.1016/j.jht.2012.06.001

Amis, A. A., Dowson, D., & Wright, V. (1979). Muscle strengths and musculo-skeletal geometry of the upper limb. *Engineering in Medicine*, *8*(1), 41–48.

An, K. N. (2003). In Vivo Force and Strain of Tendon, Ligament and Capsule. In F. Guilak, D. L. Butler, S. A. Goldstein, & D. J. Mooney (Eds.), *Functional Tissue Engineering* (pp. 96–106). New York, NY: Springer-Verlag New York Inc.

An, K. N., Kwak, B. M., Chao, E. Y., & Morrey, B. F. (1984). Determination of muscle and joint forces: A new technique to solve the indeterminate problem. *J Biomech Eng*, *106*(4), 364–367.

Antuña, S., & Barco, R. (2014). *Essentials in Elbow Surgery: A Comprehensive Approach to Common Elbow Disorders* (illustrate). Springer Science and Business Media.

Armstrong, A. (2016). Simple Elbow Dislocation. In G. S. Athwal (Ed.), *Management of elbow trauma, an issue of hand clinics 31-4, E-Book* (pp. 521–531). Elsevier Health Sciences.

Armstrong, A. D., Dunning, C. E., Faber, K. J., Duck, T. R., Johnson, J. A., & King, G. J. W. (2000). Rehabilitation of the medial collateral ligament-deficient elbow: an in vitro biomechanical study. *The Journal of Hand Surgery*, *25*(6), 1051–1057.

Arvind, C. H., & Hargreaves, D. G. (2006). Tabletop relocation test: a new clinical test for poste- rolateral rotatory instability of the elbow. *J Shoulder Elbow Surg*, *15*, 707–8.

Ascension. (2000). Flock of Birds (R) Real-time Motion Tracking. Retrieved September

5, 2017, from http://www.razor3d.co.kr/pds/ascension/flockofbirds.pdf

Atkinson, W. B., & Elftman, H. (1945). The carrying angle of the human arm as a secondary sex character. *The Anatomical Record*, *91*(1), 49–52.

Azar, F. M., Canale, S. T., & Beaty, J. H. (2016). *Campbell's Operative Orthopaedics E-Book* (13th ed.). Elsevier Health Sciences.

Babhulkar, S. (2015). *Elbow injuries* (illustrate). New Delhi, India: JP Medical Ltd.

Beam, J. W. (2011). *Orthopedic taping, wrapping, bracing & padding*. Philadelphia, PA: F.A. Davis.

Bell, S. (2008). (iii) Elbow instability, mechanism and management. *Current Orthopaedics*, *22*(2), 90–103. https://doi.org/10.1016/j.cuor.2008.04.007

Birkfellner, W., Hummel, J., Wilson, E., & Cleary, K. (2008). Tracking Devices. In T. Peters & K. Cleary (Eds.), *Image-Guided Interventions: Technology and Applications* (pp. 23–44). Springer Science and Business Media.

Canale, S. T., & Beaty, J. H. (2012). *Campbell's Operative Orthopaedics: Sports Injuries of the Shoulder and Elbow E-Book* (12th ed.). Elsevier Health Sciences.

Charalambous, C. P., & Stanley, J. K. (2008). Posterolateral rotatory instability of the elbow. *Bone & Joint Journal*, *90*(3), 272–279.

Clitherow, H. D. S., McGuire, D. T., & Bain, G. I. (2014). Lateral elbow instability. *Asia-Pacific Journal of Sports Medicine, Arthroscopy, Rehabilitation and Technology*, *1*(1), 11–18.

Cohen, M. S., & Bruno, R. J. (2001). Anatomy and Clinical Correlation. *Clin Orthop Relat Res*, *383*, 123–130.

Cohen, M. S., & Hastings II, H. (1998). Acute Elbow Dislocation : Evaluation and Management, *6*(1), 15–23.

Craig, A. B. (2013). Augmented Reality Hardware. In *Understanding Augmented Reality: Concepts and Applications* (pp. 69–123). Waltham, MA: Elsevier Inc.

de Haan, J., Schep, N. W. L., Eygendaal, D., Kleinrensink, G.-J., Tuinebreijer, W. E., & den Hartog, D. (2011). Stability of the elbow joint: relevant anatomy and clinical implications of in vitro biomechanical studies. *The Open Orthopaedics Journal*, *5*, 168–176. https://doi.org/10.2174/1874325001105010168

de Haan, J., Schep, N. W. L., Tuinebreijer, W. E., Patka, P., & den Hartog, D. (2010). Simple elbow dislocations: a systematic review of the literature. *Arch Orthop Trauma Surg*, *130*(2), 241–249.

Deland, J. T., Garg, A., & Walker, P. S. (1987). Biomechanical basis for elbow hinge-distractor design. *Clinical Orthopaedics and Related Research*, *215*, 303–312.

Dines, J., & Altchek, D. (2015). *Elbow Ulnar Collateral Ligament Injury: A Guide to Diagnosis and Treatment* (Illustrate). Springer.

Dunning, C. E., Duck, T. R., King, G. J., & Johnson, J. (2001). Simulated active control produces repeatable motion pathways of the elbow in an in vitro testing system. *J Biomech*, *34*(8), 1039–48.

Dunning, C. E., Gordon, K. D., King, G. J. W., & Johnson, J. A. (2003). Development of a motion-controlled in vitro elbow testing system. *Journal of Orthopaedic Research*, *21*(3), 405–411. https://doi.org/10.1016/S0736-0266(02)00233-4

Dunning, C. E., Zarzour, Z. D., Patterson, S. D., Johnson, J. A., & King, G. J. (2001). Ligamentous stabilizers against posterolateral rotatory instability of the elbow. *JBJS*, *83*(12), 1823–1828.

Dunning, C. E., Zarzour, Z. D. S., Patterson, S. D., Johnson, J. A., & King, G. J. W. (2001). Muscle forces and pronation stabilize the lateral ligament deficient elbow. *Clin Orthop Relat Res*, *388*, 118–124.

Eckhouse, R. H., Penny, M. A., & Maulucci, R. A. (1996). A comparison of kinematic

recording instruments. *Journal of Medical Systems*, *20*(6), 439–456.

Edelstein, J. E., & Bruckner, J. (2002). Introduction to Orthotics. In *Orthotics: A comprehensive clinical approach* (illustrate). SLACK Incorporated.

Elbestawi, M. A. (2014). Force Measurement. In J. G. Webster & H. Eren (Eds.), *Measurement, Instrumentation, and Sensors Handbook* (Second, pp. 41-3-41–6). Boca Raton, Fl: CRC Press.

Fay, K. E., Lalone, E. A., Ferreira, L., Johnson, J., & King, G. J. . (2010). Quantification of Medial collateral ligament tension in the elbow. In *56th Annual Meeting of the Orthopaedic Research Society*.

Ferreira, L. M. (2011). *Development of an Active Elbow Motion Simulator and Coordinate Systems to Evaluate Kinematics in Multiple Positions*. The University of Western Ontario.

Ferreira, L. M., Johnson, J. A., & King, G. J. W. (2010). Development of an active elbow flexion simulator to evaluate joint kinematics with the humerus in the horizontal position. *J Biomech*, *43*(11), 2114–2119.

Fess, E. F. (1995). Splints: Mechanics versus convention. *Journal of Hand Therapy*, *8*(2), 124–130.

Fornalski, S., Gupta, R., & Lee, T. Q. (2003). Anatomy and biomechanics of the elbow joint. *Techniques in Hand and Upper Extremity Surgery*, *7*(4), 168–178. Retrieved from http://ovidsp.ovid.com/ovidweb.cgi?T=JS&CSC=Y&NEWS=N&PAGE=fulltext&D=emed11&AN=24327407

Fraser, G., Pichora, J., Ferreira, L., Brownhill, J., Johnson, J., & King, G. (2008). Lateral collateral ligament repair restores the initial varus stability of the elbow: an in vitro biomechanical study. *J Orthop Trauma*, *22*(9), 615–23.

Fraser, G. S., Pichora, J., Ferreira, L. M., Brownhill, J., Johnson, J. A., & King, G. J. W.

(2008). Lateral Collateral Ligament Repair Restores the Initial Varus Stability of the Elbow: An In Vitro Biomechanical Study. *Journal of Orthopaedic Trauma*, *22*(9), 615–623.

Funk, D. A., An, K. N., Morrey, B. F., & Daube, J. R. (1987). Electromyographic analysis of muscles across the elbow joint. *Journal of Orthopaedic Research*, *5*, 529–538.

Fusaro, I., Orsini, S., Szorza, T., Rotini, R., & Benedetti, M. G. (2014). The use of braces in the rehabilitation treatment of the post-traumatic elbow. *Joints*, *2*(4), 81–86.

Fuss, F. K. (1991). The ulnar collateral ligament of the human elbow joint. Anatomy, function and biomechanics. *Journal of Anatomy*, *175*, 203–12. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/2050566

Griffin, L. Y., Kercher, J., & Shoop, J. L. (2008). Protective equipment of the upper limb in sport. In J. D. Hsu, J. W. Michael, & J. R. Fisk (Eds.), *AAOS Atlas of Orthoses and Assistive Devices* (Fourth, pp. 271–288). Philadelphia, PA: Mosby Inc.

Heo, Y. M., Yi, J. W., Lee, J. B., Lee, D. H., Park, W. K., & Kim, S. J. (2015). Unstable simple elbow dislocation treated with the repair of lateral collateral ligament complex. *Clin Orthop Surg*, *7*(2), 241–247.

Hildebrand, K. A., Patterson, S. D., & King, G. J. W. (1999). Acute elbow dislocations: simple and complex. *Orthop Clin North Am*, *30*, 63–79.

Hunt, T. R. (2016). *Operative Techniques in Hand, Wrist, and Elbow Surgery* (2nd ed.). Lippincott Williams & Wilkins.

Ilbeigi, S., & Ramezani, H. (2014). Assessment and modelling of elbow joint for analysing of muscle moment and reaction force during flexion movement with ADAMS software. *European Journal of Sports and Exercise Science*, *3*(1), 18–26.

Jacobs, M. A. (2003). Splint Classification. In M. A. Jacobs & N. M. Austin (Eds.), *Splinting the Hand and Upper Extremity: Principles and Process* (illustrate, pp. 2–

18). Lippincott Williams & Wilkins.

Jacobson, J. A. (Jon A. (2007). *Fundamentals of musculoskeletal ultrasound* (1st ed.). Philadelphia, PA: Saunders/Elsevier. Retrieved from https://books.google.ca/books?id=cFm_qnPAvV8C&pg=PA102&dq=elbow+anato my&hl=en&sa=X&ved=0ahUKEwj3vPm9s-PTAhUl0IMKHSDBCb0Q6AEIOTAE#v=onepage&q=elbow anatomy&f=false

Johnson, J. A., Rath, D. A., Dunning, C. E., Roth, S. E., & King, G. J. W. (2000). Simulation of elbow and forearm motion in vitro using a load controlled testing apparatus. *Journal of Biomechanics*, *33*(5), 635–639. https://doi.org/10.1016/S0021-9290(99)00204-3

Josefsson, P. O., Gentz, C. F., Johnell, O., & Wendeberg, B. (1987). Surgical versus non-surgical treatment of ligamentous injuries following dislocation of the elbow joint. A prospective randomized study. *JBJS*, *69*(4), 605–608.

Josefsson, P. O., Johnell, O., & Gentz, C. F. (1984). Long-term sequelae of simple dislocation of the elbow. *J Bone Joint Surg*, *66*(6), 927–930.

Josefsson, P. O., Johnell, O., & Wendeberg, B. (1987). Ligamentous injuries in dislocations of the elbow joint. *Clin Orthop Relat Res*.

Khare, G. N., Goel, S. C., Saraf, S. K., Singh, G., & Mohanty, C. (1999). New observations on carrying angle. *Indian Journal of Medical Sciences*, *53*(2), 61–67.

Kuhn, M. A., & Ross, G. (2008). Acute elbow dislocations. *Orthop Clin North Am*, *39*(2), 155–161.

Kusins, J. (2015). *Experimenal Analysis of Parameters Influencing the Bone Burring Process*. The University of Western Ontario.

Lazar, M. L., Lurski, A., Ghafurian, S., Chen, L., Uko, L., Tan, V., & Li, K. (2015). In vitro dynamic simulation of elbow motion. In *Biomedical Engineering Conference (NEBEC), 2015 41st Annual Northeast* (pp. 1–2). IEEE.

Lee, A. T., Schrumpf, M. A., Choi, D., Meyers, K. N., Patel, R., Wright, T. M., … Daluiski, A. (2013). The influence of gravity on the unstable elbow. *J Shoulder Elb Surg*, *22*(1), 81–87. https://doi.org/http://dx.doi.org/10.1016/j.jse.2012.08.012

Lee, Y. C., Eng, K., Keogn, A., McLean, J. M., & Bain, G. I. (2012). Repair of the acutely unstable elbow: use of tensionable anchors. *Techniques in Hand and Upper Extremity Surgery*, *16*(4), 225–9.

London, J. T. (1981). Kinematics of the elbow. *The Journal of Bone and Joint Surgery. American Volume*, *63*(4), 529–35.

Lunsford, T. R., & Contoyannis, B. (2008). Materials science. In J. D. Hsu, J. W. Michael, & J. R. Fisk (Eds.), *AAOS Atlas of Orthoses and Assistive Devices* (pp. 15–52). Philadelphia, PA.

Lunsford, T. R., & DiBello, T. V. (2008). Principles and components of upper limb orthosis. In J. D. Hsu, J. W. Michael, & J. R. Fisk (Eds.), *AAOS Atlas of Orthoses and Assistive Devices* (Fourth, pp. 179–190). Philadelphia, PA: Mosby Inc.

Magnusen, J. P. (2004). *Design And Fabrication Of An Elbow Motion Simulator*. University of Pittsburgh.

Manocha, R. H. K. (2016). *Optimizing the Rehabilitation of Elbow Lateral Collateral Ligament Injuries*. The University of Western Ontario.

Maripuri, S. N., Debnath, U. K., Rao, P., & Mohanty, K. (2007). Simple elbow dislocation among adults: a comparative study of two different methods of treatment. *Injury*, *38*(11), 1254–1258.

Markholf, K. L., Lamey, D., Yang, S., Meals, R., & Hotchkiss, R. (1998). Radioulnar load-sharing in the forearm. A study in cadavera. *J Bone Joint Surg Am*, *80*(6), 879–88. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/9655106

May, B. J., & Lockard, M. A. (2011). Designing and Prescribing Orthoses. In *Prosthetics & Orthotics in Clinical Practice: A case study approach* (pp. 203–220).

Philadelphia, PA: E.A. Davis Company.

McKee, M. D., Schemitsch, E. H., Sala, M. J., & O'Driscoll, S. W. (2003). The pathoanatomy of lateral ligamentous disruption in complex elbow instability. *Journal of Shoulder and Elbow Surgery*, *12*(4), 391–6.

Mehlhoff, T., Noble, P., Bennett, J., & Tullos, H. (1988). Simple dislocation of the elbow in the adult. Results after closed treatment. *J Bone Joint Surg Am*, *70*(2), 244–249.

Mica, M. C., Caekebeke, P., & Riet, R. van. (2016). Lateral collateral ligament injuries of the elbow – chronic posterolateral rotatory instability (PLRI). *EFORT Open Reviews*, *1*(12), 461–468.

Mimic, I., Kim, S.-Y., Park, I.-H., Kim, P.-T., & Jeon, I.-H. (2009). Surgical management of unstable elbow dislocation without intra-articular fracture. *Int Orthop*, *33*(4), 1141–1147.

Mirzayan, R., & Itamura, J. M. (2004). Dislocations, Fracture Dislocations and Instability. In *Shoulder and Elbow Trauma* (Illustrate, pp. 3–22). Thieme.

Morrey, B. (2000). Anatomy of the elbow joint. In *The elbow and its disorders* (3rd ed., pp. 13–42). Philadelphia, PA: Saunders/Elsevier.

Morrey, B. F., & An, K. N. (1983). Articular and ligamentous contributions to the stability of the elbow joint. *The Americal Journal of Sports Medicine*, *11*(5), 315–319.

Morrey, B. F., & An, K. N. (1985). Functional anatomy of the ligaments of the elbow. *Clin Orthop Relat Res*, *201*, 84–90.

Morrey, B. F., Askew, L. J., An, K.-N., & Chao, E. Y. (1981). A biomechanical study of normal elbow motion. *J Bone Joint Surg*, *63–A*(6), 872–877. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/7240327

Morrey, B. F., & Sanchez-Sotelo, J. (2009). *The Elbow and Its Disorders* (4th ed.).

Philadelphia: Elsevier Health Sciences.

Nestor, B. J., O'Driscoll, S., & Morrey, B. F. (1992). Ligamentous reconstruction for posterolateral rotatory instability of the elbow. *J Bone Joint Surg*, *74–A*, 1235–1241.

Nielsen, K., & Olsen, B. (1999). No stabilizing effect of the elbow joint capsule: a kinematic study. *Acta Orthop Scand*, *70*(1), 6–8.

Nordin, M., & Frankel, V. H. (2001). *Basic Biomechanics of the Musculoskeletal System* (illustrate). Lippincott Williams & Wilkins.

O'Driscoll, S., Bell, D., & Morrey, B. (1991). Posterolateral rotatory instability of the elbow. *J Bone Joint Surg Am*, *73*(3), 440–446.

O'Driscoll, S., Morrey, B. F., Korinek, S. L., & An, K. N. (1992). Elbow subluxation and dislocation. A spectrum of instability. *Clinical Orthopaedics and Related Research*, *280*, 186–197.

O'Driscoll, S. W., Jupiter, J. B., King, G. J. W., Hotchkiss, R. N., & Morrey, B. F. (2000). The unstable elbow. *Journal of Bone and Joint Surgery*, *82*(5), 724–738.

Olsen, B. S., Michael, T. V., Søjbjerg, J. O., Helmig, P., & Sneppen, O. (1966). Lateral collateral ligament of the elbow joint: Anatomy and kinematics. *Journal of Shoulder and Elbow Surgery*, *5*(2), 103–112.

Olsen, B. S., & Søjbjerg, J. O. (2003). The treatment of recurrent posterolateral instability of the elbow. *J Bone Joint Surg [Br]*, (85–B), 342–6.

Olsen, B. S., Søjbjerg, J. O., Nielson, K. K., Vaesel, M. T., Dalstra, M., & Sneppen, O. (1998). Posterolateral elbow joint instability: the basic kinematics. *J Shoulder Elbow Surg*, *7*(1), 19–29.

Osbourne, G., & Cotterill, P. (1966). Recurrent dislocation of the elbow. *Bone & Joint Journal*, *48*(2), 340–346.

Paraskevas, G., Papadopoulos, A., Papaziogas, B., Spanidou, S., Argiriadoi, H., & Gigis,

J. (2004). Study of the carrying angle of the human elbow joint in full extension: a morphometric analysis. *Surg Radiol Anat*, *26*(1), 19–23.

Pederzini, L. A., Eygendaal, D., & Denti, M. (2016). *Elbow and Sport*. (L. A. Pederzini, D. Eygendaal, & M. Denti, Eds.). Berlin, Heidelberg: Springer Berlin Heidelberg. https://doi.org/10.1007/978-3-662-48742-6

Pichora, J. E., Fraser, G. S., Ferreira, L. F., Brownhill, J. R., Johnson, J. A., & King, G. J. W. (2007). The effect of medial collateral ligament repair tension on elbow joint kinematics and stability. *The Journal of Hand Surgery*, *32*(8), 1210–1217.

Poulin, F., & Amoit, L. P. (2002). Interference during the use of an electromagnetic tracking system under OR conditions. *Journal of Biomechanics*, *35*, 733–737.

Prendergast, P. J., van der Helm, F. C., & Duda, G. N. (2005). Analysis of Muscle and Joint Loads. In V. C. Mow & R. Huiskes (Eds.), *Basic Orthopaedic Biomechanics and Mechano-Biology* (Third, pp. 29–91). Philadelphia, PA: Lippincott Williams & Wilkins.

Ramponi, D. R., & Kaufmann, J. A. (2012). Elbow injuries and fractures. *Adv Emerg Nurs J*, *34*(2), 91–99. https://doi.org/10.1097/TME.0b013e318251cdba\r01261775-201204000-00003 [pii]

Rath, D. A. (1997). *Design and Development of an Elbow Loading Apparatus and the Determination of Elbow Kinematics*. The University of Western Ontario.

Regan, W. D., Korinek, S. L., Morrey, B. F., & An, K. N. (1991). Biomechanical study of ligaments around the elbow joint. *Clin Orthop Relat Res*, *271*, 170–179.

Regan, W., & Lapner, P. C. (2006). Prospective evaluation of two diagnostic apprehension signs for postero-lateral instability of the elbow. *J Shoulder Elbow Surg*, *15*, 344–6.

Reichel, L. M., Milam, G. S., Sitton, S. E., Curry, M. C., & Mehlhoff, T. (2013). Elbow lateral collateral ligament injuries. *The Journal of Hand Surgery*, *38*(1), 184–201.

Rhyou, I. H., & Kim, Y. S. (2012). New mechanism of the posterior elbow dislocation. *Knee Surg Sports Traumatol Arthrosc*, *20*(12), 2535–2541.

Ross, G., McDevitt, E. R., Chronister, R., & Ove, P. N. (1999). Treatment of simple elbow dislocation using an immediate motion protocol. *The Americal Journal of Sports Medicine*, *27*(3), 308–311.

Rotini, R., & Marinelli, A. (2008). Elbow Brace in Physical Therapy. In A. Celli, L. Celli, & B. F. Morrey (Eds.), *Treatment of Elbow Lesions: New Aspects in Diagnosis and Surgical Techniques* (illustrate, pp. 323–332). Springer.

Sanchez-Sotelo, J., Morrey, B. F., & O'Driscoll, S. (2005). Ligamentous repair and reconstruction for posterolateral rotatory instability of the elbow. *J Bone Joint Surg [Br]*, *87–B*, 54–61.

Sanderhoff OIsen, B., V, M. T., Sejbjerg, J. O., Helmig, P., Sneppen, O., & Martin, B. (1996). Lateral collateral ligament of the elbow joint: Anatomy and kinematics. *J Shoulder Elbow Surg*, *5*, 103–12. Retrieved from https://journals-scholarsportal-info.proxy1.lib.uwo.ca/pdf/10582746/v05i0002_p1/103_lclotejaak.xml

Schenck, R. C. (Ed.). (1999). *Athletic Training and Sports Medicine* (Illustrate). Jones & Bartlett Learning.

Schimoler, P. J. (2008). *DESIGN OF A CONTROL SYSTEM FOR AN ELBOW JOINT MOTION SIMULATOR*. University of Pittsburgh.

Schippinger, G., Seibert, F. J., Steinböck, J., & Kucharczyk, M. (1999). Management of simple elbow dislocations. *Langenbeck's Archives of Surgery*, *384*(3), 294–297.

Schwab, G. H., Bennett, J. B., Woods, G. W., & Tullos, H. S. (1980). Biomechanics of elbow instability: the role of the medial collateral ligament. *Clinical Orthopaedics and Related Research*, *146*, 42–52.

Sheps, D. M., Hildebrand, K. A., & Boorman, R. S. (2004). Simple dislocations of the elbow: evaluation and treatment. *Hand Clin*, *20*, 389–404.

Shillingford, J. N., & Levine, W. N. (2015). Simple elbow dislocation. In *Shoulder and Elbow Trauma and its Complications: Volume 2: The Elbow* (pp. 3–12).

Sivananthan, S., Sherry, E., Warnke, P., & Miller, M. D. (2012). *Mercer's Textbook of Orthopaedics and Trauma* (Tenth). CRC Press.

Smith, J. P., Savoie, F. H., & Field, L. D. (2001). Posterolateral rotatory instability of the elbow. *Clinics in Sports Medicine*, *20*(1), 47–58.

Starkey, C., & Brown, S. D. (2015). *Examination of Orthopedic & Athletic Injuries* (Revised). F.A. Davis.

Stoneback, J. W., Owens, B. D., Sykes, J., Athwal, G. S., Pointer, L., & Wolf, J. M. (2012). Incidence of elbow dislocations in the United States population. *J Bone Joint Surg Am*, *94*(3), 240–245.

Szekeres, M., Chinchalkar, S. J., & King, G. J. W. (2008). Optimizing Elbow Rehabilitation After Instability. *Hand Clinics*, *24*(1), 27–38. https://doi.org/10.1016/j.hcl.2007.11.005

Tanaka, S., An, K. N., & Morrey, B. F. (1998). Kinematics of ulnohumeral joint under varus-valgus stress. *J Musculoskel Res*, *2*, 45.

Tawari, G. J. K., Lawrence, T., & Stanley, D. (2013). Surgical reconstructions for posterolateral rotatory instability of elbow using a synthetic ligament. *Shoulder & Elbow*, *5*(4), 251–255.

Terra, B. B., Silva, B. C. M., Carvalho, H. B. F. D., Dobashi, E. T., Pinto, J. A., & Ishida, A. (2011). Evolution of the carrying angle of the elbow: a clinical and radiographic study. *Acta Ortopédica Brasileira*, *19*(2), 79–82.

Upper extremity muscle anatomy muscles of the pectoral girdle and upper limbs anatomy and - upper extremity muscle anatomy. (2016). Retrieved October 3, 2017, from http://www.anatomylibrary.us/upper-extremity-muscle-anatomy/upper-extremity-muscle-anatomy-muscles-of-the-pectoral-girdle-and-upper-limbs-anatomy-and/

van Roy, P., Baeyens, J. P., Fauvart, D., Lanssiers, R., & Clarijs, J. P. (2005). Arthro-kinematics of the elbow: study of the carrying angle. *Ergonomics*, *48*(11–14), 19–21.

Wiesel, S. W. (2015). *Operative Techniques in Orthopaedic Surgery (Four Volume Set)* (2nd ed.). Lippincott Williams & Wilkins.

Wilk, K. E., Arrigo, C., & Andrews, J. R. (1993). Rehabilitation of the elbow in the throwing athlete. *The Journal of Orthopaedic and Sports Physical Therapy*, *17*(6), 305–317. https://doi.org/10.2519/jospt.1993.17.6.305

Winter, D. A. (1990). *Biomechanics and Motor Control of Human Movement* (2nd ed.). Wiley.

Wolff, A. L., & Hotchkiss, R. N. (2006). Lateral Elbow Instability: Nonoperative, Operative, and Postoperative Management. *Journal of Hand Therapy*, *19*(2), 238–244.

Yilmaz, E., Karakurt, L., Belhan, O., Bulut, M., Serin, E., & Avci, M. (2005). Variation of Carrying Angle With Age, Sex, and Special Reference to Side. *Orthopedics*, *28*(11), 1360–1363.

Chapter 2

2 Design and Development of an Experimental Hinged Elbow Orthosis for Examining *In Vitro* Kinematics and Biomechanics of the LCL-Injured Elbow

Overview

This chapter focuses on the design of an adjustable hinged elbow orthosis (HEO) to investigate the effects of hinged elbow orthosis (HEO) valgus angulation following elbow injury. The rationale behind the HEO design, components and measurements are explained. An overview of an active motion elbow simulator and kinematic data acquisition is discussed. The reliability of the elbow orthosis measurements is also examined.

2.1   Hinged Elbow Orthosis (HEO) Design Rationale

An adjustable hinged elbow orthosis (HEO) was developed to study the effect of brace (valgus) angle on the stability of the lateral collateral ligament (LCL) deficient elbow. Existing HEOs such as the Bledsoe and Mayo Clinic elbow HEOs are designed with no valgus angulation, which does not take into account the native anatomy of the elbow. In full extension, the elbow has a valgus orientation called the carrying angle, of approximately 11° to 14° and 13° to 16° in men and women, respectively (Atkinson & Elftman, 1945; Morrey & Sanchez-Sotelo, 2009; Paraskevas et al., 2004; Terra et al., 2011; van Roy, Baeyens, Fauvart, Lanssiers, & Clarijs, 2005). As the elbow is flexed, the valgus angle continuously decreases and eventually reaches a varus orientation towards full flexion (Morrey, Askew, An, & Chao, 1981).

The objective of this project was to design and fabricate a HEO with an adjustable valgus angulation in order to investigate its effects on the LCL injured elbow using *in-vitro* biomechanical testing. A secondary objective of the design was to include an adjustable width to ensure a secure fit to each arm, with minimal internal movement. The orthosis was developed to examine the kinematics and LCL loads in various arm positions and forearm rotations.

2.2 HEO Design and Function

Figure 2-1 depicts an overview of the experimental HEO designed to investigate the effect of LCL injuries. Four rigid arms (two humeral and two forearm) were designed and 3D printed from plastic. Aluminum rods, held in place by setscrews, attach the medial and lateral HEO arms to each other. The connecting hinge mechanisms on the medial and lateral sides of the elbow are comprised of a custom designed brass connector, yoke ends, and aluminum rods. Engineering drawings and specifications of all components are included in Appendix B.

Hook-and-loop fastener straps (BlueDot Trading, Poway, CA, USA) were fabricated and connected anteriorly, from the medial to lateral sides of the HEO, to secure it firmly around the arm, mimicking the clinical scenario. The interior surface was lined with

sponges for comfort by avoiding pressure points between the HEO and skin. The HEO was developed to investigate LCL injures, most often resulting from elbow dislocations and fracture-dislocations (Babhulkar, 2015; Osbourne & Cotterill, 1966; Reichel, Milam, Sitton, Curry, & Mehlhoff, 2013). The *in vitro* studies in this thesis were conducted on left arms only; therefore, the anatomical references described for the duration of this chapter are in reference to a left elbow joint.
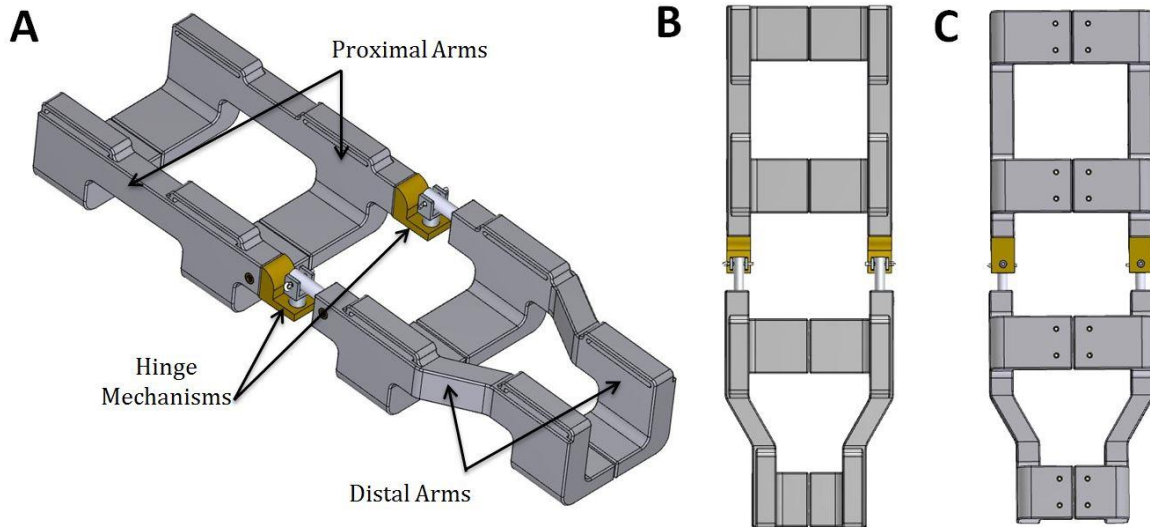
Figure 2-1: Overview of HEO Design. *Overview of the modular brace design from its isometric (A), anterior (B) and posterior (C) views. The design incorporates four arms (two humeral and two forearm) and two hinge mechanisms (medial and lateral).*

2.2.1 HEO Arms

The humeral (proximal) and forearm (distal) arms of the HEO (Figure 2-2 and Figure 2-3), were 3D printed from plastic (Tough, Formlabs Inc., Somerville, MA, USA). Designed similar to ABS plastic, this strong polymer resin has an elastic modulus of 2.7 GPA and was used to avoid metallic interference with the electromagnetic tracker previously described in Chapter 1 (Section 1.4.2.). The two medial arms were designed as mirror images of the lateral arms. To ensure rigidity of the HEO, each arm features two cuffs connecting the medial and lateral arms posteriorly. Two slots were also added anteriorly on each arm to attach the fastener straps.
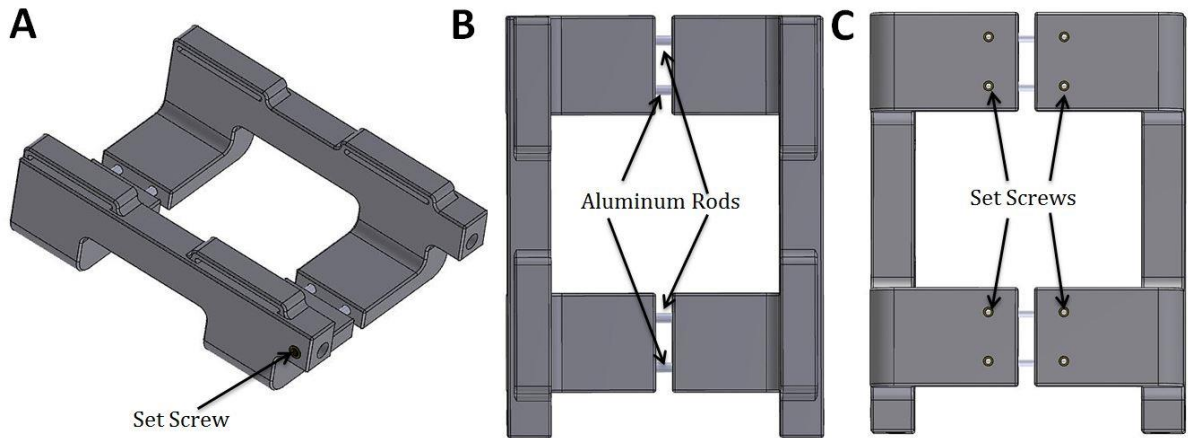
Figure 2-2: Humeral HEO Arms. *Proximally, there are two arms (medial and lateral) that surround the upper arm. Aluminum rods, held in place by brass set screws, connect the arms. Isometric (A), anterior (B) and posterior (C) views are shown.*
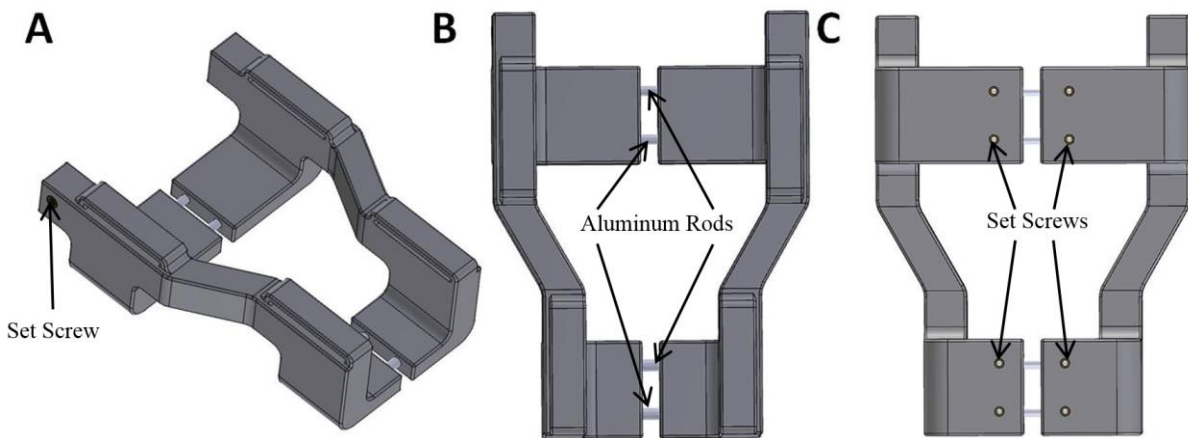
Figure 2-3: Forearm HEO Arms. *The distal end is comprised of two 3-D printed arms, tapered distally. Similar to the humeral arms, the distal arms are connected by aluminum rods that are held in place by setscrews. Isometric (A), anterior (B) and posterior (C) views are shown.*

2.2.1.1   Humeral Arms

The *medial and lateral humeral arms* were connected using two 5/16'' diameter, 66 mm long aluminum rods at each cuff. Two brass, M2.5 heat set inserts (McMaster-Carr, Elmhurst, IL, USA) were secure into the cuff. M2.5 stainless steel flat-tip set screws (McMaster-Carr, Elmhurst, IL, USA) were inserted into the heat set inserts to hold the aluminum rod securely at the desired width. To change the width of the HEO, the setscrews were loosened using an allen key, the width was adjusted, and the setscrews were retightened. At the smallest width, with the cuffs directly aligned, the width is 83 mm. Based on the anthropometric data for 40-year-old American males, the 5th percentile of the biceps diameter is approximately 86.9 mm (National Aeronautics and Space Administration, 1995). Thus, a minimum width of 83 mm was selected to ensure that the HEO would be suitable for smaller cadaveric arms. At the largest width, the width is 11.2 mm corresponding to the biceps diameter of 40-year-old American males in the 95th percentile (National Aeronautics and Space Administration, 1995). However, 80 mm rods were also machined for use in larger specimens if required.

At the distal ends of the humeral arms, a 5/16" diameter aluminum rod was inserted and secured in place by an M4 stainless steel flat-tip setscrew (McMaster-Carr, Elmhurst, IL, USA), housed by a brass M4 heat set insert for plastic (McMaster-Carr, Elmhurst, IL, USA), as previously described. The adjustability of these rods serves a dual function; 1) both rods may be moved distally to increase the length of the orthosis to accommodate longer specimens, and 2) the rods adjust the valgus angle of the HEO, as described further in Section 2.2.2 ahead.

2.2.1.2   The Forearm Arms

The *medial and lateral forearm arms* of the orthosis were designed almost identical to the proximal (humeral) arms, however, the most distal cuff was tapered to decrease the width of the HEO at the distal forearm. This design feature decreases the bulk of the orthosis and ensures a tighter fit to minimize internal arm movement. The cuffs were held together by the same mechanism as the humeral arms, comprising aluminum rods and setscrews. Similarly, a 5/16'' aluminum rod was inserted at the proximal end of both

the medial and lateral arms and two slots were designed anteriorly on each arm to attach the fastener straps.

2.2.2 Hinge Mechanism

The modular ***hinge mechanism*** illustrated in Figure 2-4 was designed to adjust the valgus angle of the HEO as well as permit elbow flexion and extension. Although only left arms were studied in this thesis for convenience, the hinge mechanism was included on both sides of the HEO to ensure that it would be suitable for both left and right arms. The medial and lateral hinge mechanisms each feature two 5/16" aluminum rods, a galvanized steel yoke end, a custom machined brass connector, a 3-mm diameter cotter pin, and an M6 stainless steel bolt. Non-ferrous metals were utilized as much as possible to avoid magnetic interference with the electromagnetic tracking system.
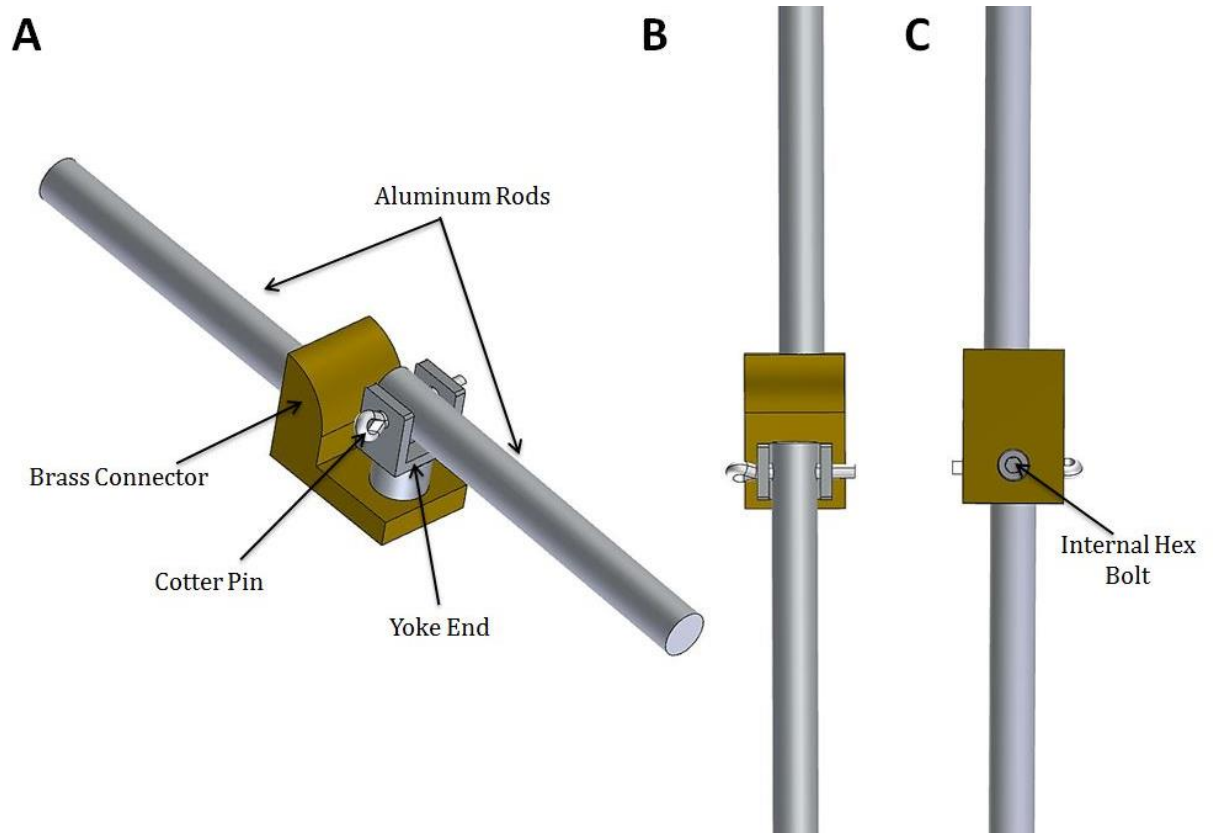
Figure 2-4: Hinge Mechanism. *An isometric (A), anterior (B) and posterior (C) view of the hinge mechanism is shown. The mechanism is comprised of two aluminum rods, a custom machined brass connector, a yoke end, a cotter pin and an internal hex bolt. This mechanism allows unrestricted flexion-extension as well as an adjustable valgus angulation of the HEO.*

To permit elbow flexion-extension, the rods were proximally attached to yoke ends via cotter pins (Figure 2-5). The distal ends of the rods were secured into the HEO forearm arms. This allows the humeral arms to stay fixed, while the proximal arms freely flex and extend.
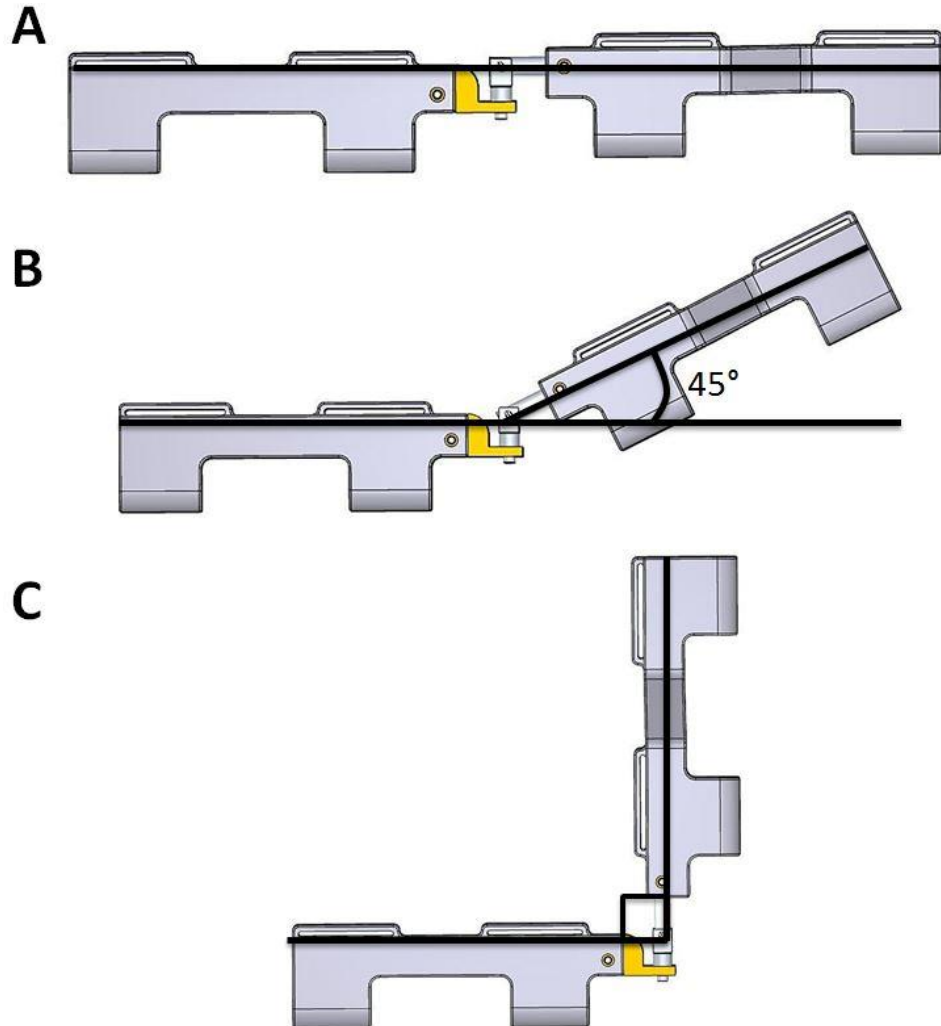
Figure 2-5: Flexion-Extension of the HEO. *The connection between the yoke end aluminum rod permits the arm to freely flex and extend. Pictured is the HEO at 0 degrees (A), 45 degrees (B) and 90 degrees (C) of flexion.*

The connection between the brass connector and yoke end permits varus-valgus angulation as demonstrated. For the purpose of this thesis, only valgus HEO angulation will be implemented, as seen in Figure 2-6. The brass connectors were secured to the yoke ends using M6 internal hex bolts. Proximally, aluminum rods were threaded into the brass components to attach the hinge mechanisms to the rigid humeral arms. During experimentation, the valgus angle of the HEO was increased from 0° to 20° at 5° intervals, by loosening the bolts as well as the setscrews on the medial side of the HEO. The HEO was then moved to its desired angle, and the bolts were retightened. During this process, the proximal and distal aluminum rods of the medial hinge were moved out of the rigid humeral and forearm arms.
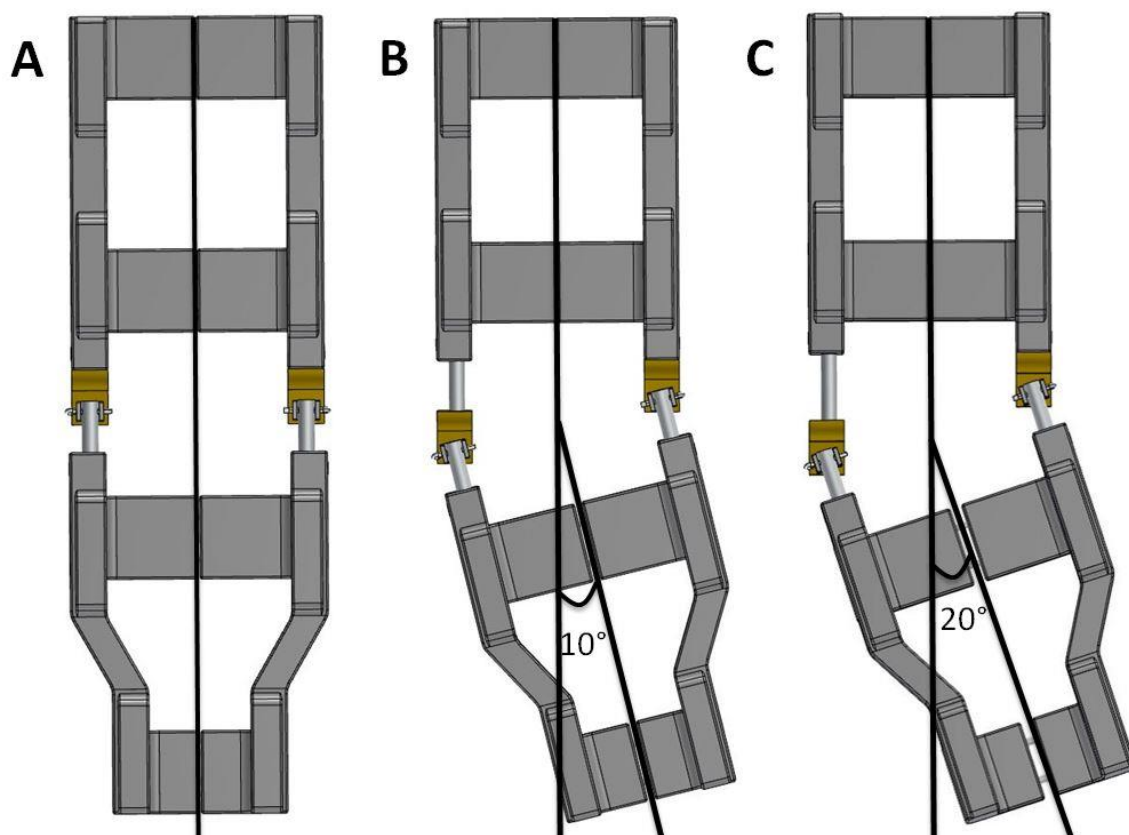
Figure 2-6: Valgus Angulation of the HEO. *The attachment of the yoke end to the brass connector allows the valgus angulation of the HEO to be adjusted and secured in place. Pictured are valgus angles of 0 degrees (A), 10 degrees (B) and 20 degrees (C) for a left elbow joint.*

2.3 Experimental Testing

An existing active motion elbow simulator was used to simulate elbow flexion and extension using a custom LabVIEW software (National Instruments Corporation, Austin, TX) (Armstrong et al., 2000; Cynthia E. Dunning, Duck, King, & Johnson, 2001; Ferreira, 2011; Johnson, Rath, Dunning, Roth, & King, 2000). The simulator is shown in Figure 2-7. To quantify the reliability of the HEO, five active and passive flexion-extension trials were conducted in the vertical dependent position with the forearm pronated on the same cadaveric specimen.

2.3.1 Specimen Preparation

One fresh-frozen cadaveric upper extremity stored at -20℃ was thawed at room temperature (22 $\pm$2℃) for approximately 18 hours. The distal tendons of the biceps, brachialis, triceps, brachioradialis, pronator teres, wrist flexors (flexor carpi radialis and flexor carpi ulnaris) and wrist extensors (extensor carpi radialis longus and extensor carpi ulnaris) were sutured with braided fishing line (Bravefishermen, Shanghai, China). In order to reproduce anatomic lines of action, alignment guides were placed near the medial and lateral epicondyles, as well as at the supracondylar ridge (Figure 2-8). The pronator teres and wrist flexors were passed subcutaneously through the medial epicondylar guide, the wrist extensors through the lateral epicondylar guide, and the brachioradialis at the supracondylar ridge. A custom designed 8 mm stainless steel rod was rigidly fixed into the medullary canal of the humerus with bone cement (Stryker, Kalamazoo, MI) (Figure 2-9). The specimen was then mounted onto a custom humeral clamp on the elbow simulator. The biceps, brachialis and triceps were connected to computer-controlled motors and the remaining muscles were connected to actuators.
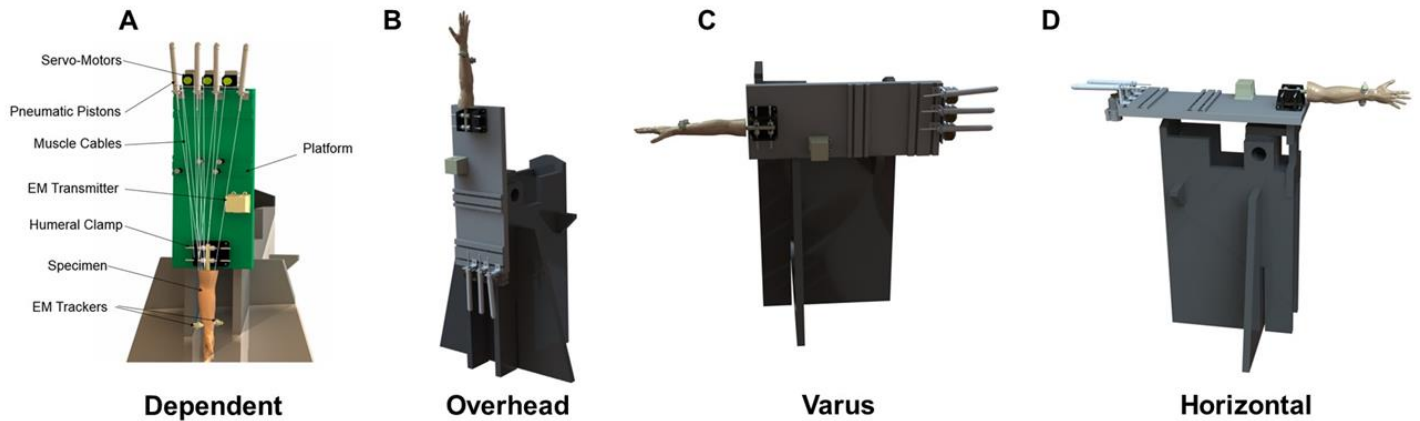
Figure 2-7: Active Elbow Motion Simulator. *An elbow simulator capable of loading various muscles using computer-controlled servomotors and pneumatic actuators to simulate active flexion-extension. The arm is rigidly connected to the simulator by a humeral clamp. A transmitter on the simulator base and an electromagnetic tracker fixed to the ulna record ulnohumeral kinematics during in vitro experimentation. This image depicts the simulator in the (A) dependent position (B) overhead position, (C) varus position and (D) horizontal position. Right arm shown (Manocha 2016).*
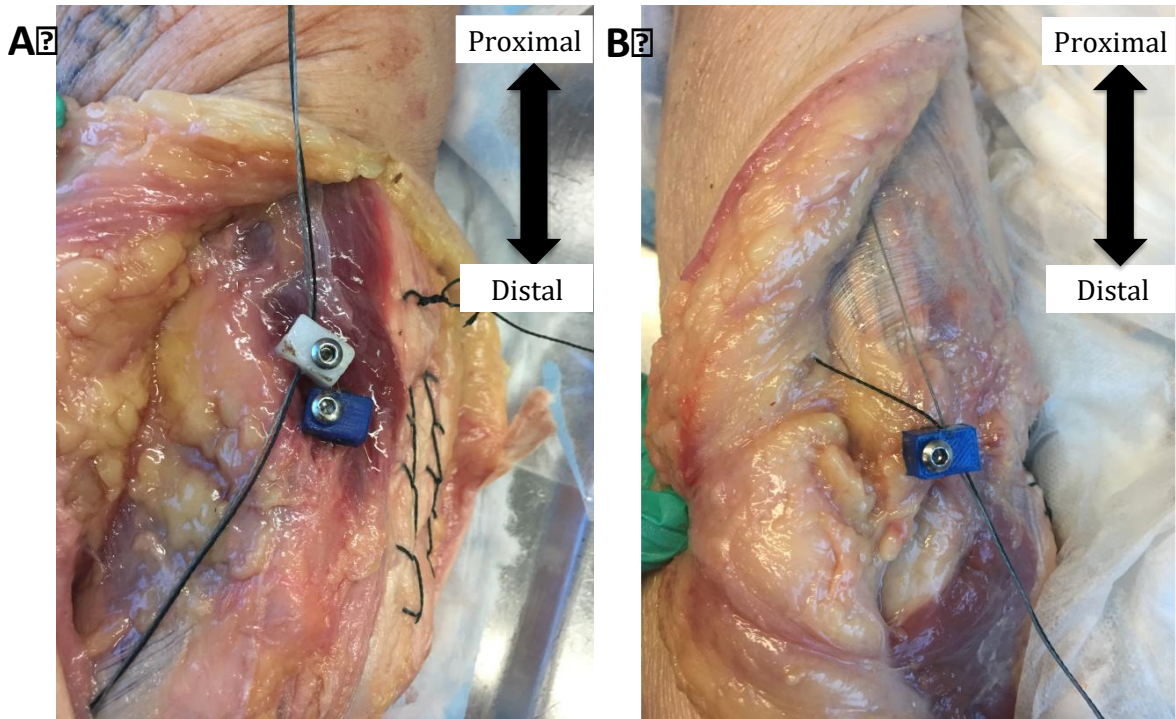
Figure 2-8: Suture Alignment Guides. *Alignment guides were placed laterally (A) and medially (B). The sutures connecting the wrist extensors and brachioradialis were passed through the lateral guides, whereas the pronator teres and wrist flexors were inserted through the medial guides.*
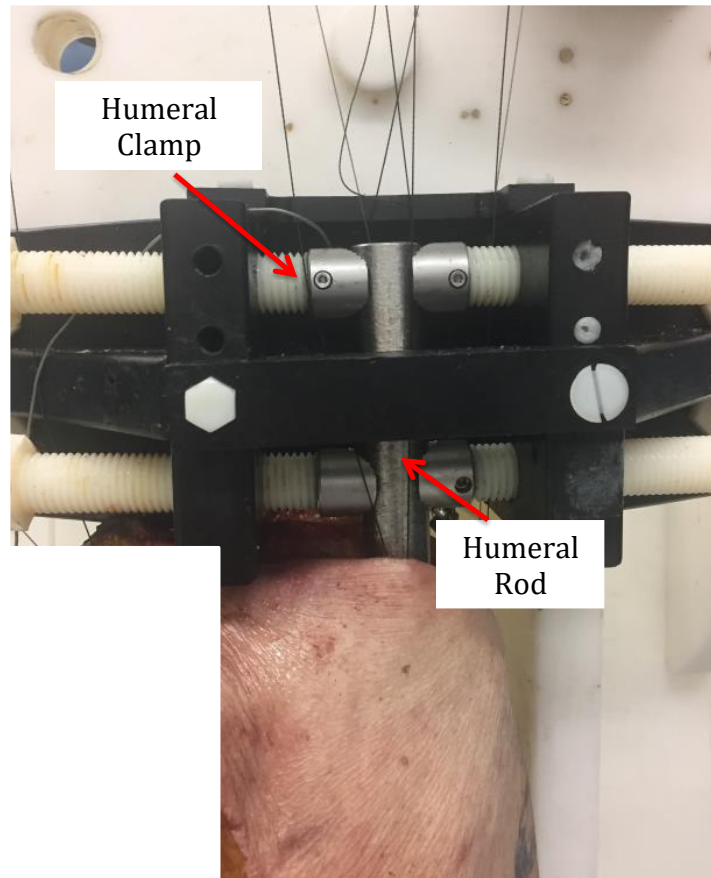
Figure 2-9: Humeral Clamping System. *A stainless steel humeral rod was cemented into the humeral canal using bone cement. The humeral rod was then ridigly fixed to a humeral clamp on the base of the simulator. Left arm shown.*

2.3.2 Testing Protocol

Five active and passive elbow flexion-extension motions were simulated with the forearm pronated, while the varus-valgus angulation and internal-external rotation ulnohumeral kinematics were simultaneously measured. Passive extension trials were conducted by the same investigator (SB) grasping the wrist, rotating the forearm into pronation, and manually extending the arm at a rate of approximately 10° per second. In order to simulate active flexion and extension, a custom designed LabVIEW program (National Instruments, Austin, TX, USA) was used for motor and actuator control as well as kinematic data collection (Dunning, Zarzour, Patterson, Johnson, & King, 2001; Ferreira, 2011; Johnson et al., 2000; Kusins, Willing, King, & Ferreira, 2016). The biceps, triceps and brachialis loads were controlled by employing the maximum voluntary contraction of each muscle based on *in vivo* electromyographic (EMG) values and cross section areas (CSA) reported (Amis, Dowson, & Wright, 1979; Funk, An, Morrey, & Daube, 1987). A 10 N tone load was also applied to the wrist flexors and extensors to stabilize the hand. A tone load of 40 – 60 N was applied to the pronator teres to maintain the forearm in pronation. Active motion trials were also conducted at a rate of approximately 10 degrees/second.

Testing was conducted on the LCL-injured elbow, which was simulated by sectioning both the LCL and common extensor origin (CEO). The interval between the anconeus and extensor carpi ulnaris was opened and the LUCL and RCL were released off the lateral epicondyle. All skin incisions were sutured during testing to keep the soft tissues moist. The HEO was placed on the arm by palpating the medial and lateral epicondyles to determine the flexion-extension axis of the elbow and aligning the center of the hinge mechanism with it. The HEO straps were then secured tightly. The five active and passive pronation trials were conducted in the vertical dependent position with the HEO at 0°. The HEO was then changed to 15° of valgus angulation and the testing sequence was repeated.

The Flock of Birds® (Ascension Technology Corporation, Burlington, VT, USA) a six-DOF electromagnetic tracking system was employed to track varus-valgus angulation (VVA) and internal-external rotation (IER) of the ulna relative to the humerus. A

transmitter was rigidly fixed to the base of the simulator, and a receiver was securely fixed to the medial ulna (Figure 2-10). The tracker mount was carefully placed to ensure that there was no soft tissue impingement or brace obstruction throughout the range of motion. After testing was completed, the wrist and elbow joints were disarticulated to create anatomic humeral and ulnar coordinate systems via digitization. A Delrin® (DuPont, Wilmington, DE, USA) stylus was used to digitize the center of the humeral shaft, the center of the trochlear groove and the center of curvature of the capitellum to create the humeral coordinate system. For the ulnar coordinate system the center and plane of the greater sigmoid notch as well as the tip of the ulnar styloid were digitized. Using the Euler Z-Y-X sequence, the relative motion of the ulna with respect to the humerus was determined and VVA and IER measurements at each extension angle were calculated. As a measure of reliability, a single measures intraclass correlation coefficient (ICC) between each of the five trials was calculated using SPSS 25.0 (SPSS Inc, Chicago, IL, USA).
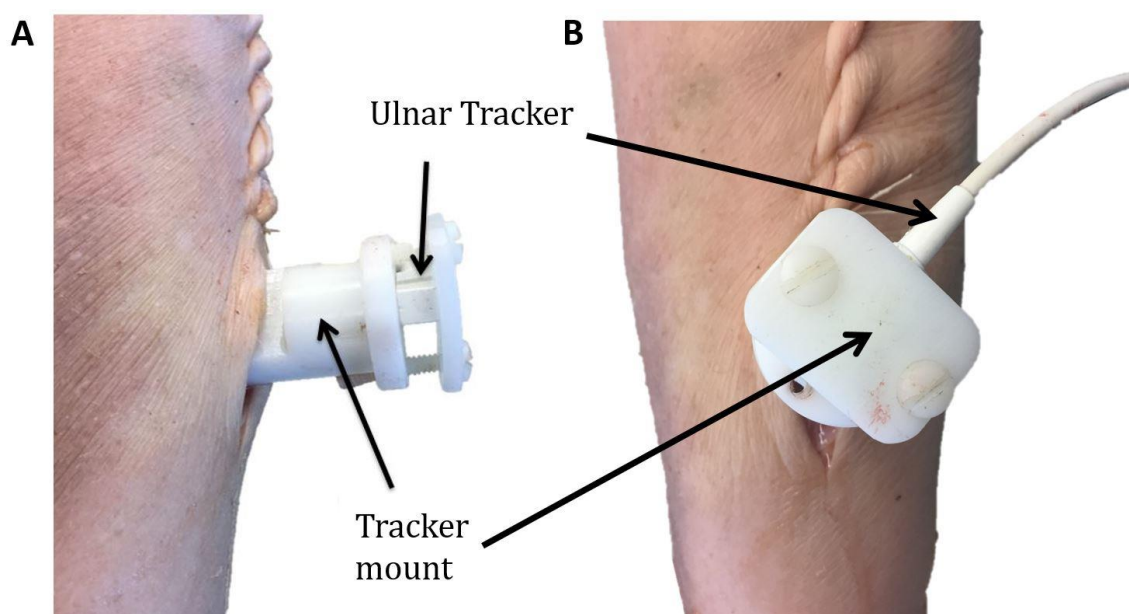
Figure 2-10: Ulnar Receiver. *A tracker mount was secured to the ulna using cortical screws. The ulnar tracker was then fixed to the mount to transmit ulnar orientation information to the electromagnetic tracking system. Anterior (A) and lateral (B) views of a left forearm are shown.*

2.4 Results

The HEO produced reliable results for both the kinematic variables (VVA and IER) analyzed. Only the results of flexion trials are presented and discussed, as the outcomes were similar during extension trials. ICC values less than 0.5, between 0.5 and 0.75, between 0.75 and 0.9 and greater than 0.9 indicated poor, moderate, good and excellent reliability (Koo & Li, 2016).

Reliability trials for VVA during active elbow flexion (Figure 2-11) resulted in single measures ICC = 0.998 with the HEO at 0° and ICC = 0.999 with the HEO at 15°. For IER measurements (Figure 2-12) the ICC values were 0.997 and 0.998 with the HEO at 0° and 15°, respectively.

Passive flexion trials produced less reliable results. VVA measurements (Figure 2-13) with the HEO at 0° and 15° were found to have ICC = 0.895 and 0.707, respectively. Reliability of the IER values were found to be 0.956 and 0.425 for HEO valgus angulation of 0° and 15°, respectively (Figure 2-14).

When the first of the five passive trials was removed and the single measures ICC values were recalculated, the reliability increased tremendously. For VVA measurements with the HEO at 0 and 15° the ICC measurements increased from 0.895 to 0.959 and from 0.707 to 0.973, respectively. ICC values for the IER measurements increased from 0.956 to 0.973 and from 0.425 to 0.984 for HEO valgus angulations of 0° and 15°, respectively.
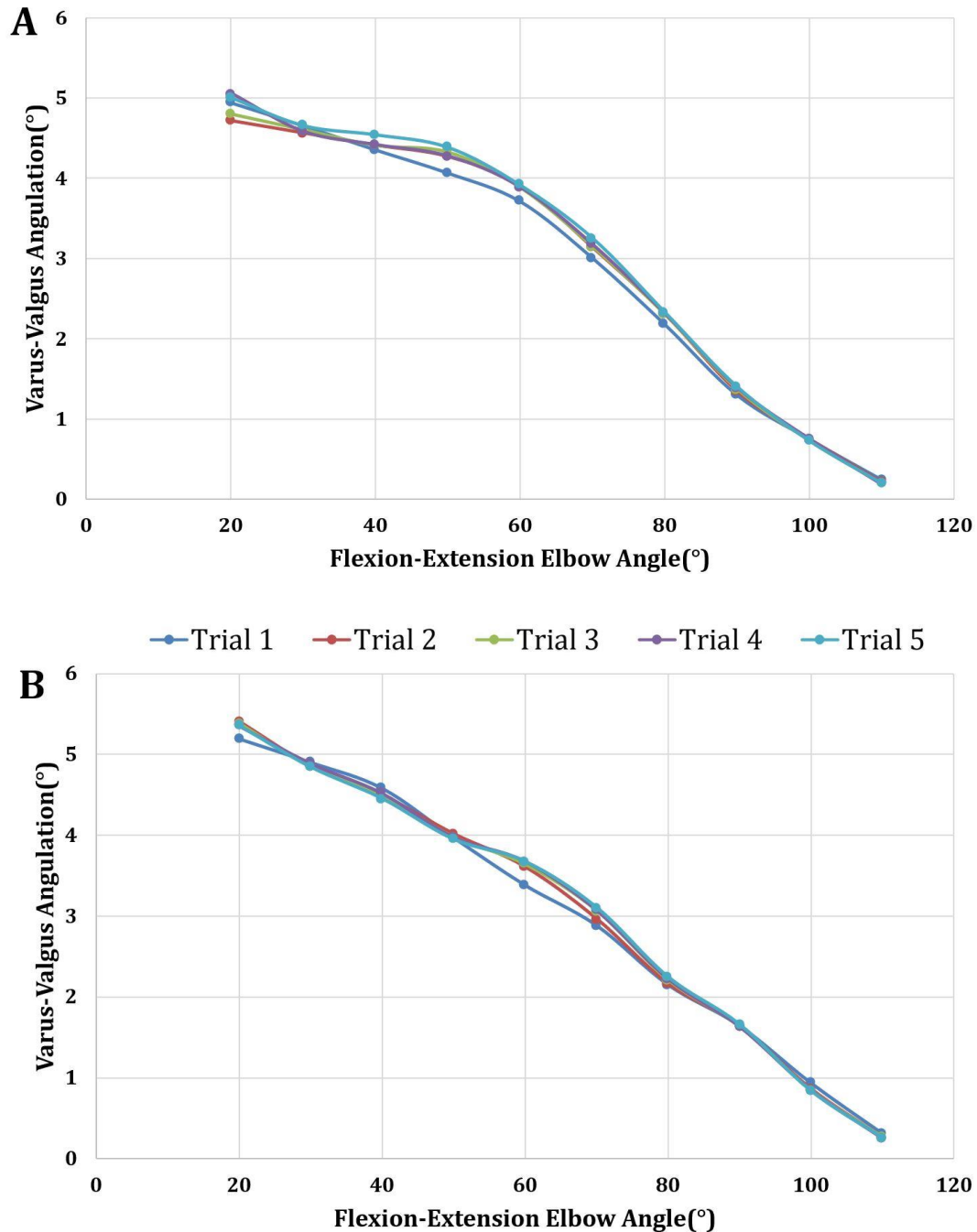
Figure 2-11: Reliability of the Varus-Valgus Angulation Measurements During Active Elbow Flexion. *The reliability of the varus-valgus angulation measurements with the HEO at 0 degrees (A) and 15 degrees (B) is plotted against the flexion-extension angle for five trials. VVA measurements an ICC = 0.998 and ICC = 0.999 for 0 and 15 degrees of valgus HEO angulation, respectively, for the five flexion motions.*
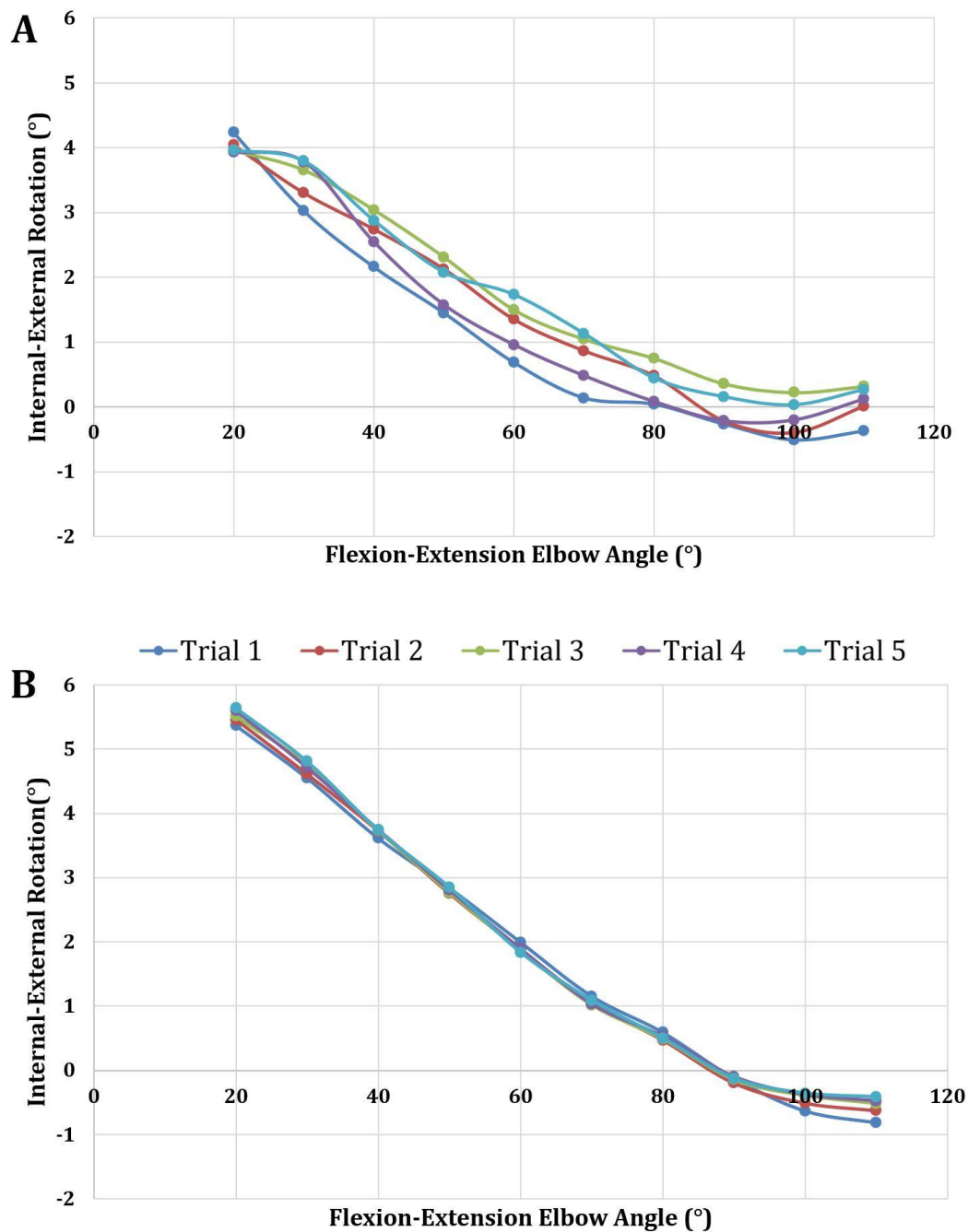
Figure 2-12: Reliability of the Internal-External Rotation Measurements During Active Elbow Flexion. *The reliability of the internal-external rotation measurements with the HEO at 0 degrees (A) and 15 degrees (B) is plotted against the flexion-extension angle for five trials. IER measurements had an ICC = 0.997 and ICC = 0.998, for the HEO at 0 and 15 degrees, respectively.*
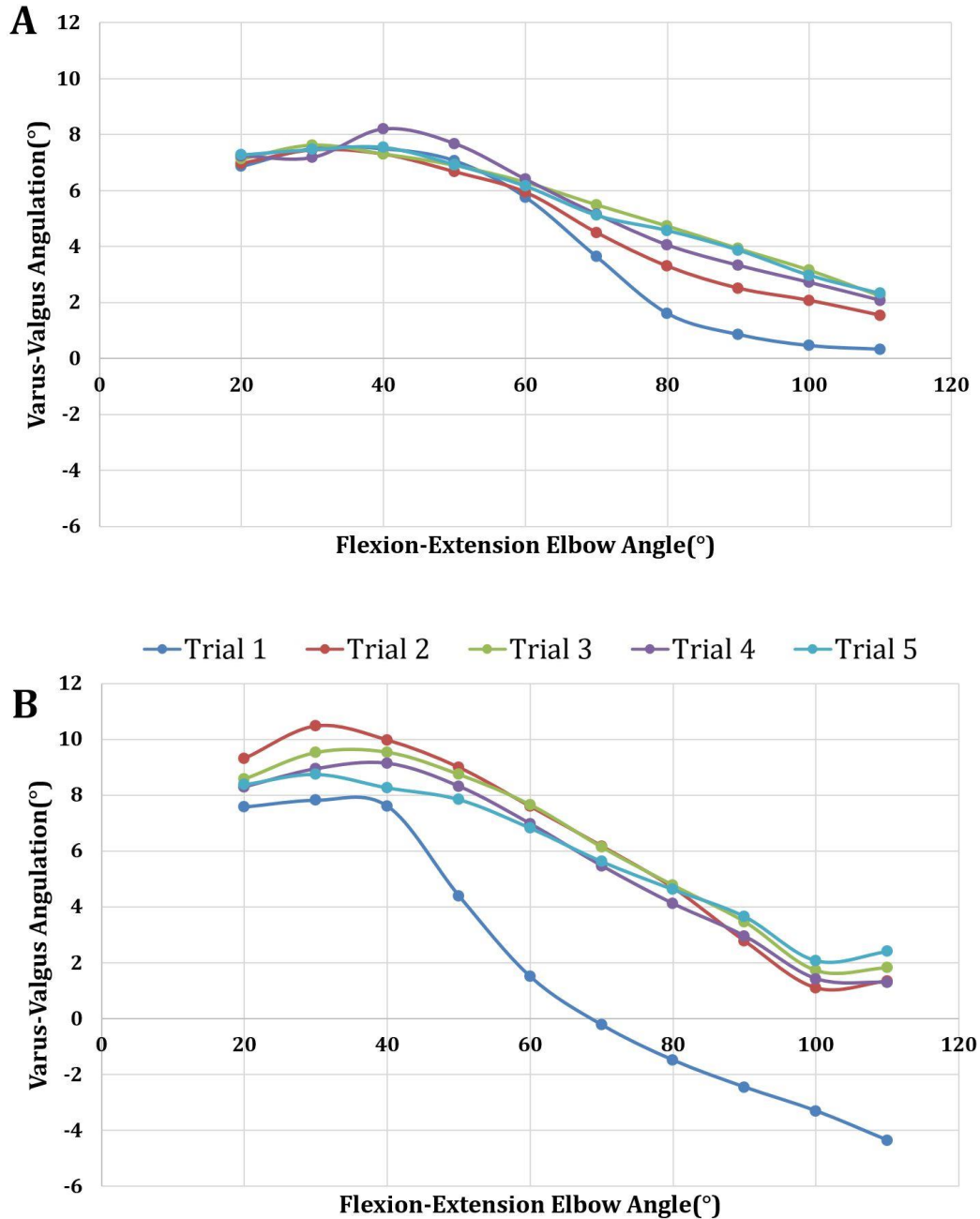
Figure 2-13: Reliability of the Varus-Valgus Angulation Measurements During Passive Elbow Flexion. *The reliability of the varus-valgus angulation measurements with the HEO at 0 degrees (A) and 15 degrees (B) is plotted against the flexion-extension angle for five trials. VVA measurements with the HEO at 0 and 15 degrees had ICC values of 0.896 and 0.707, respectively.*
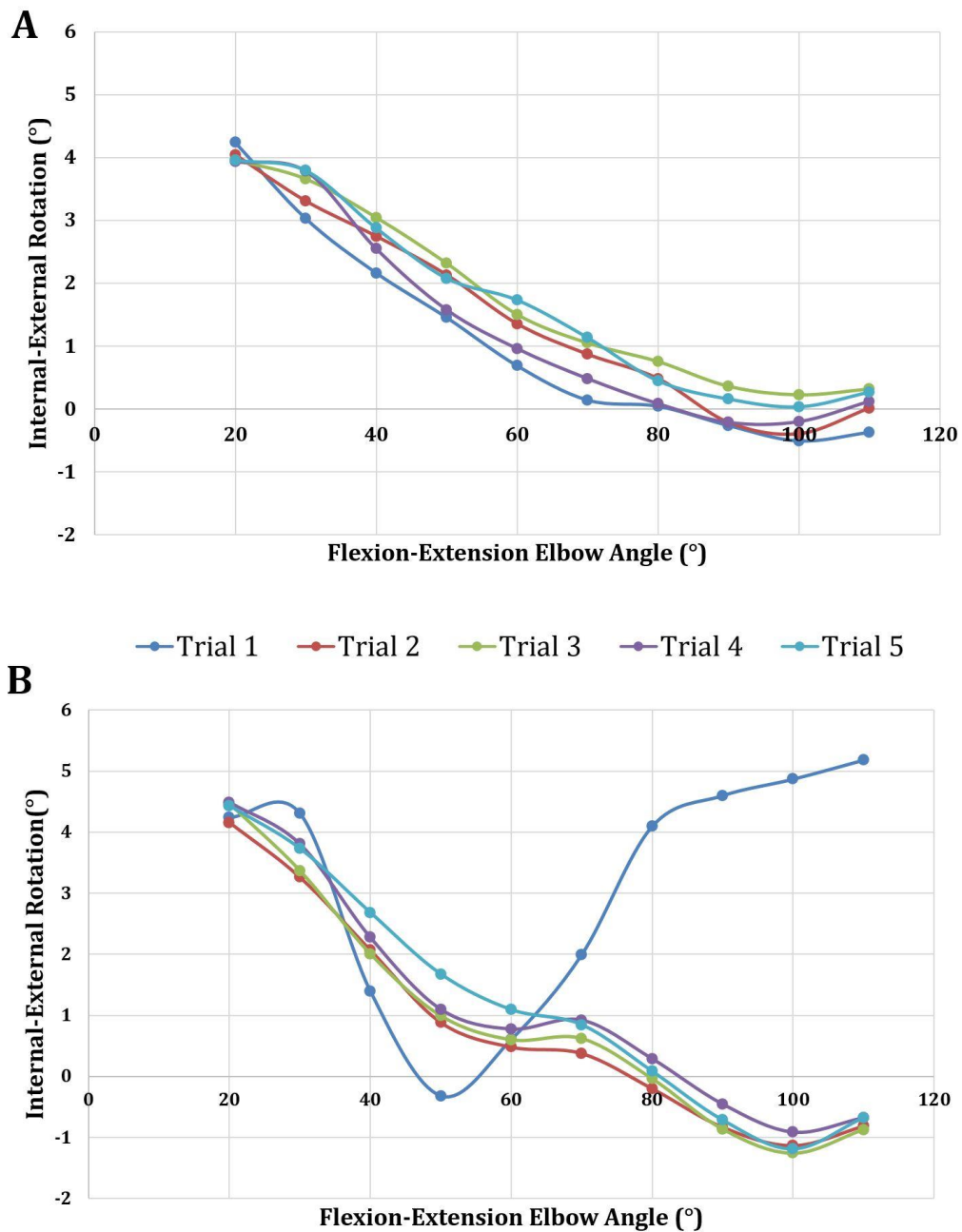
Figure 2-14: Reliability of the Internal-External Rotation Measurements during Passive Elbow Flexion. *The reliability of the internal-external rotation measurements with the HEO at 0 degrees (A) and 15 degrees (B) is plotted against the flexion-extension angle for five trials. IER measurements had ICC = 0.956 and ICC = 0.425, for the HEO at 0 and 15 degrees, respectively.*

2.4.1 Discussion and Conclusions

A modular HEO was successfully designed and developed to investigate the effects bracing on elbow stability. This is a clinically relevant issue, as HEOs are commonly employed for elbow instability without evidence for their effectiveness. The HEO design allowed for unrestricted flexion and extension, with adjustable valgus angulation. The kinematic results obtained confirmed the reliability of the VVA and IER measurements.

During active motion, the reliability of the measurements was excellent. The single measures ICC values ranged from 0.997 to 0.999, indicating minimal variation between flexion trials. This confirms the reliability of the HEO at difference valgus angles during active motion.

Passive motion proved to have less reliable kinematic results than active motion. The ICC values ranged from 0.425 to 0.956. Active motion and pronation have been shown to stabilize the LCL-injured elbow, whereas the elbow is much more unstable during passive motion (Alolabi et al., 2012; Dunning et al., 2001). This instability may have caused less reliable results than active motion. During passive motion, forces and moments applied by the investigator causes decreased reliability, as it is difficult to replicate the same trajectory during each trial, particularly when the elbow is unstable. These findings are in agreement with previous reports that demonstrated a marked reduction in scatter between trials with active motion, likely attributable to the muscle loading reducing the joint and applying more consistent forces across the elbow (Johnson et al., 2000). Without the HEO, it may be postulated that the reliability may have been even lower, since the HEO held the unstable arm tightly, restricting motion and minimizing the difference in the trials. Interestingly, removal of the first trial tremendously improved the reliability results. The ICC values for the last 4 trials, with the first trial excluded, ranged from 0.959 to 0.984, which indicates excellent reliability (Koo & Li, 2016). These results imply that during passive motion, perhaps the first trial should be excluded for the studies to be conducted herein.

A better understanding of the effects of bracing on elbow stability following LCL injury can help improve existing surgical and rehabilitation protocols. The following chapters employ *in vitro* biomechanical and kinematic experiments to investigate the effects of changing the valgus HEO angulation during various arm positions, forearm rotations, LCL repair tensions and muscle activations.

2.5 References

Alolabi, B., Gray, A., Ferreira, L. M., Johnson, J. A., Athwal, G. S., & King, G. J. W. (2012). Rehabilitation of the medial- and lateral collateral ligament-deficient elbow: An in vitro biomechanical study. *Journal of Hand Therapy*, *25*(4), 363–373. https://doi.org/10.1016/j.jht.2012.06.001

Amis, A. A., Dowson, D., & Wright, V. (1979). Muscle strengths and musculo-skeletal geometry of the upper limb. *Engineering in Medicine*, *8*(1), 41–48.

Armstrong, A. D., Dunning, C. E., Faber, K. J., Duck, T. R., Johnson, J. A., & King, G. J. W. (2000). Rehabilitation of the medial collateral ligament-deficient elbow: an in vitro biomechanical study. *The Journal of Hand Surgery*, *25*(6), 1051–1057.

Atkinson, W. B., & Elftman, H. (1945). The carrying angle of the human arm as a secondary sex character. *The Anatomical Record*, *91*(1), 49–52.

Babhulkar, S. (2015). *Elbow injuries* (illustrate). New Delhi, India: JP Medical Ltd.

Dunning, C. E., Duck, T. R., King, G. J., & Johnson, J. (2001). Simulated active control produces repeatable motion pathways of the elbow in an in vitro testing system. *J Biomech*, *34*(8), 1039–48.

Dunning, C. E., Zarzour, Z. D. S., Patterson, S. D., Johnson, J. A., & King, G. J. W. (2001). Muscle forces and pronation stabilize the lateral ligament deficient elbow. *Clin Orthop Relat Res*, *388*, 118–124.

Ferreira, L. M. (2011). *Development of an Active Elbow Motion Simulator and Coordinate Systems to Evaluate Kinematics in Multiple Positions*. The University of Western Ontario.

Fraser, G., Pichora, J., Ferreira, L., Brownhill, J., Johnson, J., & King, G. (2008). Lateral collateral ligament repair restores the initial varus stability of the elbow: an in vitro biomechanical study. *J Orthop Trauma*, *22*(9), 615–23.

Funk, D. A., An, K. N., Morrey, B. F., & Daube, J. R. (1987). Electromyographic analysis of muscles across the elbow joint. *Journal of Orthopaedic Research*, *5*, 529–538.

Johnson, J. A., Rath, D. A., Dunning, C. E., Roth, S. E., & King, G. J. W. (2000). Simulation of elbow and forearm motion in vitro using a load controlled testing apparatus. *Journal of Biomechanics*, *33*(5), 635–639. https://doi.org/10.1016/S0021-9290(99)00204-3

Koo, T.K., & Li, M.Y. (2016). A guideline of selecting and reporting intraclass correlation coefficients for reliability research. *J Chirop Med., 15(2),* 155-163.

Kusins, J., Willing, R., King, G., & Ferreira, L. (2016). Development of a computational elbow model with experimental validation of kinematics and muscle forces. *J Appl Biomech*, *32*(4), 407–14.

Morrey, B. F., Askew, L. J., An, K.-N., & Chao, E. Y. (1981). A biomechanical study of normal elbow motion. *J Bone Joint Surg*, *63–A*(6), 872–877. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/7240327

Morrey, B. F., & Sanchez-Sotelo, J. (2009). *The Elbow and Its Disorders* (4th ed.). Philadelphia: Elsevier Health Sciences.

National Aeronautics and Space Administration. (1995). Anthropometry and Biomechanics. In *Man-Systems Integration Standards NASA-STD-3000* (Revision B, Vol. 1, pp. 16–45). https://doi.org/1995

Osbourne, G., & Cotterill, P. (1966). Recurrent dislocation of the elbow. *Bone & Joint Journal*, *48*(2), 340–346.

Paraskevas, G., Papadopoulos, A., Papaziogas, B., Spanidou, S., Argiriadoi, H., & Gigis, J. (2004). Study of the carrying angle of the human elbow joint in full extension: a morphometric analysis. *Surg Radiol Anat*, *26*(1), 19–23.

Reichel, L. M., Milam, G. S., Sitton, S. E., Curry, M. C., & Mehlhoff, T. (2013). Elbow

lateral collateral ligament injuries. *The Journal of Hand Surgery*, *38*(1), 184–201.

Terra, B. B., Silva, B. C. M., Carvalho, H. B. F. D., Dobashi, E. T., Pinto, J. A., & Ishida, A. (2011). Evolution of the carrying angle of the elbow: a clinical and radiographic study. *Acta Ortopédica Brasileira*, *19*(2), 79–82.

van Roy, P., Baeyens, J. P., Fauvart, D., Lanssiers, R., & Clarijs, J. P. (2005). Arthrokinematics of the elbow: study of the carrying angle. *Ergonomics*, *48*(11–14), 19–21.

Chapter 3

3 The Effects of Brace Valgus Angulation on the Stability of the LCL Injured Elbow: An *In Vitro* Study

Overview

An in vitro kinematic study was conducted to examine the effects of varying the valgus angulation of the custom designed hinged elbow orthosis (HEO) as described and initially tested in Chapter 2. Using an elbow motion simulator, the ulnohumeral kinematics were investigated during active and passive flexion with the forearm in both pronation and supination, with the arm in the vertical dependent and varus positions. As a metric of stability, varus-valgus angulation (VVA) and internal-external rotation (IER) were quantified using an electromagnetic tracking system.

(Note: Some portions of the Introduction and Methods appear in Chapters 2 & 4, and have been repeated, in part, given that this thesis is presented in integrated article format.)

3.1   Introduction

LCL injuries of the elbow most often occur as the result of a dislocation and may cause instability (Babhulkar, 2015; Osbourne & Cotterill, 1966; Reichel, Milam, Sitton, Curry, & Mehlhoff, 2013). Simple elbow dislocations, characterized by the absence of a fracture, tend to be managed non-operatively with a closed reduction (Clitherow, McGuire, & Bain, 2014; Josefsson, Gentz, Johnell, & Wendeberg, 1987; Szekeres, Chinchalkar, & King, 2008). Early mobilization is encouraged to promote healing, prevent elbow stiffness and co-ordinate with the phases of histological healing (Fusaro, Orsini, Szorza, Rotini, & Benedetti, 2014; Mehlhoff, Noble, Bennett, & Tullos, 1988; Szekeres et al., 2008; Wilk, Arrigo, & Andrews, 1993).

Studies have found that active flexion and pronation better stabilize the LCL injured elbow compared to passive flexion and supination (Alolabi et al., 2012; Armstrong et al., 2000; Dunning, Zarzour, Patterson, Johnson, & King, 2001; Fusaro et al., 2014; Szekeres et al., 2008). Passive flexion is usually introduced 6 weeks following injury to reduce the risk of swelling, pain inhibition and ectopic ossification (Cohen & Hastings II, 1998; Dunning et al., 2001; Szekeres et al., 2008). It has also been shown that rehabilitation exercises are safe with the arm in the vertical dependent and overhead positions, but the varus arm position should be avoided (Alolabi et al., 2012; Dunning et al., 2001; Manocha, 2016; Wolff & Hotchkiss, 2006). Following reduction, a hinged elbow orthosis (HEO) or brace, is often prescribed in an effort to stabilize the elbow and encourage early motion (Hunt, 2016; O'Driscoll, Jupiter, King, Hotchkiss, & Morrey, 2000).

The Bledsoe Brace (Bledsoe Brace Systems, Grand Prairie, TX) and the Mayo Clinic Universal Elbow Brace (Aircast, Summit, NJ) as shown in Figures 1-15 and 1-16, are often recommended by surgeons following a simple elbow dislocation (Wolff & Hotchkiss, 2006). However, there is minimal research to support the efficacy of elbow HEOs following ligamentous injury. A study investigating the stabilizing effect of the Bledsoe Elbow Brace on the LCL-deficient elbow found that passive flexion in the vertical dependent position nearly doubled the ulnohumeral distraction compared with the intact elbow; however, the difference was not statistically significant (Lee et al., 2013). Another study using the Mayo Clinic Brace in the vertical dependent position, found that

the HEO did not significantly affect elbow stability during active flexion, and increased instability during passive flexion (Manocha, 2016). One explanation for the lack of effectiveness of current HEO designs is that these devices feature a straight hinge, which does not consider the native carrying angle of the elbow (approximately 11-14° in men, and 13-16° in women) (Atkinson & Elftman, 1945; Morrey & Sanchez-Sotelo, 2009; Paraskevas et al., 2004; Terra et al., 2011; van Roy, Baeyens, Fauvart, Lanssiers, & Clarijs, 2005).

The objective of this study was to examine the kinematic effects of a HEO with different valgus angles following simulated LCL injury of the elbow. A custom designed HEO (as described and evaluated in detail in Chapter 2) was utilized to modify the valgus angulation to five predetermined angles. Elbow flexion was actively and passively conducted with the arm in the vertical dependent and varus positions and with the forearm pronated and supinated. It was hypothesized that increasing the valgus angulation of the HEO would improve stability of the LCL compromised elbow, particularly in the varus arm position.

3.2   Materials and Methods

3.2.1 Specimen Preparation

Five fresh-frozen cadaveric specimens (mean age ± standard deviation: 69.6 ± 7.4) were stored at   -20℃ and thawed at room temperature (22 ± 2℃) for approximately 18 hours. Computer tomography scans of each specimen were examined to ensure that pre-existing pathologies such as arthritis were not present. The distal tendons of the biceps, brachialis, triceps, brachioradialis, pronator teres, wrist flexors (flexor carpi radialis and flexor carpi ulnaris) and wrist extensors (extensor carpi radialis longus and extensor carpi ulnaris) were sutured with braided fishing line (Bravefisherman, Shanghai, China). Guides were secured near the medial and lateral humeral epicondyles to maintain anatomical lines of action (Chapter 2, Figure 2-8). The pronator teres and wrist flexors were passed subcutaneously through the medial guide, while the wrist extensors and brachioradialis were fed through the lateral guides. A custom designed 8 mm stainless steel rod was fixed into the humeral medullary canal with bone cement (Stryker, Kalamazoo, MI, USA). This rod was then mounted onto a custom humeral clamp on the elbow simulator (Chapter 2, Figure 2-10). The biceps, brachialis and triceps sutures were connected to computer-controlled motors, while the remaining muscles were connected to actuators. A custom ulnar tracker mount was rigidly fixed with 3.5 mm cortical screws distally on the medial ulna. It was carefully placed to ensure that soft tissue impingement and HEO obstruction throughout the range of motion were not an issue. An electromagnetic tracker (Flock of Birds®, Ascension Technology Corporation, Burlington, VT, USA) was secured to the tracker mount.

3.2.2 Testing Protocol

Active flexion and extension was simulated using a custom designed LabVIEW program (National Instruments, Austin, TX, USA) to control the motors and actuators (Dunning, Zarzour, Patterson, Johnson, & King, 2001; Ferreira, 2011; Johnson et al., 2000; Kusins, Willing, King, & Ferreira, 2016). Physiologic loads of the biceps, triceps and brachialis were generated by using the maximum voluntary contraction of each muscle based on *in vivo* electromyographic (EMG) values and reported cross section areas (Amis, Dowson,

& Wright, 1979; Funk, An, Morrey, & Daube, 1987). Wrist stabilization was achieved by a 10 N tone load on the wrist flexors and extensors. Pronation was achieved by a tone load of 40 – 60 N applied to the pronator teres. If supination was desired, the ratio of muscle loading was altered to increase the load of the biceps. Passive flexion trials were conducted by the same investigator (SB) grasping the wrist, rotating the forearm into pronation or supination in full extension, and manually flexing the arm. The same forearm rotation was maintained throughout the duration of flexion and extension.

3.2.2.1   Independent Variables

Prior to testing, five active and five passive preconditioning flexion-extension motions were conducted on the intact elbow in pronation and supination. Two active and two passive elbow flexion and extension exercises were simulated during testing. Specimens were tested with the forearm pronated and supinated in both the vertical dependent and varus arm positions. LCL-injury was then simulated by sectioning both the LCL and common extensor origin (CEO) off the lateral epicondyle. All skin incisions were sutured during testing to keep the soft tissues moist. The testing protocol was repeated for the injured state. Then, the HEO was secured to the arm so that the hinge mechanism aligned with the elbow flexion-extension axis, determined by palpating the medial and lateral epicondyles. Once it was placed in a satisfactory location, the straps were tightly secured. The testing sequence was repeated with the HEO at $0°$, $5°$, $10°$, $15°$ and $20°$ of valgus angulation.

Two trials were conducted for each active and passive flexion motions. For active flexion, an average of the two trials was taken. For passive motion, only the second trial was used for data analysis. The rationale for this decision was detailed in Chapter 2 (Section 2.3). Five active flexion trials resulted in sufficiently adequate intraclass correlation coefficients (ICCs) (ICC > 0.996). During passive flexion the ICC values were much lower (between 0.425 and 0.956). However, when the first trial was omitted, all ICC values were greater than 0.959, indicating that the second to fifth trials are much more similar.

3.2.3 Data Acquisition and Analysis

The Flock of Birds® (Ascension Technology Corporation, Burlington, VT, USA) electromagnetic tracking system was utilized to quantify varus-valgus angulation (VVA) and internal-external rotation (IER) of the ulna relative to the humerus (Chapter 1, Figure 1-19). A transmitter, rigidly fixed to the base of the simulator recorded the location of the ulnar receiver throughout testing. Following testing, the wrist and elbow joints were disarticulated for digitization. A stylus was used to digitize the center of the humeral shaft, the center of the trochlear groove and the center of curvature of the capitellum to create the humeral coordinate system (Figure 3-1). For the ulnar coordinate system, the center and plane of the greater sigmoid notch as well as the tip of the ulnar styloid were digitized. Using the Euler Z-Y-X sequence, the relative motion of the ulna with respect to the humerus was determined. Using a custom LabVIEW program, VVA and IER measurements at each flexion and extension angle were calculated (Ferreira, 2011; Kusins, 2015; Manocha, 2016). For simplicity, only flexion data will be presented.

3.2.4 Statistical Analysis

The VVA and IER variables were individually analyzed using SPSS 25.0 statistics software (SPSS Inc, Chicago, IL, USA). Analyses were also completed for the vertical dependent and varus positions separately. A two-way repeated measures of variance (ANOVA) was performed to compare the effects of flexion angle and the HEO, LCL injured and intact states. If significance was found, post hoc tests were conducted to compare HEO angles, LCL injured and intact states to each other. Pairwise comparisons were employed using Bonferroni adjustments. Statistical significance was considered at $p < 0.05$.
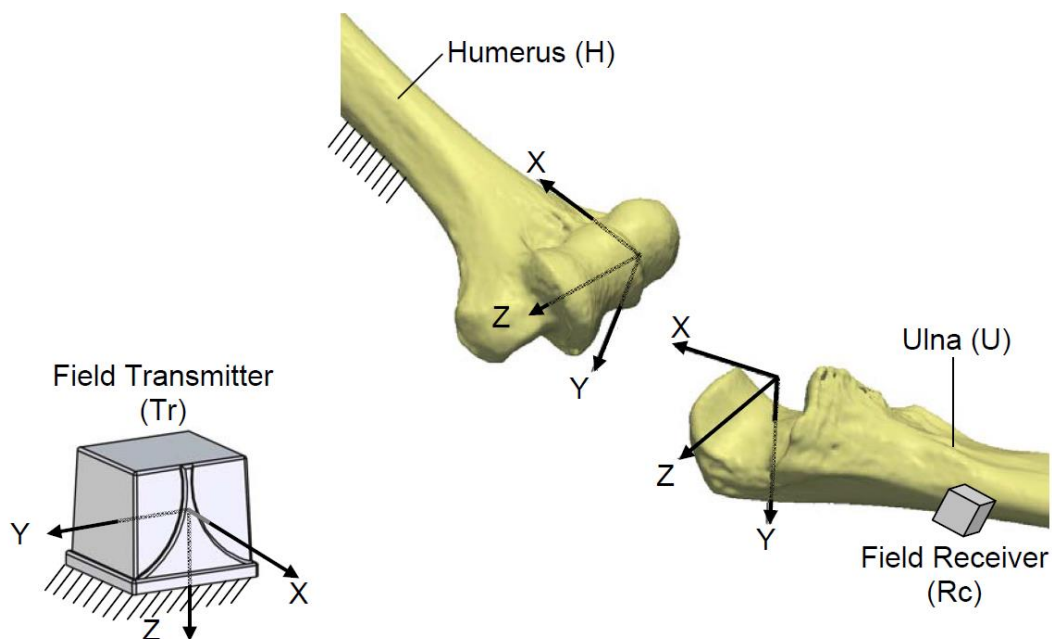
Figure 3-1: Humeral and Ulnar Coordinate Systems. *A field transmitter (Tr) is rigidly fixed to the base of the simulator and the humeral shaft, trochlear groove and center of curvature of the capitellum are digitized to determine the humeral coordinate system relative to the transmitter. An ulnar tracker (Rc), fixed to the ulna (U), is used. Digitization of the greater sigmoid notch and ulnar styloid was conducted to derive the ulnar coordinate system. The origin of the coordinate systems are located at the center of joint rotation, with the X-axis pointing proximally, the Y-axis posteriorly, and the Z-axis medially. The distal humerus and proximal ulna of a left arm are shown (Ferreira, 2011).*

3.3 Results[1]

3.3.1 Varus-Valgus Angulation (VVA) during Flexion Motion

3.3.1.1   Varus-Valgus Angulation (VVA) in the Vertical Dependent Position

During active flexion in the vertical dependent position and forearm pronated, there was no significant effect of LCL sectioning or HEO application (intact and LCL injured states, with and without a HEO) ($p = 0.24$) (Figure 3-2 and Table 3-1). The differences between each LCL injured state to the intact state ranged from 0.29° to 0.48°.

During passive flexion with the forearm pronated, there was no significant effect of LCL sectioning or HEO application ($p = 0.30$). The angular differences between the intact and LCL injured states were between 0.06° and 2.36°.

Active flexion with the forearm supinated had no significant effect of LCL sectioning or HEO application ($p = 0.17$). The differences between means of the intact to injured states was between 0.29° and 0.87°.

During passive flexion with the forearm supinated, there was no significant effect of LCL sectioning or HEO application ($p = 0.12$). The differences between the intact and LCL injured states ranged from 0.86° to 4.48°.

As documented in Appendix C, Table C-1, during active flexion with the forearm pronated, higher angles of elbow flexion significantly increased instability ($p < 0.05$). During passive flexion with the forearm pronated, there was no significant effect of flexion angle ($p > 0.05$). Similarly, there was no significant effect of flexion angle with the forearm supinated during active ($p > 0.05$) or passive flexion ($p > 0.05$).

_____

[1]For simplicity, all pairwise comparisons that depicted significance are summarized in Appendix C.
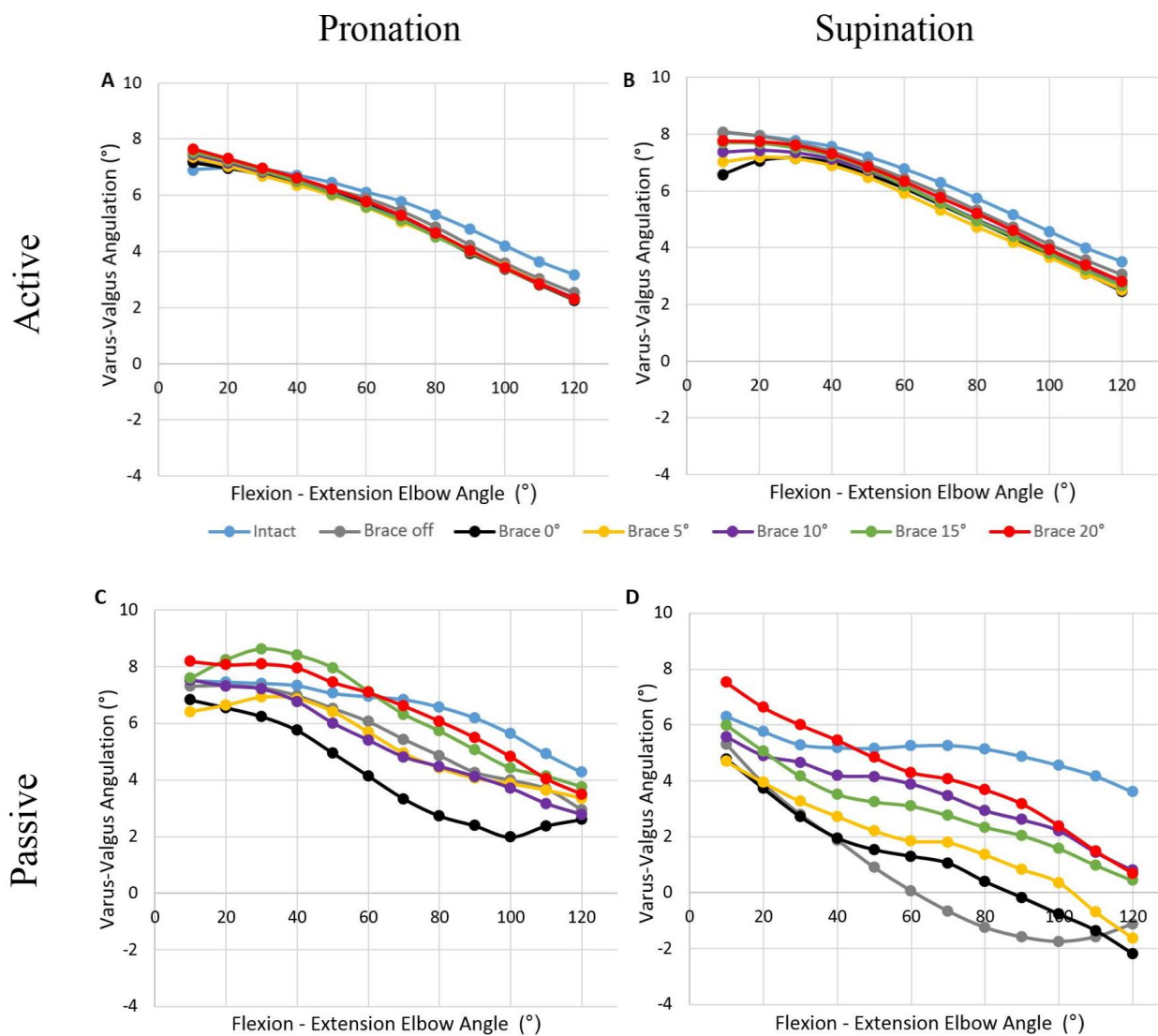
Figure 3-2: Mean Varus-Valgus Angulation during Elbow Flexion in the Vertical Dependent Arm Position. *The graph illustrates the varus-valgus angulation profiles ((+) valgus and (-) varus)) during active pronation (A) and supination (B), as well as passive pronation (C) and supination (D). Intact and injured states are shown with and without a HEO at various valgus angulations. Standard deviations were omitted for clarity but range as follows: active pronation (2.42 - 2.65), active supination (2.40 - 2.74), passive pronation (2.47 - 3.66) and passive supination (3.00 - 4.51).*

Table 3-1: Effect of HEO Angulation, Forearm Rotation and Muscle Activation on Varus-Valgus Angulation during Flexion in the Vertical Dependent Arm Position

Forearm Rotation	Muscle Activation			Mean ± SD Varus-Valgus Angle (°)	Difference (°)	p
Pronation	Active		Intact	5.58 ± 2.42	-	0.24
		LCL Injury	No HEO	5.29 ± 2.51	0.29	
			HEO 0°	5.10 ± 2.47	0.48	
			HEO 5°	5.11 ± 2.48	0.47	
			HEO 10°	5.17 ± 2.56	0.41	
			HEO 15°	5.16 ± 2.65	0.42	
			HEO 20°	5.26 ± 2.62	0.32	
	Passive		Intact	6.51 ± 2.52	-	0.30
		LCL Injury	No HEO	5.55 ± 2.47	0.96	
			HEO 0°	4.15 ± 3.66	2.36	
			HEO 5°	5.27 ± 2.98	1.24	
			HEO 10°	5.27 ± 3.43	1.24	
			HEO 15°	6.44 ± 2.94	0.07	
			HEO 20°	6.45 ± 3.72	0.06	
Supination	Active		Intact	6.22 ± 2.40	-	0.17
		LCL Injury	No HEO	5.93 ± 2.55	0.29	
			HEO 0°	5.38 ± 2.70	0.84	
			HEO 5°	5.35 ± 2.67	0.87	
			HEO 10°	5.59 ± 2.70	0.63	
			HEO 15°	5.65 ± 2.74	0.57	
			HEO 20°	5.78 ± 2.67	0.44	
	Passive		Intact	5.05 ± 2.37	-	0.12
		LCL Injury	No HEO	0.57 ± 4.51	4.48	
			HEO 0°	1.08 ± 4.50	3.97	
			HEO 5°	1.72 ± 3.00	3.33	
			HEO 10°	3.40 ± 3.28	1.65	
			HEO 15°	2.93 ± 3.40	2.12	
			HEO 20°	4.19 ± 3.11	0.86	

Positive means indicate valgus angulation and negative values indicate varus angulation. "Difference" describes the difference in varus-valgus angulation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from a two-way analysis of variance (ANOVA) with ligament state and flexion angle. The asterisk (*) indicated significance ($p < 0.05$). Abbreviations: SD – standard deviation; HEO – hinged elbow orthosis and LCL - lateral collateral ligament.

3.3.1.2 Varus-Valgus Angulation (VVA) in the Varus Position

In the varus position, during active flexion with the forearm in pronation there was a significant effect of LCL sectioning and HEO application (p = 0.01) (Figure 3-3 and Table 3-2). Without the HEO, the mean VVA difference increased 6.62° after LCL sectioning (p = 0.00). Application of the HEO did not improve the stability of the LCL deficient elbow for any HEO angulation; the elbows remained unstable compared to the intact state (HEO 0°; p = 0.03; HEO 5°, p = 0.01). However, there was a decreasing trend in the VVA difference between the intact and injured state as the valgus HEO angle increased (from 6.07° to 2.81°). At HEO angles of 10° and greater there was no significant difference between the LCL deficient elbow and the intact state (HEO 10°, p = 0.09; HEO 15°, p = 0.06, HEO 20°, p = 0.07).

With the forearm supinated during active flexion, LCL sectioning and HEO application were found to significantly increase instability (p = 0.00), therefore post-hoc tests were performed. LCL injured states without the HEO and up to an HEO valgus angulation of 15° significantly reduced stability compared to the intact state (p values from 0.00 to 0.03). Increasing the angle of the HEO trended towards reducing instability, and was not significantly worse than the intact elbow at 20° of HEO angulation (p = 0.09).

There was no effect of flexion angle during passive elbow flexion with the forearm in pronation (p = 0.08) or supination (p = 0.10).

During active flexion in pronation and supination, instability was greatest in the mid flexion range, p < 0.05 and p< 0.04, respectively (Appendix C, Table C-2).
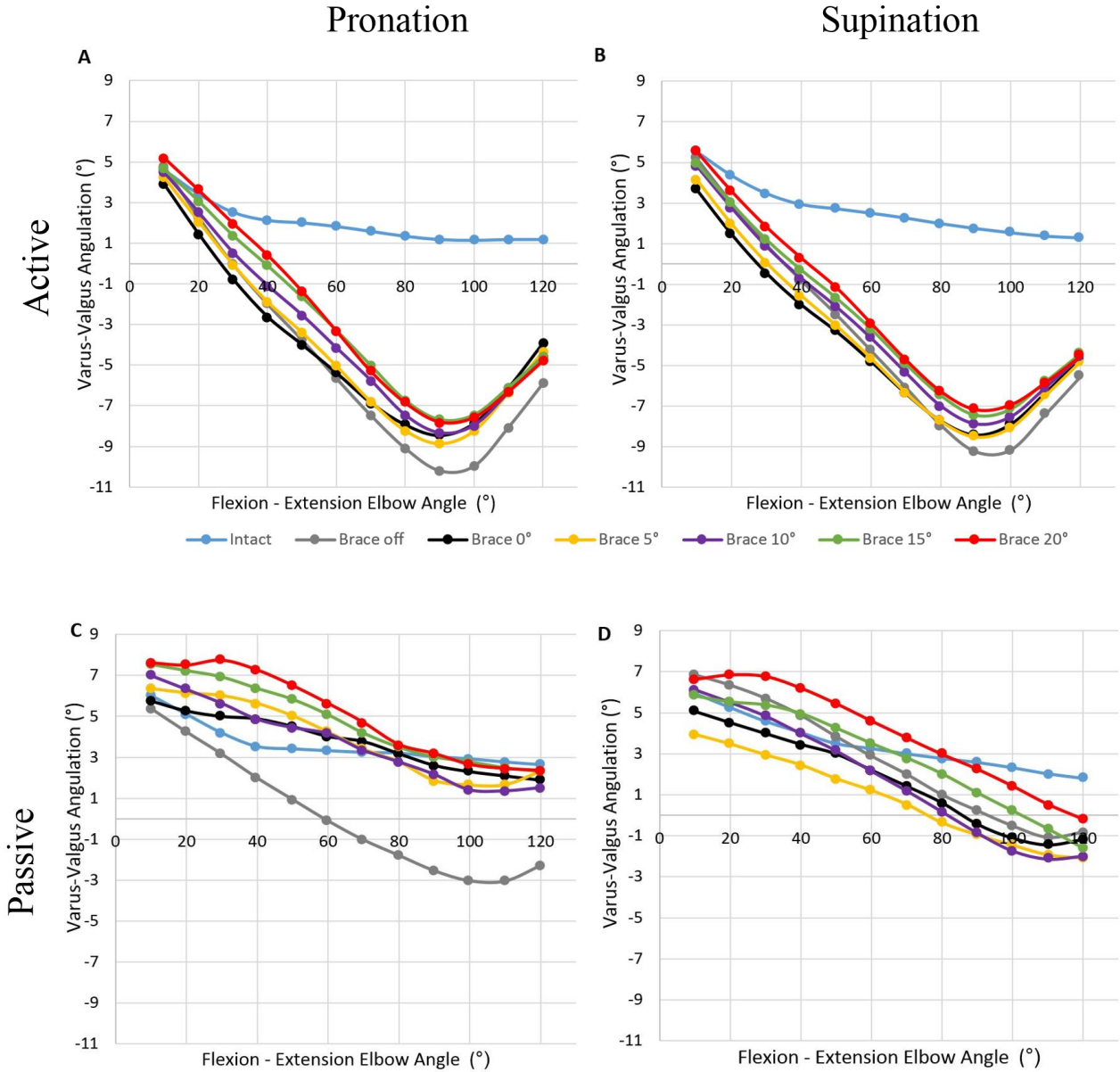
Figure 3-3: Mean Varus-Valgus Angulation during Elbow Flexion in the Varus Arm Position. *The graph illustrates the varus-valgus angulation profiles ((+) valgus and (-) varus)) during active pronation (A) and supination (B), as well as passive pronation (C) and supination (D). Intact and injured states are shown with and without a HEO at various valgus angulations. Standard deviations were omitted for clarity (active pronation: ±2.83 to ±4.97, active supination: ±2.78 to ±4.53, passive pronation: ±2.67 to ±5.27, passive supination: ±2.56 to ±5.89).*

Table 3-2: Effect of HEO Angulation, Forearm Rotation and Muscle Activation on Varus-Valgus Angulation during Flexion in the Varus Arm Position

Forearm Rotation	Muscle Activation			Mean ± SD Varus-Valgus Angle (°)	Difference (°)	p
Pronation	**Active**		Intact	2.02 ± 3.07	-	-
		LCL Injury	No HEO	-4.60 ± 2.83	6.62	0.00*
			HEO 0°	-4.05 ± 3.99	6.07	0.00*
			HEO 5°	-3.91 ± 3.89	5.93	0.00*
			HEO 10°	-3.39 ± 4.87	5.41	0.09
			HEO 15°	-2.79 ± 4.43	4.81	0.06
			HEO 20°	-2.67 ± 4.97	4.69	0.07
	Passive		Intact	3.62 ± 2.67	-	
		LCL Injury	No HEO	0.17 ± 3.78	3.45	
			HEO 0°	3.78 ± 2.94	0.16	
			HEO 5°	3.93 ± 5.27	0.31	0.08
			HEO 10°	3.76 ± 3.24	0.14	
			HEO 15°	4.78 ± 3.20	1.16	
			HEO 20°	5.11 ± 3.17	1.49	
Supination	**Active**		Intact	2.64 ± 2.78	-	-
		LCL Injury	No HEO	-3.66 ± 3.93	6.30	0.02*
			HEO 0°	-3.92 ± 3.07	6.56	0.01*
			HEO 5°	-3.76 ± 3.16	6.40	0.00*
			HEO 10°	-3.04 ± 4.17	5.68	0.03*
			HEO 15°	-2.69 ± 4.00	5.33	0.02*
			HEO 20°	-2.36 ± 4.53	5.00	0.09
	Passive		Intact	3.43 ± 2.56	-	
		LCL Injury	No HEO	2.61 ± 3.75	0.82	
			HEO 0°	1.67 ± 5.81	1.76	
			HEO 5°	0.80 ± 4.88	2.63	0.10
			HEO 10°	1.70 ± 5.89	1.73	
			HEO 15°	2.77 ± 3.55	0.66	
			HEO 20°	3.95 ± 4.15	0.52	

Positive means indicate valgus angulation and negative values indicate varus angulation. "Difference" describes the difference in varus-valgus angulation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from a two-way analysis of variance (ANOVA) with ligament state and flexion angle. The asterisk (*) indicated significance ($p < 0.05$). Abbreviations: SD – standard deviation; HEO – hinged elbow orthosis and LCL - lateral collateral ligament.

3.3.2 Internal-External Rotation during Flexion Motion

3.3.2.1 Internal-External Rotation (IER) in the Vertical Dependent Position

In the vertical dependent position, there were no significant differences in IER found after LCL sectioning or HEO application during active pronation ($p = 0.13$) and supination ($p = 0.24$) as well as passive pronation ($p = 0.22$) and supination ($p = 0.18$) (Figure 33-4 and Table 3-3).

During active flexion with the forearm pronated, there was a small but significant effect of flexion angle on elbow stability at low angles of flexion ($p < 0.05$) (Appendix C, Table C-3). There was no significant effect of flexion angle during active flexion with the forearm in supination ($p > 0.05$). There was no significant effect of flexion angle on stability during passive flexion with the forearm in pronation ($p > 0.05$) or supination ($p > 0.05$).
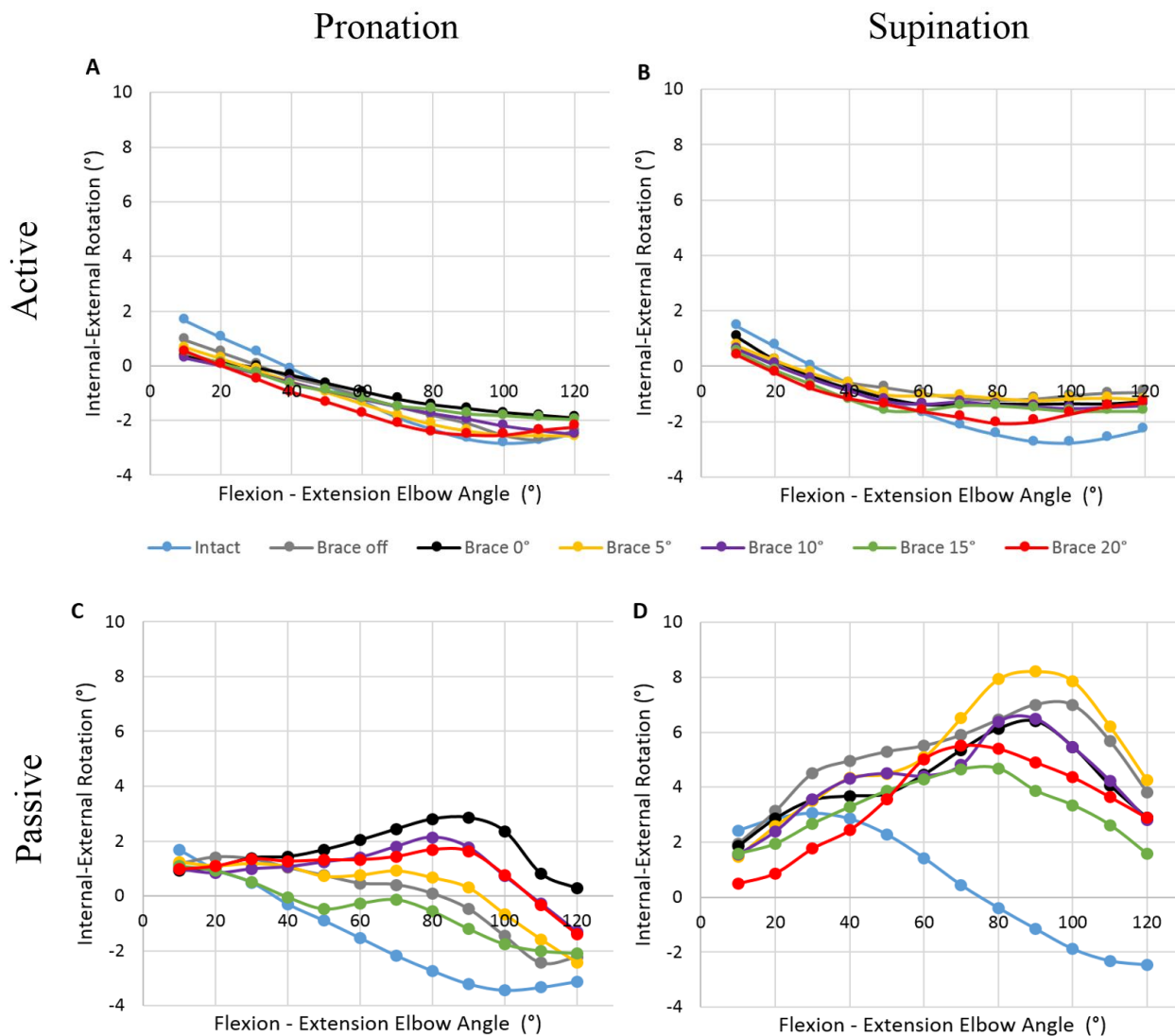
Figure 33-4: Mean Internal-External Rotation during Elbow Flexion in the Vertical Dependent Arm Position. *The graph illustrates the internal-external rotation profiles ((+) external and (-) internal)) during active pronation (A) and supination (B), as well as passive pronation (C) and supination (D). Intact and injured states are shown with and without a HEO at various valgus angulations. Standard deviations were omitted for clarity but range as follows: active pronation (5.45 – 6.26), active supination (5.43 – 6.74), passive pronation (5.19 – 7.05) and passive supination (4.34 – 6.26).*

Table 3-3: Effect of HEO Angulation, Forearm Rotation and Muscle Activation on Internal-External Rotation during Flexion in the Vertical Dependent Arm Position

Forearm Rotation	Muscle Activation			Mean ± SD Internal-External Rotation (°)	Difference (°)	p
Pronation	**Active**		Intact	-1.16 ± 5.46	-	0.13
		LCL Injury	No HEO	-1.16 ± 5.99	0.00	
			HEO 0°	-0.93 ± 6.24	0.23	
			HEO 5°	-1.34 ± 5.65	0.18	
			HEO 10°	-1.25 ± 5.97	0.09	
			HEO 15°	-1.08 ± 6.26	0.08	
			HEO 20°	-1.51 ± 5.45	0.35	
	Passive		Intact	-1.47 ± 5.19	-	0.22
		LCL Injury	No HEO	0.00 ± 5.80	1.47	
			HEO 0°	1.67 ± 6.54	0.20	
			HEO 5°	0.27 ± 5.66	1.20	
			HEO 10°	0.94 ± 6.51	0.53	
			HEO 15°	-0.50 ± 6.09	0.97	
			HEO 20°	0.92 ± 7.05	0.55	
Supination	**Active**		Intact	-1.37 ± 5.43	-	0.24
		LCL Injury	No HEO	-0.72 ± 6.73	0.65	
			HEO 0°	-0.87 ± 6.75	0.50	
			HEO 5°	-0.76 ± 6.73	0.61	
			HEO 10°	-1.00 ± 6.57	0.37	
			HEO 15°	-1.18 ± 6.53	0.19	
			HEO 20°	-1.28 ± 6.48	0.09	
	Passive		Intact	0.59 ± 5.30	-	0.18
		LCL Injury	No HEO	5.10 ± 4.39	4.51	
			HEO 0°	4.20 ± 6.26	3.61	
			HEO 5°	5.21 ± 4.34	4.62	
			HEO 10°	4.24 ± 5.82	3.65	
			HEO 15°	3.19 ± 5.49	2.60	
			HEO 20°	3.40 ± 4.41	2.81	

Positive means indicate external rotation and negative values indicate internal rotation. "Difference" describes the difference in internal-external rotation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from a two-way analysis of variance (ANOVA) with ligament state and flexion angle. The asterisk (*) indicated significance (p < 0.05). Abbreviations: SD – standard deviation; HEO – hinged elbow orthosis and LCL - lateral collateral ligament.

3.3.2.1 Internal-External Rotation (IER) in the Varus Position

In the varus position, elbow stability was significantly worse following LCL injury and HEO application during active flexion with the forearm in both pronation (p = 0.00) and supination p = 0.00) (Figure 3-5 and Table 3-4). Post-hoc test revealed that the injured state with the HEO at 5° worsened stability in pronation (p = 0.03) and supination (p = 0.04). The injured state with the HEO removed was also significantly worse with the forearm supinated (p = 0.01).

During passive flexion with the forearm supinated, there was no significant effect of the LCL injury and HEO application (p = 0.17). The differences between means of the intact to injured states was 2.95° without the HEO, and was less than 1° for all HEO angles (difference ranging between 0.14° and 0.75°). Similarly, no significance of ligament state was found during passive flexion and forearm supinated (p = 0.17). The differences between the intact and LCL injured states ranged from 0.08° (HEO 20°) to 2.85° (HEO 0°).

With respect to active pronation and supination there was greater instability in the mid flexion range (approximately 80° to 100° of flexion) (p < 0.04) (Appendix C, Table C-4). Additionally, there was no significant effect of flexion angle on elbow stability during passive flexion with the forearm in pronation or supination (p > 0.05).
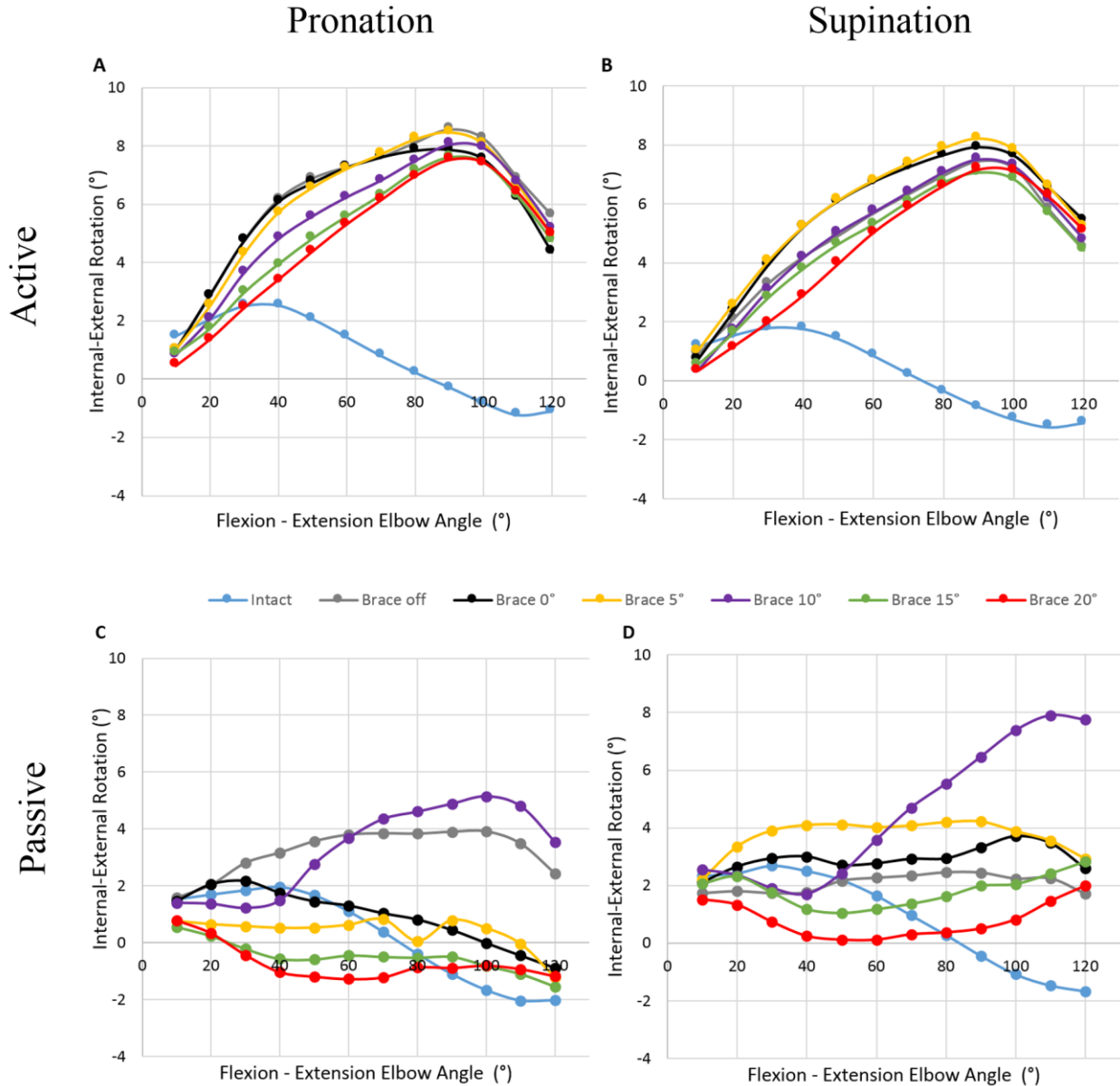
Figure 3-5: Mean Internal-External Rotation during Elbow Flexion in the Varus Position. *The graph illustrates the internal-external rotation profiles ((+) external and (-) internal)) during active pronation (A) and supination (B), as well as passive pronation (C) and supination (D). Intact and injured states are shown with and without a HEO at various valgus angulations. Standard deviations were omitted for clarity but range as follows: active pronation (5.63 – 6.27), active supination (5.59 – 6.33), passive pronation (5.66 – 6.98) and passive supination (5.55 – 7.37).*

Table 3-4: Effect of HEO Angulation, Forearm Rotation and Muscle Activation on Internal-External Rotation during Flexion in the Varus Arm Position

Forearm Rotation	Muscle Activation			Mean ± SD Internal-External Rotation (°)	Difference (°)	p
Pronation	**Active**	Intact		0.81 ± 5.63	-	-
		LCL Injury	No HEO	6.17 ± 5.64	5.36	0.09
			HEO 0°	5.84 ± 6.08	5.03	0.05
			HEO 5°	5.96 ± 5.78	5.15	0.03*
			HEO 10°	5.45 ± 6.27	4.64	0.08
			HEO 15°	4.94 ± 6.10	4.13	0.06
			HEO 20°	4.73 ± 6.02	3.92	0.19
	Passive	Intact		0.25 ± 5.66	-	
		LCL Injury	No HEO	3.20 ± 6.55	2.95	
			HEO 0°	0.93 ± 6.03	0.68	
			HEO 5°	0.39 ± 6.98	0.14	0.10
			HEO 10°	0.47 ± 6.43	0.22	
			HEO 15°	-0.50 ± 6.42	0.75	
			HEO 20°	-0.72 ± 6.13	0.47	
Supination	**Active**	Intact		0.27 ± 5.63	-	-
		LCL Injury	No HEO	4.94 ± 5.59	4.67	0.01*
			HEO 0°	5.63 ± 5.80	5.36	0.06
			HEO 5°	5.74 ± 5.72	5.47	0.04*
			HEO 10°	4.94 ± 5.92	4.67	0.09
			HEO 15°	4.63 ± 6.02	4.36	0.07
			HEO 20°	4.46 ± 6.33	4.19	0.00
	Passive	Intact		0.87 ± 5.64	-	
		LCL Injury	No HEO	2.08 ± 5.55	1.21	
			HEO 0°	2.93 ± 6.90	2.06	
			HEO 5°	3.72 ± 7.37	2.85	0.17
			HEO 10°	2.62 ± 7.07	1.75	
			HEO 15°	1.82 ± 6.23	0.95	
			HEO 20°	0.79 ± 6.31	0.08	

Positive means indicate external rotation and negative values indicate internal rotation. "Difference" describes the difference in internal-external rotation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from a two-way analysis of variance (ANOVA) with ligament state and flexion angle. The asterisk (*) indicated significance ($p < 0.05$). Abbreviations: SD – standard deviation; HEO – hinged elbow orthosis and LCL - lateral collateral ligament

3.4 Discussion

LCL injures, commonly resulting from elbow dislocations, are often treated with a hinged elbow orthosis (HEO) (de Haan et al., 2011; Fusaro et al., 2014; Reichel et al., 2013). A greater understanding of the kinematic implications of bracing is required to better diagnose and treat cases of LCL instability. Elbow kinematics have been previously reported during simulated *in vitro* LCL injury (Alolabi et al., 2012; Dunning et al., 2001; Lee et al., 2013; Manocha, 2016; Olsen, Michael, Søjbjerg, Helmig, & Sneppen, 1966). However, there has been minimal exploration into the effect of bracing (Lee et al., 2013; Manocha, 2016). This *in vitro* study investigated the effects of varying the valgus angulation of a custom designed HEO during simulated LCL injury. Active and passive flexion-extension motions were conducted in the vertical dependent and varus arm positions, during pronation and supination. The varus-valgus angulation (VVA) and internal-external rotation (IER) of the ulna relative to the humerus were reported throughout the range of motion.

In agreement with other studies, active flexion in the vertical dependent position with the forearm in pronation, with or without the HEO, is safe following LCL injury (Manocha, 2016). The VVA and IER means were within 0.5° after LCL sectioning with or without HEO application compared to the intact state. This suggests that the HEO does not increase instability in the vertical dependent position and may be used to hold the forearm in pronation, or restrict range of motion if needed.

Contrary to other studies, during active flexion with the forearm in supination and the arm oriented in the vertical dependent position there was no effect of LCL sectioning on elbow stability (Dunning et al., 2001; Manocha, 2016). One reason for this may be that prior studies may have sectioned more secondary stabilizers than the current study, which only cut the LCL, common extensor origin and the lateral half of the anterior capsule.

During active flexion with the forearm supinated in the dependent position, application of the orthosis did not significantly worsen stability. It was hypothesized that the weight of the HEO (approximately 1 kg) may have added an additional downward force of about 10 N, however in this position, it did not impact stability with active elbow flexion and the

forearm in supination. The medial-lateral support provided by the HEO may have negated the added downward force.

With regard to passive flexion in the vertical dependent position and the forearm in pronation, LCL sectioning and the HEO did not significantly impact stability. Passive flexion is usually avoided until approximately 6 weeks following injury (Alolabi et al., 2012; Dunning et al., 2001; Fraser et al., 2008), however the kinematic results would likely vary based on the moments and forces applied to the arm. For this study, passive flexion was performed by the same operator (SB), likely more consistent than those produced by different investigators.

There was no significant effect of LCL injury and HEO application during passive flexion with the forearm supinated in the vertical dependent position. However, stability was significantly worse in the mid flexion range (between 80° and 120°) with the HEO removed, with respect to the IER. This is in agreement with other studies that passive supination should be avoided (Alolabi et al., 2012; Dunning et al., 2001; Lee et al., 2013; Manocha, 2016). The orthosis at 5° of angulation also significantly worsened stability between 80° and 90° of flexion, however interestingly, there was no significant impact on stability with the HEO at 0°, 10°, 15° and 20°, suggesting that these angles may be sufficient to stabilize the elbow in this position. Orthosis angles of 15° and 20° had means that were closest to the intact state, which perhaps suggests that these angles restored stability the most. The mean carrying angle for the specimens tested was approximately 8.2° suggesting that an overcorrection of the HEO may protect the lateral structures and allow the elbow to maintain stability. The data supports this postulation, as there is an overcorrection during full extension, which causes less variation as the elbow is flexed. This could have important clinical implications on rehabilitation and bracing protocols.

LCL injury and HEO application worsened stability in the varus position up to 15° of orthosis angulation during simulated active flexion with the forearm pronated and supinated. Manocha (2016) reported similar findings for the injured state and with a HEO at 0° of angulation. During full extension, the arm is the most stable but stability

drastically decreases during flexion, with the most instability apparent between approximately 70° and 100°. This effect may be attributed to varying gravitational moments and the stabilizing effects of the interlocking articulations in full extension and flexion. The anatomical lines of action of the muscles compress the joint in full extension also, improving stability. As the elbow is flexed, the compressive effect of the muscles stabilizing the joint articulation decreases and the weight of the forearm and HEO increases the gravitational moment and contributes to increased instability. Interestingly, no significant differences were found at any flexion angles with the orthosis at 20° of valgus angulation. This may indicate that rehabilitative exercises could be safely performed with the HEO at 20° of valgus angulation.

Passive elbow flexion with the forearm in both pronation and supination and the elbow oriented in the varus position did not significantly impact stability following LCL injury and HEO application. With respect to pronation, introduction of the HEO improved stability compared the injured state without the HEO. The brace at 0° most closely followed the trend of the intact state. In supination, the brace at 20° came closest to restoring initial kinematics. However, it is difficult to draw solid conclusions from passive data, as it varies by investigator. By grasping the specimen under the wrist to achieve passive flexion, the arm was held up to some extent to allow full motion and prevent the arm from dropping. In doing so, part of weight of the arm and HEO was absorbed by the investigator, reducing the load through the LCL. If passive flexion is to be performed, it should only be done so by a trained professional during the rehabilitation of elbow LCL injuries.

With the forearm in both pronation and supination, in the vertical dependent and varus positions, the 20° HEO angles tended to improve stability, as no significant differences were found at any flexion angles, however further specimens should be tested to confirm these trends. As previously mentioned, the mean carrying angle was approximately 8.2°, suggesting that a valgus overcorrection may help improve elbow stability. Active flexion with the forearm in pronation in the vertical dependent position restored stability better than in supination, however neither differences were significant. Other *in vitro* studies have also concluded that pronation better stabilizes the LCL- deficient elbow, therefore

the HEO may be useful at maintaining pronation in this position (Armstrong et al., 2000; Dunning et al., 2001; Manocha, 2016).

For this study, the first passive trial was omitted and only the data obtained from the second trial was used for analysis. As discussed in Chapter 2 (Section 2.3), the reliability during passive flexion trials was much greater with the first trial removed. This indicated that the second to fourth trials were more similar to each other than to the first trial and validates the decision to exclude the first trial.

The study presented in this chapter has several strengths. No other studies have analyzed the effect of adjusting the carrying angle of a HEO on stability. These observations have important implications for rehabilitation protocols. This study shows that an overcorrection using a more valgus HEO angle may be useful to improve stability of the LCL deficient elbow. Our data also shows that in the setting of LCL insufficiency, greater angles of flexion should be avoided as this increased instability. Only two other studies (Lee et al., 2013; Manocha, 2016) have investigated the efficacy of HEOs on the LCL injured elbow, however both used HEOs with no valgus angulation. This was also the second study to investigate active flexion with a HEO in the vertical dependent and varus positions, and similar results were found (Manocha, 2016).

Both VVA and IER were analyzed in this study. When analyzing IER data in the varus position during active flexion, there was no significant difference in stability found with the HEO removed when the forearm was pronated. However, analysis of the VVA data suggests that stability was significantly worsened during active flexion with the arm in pronation and the elbow in the vertical dependent position. This highlights the importance of considering multiple parameters to quantify elbow stability. The only other active flexion study investigating the effectiveness of HEOs only reported IER to quantify stability; our data suggests that VVA should also be considered for future investigations (Manocha, 2016).

Another strength of this study is that a custom designed HEO was used. The kinematic measurements obtained with the HEO were evaluated to ensure reliability (Chapter 2, Section 2.3). During simulated active flexion, tone loads on the brachioradialis, pronator

teres (during pronation only) and wrist flexors and extensors were employed to mimic the *in vivo* state. It is important to include these loads, since the brachioradialis contributes to elbow flexion, and the wrist flexors and extensors contribute to elbow stability (King, Morrey, & An, 1993). All skin incisions were sutured for the duration of the testing day to maintain viscoelastic behavior of the soft tissues. Highly accurate electromagnetic tracking was utilized for real-time joint angle feedback and allowed accurate kinematic markers of stability to be calculated. Lastly, vertical dependent and varus arm positions were selected, as they are among the positions where patients spend the most time during activities of daily living (Morrey, Askew, An, & Chao, 1981).

This study is not without limitations. A preliminary power analysis using 2 degrees as the desired detectable range for significance, indicated that 4 specimens would be sufficient to detect kinematic elbow instability. Regardless, our sample size was perhaps relatively small, although statistical power was not insufficient. However, we were unable to demonstrate that the custom designed HEO with a greater valgus angulation significantly reduced elbow instability after LCL injuries as we had expected. Certainly with the arm in the varus orientation the brace had little effect and this position should continue to be avoided by patients after LCL injuries. The overall observed statistical power for a comparison of LCL injured states and flexion angle was 0.991 and 0.987 for VVA and IER, respectively, indicating that enough specimens were tested for the comparisons reported. In order to analyze other variables such as muscle activation or forearm rotation, more specimens are required, however, for this study, the power is sufficient. Another limitation of this study is that the straps of the HEO were tightly secured to arm, perhaps representing a tighter than average tension used clinically. However, it was necessary to ensure consistency of the hinge axis placement and ensure that the HEO did not slide down the arm. Furthermore, the lack of muscle contraction and proprioceptive feedback from the use of a brace may affect the performance of a HEO when used clinically. Due to the nature of cadaveric studies, the specimens were elderly and likely had decreased muscle mass relative to a younger population, which may have effected stability. However, in a qualitative sense, unacceptable laxity between the HEO and arm was not noted. This study was conducted on male specimens only. It would be interesting to correlate these findings among women, as the large weight of the larger male arms

(mean arm weight: 14.7 ± 3.0 lbs) would have increased the downward force and may have affected these findings by increasing ulnohumeral distraction. Also, the difference in carrying angle between males and females may be a factor, although perhaps not markedly. Finally, no simulated object lifting was performed, although lifting with the arm is not recommended after ligament injuries in patients while in the rehabilitation phase.

3.5 Conclusion

The current study supports the hypothesis that increasing the valgus angulation of the HEO increases stability for the activities assessed herein, however further studies are required to confirm the hypotheses. To our knowledge, this is the first study investigating the effect of varying the valgus angulation of a HEO following LCL injury. The degree of flexion proved to have an effect on stability, with the most instability occurring around 80° to 100° of flexion. In the vertical dependent position, active flexion with the forearm in both pronation and supination was deemed safe for rehabilitation with and without the use of an orthosis. Passive flexion with the forearm in pronation also did not cause significant instability. In this position, passive flexion with the forearm in supination improved stability at a 20° HEO angle. Active flexion in the varus position caused instability after LCL sectioning up to 15° of HEO angle, however there was a trend towards improved stability as the HEO angle increased and no significant differences were detected with the orthosis angled at 20°. During passive flexion, no significant changes in stability were found, irrespective of HEO angle, however further studies are required. Generally, 20° was the optimal valgus angle for all injured states investigated. Further clinical studies are needed to increase our understanding of the importance of the HEO angle on patient outcomes.

3.6 References

Alolabi, B., Gray, A., Ferreira, L. M., Johnson, J. A., Athwal, G. S., & King, G. J. W. (2012). Rehabilitation of the medial- and lateral collateral ligament-deficient elbow: An in vitro biomechanical study. *Journal of Hand Therapy*, *25*(4), 363–373. https://doi.org/10.1016/j.jht.2012.06.001

Amis, A. A., Dowson, D., & Wright, V. (1979). Muscle strengths and musculo-skeletal geometry of the upper limb. *Engineering in Medicine*, *8*(1), 41–48.

Armstrong, A. D., Dunning, C. E., Faber, K. J., Duck, T. R., Johnson, J. A., & King, G. J. W. (2000). Rehabilitation of the medial collateral ligament-deficient elbow: an in vitro biomechanical study. *The Journal of Hand Surgery*, *25*(6), 1051–1057.

Atkinson, W. B., & Elftman, H. (1945). The carrying angle of the human arm as a secondary sex character. *The Anatomical Record*, *91*(1), 49–52.

Babhulkar, S. (2015). *Elbow injuries* (illustrate). New Delhi, India: JP Medical Ltd.

Clitherow, H. D. S., McGuire, D. T., & Bain, G. I. (2014). Lateral elbow instability. *Asia-Pacific Journal of Sports Medicine, Arthroscopy, Rehabilitation and Technology*, *1*(1), 11–18.

Cohen, M. S., & Hastings II, H. (1998). Acute Elbow Dislocation : Evaluation and Management, *6*(1), 15–23.

de Haan, J., Schep, N. W. L., Eygendaal, D., Kleinrensink, G.-J., Tuinebreijer, W. E., & den Hartog, D. (2011). Stability of the elbow joint: relevant anatomy and clinical implications of in vitro biomechanical studies. *The Open Orthopaedics Journal*, *5*, 168–176. https://doi.org/10.2174/1874325001105010168

Dunning, C. E., Zarzour, Z. D. S., Patterson, S. D., Johnson, J. A., & King, G. J. W. (2001). Muscle forces and pronation stabilize the lateral ligament deficient elbow. *Clin Orthop Relat Res*, *388*, 118–124.

Ferreira, L. M. (2011). *Development of an Active Elbow Motion Simulator and Coordinate Systems to Evaluate Kinematics in Multiple Positions*. The University of Western Ontario.

Fraser, G. S., Pichora, J., Ferreira, L. M., Brownhill, J., Johnson, J. A., & King, G. J. W. (2008). Lateral Collateral Ligament Repair Restores the Initial Varus Stability of the Elbow: An In Vitro Biomechanical Study. *Journal of Orthopaedic Trauma*, *22*(9), 615–623.

Funk, D. A., An, K. N., Morrey, B. F., & Daube, J. R. (1987). Electromyographic analysis of muscles across the elbow joint. *Journal of Orthopaedic Research*, *5*, 529–538.

Fusaro, I., Orsini, S., Szorza, T., Rotini, R., & Benedetti, M. G. (2014). The use of braces in the rehabilitation treatment of the post-traumatic elbow. *Joints*, *2*(4), 81–86.

Hunt, T. R. (2016). *Operative Techniques in Hand, Wrist, and Elbow Surgery* (2nd ed.). Lippincott Williams & Wilkins.

Johnson, J. A., Rath, D. A., Dunning, C. E., Roth, S. E., & King, G. J. W. (2000). Simulation of elbow and forearm motion in vitro using a load controlled testing apparatus. *Journal of Biomechanics*, *33*(5), 635–639. https://doi.org/10.1016/S0021-9290(99)00204-3

Josefsson, P. O., Gentz, C. F., Johnell, O., & Wendeberg, B. (1987). Surgical versus non-surgical treatment of ligamentous injuries following dislocation of the elbow joint. A prospective randomized study. *JBJS*, *69*(4), 605–608.

King, G. J. ., Morrey, B., & An, K. N. (1993). Stabilizers of the Elbow. *J. Shoulder Elb. Surg*, *2*(3), 165–174.

Kusins, J. (2015). *Experimenal Analysis of Parameters Influencing the Bone Burring Process*. The University of Western Ontario.

Kusins, J., Willing, R., King, G., & Ferreira, L. (2016). Development of a computational elbow model with experimental validation of kinematics and muscle forces. *J Appl Biomech*, *32*(4), 407–14.

Lee, A. T., Schrumpf, M. A., Choi, D., Meyers, K. N., Patel, R., Wright, T. M., … Daluiski, A. (2013). The influence of gravity on the unstable elbow. *J Shoulder Elb Surg*, *22*(1), 81–87. https://doi.org/http://dx.doi.org/10.1016/j.jse.2012.08.012

Manocha, R. H. K. (2016). *Optimizing the Rehabilitation of Elbow Lateral Collateral Ligament Injuries*. The University of Western Ontario.

Mehlhoff, T., Noble, P., Bennett, J., & Tullos, H. (1988). Simple dislocation of the elbow in the adult. Results after closed treatment. *J Bone Joint Surg Am*, *70*(2), 244–249.

Morrey, B. F., Askew, L. J., An, K.-N., & Chao, E. Y. (1981). A biomechanical study of normal elbow motion. *J Bone Joint Surg*, *63–A*(6), 872–877. Retrieved from http://www.ncbi.nlm.nih.gov/pubmed/7240327

Morrey, B. F., & Sanchez-Sotelo, J. (2009). *The Elbow and Its Disorders* (4th ed.). Philadelphia: Elsevier Health Sciences.

O'Driscoll, S. W., Jupiter, J. B., King, G. J. W., Hotchkiss, R. N., & Morrey, B. F. (2000). The unstable elbow. *Journal of Bone and Joint Surgery*, *82*(5), 724–738.

Olsen, B. S., Michael, T. V., Søjbjerg, J. O., Helmig, P., & Sneppen, O. (1966). Lateral collateral ligament of the elbow joint: Anatomy and kinematics. *Journal of Shoulder and Elbow Surgery*, *5*(2), 103–112.

Osbourne, G., & Cotterill, P. (1966). Recurrent dislocation of the elbow. *Bone & Joint Journal*, *48*(2), 340–346.

Paraskevas, G., Papadopoulos, A., Papaziogas, B., Spanidou, S., Argiriadoi, H., & Gigis, J. (2004). Study of the carrying angle of the human elbow joint in full extension: a morphometric analysis. *Surg Radiol Anat*, *26*(1), 19–23.

Reichel, L. M., Milam, G. S., Sitton, S. E., Curry, M. C., & Mehlhoff, T. (2013). Elbow lateral collateral ligament injuries. *The Journal of Hand Surgery*, *38*(1), 184–201.

Szekeres, M., Chinchalkar, S. J., & King, G. J. W. (2008). Optimizing Elbow Rehabilitation After Instability. *Hand Clinics*, *24*(1), 27–38. https://doi.org/10.1016/j.hcl.2007.11.005

Terra, B. B., Silva, B. C. M., Carvalho, H. B. F. D., Dobashi, E. T., Pinto, J. A., & Ishida, A. (2011). Evolution of the carrying angle of the elbow: a clinical and radiographic study. *Acta Ortopédica Brasileira*, *19*(2), 79–82.

van Roy, P., Baeyens, J. P., Fauvart, D., Lanssiers, R., & Clarijs, J. P. (2005). Arthro-kinematics of the elbow: study of the carrying angle. *Ergonomics*, *48*(11–14), 19–21.

Wilk, K. E., Arrigo, C., & Andrews, J. R. (1993). Rehabilitation of the elbow in the throwing athlete. *The Journal of Orthopaedic and Sports Physical Therapy*, *17*(6), 305–317. https://doi.org/10.2519/jospt.1993.17.6.305

Wolff, A. L., & Hotchkiss, R. N. (2006). Lateral Elbow Instability: Nonoperative, Operative, and Postoperative Management. *Journal of Hand Therapy*, *19*(2), 238–244.

Chapter 4

4   The Effects of Brace Valgus Angulation on LCL Loads following Simulated LCL Repair: An *In Vitro* Study

Overview

An in vitro biomechanical study was conducted to examine the effects of varying the valgus angulation of a custom hinged elbow orthosis (HEO), as described in Chapter 2, on LCL tension. Lateral collateral ligament (LCL) repair tension was quantified using a load cell and custom clamping system. Simulated active and passive motion was conducted with the forearm pronated and supinated, in the vertical dependent and varus positions. LCL load throughout the range of motions tested was quantified.

(Note: Some portions of the Introduction and Methods appear in Chapters 2 & 3, and have been repeated, in part in this chapter, given that this thesis is presented in integrated article format.)

4.1   Introduction

The elbow is the second most commonly dislocated joint among adults, commonly disrupting the lateral collateral ligament (LCL) (Babhulkar, 2015; Kuhn & Ross, 2008; Osbourne & Cotterill, 1966; Reichel, Milam, Sitton, Curry, & Mehlhoff, 2013). While elbow dislocations and LCL injuries are commonly managed non-operatively with a closed reduction, surgical repair may be recommended in cases of recurrent dislocation or persistent instability (Charalambous & Stanley, 2008; Clitherow, McGuire, & Bain, 2014; Josefsson, Gentz, Johnell, & Wendeberg, 1987; Mica, Caekebeke, & Riet, 2016; Sheps, Hildebrand, & Boorman, 2004; Szekeres, Chinchalkar, & King, 2008). One study found that following a simple elbow dislocation, there was no significant difference between operative and non-operative treatment (Josefsson et al., 1987). However, a long term flow-up study found that 60% of patients treated non-operatively reported symptoms such as decreased range of motion and residual pain (Mehlhoff, Noble, Bennett, & Tullos, 1988).

It has been shown that following an elbow dislocation, rehabilitation is safe during active motion in vertical dependent or overhead arm positions, whereas the passive motion and the varus position are usually avoided (Alolabi et al., 2012; A. D. Armstrong et al., 2000; Dunning, Zarzour, Patterson, Johnson, & King, 2001; Fusaro, Orsini, Szorza, Rotini, & Benedetti, 2014; Szekeres et al., 2008). It has also been shown that pronation better stabilizes the LCL deficient elbow compared to supination (Dunning et al., 2001).

Following a dislocation, the LCL may be repaired of reconstructed depending on the quality of the remaining ligament (Bernard F. Morrey & Sanchez-Sotelo, 2009; Sheps et al., 2004; Smith, Savoie, & Field, 2001). Although elbow dislocations disrupt both the lateral collateral ligament (LCL) and medial collateral ligament (MCL), injury to the lateral side is often worse because the common extensor origin (CEO) is usually disrupted, whereas the common flexor origin remains intact (Alolabi et al., 2012; A. Armstrong, 2016; Bell, 2008; Josefsson, Johnell, & Gentz, 1984; McKee, Schemitsch, Sala, & O'Driscoll, 2003; O'Driscoll, Morrey, Korinek, & An, 1992). Thus, an LCL repair or reconstruction is usually sufficient to achieve stability (Heo et al., 2015). In acute cases, where an LCL repair is appropriate, it is often completed using a humeral

bone tunnel or suture anchor (Wiesel, 2015). Fraser et al., (2008) investigated 20, 40 and 60 N LCL repair tensions and found that 20 N or less was sufficient to restore stability.

If elbow instability persists following a closed reduction of a dislocation or after surgical repair of the LCL, a hinged elbow orthosis (HEO) such as the Bledsoe Brace (Bledsoe Brace Systems, Grand Prairie, TX) and the Mayo Clinic Universal Elbow Brace (Aircast, Summit, NJ) (Chapter 1, Figures 1-15 and 1-16) may be recommended (Sheps et al., 2004; Wolff & Hotchkiss, 2006). However, to our knowledge, there is no research to validate the efficacy of HEOs following LCL injury or repair, particularly with regard to protection of loads in the ligament LCL. Additionally, current HEOs are designed with a straight hinge that does not account for the native carrying angle of the arm; the effect of a more anatomically designed HEO on LCL loading has not been investigated.

The patterns of elbow ligament tensions across the range of motion may provide vital insight that could be used to optimize treatment and rehabilitation protocols. *In vitro* ligament loads have been examined for the MCL using a buckle transducer (Fay, Lalone, Ferreira, Johnson, & King, 2010), however to our knowledge, loads in the LCL have not been investigated, possibly due to its more complex anatomical structure.

The objective of this study was to bridge this knowledge gap by examining the efficacy of HEOs at various valgus angulations following simulated LCL repair. A modular HEO (described and evaluated in Chapter 2) was adjusted to five predetermined valgus angulations. Active and passive flexion-extension motions were conducted with the forearm pronated and supinated and the load through the LCL was simultaneously measured using a load cell. Trials were completed with the arm in vertical dependent and varus positions. It was hypothesized that greater angles of HEO valgus angulation would decrease the load through the LCL.

4.2   Materials and Methods

4.2.1 Specimen Preparation

Five fresh-frozen cadaveric specimens (mean age ± standard deviation: 68.6 ± 8.0) were stored at   -20℃ and thawed at room temperature (22 ± 2℃) for approximately 18 hours. The distal tendons of the biceps, brachialis, triceps, brachioradialis, pronator teres, wrist flexors (flexor carpi radialis and flexor carpi ulnaris) and wrist extensors (extensor carpi radialis longus and extensor carpi ulnaris) were sutured with braided fishing line (Bravefisherman, Shanghai, China). The pronator teres and wrist flexors were passed through medial alignment guides, while the wrist extensors and brachioradialis were fed through lateral guides (Chapter 2, Figure 2-8). A stainless steel rod was fixed into the humeral medullary canal with bone cement (Stryker, Kalamazoo, MI, USA), then mounted onto the elbow simulator via a humeral clamp (Chapter 2, Figure 2-10). The biceps, brachialis and triceps sutures were connected to computer-controlled motors, while the remaining muscles were connected to actuators. An ulnar tracker mount was fixed on the medial ulna to house an electromagnetic tracker (Flock of Birds®, Ascension Technology Corporation, Burlington, VT, USA).

4.2.2 Testing Protocol

A custom designed LabVIEW program (National Instruments, Austin, TX, USA) for motor and actuator control (Dunning, Zarzour, Patterson, Johnson, & King, 2001; Ferreira, 2011; Johnson et al., 2000; Kusins, Willing, King, & Ferreira, 2016) was used to simulate active motion. During active flexion, the biceps and brachialis were designated as the main flexors for supination and pronation, respectively. Tone loads were applied to the brachioradialis, pronator teres (during pronation only), and wrist flexors and extensors. Passive motion trials were conducted by the same investigator (SB) rotating the forearm into pronation or supination by holding the wrist, and manually flexing the arm. Prior to testing, five active and five passive preconditioning flexion-extension motions were conducted on the intact elbow in pronation and supination.

4.2.3 Ligament Tension System

LCL injury was simulated by sectioning the LCL and common extensor origin (CEO) off the lateral epicondyle. A humeral bone tunnel was created from the axis of motion of the elbow indicated by the center of curvature of the capitellum and extending to exit at the lateral supracondylar ridge (Fraser et al., 2008). The LCL was sutured with a Hi-Fi® #2 non-absorbable surgical suture (Conmed, Utica, NY, USA) and the sutures were passed through this bone tunnel and through an alignment guide (Chapter 2, Figure 2-8).

Two load cell attachments (Figure 4-1) were designed and machined from aluminum. Each cylindrical load cell attachment was threaded (#4-40 UNC 40) and screwed onto both ends of a uniaxial load cell (Chapter 1, Section 1.4.3, Figure 1-20) (Honeywell, Golden Valley, MN, USA).
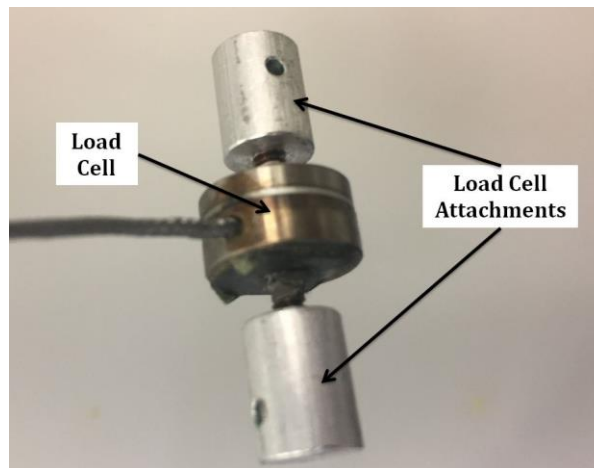
Figure 4-1: Load Cell Attachments. *Two load cell attachments were machined from aluminum and threaded onto both ends of a load cell.*

A clamping mechanism (Figure 4-2) was developed and rigidly fixed to the base of the simulator. A carriage bolt was mounted and fixed to the simulator using a nut. A hole was drilled through the middle of the bolt and a nut and washer was placed on either side of it. The end of the LCL suture was tied to braided fishing line (Bravefisherman, Shanghai, China), which was then secured to one end of the load cell (Figure 4-3). Fishing line was also tied to the other end of the load cell and fed through the hole in the clamping mechanism. Therefore, the suture could be pulled manually (by the investigator (SB)) to achieve the desired LCL tension. It was then secured at the chosen tension with the two washers and two nuts of the clamp, on either side of the line. The load cell, connected to a data acquisition (DAQ) system (National Instruments, Austin, TX, USA), provided load measurements in real time.

Before each flexion trial, with the arm fully extended in the vertical dependent or varus position, the LCL was tensioned to 20 ± 1 N. Active and passive elbow flexion and extension exercises were conducted with the forearm pronated and supinated after LCL sectioning and repair. Then, the HEO was secured to the arm, and the protocol was repeated for five different HEO valgus angulations (0°, 5°, 10°, 15° and 20°).

LCL load across the range of motion was recorded using a custom LabVIEW program (National Instruments, Austin, TX, USA). Two trials were conducted for each active and passive flexion motions. For active motion, an average of the two trials was taken. For passive motion, only the second trial was used for data analysis. This decision is based on the findings discussed in Chapter 2, Section 2.3. Active flexion resulted in intraclass correlation coefficient (ICC) values greater than 0.996, which was sufficient. During passive flexion the ICC values were lower (between 0.425 and 0.956), but increased to ICC > 0.959 when the first trial was omitted. This indicated that the second to fifth trials were more similar to each other than to the first trial. For simplicity, only flexion data is presented for this study.
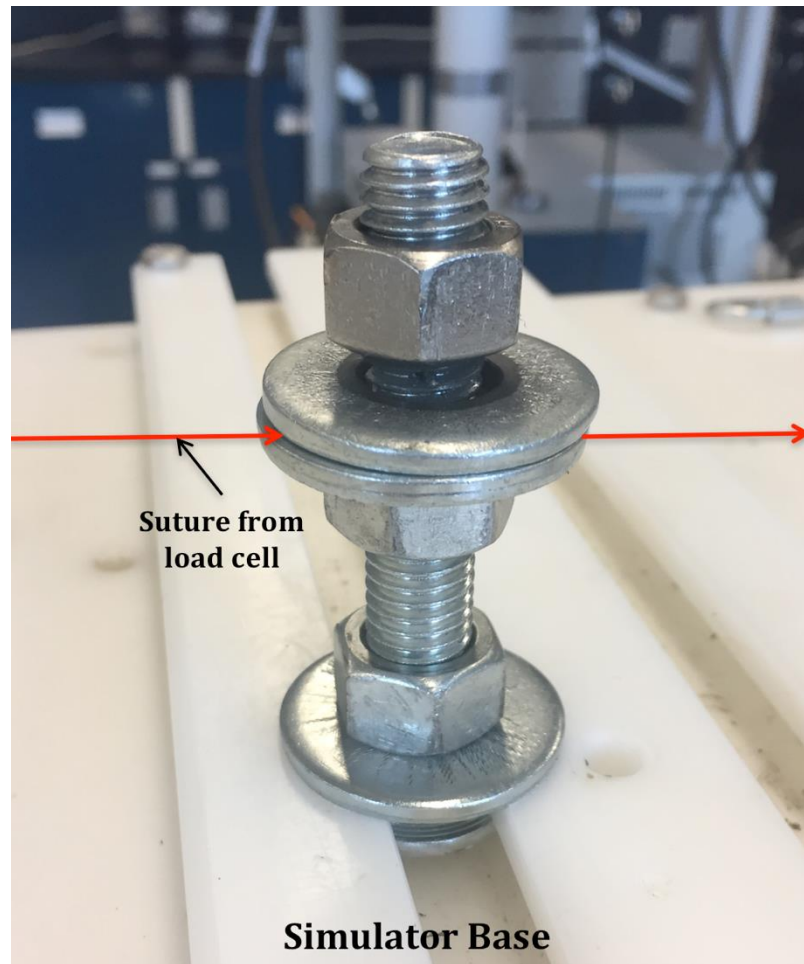
Figure 4-2: Custom Clamping Mechanism. *Using a combination of nuts and washers, a clamping system was designed to secure a bolt to the base of the simulator. A suture connected to the LCL (on the left side) was fed through the clamp and secured in place between the washers at the chosen tension as measured by the load cell (on the right side).*
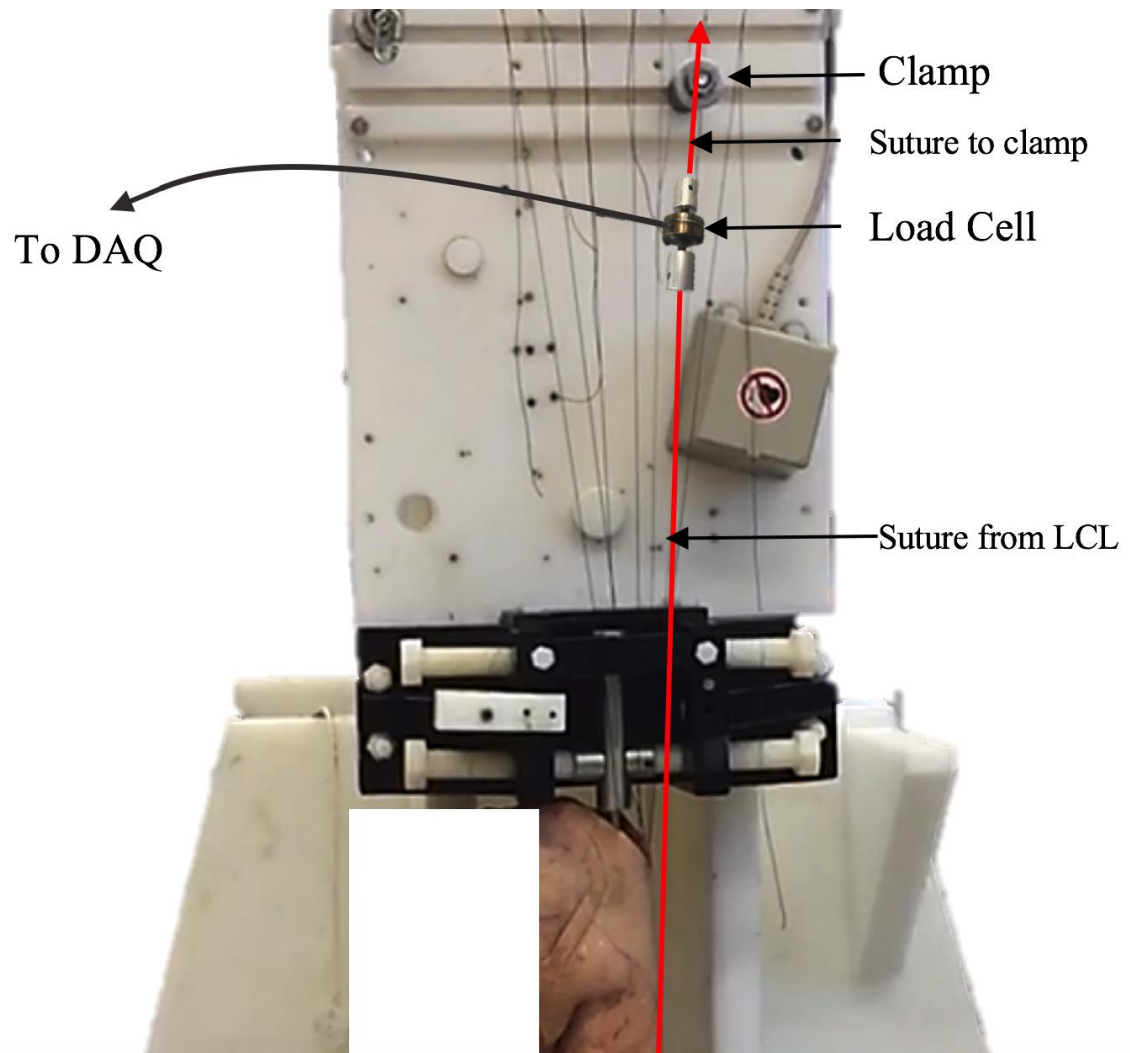
Figure 4-3: Simulated LCL Repair Setup. *A suture connected the LCL to one end of a load cell. A suture was also connected at the other end of the load cell and was fed through a clamping mechanism. To achieve the desired LCL repair tension the suture was pulled until the desired load was reached, and the clamp was tightened to secure the suture line in place. Shown is a left arm with the simulator in the vertical dependent arm position. (The tissue at the amputation site has been masked).*

4.2.4 Statistical Analysis

LCL loads analyzed using SPSS 25.0 statistics software (SPSS Inc, Chicago, IL, USA). Analyses were completed for the vertical dependent and varus positions separately. An analysis was also done to investigate the statistical power of the study. A two-way repeated measures of variance (ANOVA) was performed to compare the effect of flexion angle and the LCL injured state with and without the HEO. If significance was found, post hoc tests were conducted. Pairwise comparisons were employed using Bonferroni adjustments. Statistical significance was considered at $p < 0.05$.

4.3   Results

In the vertical dependent position, with the forearm pronated there was no significant difference in LCL tension (with and without an HEO) during active ($p = 0.33$) or passive ($p = 0.38$) motion (Figure 4-4 and Table 4-1). During active and passive motion, the mean loads during flexion varied within a range of 3.40 and 4.08 N, respectively. Similarly, there was no significant difference in LCL loads during active ($p = 0.65$) and passive motion ($p = 0.28$) with the forearm supinated. Here, the mean loads differed within 1.65 and 3.10 N for active and passive motion, respectively. There was no effect of flexion angle on the LCL loads for both active and passive motion with the forearm in pronation ($p > 0.05$) (Appendix C, Table C-5); LCL tensions were consistent throughout the range of motion. During active flexion and forearm supinated there was a significant change in LCL load in the mid flexion range ($p = 0.03$) with the HEO at $0°$, but no significant effects were found during passive motion ($p > 0.05$).

With the arm in the varus arm position, no significant difference in LCL tension was found with and without an HEO during active ($p = 0.65$) or passive ($p = 0.28$) motion with the forearm pronated (Figure 4-5 and Table 4-2). During active and passive motion, the mean loads during flexion varied within 2.64 and 5.80 N, respectively. There was also no significant difference in LCL tension found with and without an HEO during active ($p = 0.13$) and passive motion ($p = 0.39$) with the forearm supinated. The mean loads for supination varied within 1.85 and 3.15 N for active and passive motion, respectively. There was no effect of flexion angle on the LCL loads for both active and passive motion with the forearm in pronation and supination ($p > 0.05$) (Appendix C, Table C-5).

The statistical power of the study when comparing the LCL injured state to the flexion angle as the result of a four-way repeated measures analysis of variance (ANOVA) was found to be 0.437.
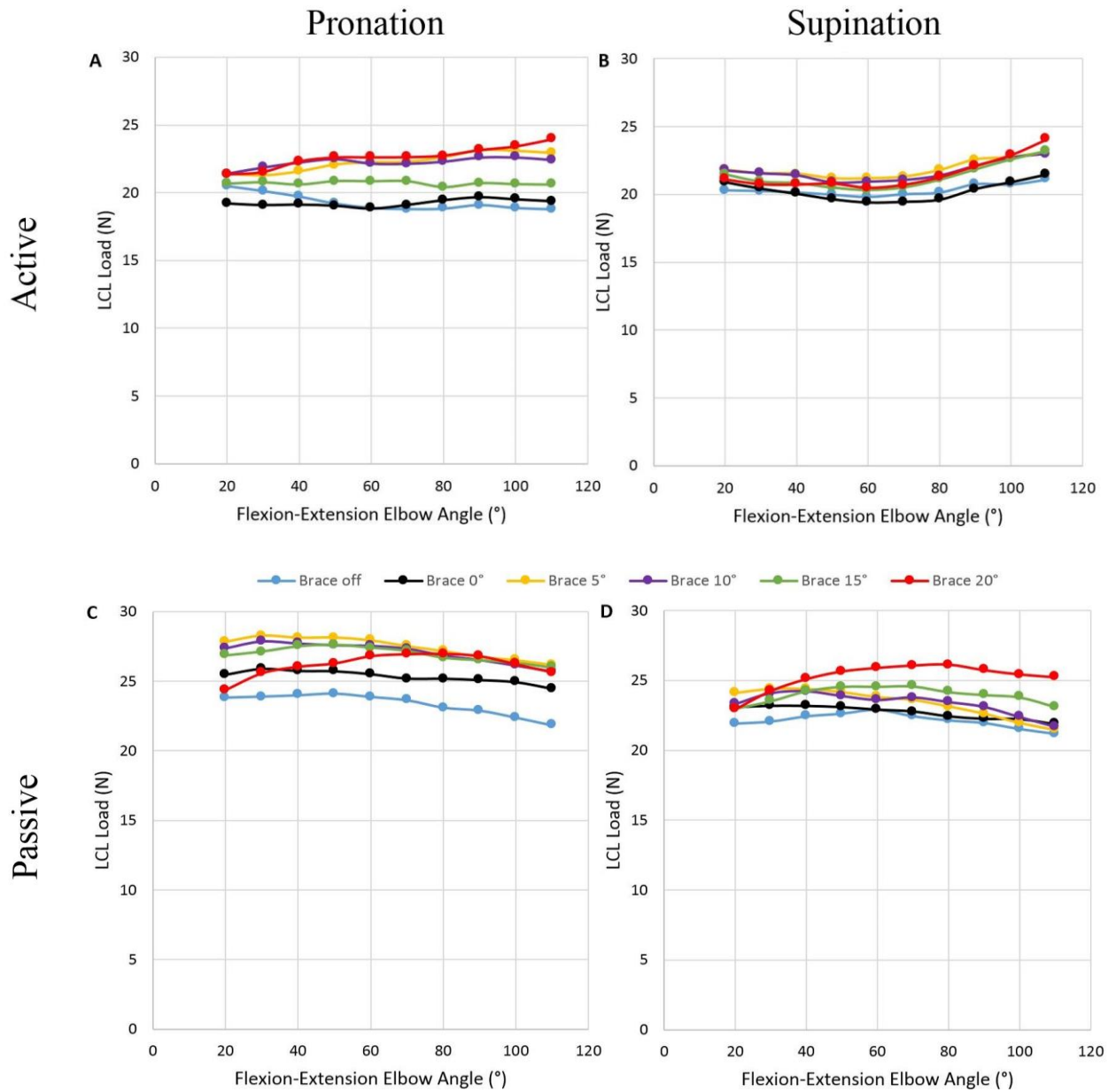
Figure 4-4: Mean LCL Tension during Elbow Flexion in the Vertical Dependent Arm Position. *The graph illustrates the tension profiles during active pronation (A) and supination (B), as well as passive pronation (C) and supination (D). LCL injured states are shown with and without a HEO at various valgus angulations. Standard deviations were omitted for clarity but range as follows: active pronation (2.77 – 10.1), active supination (4.31 – 6.25), passive pronation (4.67 – 8.76) and passive supination (3.35 – 6.14).*

Table 4-5: Effect of Brace Angulation, Forearm Rotation and Muscle Activation on LCL Tension during Flexion in the Vertical Dependent Arm Position

Forearm Rotation	Muscle Activation	LCL Injury	Mean Load ± SD (N)	p
Pronation	Active	No HEO	19.28 ± 4.48	0.33
		HEO 0°	19.23 ± 2.77	
		HEO 5°	22.25 ± 8.63	
		HEO 10°	22.22 ± 10.1	
		HEO 15°	20.71 ± 6.43	
		HEO 20°	22.63 ± 7.27	
	Passive	No HEO	23.37 ± 4.67	0.38
		HEO 0°	25.33 ± 4.92	
		HEO 5°	27.45 ± 8.76	
		HEO 10°	27.07 ± 7.92	
		HEO 15°	26.93 ± 7.16	
		HEO 20°	26.16 ± 7.87	
Supination	Active	No HEO	20.33 ± 4.54	0.65
		HEO 0°	20.23 ± 4.31	
		HEO 5°	21.88 ± 5.89	
		HEO 10°	21.00 ± 5.62	
		HEO 15°	21.34 ± 7.59	
		HEO 20°	21.48 ± 6.25	
	Passive	No HEO	22.13 ± 4.08	0.28
		HEO 0°	22.70 ± 4.09	
		HEO 5°	22.36 ± 3.35	
		HEO 10°	23.35 ± 4.72	
		HEO 15°	23.95 ± 6.11	
		HEO 20°	25.23 ± 6.14	

p-values describe the significance of the injured states resulting from a two-way analysis of variance (ANOVA) with LCL injured state and flexion angle. The asterisk (*) indicates significance (p < 0.05). Abbreviations: SD – standard deviation; HEO – hinged elbow orthosis and LCL - lateral collateral ligament.
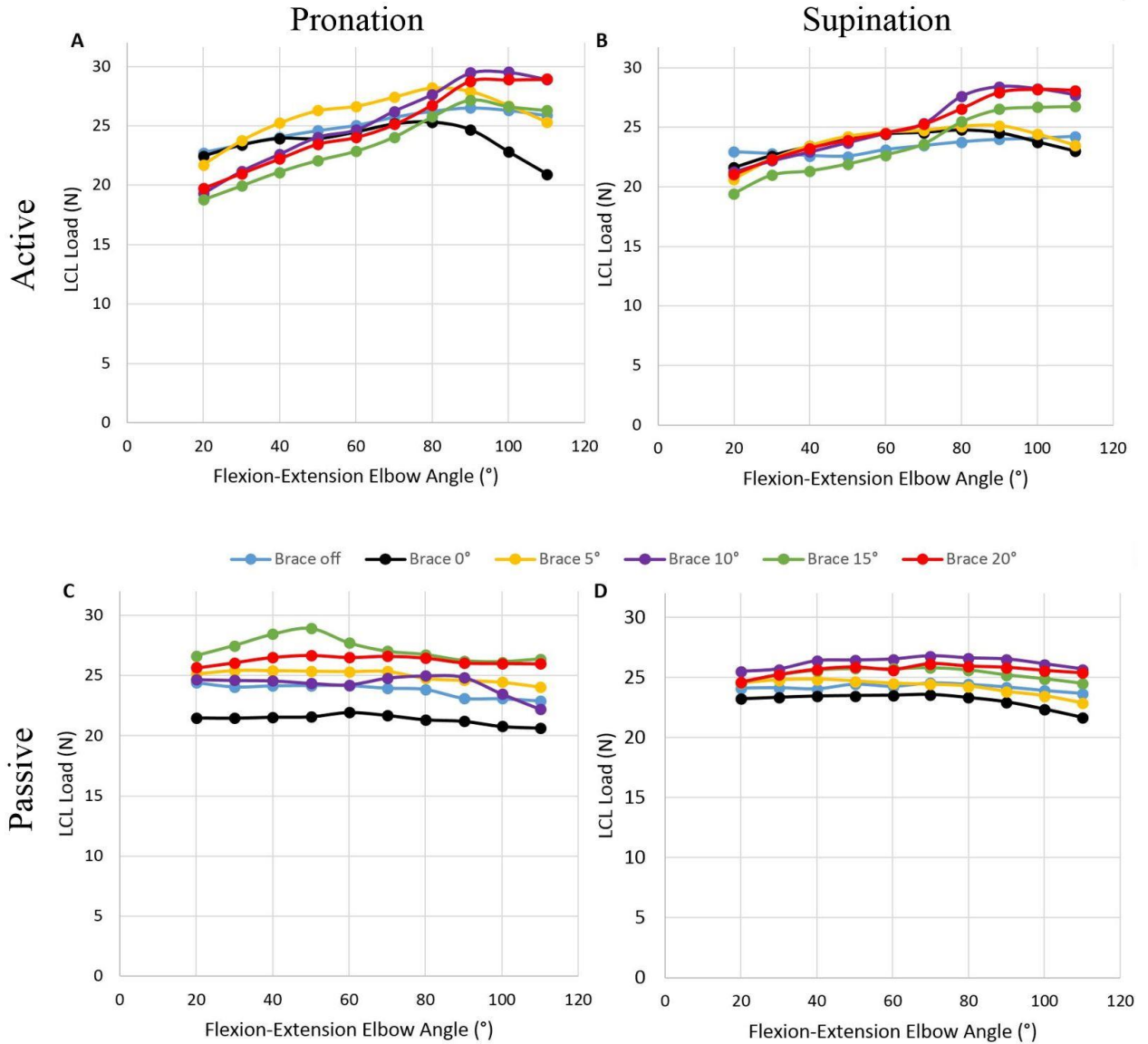
Figure 4-5: Mean LCL Tension during Elbow Flexion in the Varus Arm Position. *The graph illustrates the tension profiles during active pronation (A) and supination (B), as well as passive pronation (C) and supination (D). LCL injured states are shown with and without a HEO at various valgus angulations. Standard deviations were omitted for clarity but range as follows: active pronation (3.89 – 7.99), active supination (1.53 – 9.32), passive pronation (2.41 – 6.94) and passive supination (2.60 – 9.89).*

Table 4-6: Effect of Brace Angulation, Forearm Rotation and Muscle Activation on LCL Tension during Flexion in the Vertical Dependent Arm Position

Forearm Rotation	Muscle Activation	LCL Injury	Mean Load ± SD (N)	p
Pronation	Active	No HEO	25.05 ± 7.02	0.07
		HEO 0°	23.72 ± 7.18	
		HEO 5°	25.92 ± 7.99	
		HEO 10°	25.36 ± 7.26	
		HEO 15°	23.28 ± 7.02	
		HEO 20°	24.90 ± 3.89	
	Passive	No HEO	23.79 ± 4.20	0.51
		HEO 0°	21.38 ± 1.53	
		HEO 5°	25.01 ± 4.98	
		HEO 10°	24.29 ± 6.33	
		HEO 15°	27.18 ± 9.32	
		HEO 20°	26.27 ± 8.51	
Supination	Active	No HEO	23.37 ± 4.47	0.13
		HEO 0°	23.70 ± 4.97	
		HEO 5°	23.82 ± 5.32	
		HEO 10°	25.22 ± 6.94	
		HEO 15°	23.56 ± 3.33	
		HEO 20°	25.12 ± 2.41	
	Passive	No HEO	24.17 ± 4.12	0.39
		HEO 0°	23.08 ± 2.60	
		HEO 5°	24.22 ± 3.21	
		HEO 10°	26.23 ± 9.89	
		HEO 15°	25.25 ± 7.70	
		HEO 20°	25.59 ± 6.37	

p-values describe the significance of the injured states resulting from a two-way analysis of variance (ANOVA) with LCL injured state and flexion angle. The asterisk (*) indicates significance (p < 0.05). Abbreviations: SD – standard deviation; HEO – hinged elbow orthosis and LCL - lateral collateral ligament.

4.4   Discussion

LCL injuries commonly produce elbow instability, and in some cases, may be treated with a ligament repair (Charalambous & Stanley, 2008; Mica et al., 2016; Osbourne & Cotterill, 1966; Sheps et al., 2004). If the elbow remains unstable following a repair, a hinged elbow orthosis (HEO) may be recommended to 'protect' the repair and maintain elbow stability during the healing process (Sheps et al., 2004). It has been previously shown that an LCL repair tension of 20 N or less is sufficient to restore native kinematics, however the load change throughout the range of motion has not been reported (Fraser et al., 2008). To our knowledge, the efficacy of HEOs following LCL repair, as well as the variation in the valgus angulation of an HEO, have not been investigated. A greater understanding of the effect of HEOs on LCL tension is needed to validate (or refute) current treatment protocols. This *in vitro* study investigated the effects of varying the valgus angulation of a custom designed HEO during a simulated LCL repair. The LCL loads were quantifed throughout the range of motion. Active and passive trials were simulated in the vertical dependent and varus arm positions, with the forearm maintained in both pronation and supination.

In both the vertical dependent and varus arm positions, there were no significant changes in the LCL loads as the HEO angle was altered. In the vertical dependent arm position, the LCL loads varied by no greater that 5 N regardless of the angulation of the brace, therefore it is not surprising that statistically significant differences were not found. During active motion in the varus position there is a slight increase in LCL loading in greater flexion, which may be speculated to be the result of the changing muscle lines of action, ligament length, and variable tensioning of other soft-tissues crossing the joint (*viz.* capsule, etc), however the differences were not significant. In full extension, the elbow joint is compressed by the surrounding muscles, but during flexion, the direction of the muscle lines of action vary and the weight of the forearm and HEO increases the gravitational moment. During passive motion in the varus position LCL loading is relatively unchanged throughout flexion. This is expected, since the investigator (SB) likely minimized the effects of gravity and the weight of the HEO by mildly holding the arm up during passive trials.

It may be postulated that there may not have been much load variation across the range of motion due to the structure of the LCL. It was previously reported that the LCL is isometric (Morrey & An, 1985), which would have explained the lack of change in LCL loads during flexion, however more recent studies have concluded otherwise. The radial collateral ligament (RCL) of the LCL is has been shown to be relatively isometric (Hackl, Bercher, Wegmann, Müller, & Dargel, 2016; Morimoto et al., 2007), however Hackl et al. (2016) found that the anterior portion of the RCL is tight in full extension and loosens as the elbow is flexed. Conversely, studies agree that the LUCL is loose in extension and tightens as the elbow is flexed (Hackl et al., 2016; Morimoto et al., 2007; Wavreille, Seraphin, Chantelot, Matachandise, & Fontaine, 2008). Since both the RCL and LUCL were sutured, and they have opposite effects, they may have balanced out the load throughout the range of motion. Therefore, in extension the RCL may have accepted most of the load, with majority of the load being passed to the LUCL in flexion. Furthermore since our LCL 'repair' was placed at the axis of motion of the elbow it would be expected that the LCL would behave isometrically and may not fully represent ligament loading in the native situation.

At all angles of HEO valgus angulation, the mean loads were not significantly different. It was hypothesized that increasing valgus angulation of the HEO would reduce the load through the LCL, however this was not found. This may be due to a variety of factors as the biomechanical changes across the elbow are likely complex as the HEO is added. While there appears to be little change in support of the LCL (*i.e.* no decrease in tension was found) as the HEO valgus angle increased, this may be attributed to changes in the rotation angle (IER from Chapter 3) acting less favorable as the HEO is increased. Hence, while the valgus angulation of the ulna relative to the humerus increases with greater HEO valgus angulation, any rotation that is coupled with the increased HEO angle that tends to lengthen the LCL could act to negate a potential favorable reduction in ligament tension with increased HEO valgus. This is one potential mechanism, although it is only a postulation as the biomechanics of load transfer at the elbow is highly complex and indeterminate.

The addition of the HEO did not increase the LCL loads as expected from the weight of the HEO. The most likely reason for this may be that the LCL tension was adjusted with the HEO on the arm. Thus, the weight of the brace may have been accounted when the LCL was initially tensioned. This does not represent clinical practice, as a surgical repair could not be accomplished with a HEO applied. Unfortunately, due to the repeated measures design of this study the HEO was not removed between trials to ensure consistency of hinge placement and strap tightness. To alternate between pronation and supination the two distal straps were loosened, but the proximal two remained tightly strapped to the upper arm.

A main strength of this study is that it was the first known study reporting on the change in LCL loads across the range of motion. While this has been done for the MCL using a buckle transducer, the anatomical shape of the LCL would not allow the use of this measurement method (Fay et al., 2010). Another study simulating LCL repair tensions, achieved the desired tension using an actuator, however, the load changes during motion were not quantified (Fraser et al., 2008). Therefore, this study presents a novel method to examine LCL loads *in vitro* and may be useful for future studies. Another strength of this study was the use of a custom designed, novel HEO that allowed the valgus angulation to be modified quickly and effectively. The reliability of the HEO was previously confirmed in Chapter 2. Additionally, this is only the second study investigating HEOs during active motion.

With regard to the testing protocol, there were also several strengths. All incisions were sutured for the duration of the day to preserve the hydration of the soft tissues, allowing them to maintain their viscoelastic behavior. The base of the custom elbow simulator was also able rotate to simulate the desired rehabilitative arm positions. Tone loads were employed to several muscles to better simulate *in vivo* loading. Finally, the LCL repair technique featuring a humeral bone tunnel that was employed for this study is commonly used in clinical practice.

There were some limitations to this study, several of which have been noted in Chapter 3. The post-hoc statistical power of this study comparing the injured state to the flexion

angle was 0.437. This suggests that perhaps a greater sample size is required to accurately draw conclusions. However, while the results of the current study were not statistically significant, we have developed a novel model to study LCL tension *in vitro*. Furthermore, some of the specimens were quite obese limiting flexion range we could study due to impingement of fat caused by the brace straps. Another limitation was that there was limited range of motion for some specimens, likely because the specimens were older. Therefore, the LCL could not be tensioned at the same joint angle for each specimen. Tensioning was performed with the arm in full extension, where this joint angle varied by specimen, but was kept consistent within each specimen. Currently, there is no gold standard for the appropriate orientation of the elbow to surgically perform a repair, nor is there a target LCL repair tension. With the new repair simulation technique presented in here, further investigations may perhaps be performed to optimize surgical repair methods.

4.5   Conclusion

Optimal rehabilitation of the LCL injured elbow following ligamentous injury or repair has not been sufficiently studied nor reported in literature. This study aimed to provide biomechanical insight into the load trends following a repair with and without an HEO and different valgus angulations. An innovative technique to simulate an LCL repair and simultaneously measure LCL loads was employed making this a novel contribution to existing biomechanical literature. Increasing the valgus angulation of the HEO did not decrease the loads in the LCL across the range of motion suggesting that the addition of a brace to protect an LCL injury or surgical repair may not be efficacious.

4.6   References

Alolabi, B., Gray, A., Ferreira, L. M., Johnson, J. A., Athwal, G. S., & King, G. J. W. (2012). Rehabilitation of the medial- and lateral collateral ligament-deficient elbow: An in vitro biomechanical study. *Journal of Hand Therapy*, *25*(4), 363–373. https://doi.org/10.1016/j.jht.2012.06.001

Armstrong, A. (2016). Simple Elbow Dislocation. In G. S. Athwal (Ed.), *Management of elbow trauma, an issue of hand clinics 31-4, E-Book* (pp. 521–531). Elsevier Health Sciences.

Armstrong, A. D., Dunning, C. E., Faber, K. J., Duck, T. R., Johnson, J. A., & King, G. J. W. (2000). Rehabilitation of the medial collateral ligament-deficient elbow: an in vitro biomechanical study. *The Journal of Hand Surgery*, *25*(6), 1051–1057.

Babhulkar, S. (2015). *Elbow injuries* (illustrate). New Delhi, India: JP Medical Ltd.

Bell, S. (2008). (iii) Elbow instability, mechanism and management. *Current Orthopaedics*, *22*(2), 90–103. https://doi.org/10.1016/j.cuor.2008.04.007

Charalambous, C. P., & Stanley, J. K. (2008). Posterolateral rotatory instability of the elbow. *Bone & Joint Journal*, *90*(3), 272–279.

Clitherow, H. D. S., McGuire, D. T., & Bain, G. I. (2014). Lateral elbow instability. *Asia-Pacific Journal of Sports Medicine, Arthroscopy, Rehabilitation and Technology*, *1*(1), 11–18.

Dunning, C. E., Zarzour, Z. D. S., Patterson, S. D., Johnson, J. A., & King, G. J. W. (2001). Muscle forces and pronation stabilize the lateral ligament deficient elbow. *Clin Orthop Relat Res*, *388*, 118–124.

Fay, K. E., Lalone, E. A., Ferreira, L., Johnson, J., & King, G. J. W. (2010). Quantification of Medial collateral ligament tension in the elbow. In *56th Annual Meeting of the Orthopaedic Research Society*.

Ferreira, L. M. (2011). *Development of an Active Elbow Motion Simulator and Coordinate Systems to Evaluate Kinematics in Multiple Positions*. The University of Western Ontario.

Fraser, G., Pichora, J., Ferreira, L., Brownhill, J., Johnson, J., & King, G. J.W. (2008). Lateral collateral ligament repair restores the initial varus stability of the elbow: an in vitro biomechanical study. *J Orthop Trauma*, *22*(9), 615–23.

Fraser, G. S., Pichora, J., Ferreira, L. M., Brownhill, J., Johnson, J. A., & King, G. J. W. (2008). Lateral Collateral Ligament Repair Restores the Initial Varus Stability of the Elbow: An In Vitro Biomechanical Study. *Journal of Orthopaedic Trauma*, *22*(9), 615–623.

Fusaro, I., Orsini, S., Szorza, T., Rotini, R., & Benedetti, M. G. (2014). The use of braces in the rehabilitation treatment of the post-traumatic elbow. *Joints*, *2*(4), 81–86.

Hackl, M., Bercher, M., Wegmann, K., Müller, L. P., & Dargel, J. (2016). Functional anatomy of the lateral collateral ligament of the elbow. *Archives of Orthopaedic and Trauma Surgery*, *136*(7), 1031–1037. https://doi.org/10.1007/s00402-016-2479-8

Heo, Y. M., Yi, J. W., Lee, J. B., Lee, D. H., Park, W. K., & Kim, S. J. (2015). Unstable simple elbow dislocation treated with the repair of lateral collateral ligament complex. *Clin Orthop Surg*, *7*(2), 241–247.

Johnson, J. A., Rath, D. A., Dunning, C. E., Roth, S. E., & King, G. J. W. (2000). Simulation of elbow and forearm motion in vitro using a load controlled testing apparatus. *Journal of Biomechanics*, *33*(5), 635–639. https://doi.org/10.1016/S0021-9290(99)00204-3

Josefsson, P. O., Gentz, C. F., Johnell, O., & Wendeberg, B. (1987). Surgical versus non-surgical treatment of ligamentous injuries following dislocation of the elbow joint. A prospective randomized study. *JBJS*, *69*(4), 605–608.

Josefsson, P. O., Johnell, O., & Gentz, C. F. (1984). Long-term sequelae of simple dislocation of the elbow. *J Bone Joint Surg*, *66*(6), 927–930.

Kuhn, M. A., & Ross, G. (2008). Acute elbow dislocations. *Orthop Clin North Am*, *39*(2), 155–161.

Kusins, J., Willing, R., King, G., & Ferreira, L. (2016). Development of a computational elbow model with experimental validation of kinematics and muscle forces. *J Appl Biomech*, *32*(4), 407–14.

McKee, M. D., Schemitsch, E. H., Sala, M. J., & O'Driscoll, S. W. (2003). The pathoanatomy of lateral ligamentous disruption in complex elbow instability. *Journal of Shoulder and Elbow Surgery*, *12*(4), 391–6.

Mehlhoff, T., Noble, P., Bennett, J., & Tullos, H. (1988). Simple dislocation of the elbow in the adult. Results after closed treatment. *J Bone Joint Surg Am*, *70*(2), 244–249.

Mica, M. C., Caekebeke, P., & Riet, R. van. (2016). Lateral collateral ligament injuries of the elbow – chronic posterolateral rotatory instability (PLRI). *EFORT Open Reviews*, *1*(12), 461–468.

Morimoto, H., Murase, T., Arimitsu, S., Oka, K., Yoshikawa, H., & Sugamoto, K. (2007). The in vivo isometric point of the lateral ligament of the elbow. *J Bone Joint Surg Am*, *89*(9), 2011–7.

Morrey, B. F., & An, K. N. (1985). Functional anatomy of the ligaments of the elbow. *Clin Orthop Relat Res*, *201*, 84–90.

Morrey, B. F., & Sanchez-Sotelo, J. (2009). *The Elbow and Its Disorders* (4th ed.). Philadelphia: Elsevier Health Sciences.

O'Driscoll, S., Morrey, B. F., Korinek, S. L., & An, K. N. (1992). Elbow subluxation and dislocation. A spectrum of instability. *Clinical Orthopaedics and Related Research*, *280*, 186–197.

Osbourne, G., & Cotterill, P. (1966). Recurrent dislocation of the elbow. *Bone & Joint Journal*, *48*(2), 340–346.

Reichel, L. M., Milam, G. S., Sitton, S. E., Curry, M. C., & Mehlhoff, T. (2013). Elbow lateral collateral ligament injuries. *The Journal of Hand Surgery*, *38*(1), 184–201.

Sheps, D. M., Hildebrand, K. A., & Boorman, R. S. (2004). Simple dislocations of the elbow: evaluation and treatment. *Hand Clin*, *20*, 389–404.

Smith, J. P., Savoie, F. H., & Field, L. D. (2001). Posterolateral rotatory instability of the elbow. *Clinics in Sports Medicine*, *20*(1), 47–58.

Szekeres, M., Chinchalkar, S. J., & King, G. J. W. (2008). Optimizing Elbow Rehabilitation After Instability. *Hand Clinics*, *24*(1), 27–38. https://doi.org/10.1016/j.hcl.2007.11.005

Wavreille, G., Seraphin, J., Chantelot, C., Matachandise, X., & Fontaine, C. (2008). No Title. *Clin Biomech (Bristol, Avon)*, *23*, 193–202.

Wiesel, S. W. (2015). *Operative Techniques in Orthopaedic Surgery (Four Volume Set)* (2nd ed.). Lippincott Williams & Wilkins.

Wolff, A. L., & Hotchkiss, R. N. (2006). Lateral Elbow Instability: Nonoperative, Operative, and Postoperative Management. *Journal of Hand Therapy*, *19*(2), 238–244.

Chapter 5

5 General Discussions, Future Directions, and Conclusions

Overview

This chapter summarizes the objectives and hypothesis stated in Chapter 1, as well as the important outcomes from the studies presented in Chapters 2, 3 and 4. The strengths and limitations of this body of work are reviewed and the impact of this work on future research is discussed.

5.1 Summary

The purpose of this research was to investigate the current rehabilitation and surgical protocols employed following LCL injures. This work advances literature through the design and testing of a modular orthosis to quantify the kinematics and biomechanics of the elbow following LCL injury. These findings present additional information regarding bracing and specifically with adjusting the valgus angulation. Additionally, active and passive motion data, with the arm in clinically relevant arm positions and forearm rotations is presented. The specific objectives outlined in Chapter 1 have been fulfilled as described in Chapters 2, 3 and 4.

As presented in Chapter 1, Section 1.6, the specific objectives of this thesis were as follows:

1. To design and construct an adjustable hinged elbow orthosis (HEO) for examining *in vitro* kinematics and biomechanics of the elbow following LCL injury;

2. To determine the effects of changing the valgus angulation of the hinged elbow orthosis on elbow kinematics during simulated LCL injury;

3. To quantify the LCL load throughout simulated active and passive elbow flexion-extension, with the arm in dependent and varus orientations and with the forearm pronated and supinated.

The hypotheses and findings described in Chapters 2, 3 and 4 are reviewed summarized in the following sections.

5.1.1 Chapter 2 – Design and Development of an Experimental Hinged Elbow Orthosis for Examining *In Vitro* Kinematics and Biomechanics of the LCL-Injured Elbow

The first objective of this thesis was to design and fabricate an adjustable HEO. Four orthosis arms (two humeral and two forearm) and two modular hinge mechanisms were incorporated into the device. The orthosis arms were tightly secured to the arm using hook and loop fasteners and were connected posteriorly with aluminum rods and set screws. Between the proximal and distal sets of orthosis arms, the dual hinge mechanism allowed incremental changes to the valgus angulation, while permitting unrestricted flexion-extension. To confirm the reliability of the orthosis, *in vitro* testing on one specimen was conducted in the vertical dependent arm position with the forearm pronated. Five active and passive flexion exercises were performed with the cadaveric arm mounted on a custom elbow motion simulator. An electromagnetic tracking system was used to quantify the varus-valgus angulation (VVA) and internal-external rotation (IER).

The reliability of the HEO with the active motion simulator was confirmed using the single measures intraclass correlation coefficient (ICC) values. During active motion and forearm pronated, the ICC values ranged between 0.997 and 0.999 indicating highly reliable results. During passive motion, the ICC values ranged from 0.425 to 0.956. However, when the first trial was omitted the ICC values increased to between 0.959 and 0.984, which is sufficiently reliable. Therefore, for the duration of experimentation described in Chapters 3 and 4, the first trial was excluded for passive flexion. Moreover, kinematic measurements obtained following HEO application during future *in vitro* investigations can be considered reliable.

5.1.2 Chapter 3 – The Effects of Brace Valgus Angulation on the Stability of the LCL Injured Elbow: An *In Vitro* Study

The second objective of this work was to quantify the kinematic effects of altering the valgus angulation of the HEO during simulated LCL injury. This was achieved using an active elbow motion simulator and the HEO described in Chapter 2. Orthosis valgus angles between 0° and 20° were tested at 5° intervals. Testing was performed on the intact elbow and the LCL injured elbow with and without application of the HEO. Active and passive flexion-extension was simulated in the vertical dependent and varus arm positions with the forearm pronated and supinated.

It was hypothesized that active flexion and pronation would provide more stability than passive motion and supination **(Hypothesis #1)**. This hypothesis held true in the vertical dependent position, however the opposite was found in the varus arm position. Stability was significantly worse in the varus position for most injured states investigated. It may be postulated that this could be the result of human manipulation of the elbow, as the investigator (SB) likely offset the weight of the brace.

It was also theorized that the orthosis would provide more stability following LCL injury in both arm positions investigated **(Hypothesis #2)**. With respect to the vertical dependent position, the arm was found to be stable during active motion with the arm pronated and supinated. Throughout passive flexion, greater angulations of the HEO improved stability, however not significantly. Therefore, the orthosis did not provide any additional stability, negating the second hypothesis in the vertical dependent position. In the varus position, the theory was confirmed during active motion, as no significant kinematic differences in stability were found as the HEO angulation was increased.

It was also postulated that increasing the valgus angle of the orthosis would improve stability in extension, but cause greater instability with the elbow in flexion **(Hypothesis #3)**. This hypothesis was confirmed, as most instability was found at flexion angles greater than 80°. However, the same trends were found with the orthosis removed. This

contradicts current literature and rehabilitation protocols where full extension is often avoided post-injury, but full flexion is permitted.

5.1.3 Chapter 4 – The Effects of Brace Valgus Angulation on LCL Loads following Simulated LCL Repair: An *In Vitro* Study

The third objective was to quantify LCL loads during simulated active and passive motion. This was accomplished through a custom clamping system comprised of nuts, washers and a bolt. The LCL was sutured and connected to one end of a load cell. A suture connected at the other end was: fed through the clamp, pulled until a 20 N tension was achieved, and then clamped. Active and passive flexion exercises were simulated with the arm in the vertical dependent and varus arm positions and with the forearm pronated and supinated. The modular HEO designed in Chapter 2 was then positioned on the arm and the protocol was repeated with the orthosis at 0°, 5°, 10°, 15°, and 20° of valgus angulation.

It was hypothesized that the orthosis would improve stability of the LCL repaired elbow (**Hypothesis #2**), however it was observed that the orthosis did not have a significant effect on stability at any HEO valgus angles.

It was also postulated that the HEO would continuously decrease LCL loads as the valgus angulation was increased (**Hypothesis #4**). It was found that increasing the valgus angulation of the HEO did not have a significant effect on LCL load. This may be due to the complex anatomy and kinematics of the joint. However, the power of the study (0.437) also suggests that additional specimens are required.

5.2   Strengths and Limitations

The kinematic and biomechanical work presented in this thesis has various strengths and limitations. Currently, there is minimal literature examining the efficacy of HEOs, with only one other study considering active motion (Lee et al., 2013; Manocha, 2016). This study supplemented the inadequate research on this topic and was the first study to investigate varying the valgus angle of the orthosis. To our knowledge, this has never been reported. The studies presented also simulated multiple muscle activations, arm positions, and forearm rotations, increasing the robustness of the research. Despite the small specimen size, the study presented in Chapter 3 had adequate power, likely due to a highly reproducible motion simulator and exceptionally reliable HEO. Unfortunately, insufficient power was found for the results discussed in Chapter 4. However, a novel LCL repair simulation technique was introduced. Thus far, no studies have examined the *in vitro* LCL loads following a simulated repair. Additionally, throughout the duration of the day, all incisions were sutured to retain tissue hydration and preserve the viscoelastic behaviour of the soft tissues.

Similar to all cadaveric *in* vitro studies, a limitation of this work is that muscle loads were simulated as an estimation of *in vivo* loads. The loading ratios were modeled from previous literature (Amis, Dowson, & Wright, 1979; Funk, An, Morrey, & Daube, 1987), however it may not accurately represent the complex physiologic loading. Tone loads were also applied to increase the clinical relevancy of the results. However, they also may not replicate *in vivo* loads. Additionally, application of the HEO was likely tighter than would be comfortable for a patient and may have potentially enhanced its efficacy. The diameter of the arm also increases with muscle activation, which was not taken into account for this study. Finally, due to the nature of cadaveric experimentation, the range of motion of the specimens was limited and full extension was unachievable.

5.3 Current and Future Directions

The work presented from Chapters 2 to 4 successfully achieved the specific objectives presented in Chapter 1 of this thesis. This research contributes to a greater understanding of the clinical and surgical implications of bracing on the LCL injured elbow. A novel HEO and innovative LCL loading method were also designed. However, future work is required to further investigate the complex kinematics and biomechanics of the LCL injured elbow and orthosis applications to optimize current rehabilitation and surgical protocols.

This study was the first to investigate changing the valgus angulation of a HEO. The orthosis was angled from 0° to 20° at 5° increments. In the varus position, a 20° angulation seemed to restore the native kinematics (Chapter 3), however additional increases in the valgus angulation may further increase or decrease stability and should be investigated. In terms of biomechanics (Chapter 4), it was hypothesized that increasing the valgus angulation would decrease the load through the LCL; however, this was not realized. Greater valgus angulation may be required to further investigate the complex biomechanics and kinematics of the joint.

Future work should include additional arm positions that are often employed during rehabilitation. The elbow is also commonly rehabilitated in the horizontal and overhead arm positions, which were not included in this study, partly due to simulation complexity. However, the loads and kinematics of the elbow joint may be altered in these positions and should be included in future work.

Future efforts should also include activation of the supinator muscle during simulated active motion. In full extension, supination was difficult to achieve with the biceps alone, as the biceps supinates as the elbow flexes. Inclusion of a tone load for the supinator, similar to the pronator, would further enhance *in vivo* modelling of physiologic loads. As well, additional cadaveric specimens are required to increase the power of the study described in Chapter 4. While the power was sufficient to achieve significance in Chapter 3, additional specimens must be tested to validate the results.

5.4   Conclusion

There is limited knowledge on the efficacy of HEOs following LCL injury, although they are often advised as part of a rehabilitation protocol. Furthermore, the biomechanical and kinematic consequences of modifying the valgus angulation of the orthosis have not been investigated. The research completed as part of this thesis provides a greater understanding of the impacts of bracing on elbow loads and kinematics.

A hinged elbow orthosis may decrease elbow instability, however further experimentation is required. Based on the kinematic investigation, it is hypothesized that increasing the valgus angulation of the HEO improved stability, particularly in the varus position, although stability was not restored. Active motion in the varus position should still be avoided. However, if required, an orthosis at 20° of valgus angulation should be worn. Generally, application of an orthosis at any valgus angulation does not worsen stability. However, a HEO is not helpful at maintaining stability at flexion angles greater than 80°.

In this study, passive motion was found to be safe, likely as a result of the investigator (SB) manually flexing the arm. This suggests that passive motion may be performed safely, if implemented by a trained professional. Based on the biomechanical investigation, no significant changes in LCL tension were found at any HEO angulation, indicating that while the orthosis does not seem to improve stability, it also does not worsen it.

Despite the limitation of correlating this work to clinical populations, this thesis improves the knowledge of elbow bracing and may assist clinicians and surgeons in the improvement of surgical and rehabilitative protocols, as well as orthosis design.

5.5 References

Amis, A. A., Dowson, D., & Wright, V. (1979). Muscle strengths and musculo-skeletal geometry of the upper limb. *Engineering in Medicine*, *8*(1), 41–48.

Funk, D. A., An, K. N., Morrey, B. F., & Daube, J. R. (1987). Electromyographic analysis of muscles across the elbow joint. *Journal of Orthopaedic Research*, *5*, 529–538.

Lee, A. T., Schrumpf, M. A., Choi, D., Meyers, K. N., Patel, R., Wright, T. M., … Daluiski, A. (2013). The influence of gravity on the unstable elbow. *J Shoulder Elb Surg*, *22*(1), 81–87. https://doi.org/http://dx.doi.org/10.1016/j.jse.2012.08.012

Manocha, R. H. K. (2016). *Optimizing the Rehabilitation of Elbow Lateral Collateral Ligament Injuries*. The University of Western Ontario.

.

Appendix A – Glossary

Active range of motion (AROM): The range of motion through which a patient moves his or her joint by autonomously activating adjacent muscles.

Anterior: Movement towards the front of the body

Brace: See Orthosis.

Carrying angle: The acute angle formed by the long axis of the humerus and the long axis of the ulna. It averages 10 to 15° in men and 15 to 20° in women.

Common forearm extensors: A group of muscles arising from a common origin located at the lateral epicondyle of the humerus.

Common forearm flexors: A group of muscles arising from a common origin located at the medial epicondyle of the humerus.

Complex elbow dislocation: An injury that destabilizes the elbow because of damage to the ligamentous structures and fracture through the bones of the elbow joint.

Control box: In motion analysis, a device that processes the relative strengths of the transmitted and received signal(s) and, usually in conjunction with a computer, delivers desired motion output.

Distal: Movement further away from a structure's origin.

Extension: Movement about a joint that increases the angle between the bones forming that joint.

Flexion: Movement about a joint that decreases the angle between the bones forming that joint.

Hinged elbow orthosis (HEO): A prefabricated orthosis with 2 Velcro hook and loop straps at the arm and 2 Velcro hook and loop straps at the forearm. A metal stay is aligned axially on the medial and lateral sides of the arm and forearm. There is a metal hinge at the elbow flexion-extension axis into which pins can be inserted to limit flexion-extension range of motion. The device has no energy-storing components. This device is often used to reduce instability following ligamentous and/or bony elbow injury.

In vitro: Adjective describing the study of a natural process using a laboratory model of that process. In kinematic analyses, this often involves using a specialized device to move a cadaveric joint and observing the resulting joint motion.

In vivo: Adjective describing the study of a process occurring in a living organism. In kinematic analyses, this often involves observing a human moving a joint naturally.

Kinematics: The mechanical study of the motion of points, objects, and groups of objects, without reference to the forces that result in that motion.

Lateral: Movement away from the median sagittal plane.

Load-controlled simulation: In vitro cadaveric simulation of active joint motion whereby a set of desired force(s) is directed through the tendon(s) of selected muscle(s).

Medial: Movement towards the median sagittal plane.

Motion-controlled simulation: In vitro cadaveric simulation of active joint motion whereby a joint is moved at a prespecified rate through changing force(s) through the tendon(s) of selected muscle(s).

Orientation: The angular or rotational position of an object in 3-dimensional space.

Orthosis: An externally applied device used to modify the structural and functional characteristics of the neuromuscular and/or skeletal systems.

Passive range of motion (PROM): The range of motion of a joint by an external force, usually provided by an allied healthcare member, without any voluntary muscular effort from the patient.

Position: The location of an object in 3-dimensional space.

Posterior: Movement towards the back of the body.

Posterolateral rotatory instability (PLRI): A clinical condition whereby elbow lateral collateral ligament insufficiency results in posterolateral subluxation of the radial head relative to the capitellum and external rotation of the proximal ulna relative to the humerus.

Pronation: Rotation of the forearm such that the palm faces posteriorly when the humerus is dependent.

Proximal: Movement closer to a structure's origin

Range of motion (ROM): The full arc of potential movement of a joint, usually measured in degrees.

Receiver: A device, usually attached to an object being tracked for motion analysis purposes, which senses a signal that has been sent by a transmitter.

Simple elbow dislocation: An injury that destabilizes the elbow because of damage to the ligamentous structures, without associated fracture.

Simulated active range of motion: This occurs when a machine is used to enact forces on tendons of a cadaver to enable movement of a joint.

Supination: Rotation of the forearm such that the palm faces anteriorly when the humerus is dependent.

Transmitter: A device, usually fixed to some location in the operating environment that generates a signal for the purposes of motion tracking.

Ulno-humeral external rotation: Rotation of the ulna about its own long axis away from the midline, relative to the humerus.

Ulno-humeral internal rotation: Rotation of the ulna about its own long axis towards the midline, relative to the humerus.

Valgus: Angulation of a joint such that the distal segment is oriented away from the midline, as compared to the proximal segment.

Varus: Angulation of a joint such that the distal segment is oriented towards the midline, as compared to the proximal segment.

Appendix B –Engineering Design Drawings
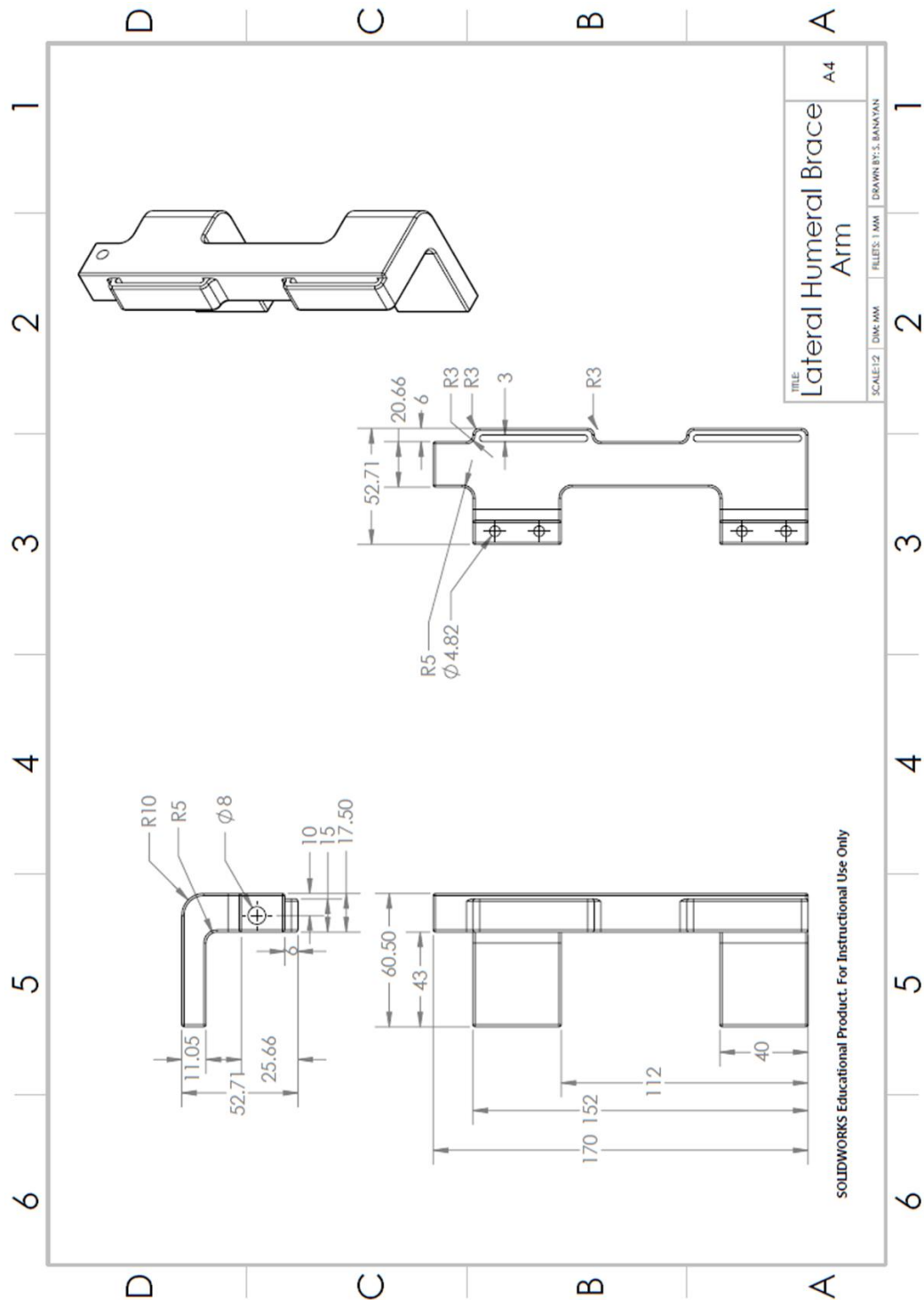
Figure B-1: Lateral Humeral Brace Arm. *The 3D printed HEO arm rests on the lateral side of the upper arm.*
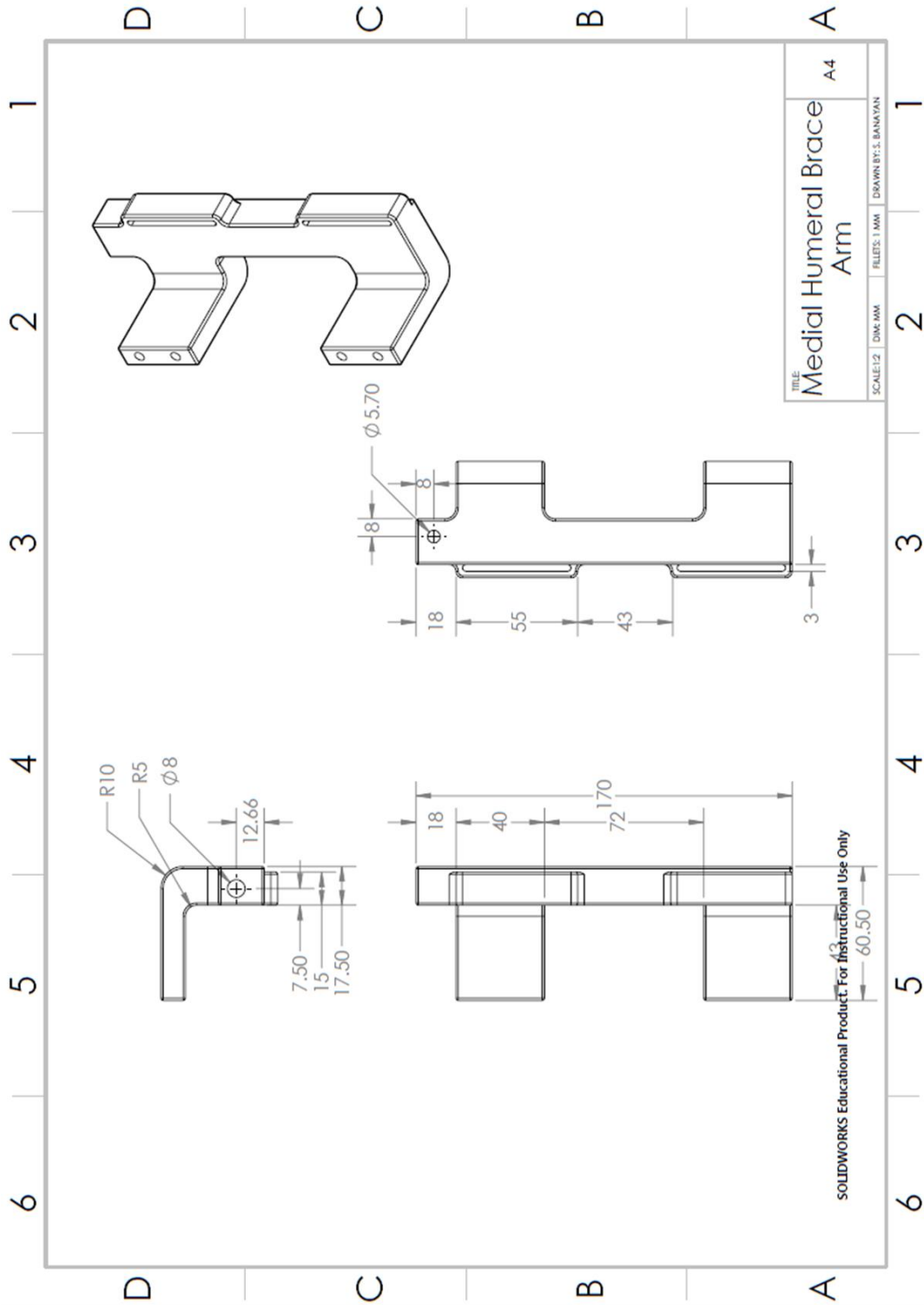
Figure B-2: Medial Humeral Brace Arm. *The 3D printed HEO arm rests on the medial side of the upper arm.*
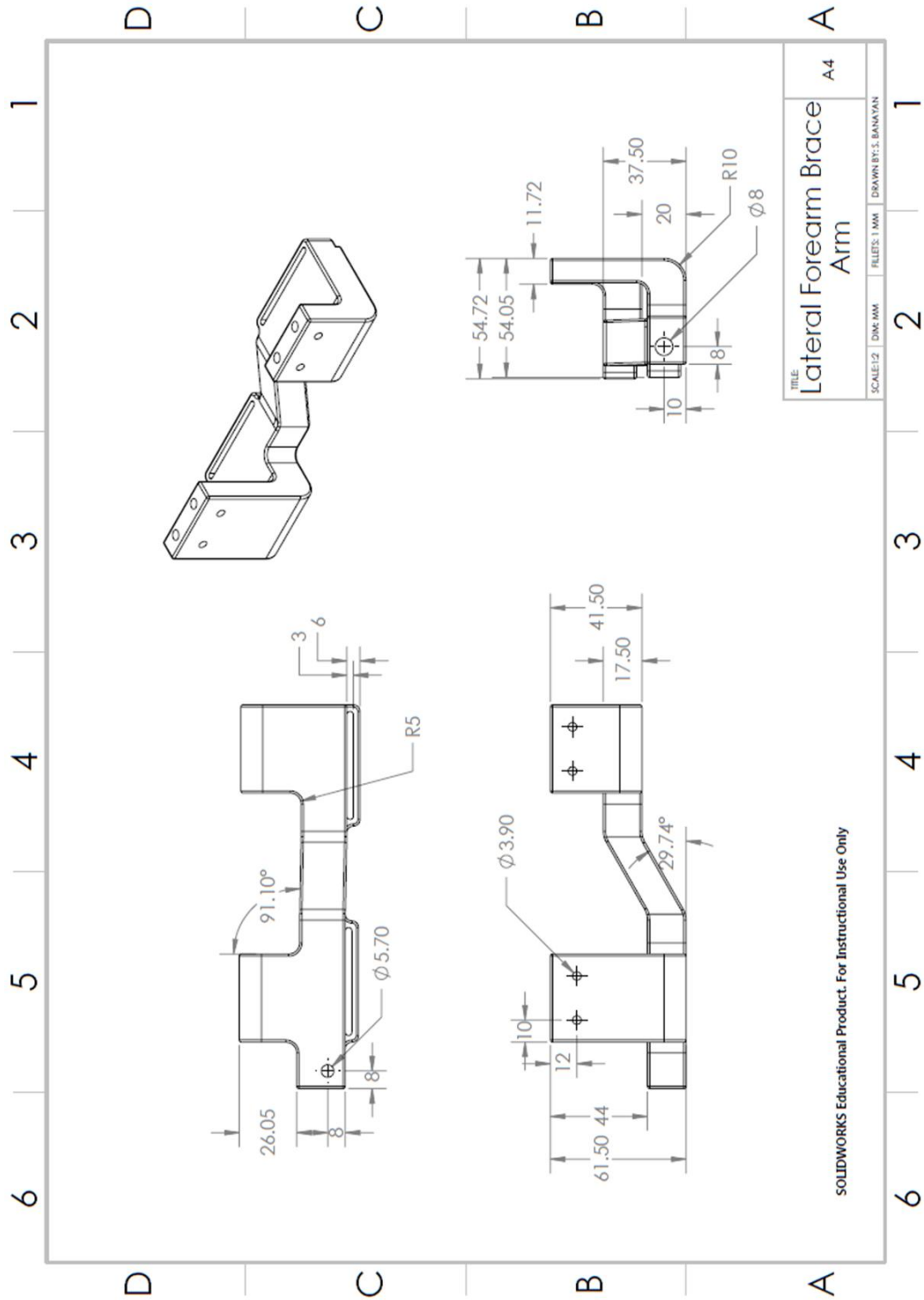
Figure B-3: Lateral Forearm Brace Arm. *The 3D printed HEO arm rests on the lateral side of the forearm.*
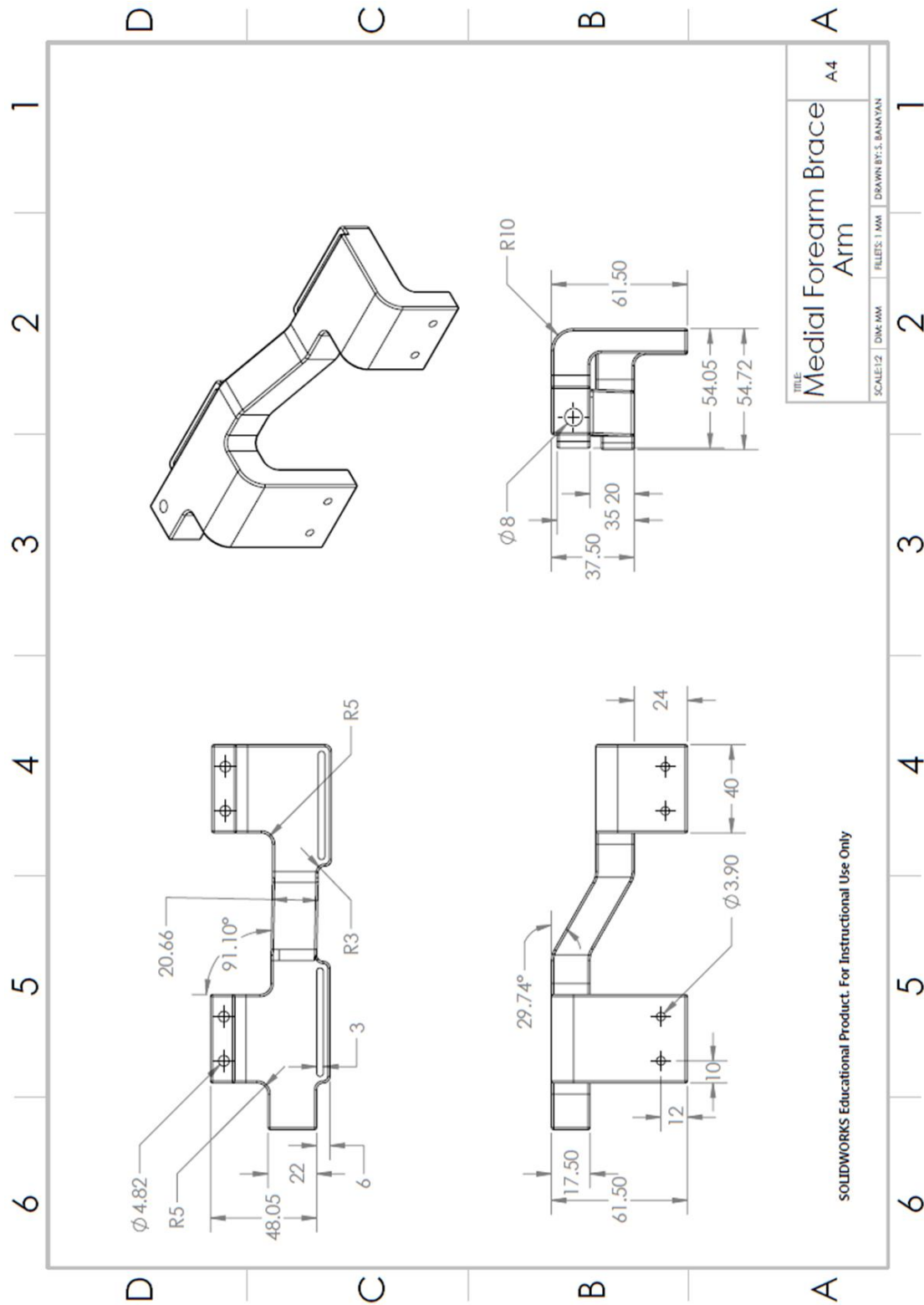
Figure B-4: Medial Forearm Brace Arm. *The 3D printed HEO arm rests on the medial side of the forearm.*
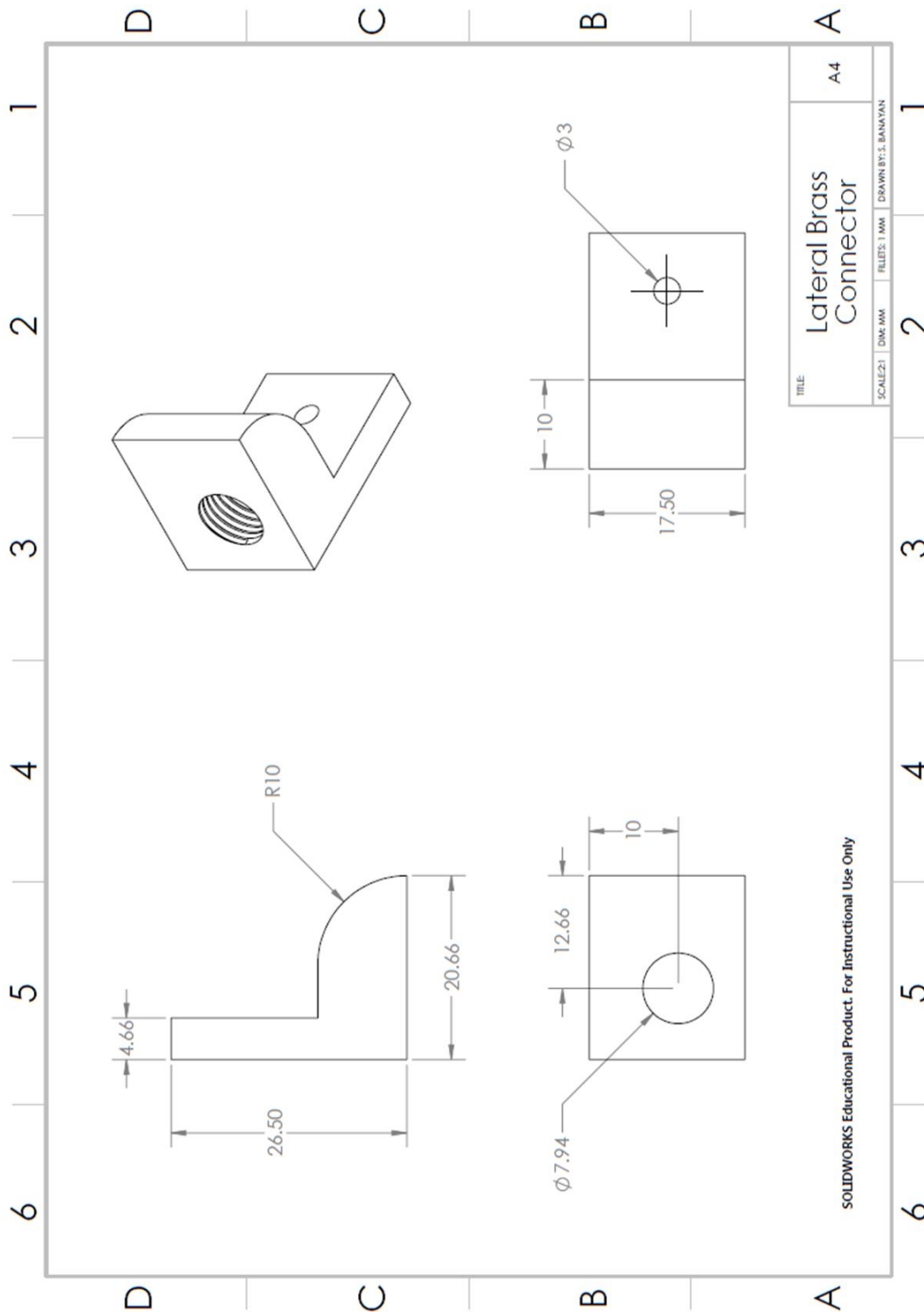
Figure B-5: Lateral Connector. *The lateral brass connector is used to connect the lateral humeral arm to the hinge mechanism. It is also part of the mechanism that allowed HEO valgus angulation.*
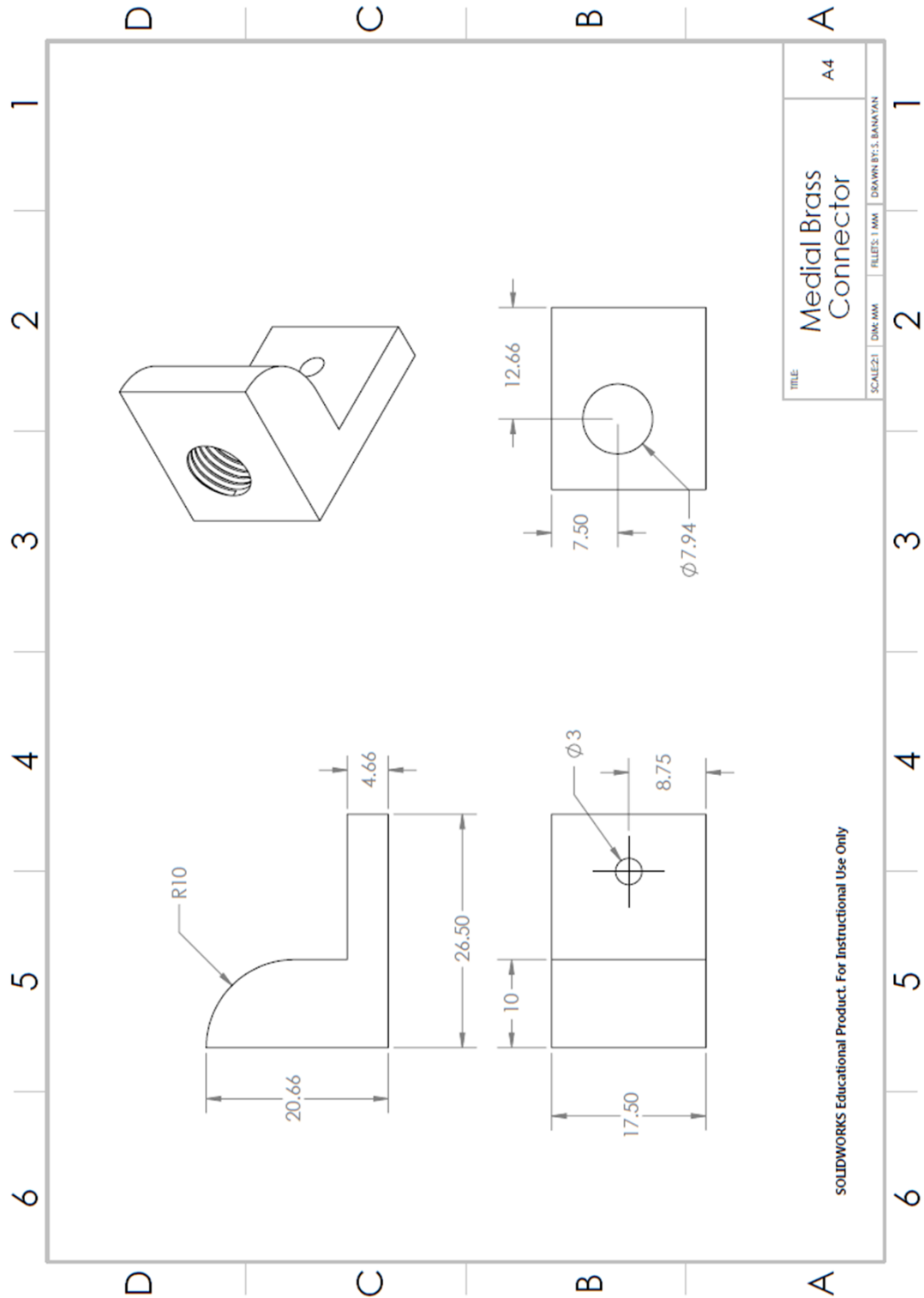
Figure B-6: Medial Connector. *The medial brass connector is used to connect the medial humeral arm to the hinge mechanism. It is also part of the mechanism that allowed HEO valgus angulation.*

153
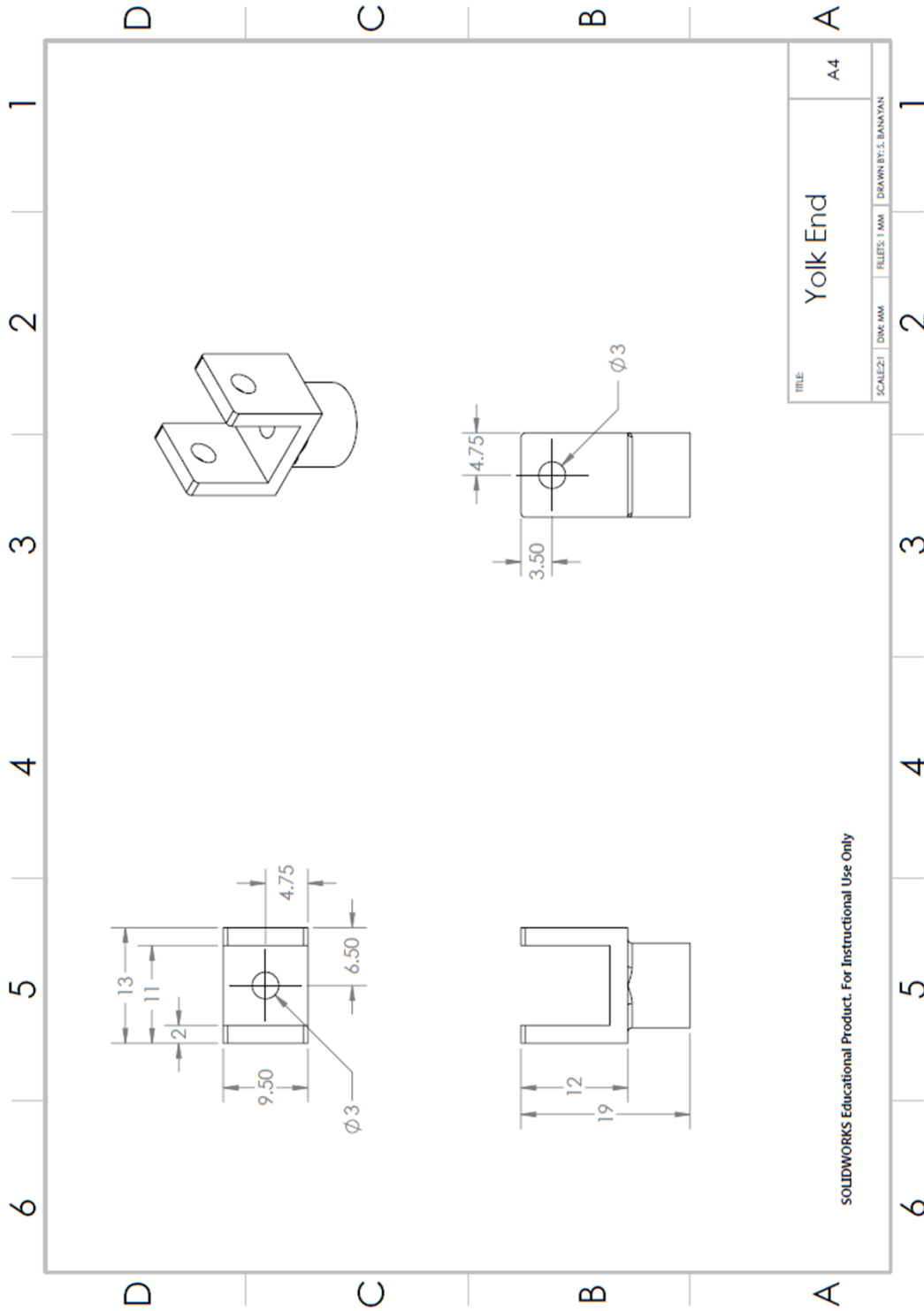
Figure B-7: Yoke End. *The yoke end is used to permit flexion-extension as well as valgus angulation of the HEO.*
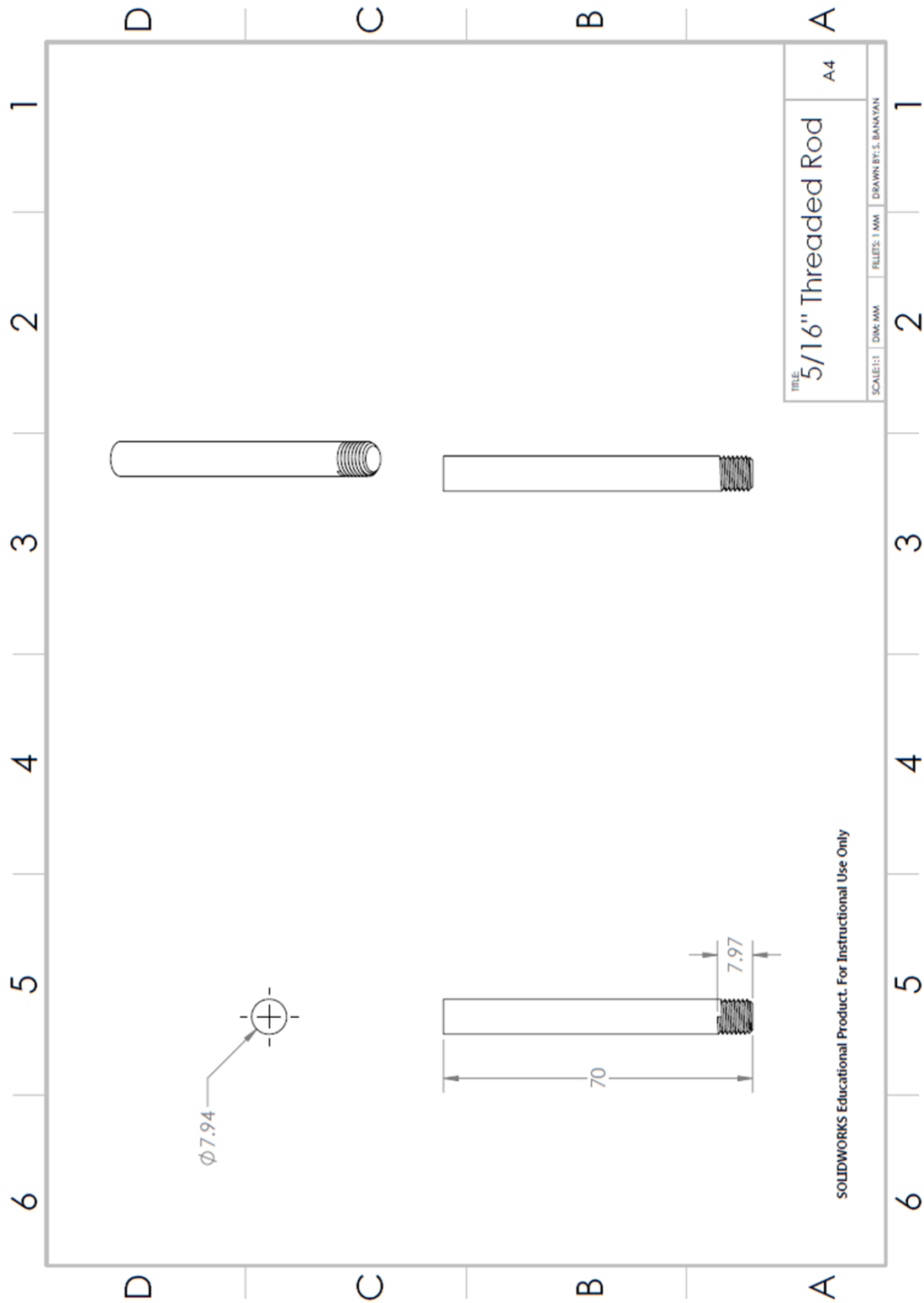
Figure B-8: 5/16'' Threaded Rod. *5/16'' threaded rods were used to connect the proximal HEO arms to the brass connectors.*
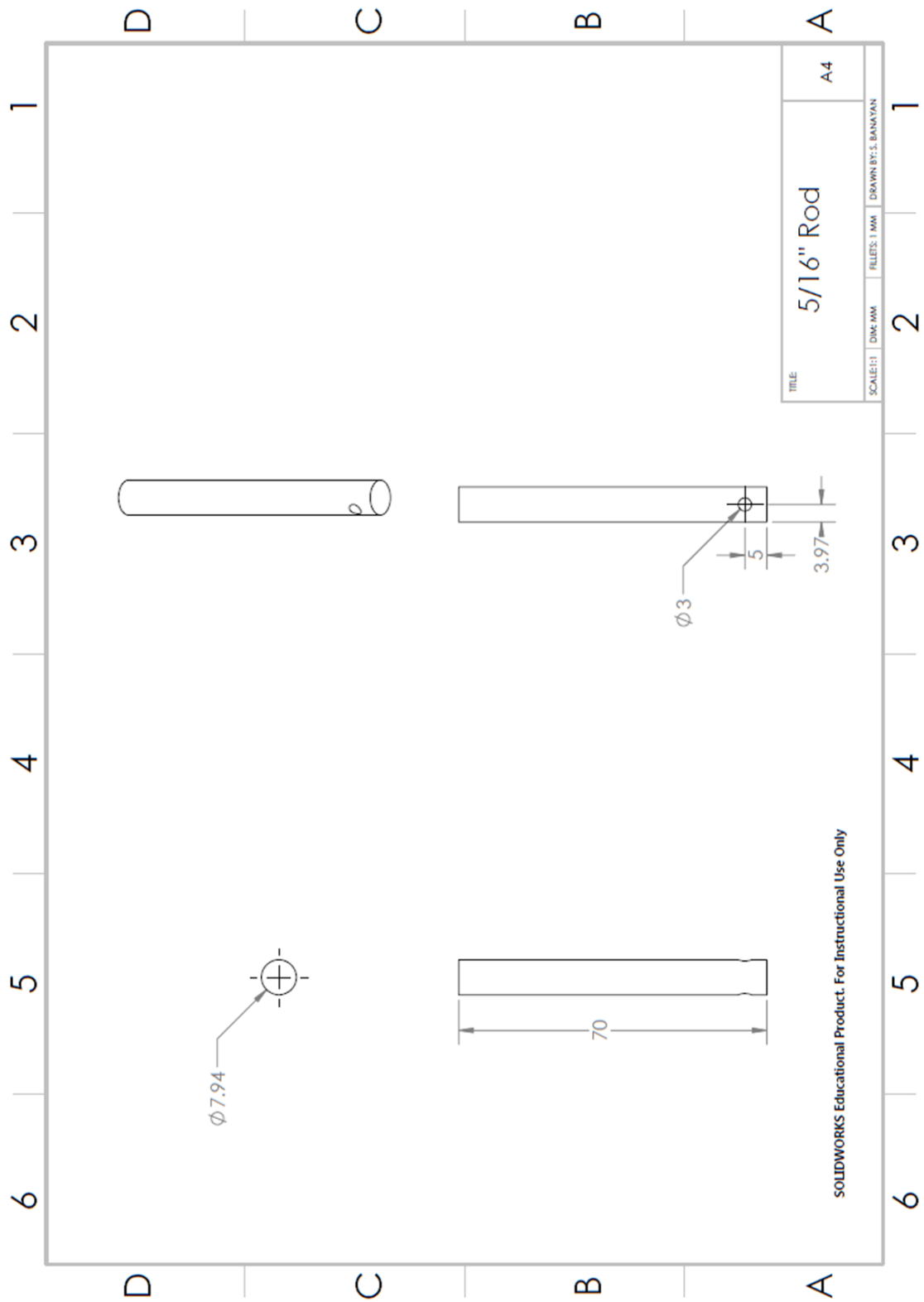
Figure B-9: 5/16'' Rod. *5/16'' rods were used to connect the distal HEO arms to the yoke end to permit HEO flexion-extension.*
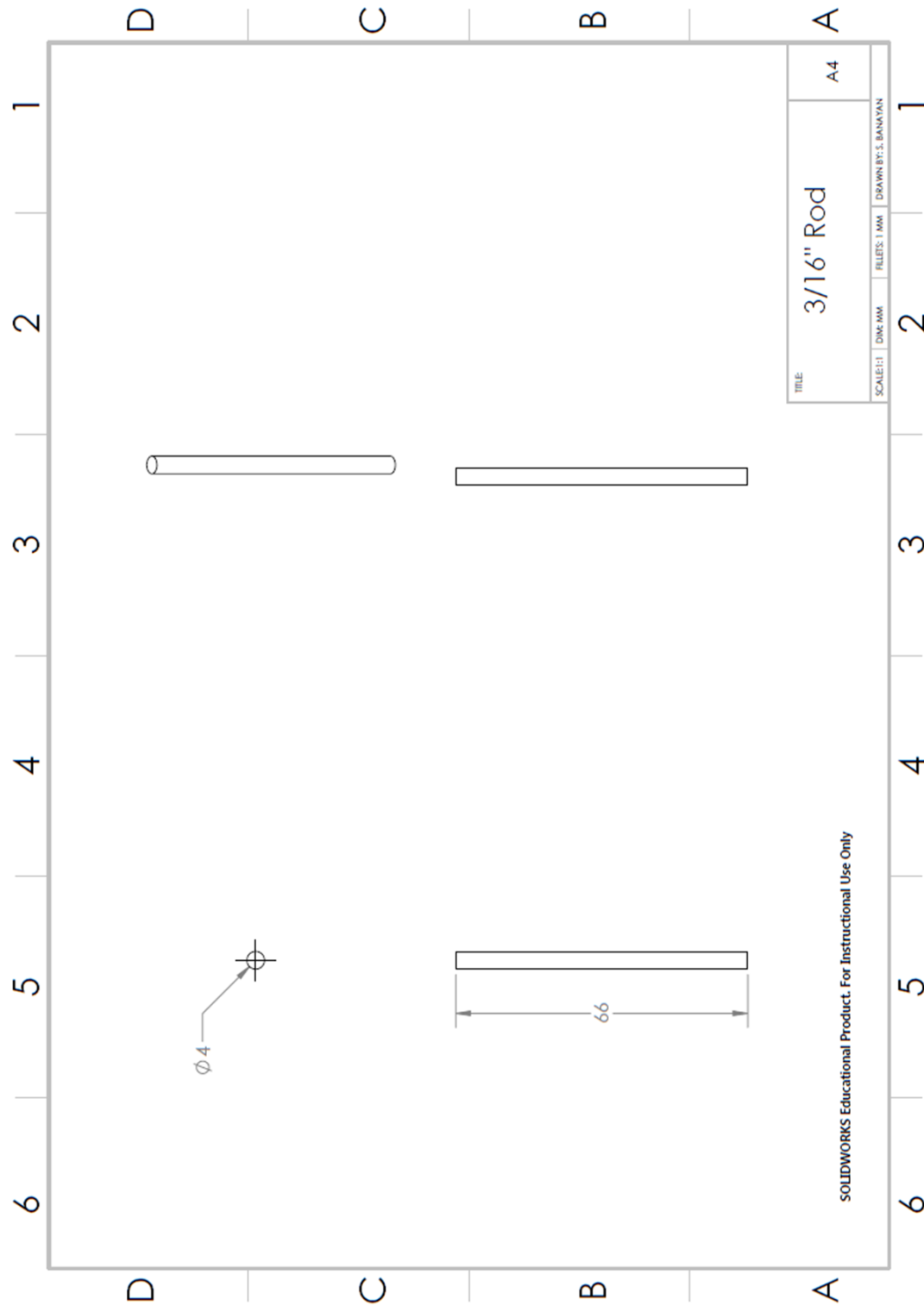
Figure B-10: 3/16'' Rod. *3/16'' rods were used to connect the medial HEO arms to its lateral (mirrored) counterpart.*

Appendix C –Pairwise Comparisons

Table C-1: Pairwise Comparisons for Varus-Valgus Angulation (VVA) in the Vertical Dependent Position

Muscle Activation	Forearm Rotation	LCL Injured State	Flexion Angle (°)	Mean Difference (compared to intact) (°)	Standard Deviation	p
Active	**Pronation**	HEO off	120	0.45	0.10	0.012
		HEO 0°	110	0.83	0.12	0.002
			120	0.92	0.26	0.031
		HEO 10°	120	0.84	0.27	0.049

"Mean Difference" describes the difference in varus-valgus angulation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from pairwise comparisons from a four-way analysis of variance (ANOVA) with muscle activation, forearm rotation, ligament state and flexion angle. Only significant values were stated ($p < 0.05$). Abbreviations: HEO – hinged elbow orthosis and LCL - lateral collateral ligament

Table C-2: Pairwise Comparisons for Varus-Valgus Angulation (VVA) in the Varus Position

Muscle Activation	Forearm Rotation	LCL Injured State	Flexion Angle (°)	Mean Difference (compared to intact) (°)	Standard Deviation	p
Active	Pronation	HEO off	60	7.49	1.69	0.012
			70	9.07	1.51	0.004
			80	10.46	1.14	0.001
			90	11.40	0.60	0.000
			100	11.13	1.65	0.002
			110	9.26	2.88	0.042
		HEO 0°	50	6.01	1.89	0.044
			60	7.21	1.72	0.015
			70	8.46	1.46	0.004
			80	9.30	1.87	0.001
			90	9.66	3.00	0.002
		HEO 5°	40	4.01	0.98	0.016
			50	5.40	0.95	0.005
			60	6.88	0.97	0.009
			70	8.39	1.22	0.002
			80	9.58	1.60	0.004
			90	10.06	2.45	0.016
	Supination	HEO off	70	8.36	2.64	0.044
			80	9.93	2.47	0.018
			90	11.00	1.98	0.005
			100	10.70	1.54	0.002
			110	8.77	2.54	0.032
		HEO 0°	60	7.32	2.15	0.034
			70	8.66	1.71	0.007
			80	9.74	1.24	0.001
			90	10.18	1.43	0.002
			100	9.44	1.93	0.008
			110	7.72	2.20	0.030
		HEO 5°	50	5.76	1.68	0.033
			60	7.16	1.50	0.009
			70	8.62	1.21	0.002
			80	9.75	1.117	0.001
			90	10.28	1.29	0.001
			100	9.61	1.69	0.005
			110	7.85	1.93	0.017
			120	5.86	1.86	0.045
			60	6.17	1.82	0.034

	70	7.64	2.12	0.021
	80	9.00	2.48	0.026
HEO 10°	90	9.64	2.74	0.034
	100	9.07	2.68	0.034
	110	7.46	2.02	0.025
	120	5.86	1.86	0.049
	60	5.71	1.67	0.033
	70	7.21	1.86	0.021
	80	8.48	2.15	0.019
HEO 15°	90	9.20	2.59	0.029
	100	8.69	2.62	0.037
	110	7.16	2.04	0.030
	120	5.71	1.85	0.049

"Mean Difference" describes the difference in varus-valgus angulation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from pairwise comparisons from a four-way analysis of variance (ANOVA) with muscle activation, forearm rotation, ligament state and flexion angle. Only significant values were stated (p < 0.05). Abbreviations: HEO – hinged elbow orthosis and LCL - lateral collateral ligament

Table C-3: Pairwise Comparisons for Internal-External Rotation (IER) in the Vertical Dependent Position

Muscle Activation	Forearm Rotation	LCL Injured State	Flexion Angle (°)	Mean Difference (compared to intact) (°)	Standard Deviation	p
Active	Pronation	HEO 5°	10	0.96	0.26	0.024
			20	0.78	0.23	0.038
Passive	Supination	HEO off	80	6.84	2.16	0.044
			90	8.15	2.20	0.025
			100	8.87	1.90	0.010
			110	8.00	0.77	0.000
			120	6.28	1.52	0.016
		HEO 5°	80	8.33	2.46	0.034
			90	9.37	2.68	0.030

"Mean Difference" describes the difference in internal-external rotation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from pairwise comparisons from a four-way analysis of variance (ANOVA) with muscle activation, forearm rotation, ligament state and flexion angle. Only significant values were stated ($p < 0.05$). Abbreviations: HEO – hinged elbow orthosis and LCL - lateral collateral ligament

Table C-4: Pairwise Comparisons for Internal-External Rotation (IER) in the Varus Position

Muscle Activation	Forearm Rotation	LCL Injured State	Flexion Angle (°)	Mean Difference (compared to intact) (°)	Standard Deviation	p
Active	Pronation	HEO off	80	7.88	2.20	0.028
			90	8.84	2.13	0.016
			100	9.06	1.95	0.010
		HEO 0°	60	5.82	1.85	0.045
			70	6.81	1.95	0.031
			80	7.61	2.29	0.037
		HEO 5°	60	5.78	1.60	0.027
			70	6.92	1.72	0.018
			80	8.01	1.84	0.013
			90	8.76	2.13	0.016
			100	8.89	2.34	0.022
		HEO 10°	70	6.01	1.93	0.047
			80	7.26	2.23	0.040
			90	8.32	2.64	0.045
			100	8.75	2.82	0.048
		HEO 15°	80	6.88	2.11	0.040
			90	7.88	2.42	0.040
			100	8.25	2.57	0.042
			110	7.50	2.23	0.035
	Supination	HEO off	70	6.09	1.82	0.036
			80	7.29	1.89	0.021
			90	8.33	1.88	0.012
			100	8.53	1.32	0.003
			110	7.38	2.17	0.034
		HEO 0°	70	7.01	2.21	0.044
			80	8.00	2.20	0.026
			90	8.81	2.30	0.021
			100	8.96	2.35	0.022
		HEO 5°	70	7.13	2.09	0.033
			80	8.23	2.25	0.025
			90	9.09	2.39	0.022
			100	9.16	2.26	0.017
		HEO 10°	100	8.60	2.58	0.036
			110	7.71	2.44	0.045
		HEO 15°	90	7.94	2.58	0.049
			100	8.15	2.41	0.034
			110	7.27	2.05	0.029

"Mean Difference" describes the difference in internal-external rotation angle between the intact and ligament injured state. p-values describe the significance of the ligament states resulting from pairwise comparisons from a four-way analysis of variance (ANOVA) with muscle activation, forearm rotation, ligament state and flexion angle. Only significant values were stated ($p < 0.05$). Abbreviations: HEO – hinged elbow orthosis and LCL - lateral collateral ligament

Table C-5: Pairwise Comparisons for LCL Loads

Arm Position	Muscle Activation	Forearm Rotation	LCL Injured State	Flexion Angle (°)	Mean Difference (N)	Standard Deviation	p
Vertical Dependent	Active	Supination	HEO 0°	50	2.65	0.82	0.029
			HEO 0°	40	2.29	0.72	0.030

"Mean Difference" describes the difference in LCL tension between the injured state without an HEO and injured state with the specified HEO. p-values describe the significance of the ligament states resulting from pairwise comparisons from a four-way analysis of variance (ANOVA) with muscle activation, forearm rotation, ligament state and flexion angle. Only significant values were stated ($p < 0.05$). Abbreviations: HEO – hinged elbow orthosis and LCL - lateral collateral ligament

Sara Banayan

Education

2015 – 2017	**MESc, Biomedical Engineering**
	Western University, London Ontario
	Project Title: The Kinematic and Biomechanical Effects of Bracing on the Rehabilitation of the LCL Injured Elbow
	Supervisors: Dr. James A. Johnson, Dr. Graham J.W. King
2011 – 2015	**BEng, Biomedical Engineering**
	University of Guelph, Guelph Ontario
	Graduated with distinction

Research Funding, Scholarships and Awards

2016	Lawson Internal Research Fund
2016	Western Graduate Research Scholarship
2012-2015	William and Nona Heaslip Scholarship ($45 000 annually)
2011 – 2015	Dean's Honours List, University of Guelph

Memberships

2017 – present	**Engineering Intern (EIT)** Professional Engineers Ontario

Technical Experience

- Experience with the use of the following programs: SolidWorks, Materialise Mimics, AutoCAD, LabView, SPSS, MATLAB and Microsoft Office
- Coding knowledge in C++ as well as experience programming Arduino and Teensy microcontrollers

Professional Experience

Oct 2016 – Oct 2017   **Office Manager,** Professional Engineers Ontario (PEO) Western Regional Office, London Ontario

Supervisor: Dr. Ralph Buchal
- Organize and execute PEO Regional Events
- Educate undergraduate and graduate students on obtaining Engineering Intern (EIT) or Professional Engineer (P.Eng) licenses

Sept 2016 – Apr 2017   **Operations Manager**, Prep101, London Ontario
- Managed a team of over 30 staff, including scheduling, training and mentoring
- Responsible for coordinating materials, instructors, staff, students and payments

May – Aug 2015   **Undergraduate Research Assistant** University of Guelph, Guelph Ontario

Supervisor: Dr. Emily Chiang
- Researched environmental factors affecting algal blooms in Southern Ontario Lakes
- Conducted lab work to investigate the factors affecting the proliferation of cyanobacteria

Teaching Experience

Sept –Dec 2016/17   **Guest Lecturer and Teaching Assistant,** CBE2214 "Thermodynamics"

University of Western Ontario, London Ontario

Supervisor: Dr. Sohrab Rohani
- Lectured 100+ students on the mass and energy balance equations for pumps, turbines, compressors, nozzles and diffusers
- Taught tutorials relating to thermodynamic laws, steam tables, thermodynamic cycles as well as other thermodynamic concepts

May 2016 – present	**High School and University Tutor,** Mathematics and Physics

Tutor Doctor, London Ontario
- Customize individual lesson plans unique to each student
- Devise alternative methods to explain difficult concepts

Jan-Apr 2016/17	**Guest Lecturer and Teaching Assistant,** CBE2291 "Computational Methods for Engineers"

University of Western Ontario, London Ontario

Supervisor: Dr. Lars Rehmann
- Guest lectured 70+ students on methods to solve linear algebra equations using MATLAB
- Lead bi-weekly tutorials with strong focus on chemical and biochemical engineering concepts and MATLAB applications
- Grade assignments and exams

Projects and Presentations

Jan 2016 – present	**Investigating the Effect of Bracing on the LCL-Compromised Elbow**

- Designed an adjustable brace with varying varus-valgus axis using SolidWorks to investigate the effects of hinged elbow orthoses following ligamentous injury
- Investigated various arm positions, forearm rotations, ligament repair techniques and brace angles

Jan 2017	**Optimizing the Rehabilitation of Elbow Dislocations**

University of Western Ontario, London Ontario
- Discussed the effects of ligamentous elbow injuries on doctor and surgeon recommended rehabilitation positions and practices
- Presented research to Engineering faculty and graduate students

Dec 2015	**Rehabilitation Therapy Band**

University of Guelph, Guelph Ontario
- Designed a lower limb rehabilitation band combining electrical, mechanical and systems components to improve quadriceps muscle strength
- Analyzed strengthening effects using a BioDex machine

Dec 2015	**Drive Alert Feedback System**

University of Guelph, Guelph Ontario
- Programmed a Teensy3.2 microcontroller to filter and analyze electromyogram (EMG), electrocephalogram (EEG) and pulse sensor signals and determine driver alertness

Apr 2015	**Astaxanthin Extraction from Haematococcus Pluvialis for Cosmetic Applications**

University of Guelph, Guelph Ontario
- Designed a process to optimize Haematococcus Pluvialis growth and extraction through improvements in purification and form of extraction

Apr 2015	**Plug Flow Anaerobic Digester**

University of Guelph, Guelph Ontario
- Designed an vertical plug flow anaerobic digester for the University of Guelph to improve diversion of campus waste to landfills
- Design parameters including physical dimensions, insulation, pH regulation, hardware and heating, among others, were considered

Publications

Haque, F., Banayan, S., Yee, J., Chiang, Y.W. (2017) Extraction and applications of cyanotoxins and other cyanobacterial secondary metabolites. Chemosphere: 183, 164-175.