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Optimizing the Rehabilitation of Elbow Lateral Collateral Ligament Injuries

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Graduate Program in Medical Biophysics

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Abstract

Elbow lateral collateral ligament (LCL) injuries frequently arise following trauma, and can result in disabling instability. Typically such injuries are managed with immobilization followed by a graduated exercise regime; however there is minimal biomechanical evidence to support current treatment protocols. This investigation examines the *in vitro* effectiveness of several rehabilitation techniques using a custom elbow motion simulator. It was found that active range of motion is safest in the overhead position ($n = 7$). Early motion in this position may reduce the incidence of elbow stiffness without compromising ligament healing following LCL injury. Forearm pronation and active motion stabilize the LCL-deficient elbow, while varus positioning worsens instability. It was also found that a hinged elbow orthosis did not significantly improve *in vitro* elbow stability following LCL injury ($n = 7$). However, such orthoses may be useful in keeping the forearm in the more stable pronated position. Future research directions are proposed, with suggestions on applying this methodology to other elbow injuries.

Keywords

elbow, lateral collateral ligament, instability, posterolateral rotatory instability, biomechanics, rehabilitation, overhead motion protocol, varus, orthosis, brace, active motion, range of motion

Co-Authorship Statement

Chapter One

Ranita Manocha – sole author

Chapter Two

Ranita Manocha – study design, specimen preparation, data collection and analysis, statistical analysis, wrote manuscript

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George Athwal – specimen preparation

James Johnson – study design, reviewed manuscript

Graham King – study design, specimen preparation, reviewed manuscript

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George Athwal – specimen preparation

James Johnson – study design, reviewed manuscript

Graham King – study design, specimen preparation, reviewed manuscript

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Table of Contents

Abstract	i
Co-Authorship Statement.....	ii
Acknowledgments.....	iii
Table of Contents	iv
List of Figures	viii
List of Tables	x
List of Appendices	xi
List of Abbreviations	xii
Chapter 1	1
1 Introduction	1
1.1 Elbow Anatomy	1
1.1.1 <i>Elbow Osteology</i>	1
1.1.2 <i>The Capsule and Ligaments</i>	1
1.1.3 <i>Muscles</i>	5
1.2 Elbow Kinematics	9
1.3 Elbow Stability.....	18
1.3.1 <i>Static Stabilizers</i>	18
1.3.2 <i>Dynamic Stabilizers</i>	19
1.4 Lateral Collateral Ligament Injury	20
1.4.1 <i>Posterolateral Rotatory Instability (PLRI)</i>	23
1.5 Management of Lateral Collateral Ligament Instability.....	25
1.5.1 <i>Muscle Activation</i>	26
1.5.2 <i>Arm Position</i>	26
1.5.3 <i>Forearm Position</i>	29

1.5.4	<i>Orthoses</i>	29
1.6	Design Principles in Hinged Elbow Orthoses.....	32
1.7	Upper Limb Biomechanical Testing.....	32
1.7.1	<i>Joint Motion Simulation Techniques</i>	32
1.7.2	<i>Kinematic Assessment</i>	35
1.7.3	<i>In Vitro Elbow Motion Simulation</i>	40
1.8	Study Rationale.....	42
1.9	Objectives	43
1.10	Hypotheses	44
1.11	Thesis Overview	44
1.12	References	46
Chapter 2	56
2	Overhead Rehabilitation in Lateral Elbow Injuries	56
2.1	Introduction.....	57
2.2	Methods.....	59
2.3	Results.....	65
2.4	Discussion.....	70
2.5	Conclusion	72
2.6	Acknowledgements.....	73
2.7	References.....	74
Chapter 3	78
3	Effectiveness of Bracing in Elbow Lateral Collateral Ligament Injuries	78
3.1	Introduction.....	79
3.2	Methods.....	80
3.3	Results.....	81

3.3.1	<i>Dependent Position</i>	81
3.3.2	<i>Overhead Position</i>	82
3.3.3	<i>Horizontal Position</i>	82
3.3.4	<i>Varus Position</i>	83
3.4	Discussion	83
3.5	Conclusion	94
3.6	Acknowledgements	95
3.7	References	96
Chapter 4	99
4	General Discussion, Conclusions, and Future Directions	99
4.1	Summary of Hypotheses and Clinical Relevance	99
4.1.1	<i>Instability with Extent of Lateral Soft Tissue Injury</i>	99
4.1.2	<i>Arm and Forearm Position in the Rehabilitation of Elbow Lateral Collateral Ligament Injuries</i>	100
4.1.3	<i>Bracing in the Rehabilitation of Elbow Lateral Collateral Ligament Injuries</i>	100
4.2	Strengths and Limitations	100
4.3	Future Directions	102
4.3.1	<i>Applying Methodology to Other Clinical Paradigms</i>	102
4.3.2	<i>Expansion of Lateral Collateral Ligament Injury Rehabilitation Research Paradigms</i>	102
4.4	Conclusion	103
4.5	References	105
Appendix A – Glossary	106
Appendix B – Appendix to Chapter 3	109
B.1	Impact of Hinged Elbow Orthosis in the Intact Elbow	109

B.2 Power for Detecting Differences in Elbow State	109
Appendix C – Copyright Releases	112
Curriculum Vitae	128

List of Figures

Figure 1-1 - Elbow osteology.	2
Figure 1-2 - Osteology of the ulnohumeral joint.	3
Figure 1-3 - The lateral collateral ligament of the elbow.	4
Figure 1-4 - The medial collateral ligament of the elbow.	6
Figure 1-5 - Elbow motions.	7
Figure 1-6 - Elbow flexors of the anterior arm.	8
Figure 1-7 - Elbow extensors of the posterior arm.	10
Figure 1-8 - Flexor-pronator muscles of the anterior forearm.	11
Figure 1-9 - Extensor-supinator muscles of the posterior forearm.	12
Figure 1-10 - Dynamic screw displacement axis changes during elbow flexion.	14
Figure 1-11 - Flexion-extension axis of the elbow joint.	15
Figure 1-12 - Carrying angle.	16
Figure 1-13 - Kinematic references for the elbow.	17
Figure 1-14 - Disruption of the circle of Horii with increasing elbow instability	21
Figure 1-15 - O'Driscoll stages of elbow instability.	22
Figure 1-16 - Posterolateral rotatory instability.	24
Figure 1-17 - Gravity-loaded humerus positions.	27
Figure 1-18 - Theoretical elbow joint compressive forces in overhead position.	28
Figure 1-19 - Mayo Clinic Elbow Brace.	30

Figure 1-20 - Schematic outlining general operation of motion tracking systems.	36
Figure 1-21 - Flock of Birds® electromagnetic tracking system.....	39
Figure 2-1 - Custom elbow motion simulator in multiple positions.	60
Figure 2-2 - Custom humeral clamp.	61
Figure 2-3 - Determination of ulnar and humeral joint coordinate systems.	64
Figure 2-4 - Mean ulnohumeral kinematic profiles during elbow extension with forearm pronated.....	66
Figure 2-5 - Mean ulnohumeral kinematic profiles during elbow extension with forearm supinated.	68
Figure 3-1 - Mean ulnohumeral rotation with arm dependent.	84
Figure 3-2 - Mean ulnohumeral rotation with arm overhead.....	85
Figure 3-3 - Mean ulnohumeral rotation with arm horizontal.	86
Figure 3-4 - Mean ulnohumeral rotation with arm varus.....	87

List of Tables

Table 2-1 - Effect of arm position and muscle activation on elbow stability during extension with forearm pronated.	67
Table 2-2 - Effect of arm position and muscle activation on elbow stability during extension with forearm supinated.	69
Table 3-1 - Impact of hinged elbow orthosis on elbow stability during extension with forearm pronated.	88
Table 3-2 - Impact of hinged elbow orthosis on elbow stability during extension with forearm supinated.	89
Table 3-3 - Pairwise comparisons for significant effects of elbow state on ulnohumeral rotation during elbow extension.	90
Table B-1 - Impact of hinged elbow orthosis on ulnohumeral rotation in the intact elbow.	110
Table B-2 - Power analysis for repeated measures ANOVAs in Chapter 3.	111

List of Appendices

Appendix A – Glossary.....	106
Appendix B – Appendix to Chapter 3.....	109
B.1 Impact of Hinged Elbow Orthosis in the Intact Elbow.....	110
B.2 Power for Detecting Differences in Elbow State.....	111
Appendix C – Copyright Releases.....	112

List of Abbreviations

AAROM: active-assisted range of motion

ACL: anterior cruciate ligament (of the knee)

ADL: activity of daily living

ANOVA: analysis of variance

AROM: active range of motion

CEO: common extensor origin

CSA: cross-sectional area

ECRB: extensor carpi radialis brevis

ECRL: extensor carpi radialis longus

EDC: extensor digitorum communis

EMG: electromyography

FEM: finite element model

FCR: flexor carpi radialis

FCU: flexor carpi ulnaris

HEO: hinged elbow orthosis

HULC: Roth-McFarlane Hand and Upper Limb Centre

JCS: joint coordinate system

LCL: lateral collateral ligament (of the elbow)

LUCL: lateral ulnar collateral ligament

MCL: medial collateral ligament (of the elbow)

PLRI: posterolateral rotatory instability

RF: radiofrequency

RMS: root-mean-square

ROM: range of motion

SD: standard deviation

STA: soft tissue artifact

Chapter 1

1 Introduction

OVERVIEW: *This chapter reviews the anatomy of the elbow joint and its supporting capsular and ligamentous structures; normal elbow kinematics; mechanisms of injury to the lateral collateral ligament (LCL) of the elbow; management of LCL injuries, with special reference to bracing; and general principles of upper limb biomechanical testing. The rationale, objectives, and hypotheses pertaining to the thesis are also outlined.*

1.1 Elbow Anatomy

1.1.1 *Elbow Osteology*

The elbow joint is formed by the convergence of three bones: the humerus, the radius, and the ulna (Morrey, 2000a). Figures 1-1 and 1-2 outline important bony landmarks that enable the more proximal humerus to articulate with the more distal radius and ulna to form the three articulations of the elbow joint (Morrey, 2000a; Stroyan & Wilk, 1993). The trochlea of the distal medial humerus articulates with the greater sigmoid notch of the proximal ulna, forming the ulnohumeral joint. The capitellum of the distal lateral humerus articulates with the radial head, forming the radio-capitellar joint. The proximal radius and the lesser sigmoid notch of the proximal ulna articulate to form the proximal radioulnar joint (Morrey, 2000a).

1.1.2 *The Capsule and Ligaments*

The elbow joint is stabilized by the lateral and medial collateral ligaments and by the elbow joint capsule (Morrey, 2000a; Szekeres *et al.*, 2008). The lateral collateral ligament (LCL) is a Y-shaped structure that consists of the lateral ulnar collateral ligament (LUCL), annular ligament, and the radial collateral ligament (Figure 1-3) (King *et al.*, 1993b; Olsen *et al.*, 1996). The LUCL originates on the lateral epicondyle and inserts on the supinator crest of

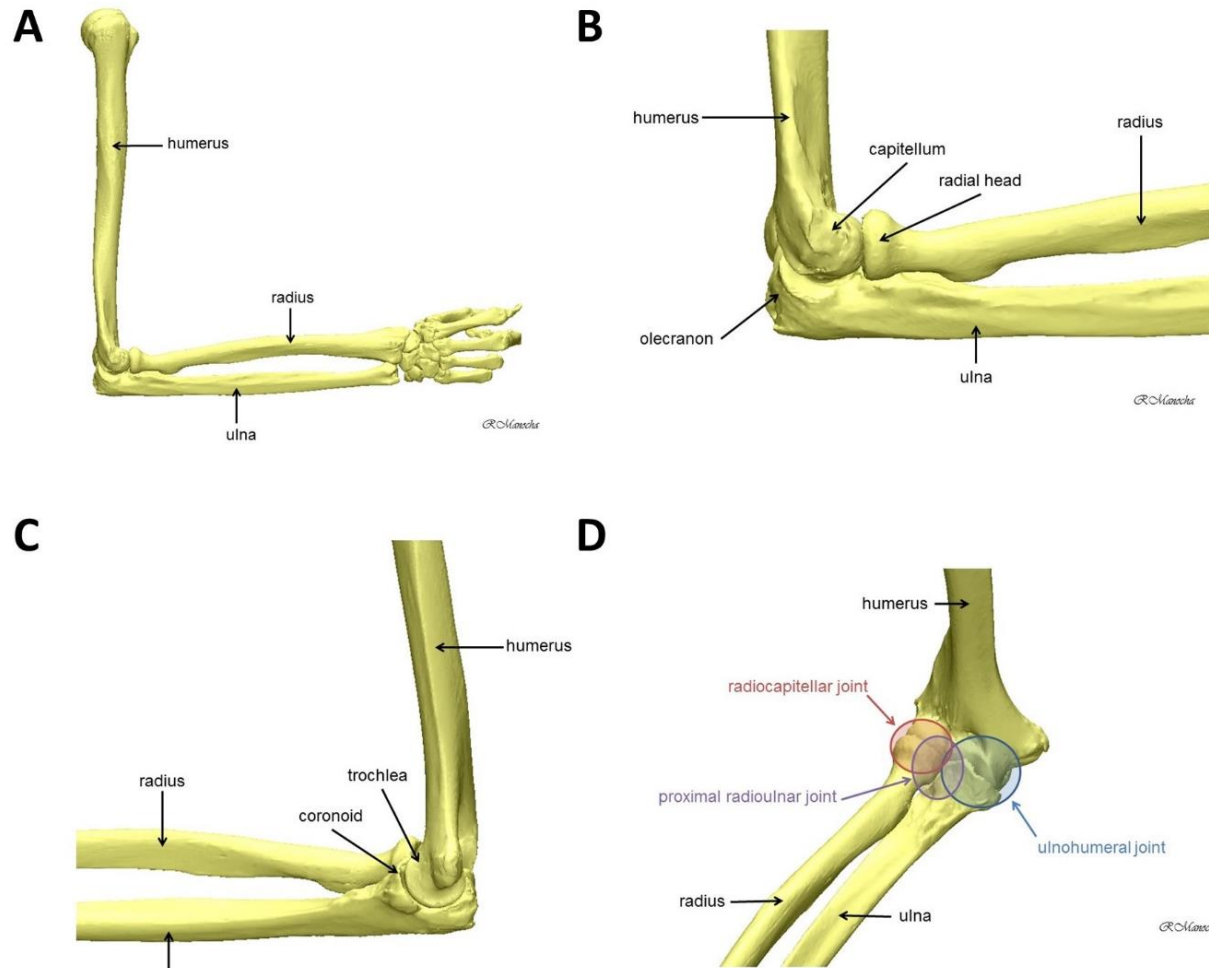


Figure 1-1 - Elbow osteology.

A: Lateral view of right upper extremity. **B:** Lateral view of elbow. **C:** Medial view of elbow. **D:** The three articulations of the elbow. The elbow joint is formed by the convergence of three bones: the humerus, the radius, and the ulna. The trochlea of the distal humerus articulates with the olecranon and coronoid of the proximal ulna, forming the ulnohumeral joint. The capitellum of the distal humerus articulates with the radial head, forming the radiocapitellar joint. The proximal radius and ulna articulate to form the proximal radioulnar joint.

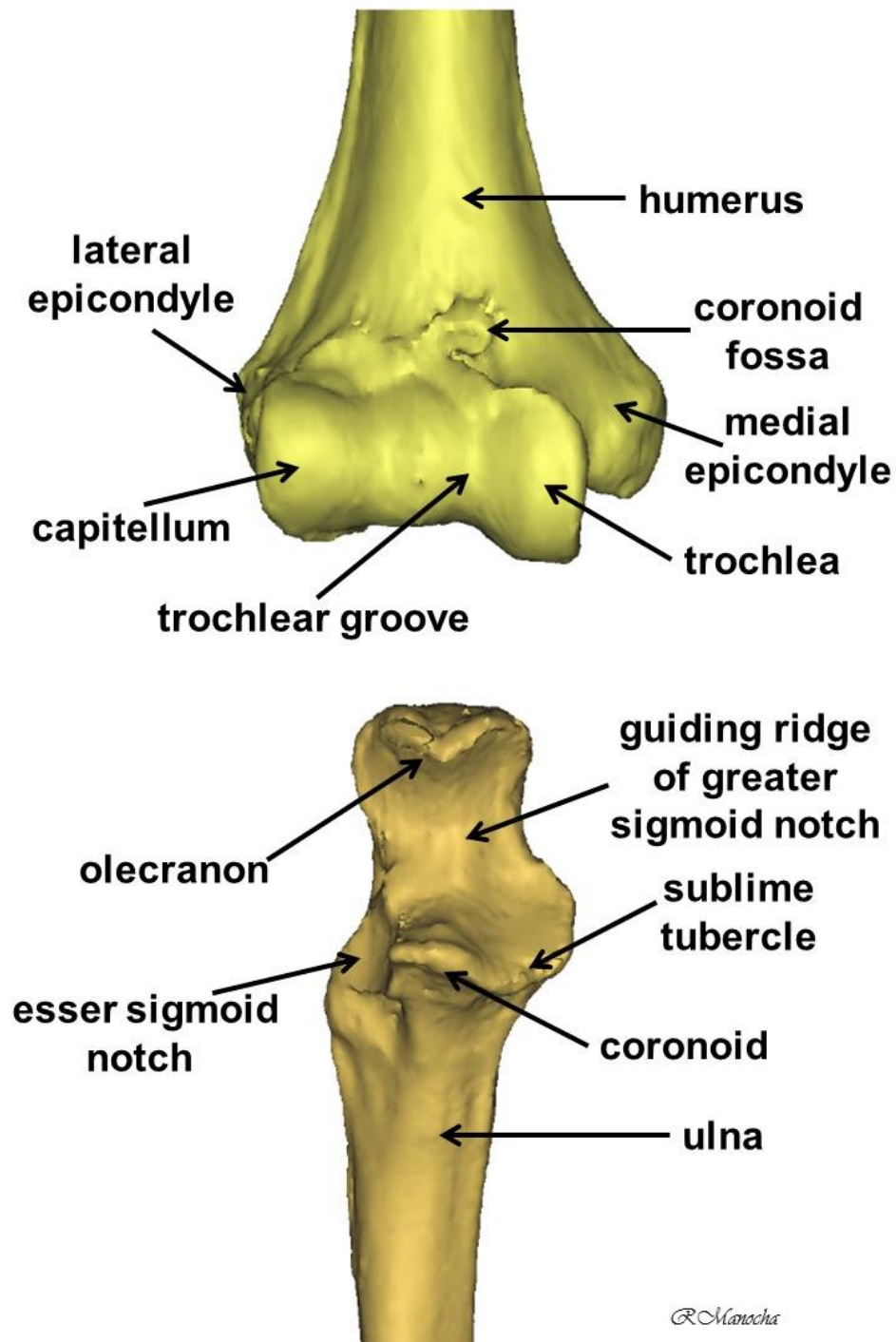


Figure 1-2 - Osteology of the ulnohumeral joint.

The complementary structures of the articular surfaces of the humerus and ulna allow for stability during elbow motion. During flexion, the guiding ridge of the greater sigmoid notch glides in the trochlear groove and at terminal flexion the coronoid enters the coronoid fossa.

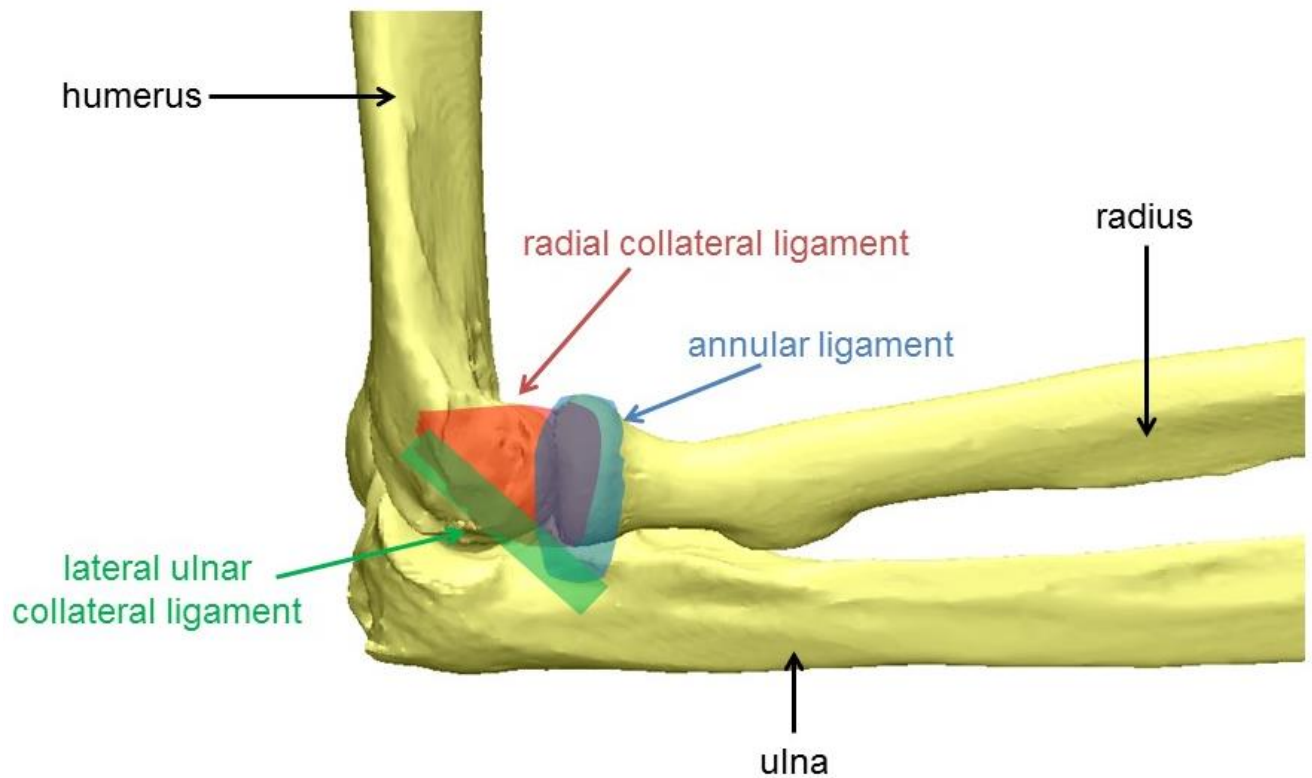


Figure 1-3 - The lateral collateral ligament of the elbow.

This lateral view of the right elbow shows the components of the lateral collateral ligament (LCL): the lateral ulnar collateral ligament, the annular ligament, and the radial collateral ligament.

the ulna, with some fibres passing through the annular ligament (Olsen *et al.*, 1996, Morrey & An, 1985). The annular ligament is oriented circumferentially around the radial head, and originates and inserts on the anterior and posterior margins of the lesser sigmoid notch of the ulna, respectively (King *et al.*, 1993b). The radial collateral ligament originates on the lateral epicondyle of the humerus and inserts into the annular ligament. The LCL tends to be closely apposed and invested with the overlying common extensor muscle origin and the deeper lateral joint capsule (Olsen *et al.*, 1996). The impact of the LCL on elbow stability is discussed further below (see Section 1.3.1.1). The medial collateral ligament (MCL) is a triangular-shaped ligament that consists of an anterior bundle, posterior bundle, and transverse ligament (Figure 1-4) (Fuss, 1991; Pribyl *et al.*, 1999). The anterior and posterior bundles originate on the medial epicondyle. The linear anterior bundle inserts on the sublime tubercle of the ulna, whereas the fan-shaped posterior bundle inserts on the trochlear notch of the ulna. The transverse ligament is inconsistently present. If it exists, it originates on the medial tip of the olecranon and inserts on the inferior medial coronoid process. The contribution of the MCL to elbow stability is briefly reviewed in Section 1.3.1.

The elbow joint capsule is composed of synovial membrane that covers the three articulations that form the elbow joint (King *et al.*, 1993b; Morrey, 2000a; Stroyan & Wilk, 1993). The anterior portion originates proximally above the coronoid and radial fossae. Distally, it attaches to the anterior coronoid and the annular ligament. The posterior capsule attaches proximally above the olecranon fossa and distally along the trochlea, the greater sigmoid notch, and the annular ligament (King *et al.*, 1993b; Morrey, 2000a). The anterior joint capsule becomes taut in elbow extension, whereas the posterior capsule becomes taut in flexion (King *et al.*, 1993b).

1.1.3 *Muscles*

There are four groups of muscles that surround the elbow (Stroyan & Wilk, 1993). These muscles act to flex and extend the elbow, pronate and supinate the forearm (Figure 1-5), and flex and extend the wrist and fingers (King *et al.*, 1993b). The primary elbow flexors cause flexion of the elbow and include the biceps brachii, brachialis, and brachioradialis (Figure 1-6). The biceps brachii is also the primary forearm supinator (Basmajian & Latif,

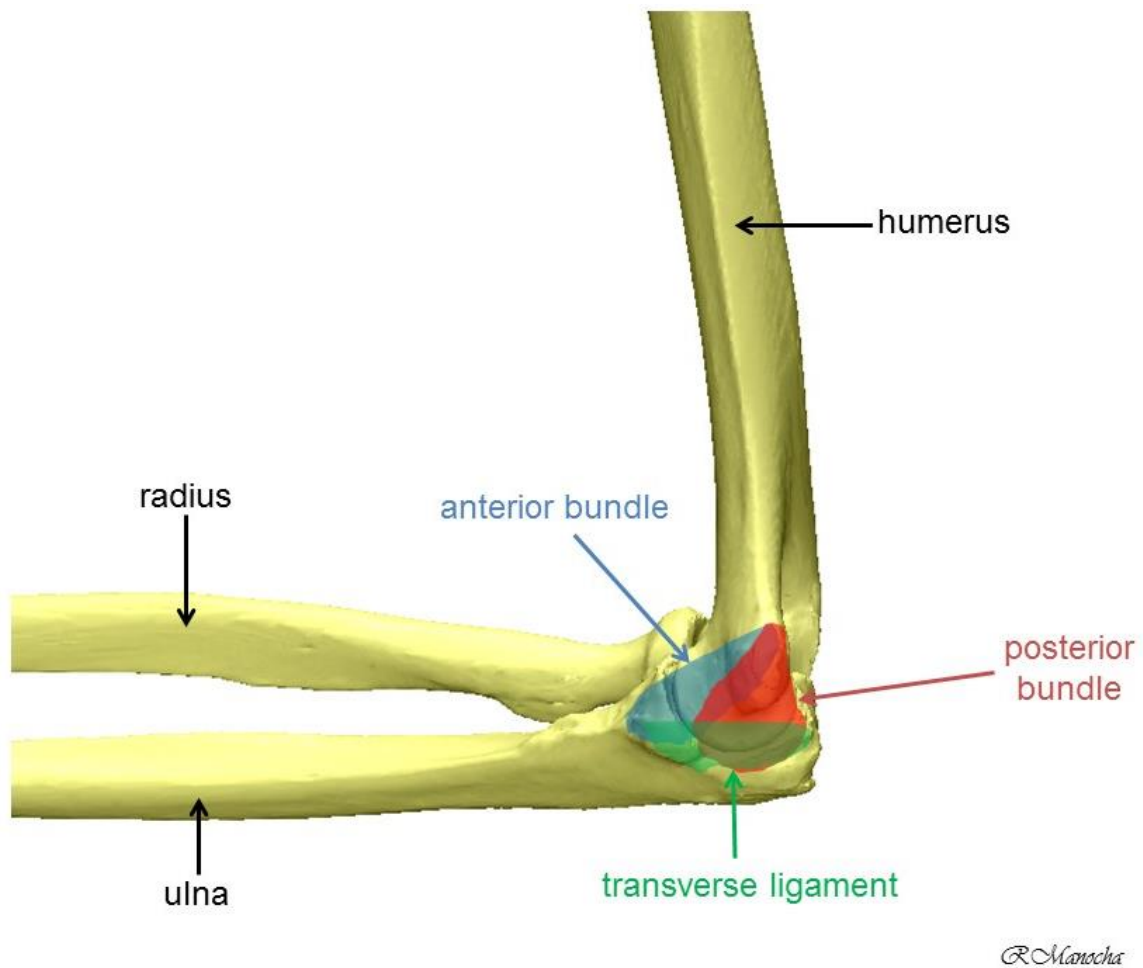


Figure 1-4 - The medial collateral ligament of the elbow.

This medial view of the right elbow shows the components of the medial collateral ligament (MCL): the anterior bundle, the posterior bundle and the transverse ligament.

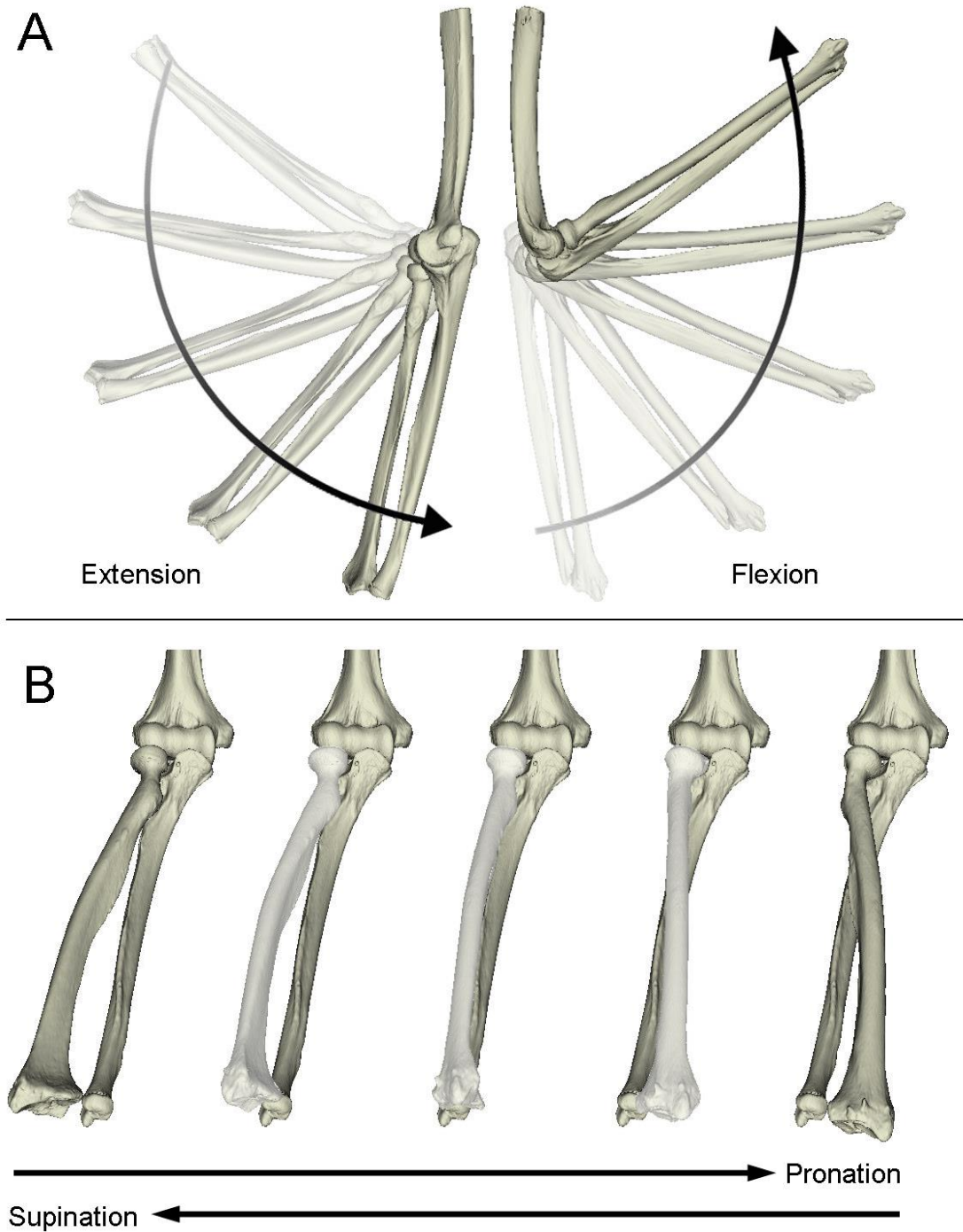


Figure 1-5 - Elbow motions.

A: Lateral view of the elbow, showing extension (left), which is normally to 0° , and flexion (right), which is normally to 145° . **B:** Anterior view of the elbow, showing forearm supination (left), which is normally to 85° , and pronation (right), which is normally to 75° . During supination, the radius rotates about a relatively stationary ulna. Right upper extremity shown. (*Reproduced with permission: Ferreira LM, 2011*).

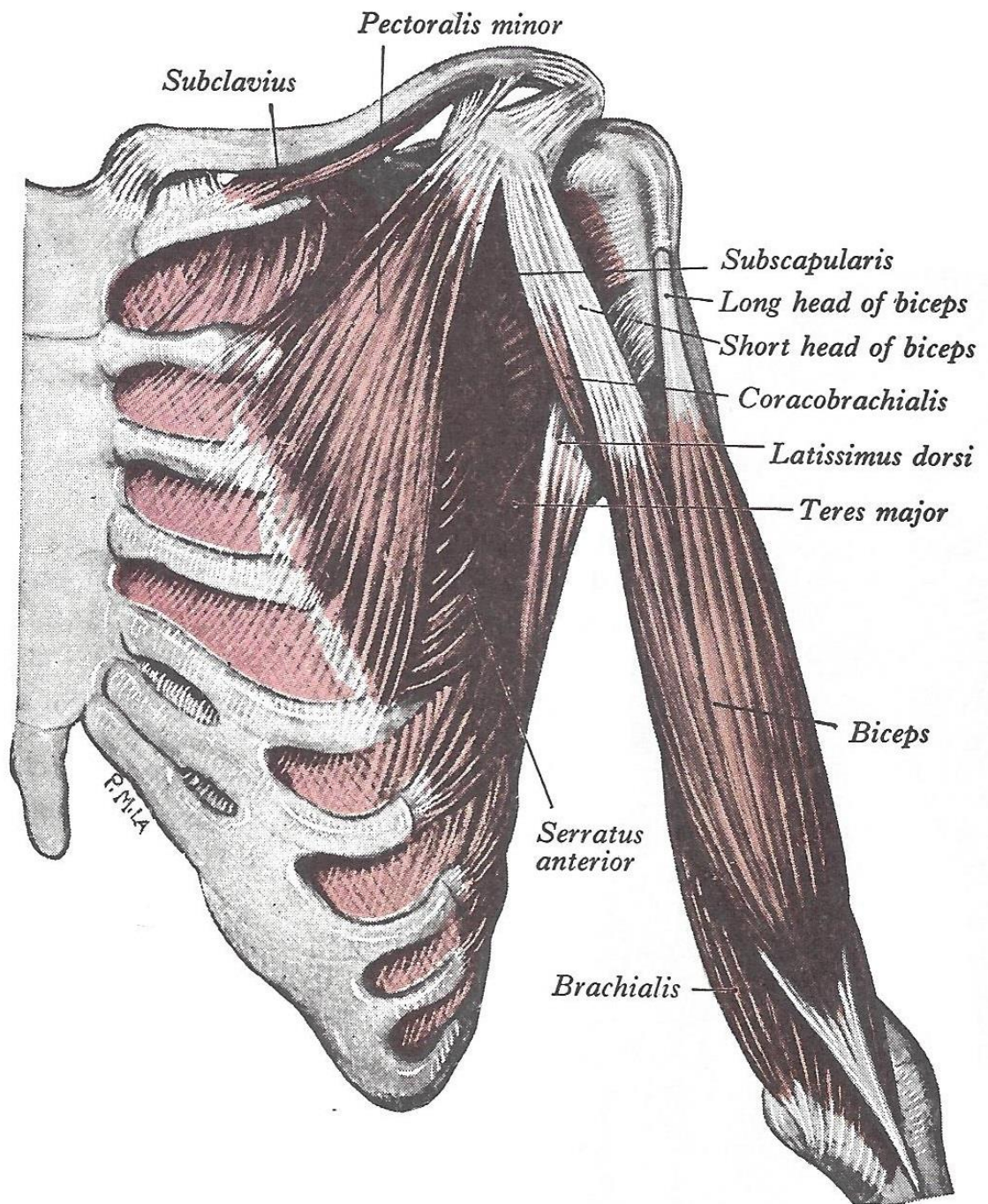


Figure 1-6 - Elbow flexors of the anterior arm.

The elbow flexors located in the anterior arm include the more superficial biceps brachii and the deeper brachialis. The brachioradialis (not shown, see Figure 1-8), located in the forearm, also enables elbow flexion. Biceps brachii is also the primary forearm supinator.

(Reproduced with permission: Salmon S, ed., 1995).

1957). The brachialis lies deep to the biceps, originating on the anterior distal humerus and inserting on the ulnar tuberosity and coronoid process (Morrey, 2000a). The brachioradialis originates along the lateral supracondylar ridge of the humerus and inserts into the base of the radial styloid, enabling elbow flexion in mid-pronation (Morrey, 2000a).

The elbow extensors, located in the posterior arm, enable elbow extension. The triceps brachii is the main elbow extensor, although anconeus plays a minimal role (Figure 1-7) (Morrey, 2000a). The long head of the triceps originates at the infraglenoid process of the scapula, whereas the medial and lateral heads originate from the posterior aspect of the humerus. These three heads merge to insert on the olecranon process of the ulna.

The flexor-pronator forearm muscles (Figure 1-8) are located in the anterior forearm and originate from a common flexor tendinous origin on the medial epicondyle (Morrey, 2000a). The most superficial muscles of this group include the flexor carpi radialis (FCR), palmaris longus, and flexor carpi ulnaris (FCU), all of which enable wrist flexion, and the pronator teres, which is the primary pronator of the forearm.

The extensor-supinator forearm muscles (Figure 1-9) originate from a common extensor tendinous origin (CEO) located on the lateral epicondyle. The largest muscles of this group include the extensor carpi radialis longus (ECRL), extensor carpi radialis brevis (ECRB), extensor digitorum communis (EDC), and supinator. The ECRL, ECRB, and EDC enable wrist extension, and the EDC also enables extension of the second to fifth fingers. The supinator lies deep to the other extensor muscles and performs forearm supination. It inserts on the lateral surface of the radius (Morrey, 2000a).

1.2 Elbow Kinematics

The ulnohumeral articulation of the elbow is responsible for elbow flexion and extension (An & Morrey, 2000; King *et al.*, 1993b; Schwab *et al.*, 1980; Stroyan & Wilk, 1993). The radiocapitellar and proximal radioulnar joints enable forearm pronation and supination (King *et al.*, 1993b; Morrey, 2000a; Schwab *et al.*, 1980; Stroyan & Wilk, 1993). During forearm rotation, the proximal radius pivots about its own centre. Distally, the radius rotates about the stationary ulna, crossing volarly in full pronation (An & Morrey, 2000).

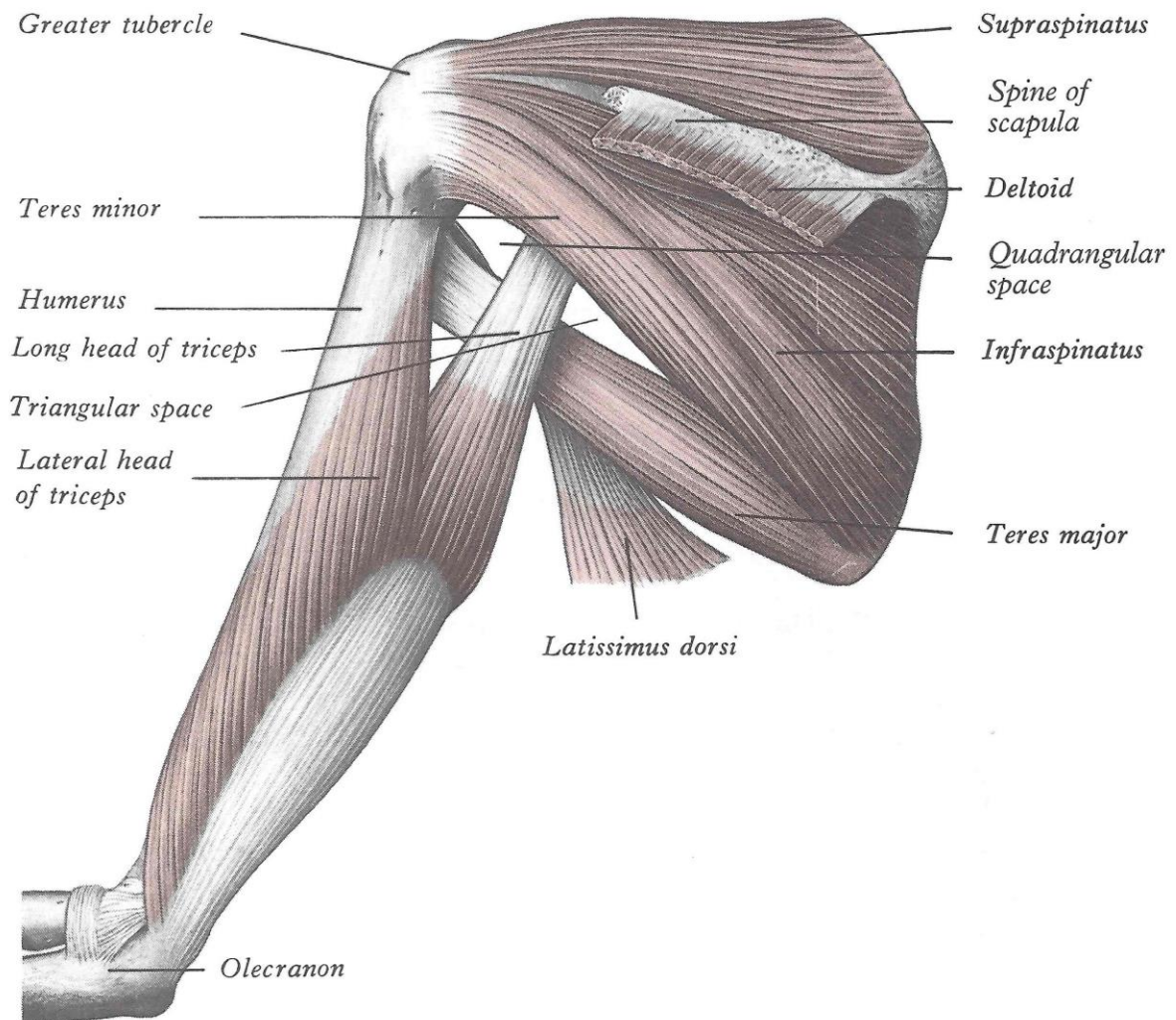


Figure 1-7 - Elbow extensors of the posterior arm.

The triceps brachii, located in the posterior arm, is the main elbow extensor. Anconeus (not shown) also enables elbow extension. (*Reproduced with permission: Salmon S, ed., 1995*).

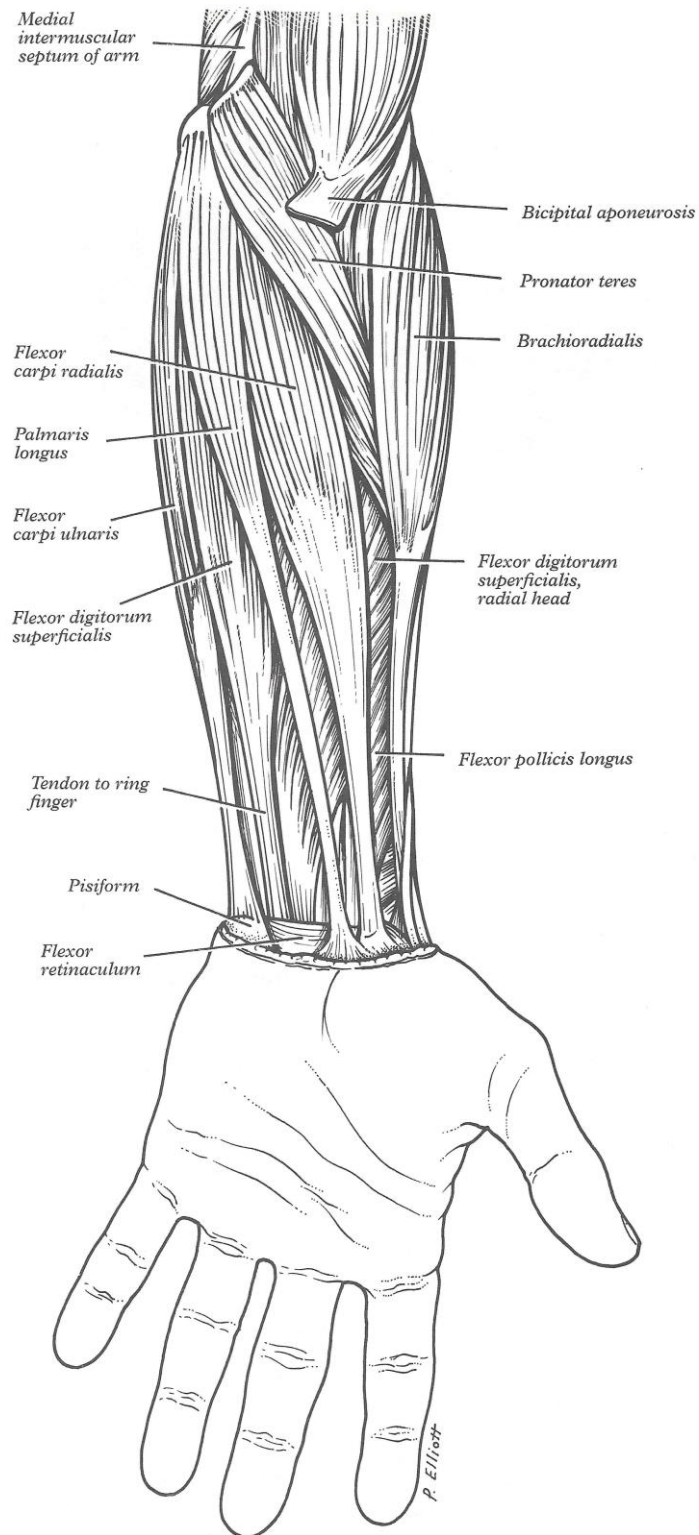


Figure 1-8 - Flexor-pronator muscles of the anterior forearm.

These muscles of the anterior compartment of the forearm originate at the medial epicondyle and enable wrist flexion and pronation. (*Reproduced with permission: Salmon S, ed., 1995*).

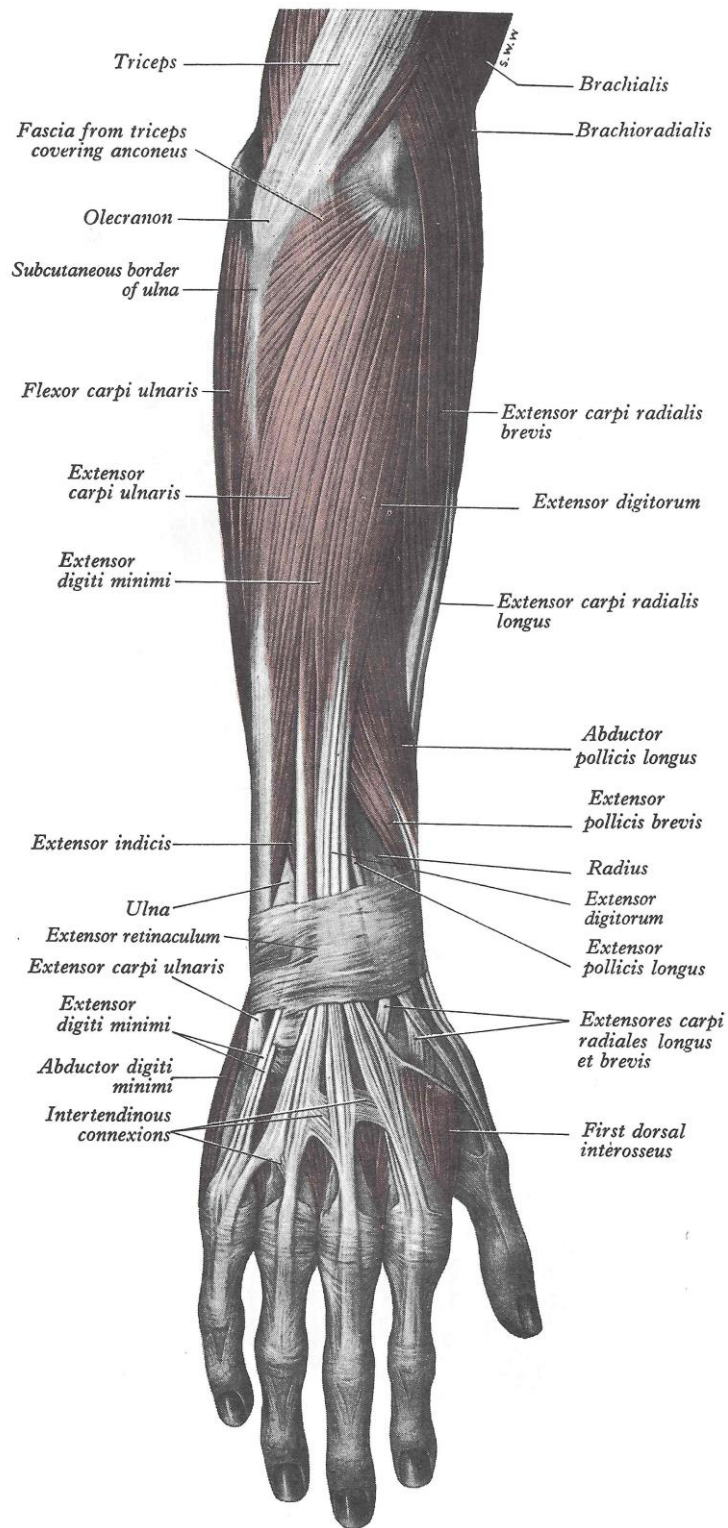


Figure 1-9 - Extensor-supinator muscles of the posterior forearm.

The extensor-supinator forearm muscles originate at a common extensor origin located on the lateral epicondyle. The extensor carpi radialis longus (ECRL), extensor carpi radialis brevis (ECRB), and extensor digitorum communis (EDC) enable wrist extension. EDC also enables extension of the second to fifth fingers. The supinator enables forearm supination. (*Reproduced with permission: Salmon S, ed., 1995*).

Normal forearm rotation averages from 75° of pronation to 85° of supination (Morrey, 2000a).

The elbow joint is described as a trochoginglymoid or “loose hinge” joint (Morrey & Chao, 1976). Throughout most of the flexion arc, the joint permits motion primarily in the flexion-extension plane (Duck *et al.*, 2003b; Morrey & Chao, 1976). However, at extremes of the flexion arc, the humerus rotates axially relative to the ulna. When the elbow starts to flex from a fully extended position, the humerus internally rotates, and when full flexion is approached the humerus externally rotates, relative to the ulna (Figure 1-10). This is independent of forearm rotation and it causes the elbow to move from a valgus to a varus position as it flexes. The flexion-extension axis of the elbow is anterior to the humeral shaft. It passes through the centres of the capitellum and trochlea, and is angled 6-8° valgus with respect to the medial-lateral axis of the humerus (Figure 1-11) (Amis *et al.*, 1979; An & Morrey, 2000). Normal range of motion is typically 0° of extension to 140° of flexion.

The carrying angle of the elbow is defined as the acute angle formed by the long axis of the humerus and the long axis of the ulna (Figure 1-12). It averages 10 to 15° in men and 15 to 20° in women (An & Morrey, 2000). The varus and valgus angles of the ulna relative to the humerus have also been described, and can be helpful in the assessment of elbow stability (Armstrong *et al.*, 2000; Dunning *et al.*, 2001b; Dunning *et al.*, 2001c; Pomianowski *et al.*, 2001). When the humeral and ulnar coordinate systems are coincident, the varus angle describes the adducted angular deviation of the ulnar long axis from the humeral long axis in the coronal plane, and the valgus angle describes the abducted angular deviation of the ulna relative to the humerus in the same plane (Ferreira, 2011) (Figure 1-13). The internal or external rotation of the ulna relative to the humerus have also been used to describe functional elbow stability (Armstrong *et al.*, 2000; Cohen *et al.*, 1997; Dunning *et al.*, 2001b; Dunning *et al.*, 2001c). This measure is defined as rotation of the ulna about its own long axis, with respect to the humerus (Ferreira, 2011). O’Driscoll *et al.* have previously shown that a small amount of external rotation of the ulna occurs with supination and internal rotation of the ulna occurs with pronation (1991). Linear translation of the ulna relative to the humerus has also been described and can occur in the proximal/distal, anterior/posterior, and medial/lateral directions.

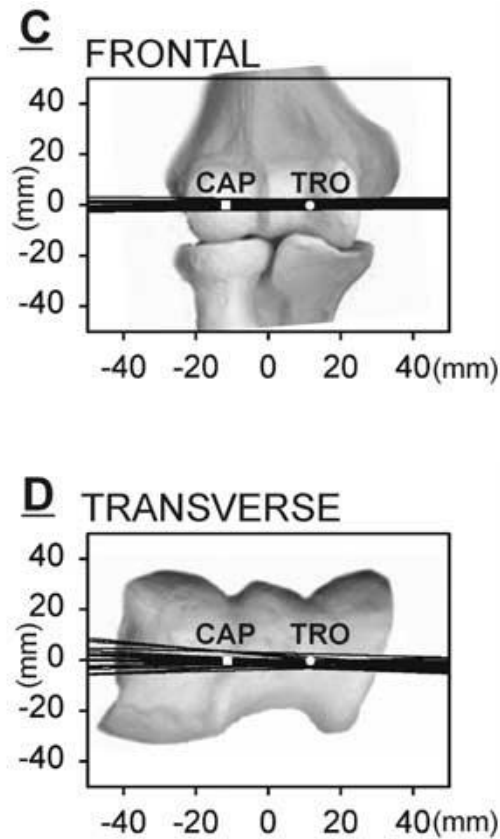


Figure 1-10 - Dynamic screw displacement axis changes during elbow flexion.

Lines representing screw displacement axes changing throughout motion for a single specimen during supinated active flexion are shown superimposed on the distal humerus in the frontal (C) and transverse (D) planes. The humerus internally rotates when the elbow is fully extended and tends to externally rotate during full flexion. Abbreviations:

CAP, capitellum; TRO, trochlea. (*Reproduced with permission: Duck, 2003b*).

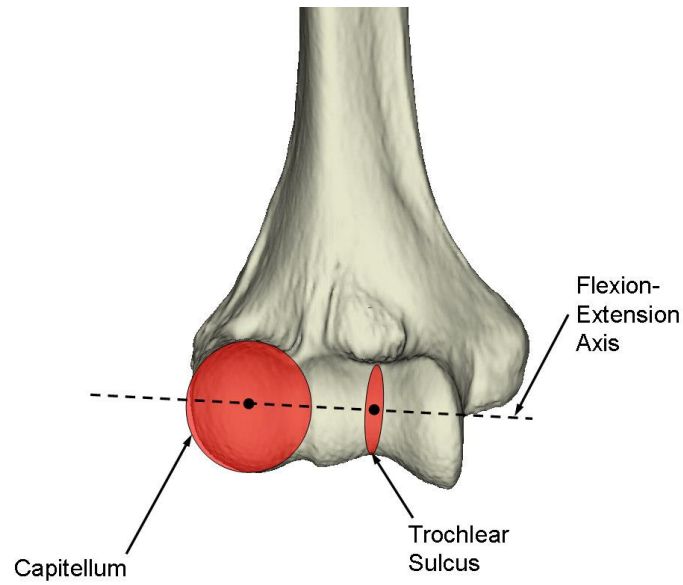
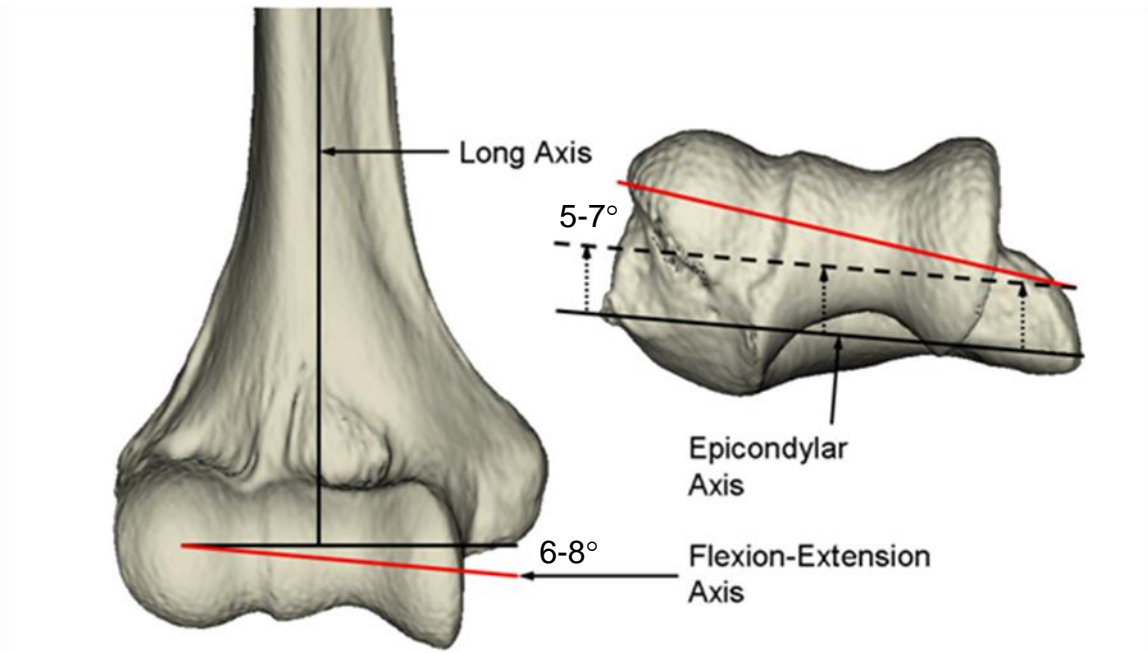
A**B**

Figure 1-11 - Flexion-extension axis of the elbow joint.

A: The elbow flexion-extension axis passes through the centre of the capitellum and the centre of the trochlea. **B:** This axis is 6-8° valgus and 5-7° internally rotated with respect to the long axis of the humerus. (*Reproduced with permission: Ferreira LM, 2011*).

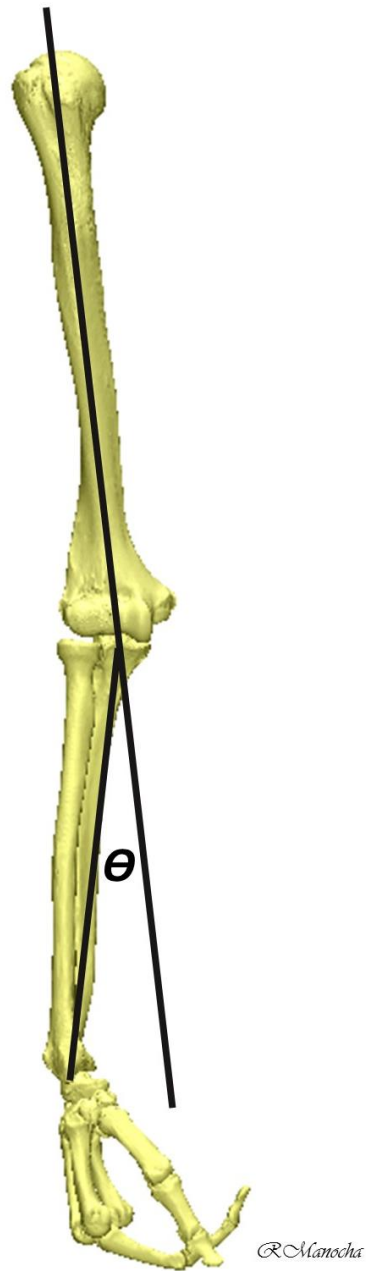


Figure 1-12 - Carrying angle.

The carrying angle of the elbow (θ) is defined as 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.

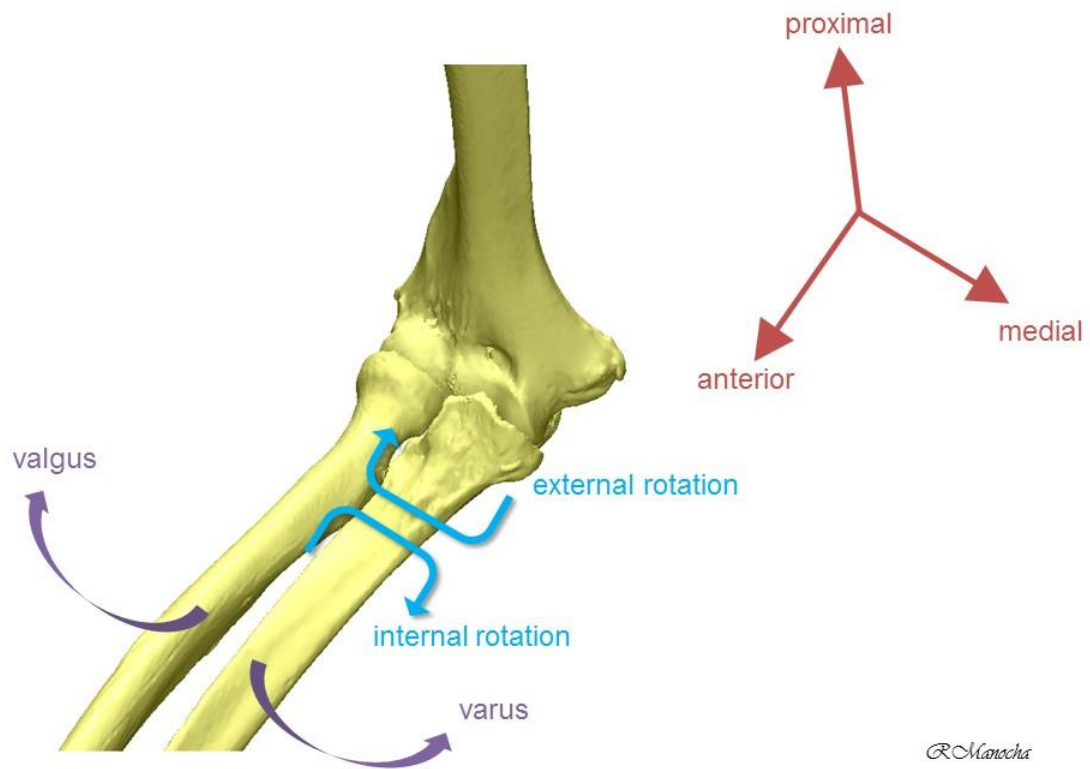


Figure 1-13 - Kinematic references for the elbow.

Several kinematic descriptors of elbow motion exist. Linear translation of the ulna relative to the humerus can occur in the proximal/distal, anterior/posterior, and medial/lateral directions (red). Varus and valgus motions can occur in the coronal plane (purple). Internal and external rotation of the ulna about its own long axis relative to the humerus has also been described (blue).

1.3 Elbow Stability

The combination of bones, ligaments, capsule, and muscles around the elbow joint confer static and dynamic stability. The relative contribution of each of these structures to joint stability depends on muscle activation, arm position, and forearm position (King *et al.*, 1993b). Damage to any of these structures could alter elbow kinematics resulting in negative short- and long-term consequences.

1.3.1 *Static Stabilizers*

The osseous articulations, ligaments, and joint capsule described above confer static stability to the elbow joint by increasing apposition of the articular surfaces (King *et al.*, 1993b). The complementary structures of the articular surfaces provide stability during elbow motion. During elbow flexion, the guiding ridge of the greater sigmoid notch glides in the trochlear groove and the oval dish-shaped radial head articulates with the spherical capitellum. The proximal portion of the greater sigmoid notch contributes to 80% of resistance to valgus stress whereas the distal portion of the notch provides 65% of the resistance to varus stress (An *et al.*, 1986). At terminal flexion, the coronoid enters the coronoid fossa and the radial head enters the radial fossa. At terminal extension, the olecranon enters the olecranon fossa. The coronoid prevents posterior subluxation of the elbow joint, particularly with the elbow extended. The anteromedial coronoid also resists varus stress. The radial head articulates with the lesser sigmoid notch during forearm pronation and supination (Hotchkiss and Weiland, 1987; King *et al.*, 1993b; Morrey, 2000a).

The MCL primarily resists valgus loading of the elbow (Hotchkiss & Weiland, 1987; Morrey *et al.*, 1991). The anterior bundle is the primary restraint to valgus stress (Morrey *et al.*, 1991; Safran *et al.*, 2005; Søjbjerg *et al.*, 1987) and when this constraint is sectioned all elbows become unstable (Hotchkiss and Weiland, 1987). The posterior bundle acts as a secondary stabilizer during valgus stress and the transverse ligament is felt to be of minimal functional significance (Morrey *et al.*, 1991; Safran *et al.*, 2005; Stroyan & Wilk, 1993). The radial head is an important secondary stabilizer against valgus stress when the anterior bundle of the MCL is absent; however it provides only minimal joint stability when the

MCL is intact (Hotchkiss & Weiland, 1987; King *et al.*, 1999; Morrey *et al.*, 1991). There are few activities besides throwing and traumatic injuries that expose the MCL to loads that can lead to symptomatic instability (Morrey, 2000b).

1.3.1.1 *Functional Anatomy of the Lateral Collateral Ligament*

The LCL stabilizes the elbow against varus and posterolateral rotational loads (King *et al.*, 2002; Morrey & An, 1983, Olsen *et al.* 1996). The LUCL is often reported to be the primary stabilizer against posterolateral rotational loads, preventing subluxation of the radial head in the posterior and lateral directions (O'Driscoll *et al.*, 1992; Olsen *et al.*, 1996). However, a subsequent studies have suggested that the radial and lateral ulnar collateral ligaments contribute equally to posterolateral stability, and that complete instability results only when both ligaments as well as the overlying extensor musculature are sectioned (Dunning *et al.*, 2001c; McAdams *et al.*, 2005). The annular ligament stabilizes the proximal radius to the ulna during forearm rotation (Søbjerg *et al.*, 1987).

Morrey and An examined cadaveric specimens to determine the degree of varus stability provided by static stabilizers of the elbow (B. F. Morrey & An, 1983). In full extension, the LCL provides 15% of restraint against varus stress, whereas the joint capsule and bony articulation contribute 30% and 55% respectively. With the elbow flexed to 90°, 75% of joint stability comes from osseous anatomy, followed by 13% from the anterior capsule and 9% from the LCL. In full extension, bony congruency resists 55% of varus stress; 32% is then provided by the anterior capsule and 14% by the LCL. Thus the anterior capsule is an important stabilizer of the elbow to varus stress in the extended elbow (King *et al.*, 1993b). The posterolateral capsule appears to have minimal mechanical resistance to varus stress (Olsen *et al.*, 1996). As most activities of daily living (ADLs) load the elbow in a varus fashion (Morrey *et al.*, 1981), the LCL is felt to be more functionally important than the MCL of the elbow (King *et al.*, 1993b).

1.3.2 *Dynamic Stabilizers*

The muscles that cross the elbow joint provide dynamic stability. As the resultant vector of their joint reaction forces compresses the articular surfaces, the contact area of the elbow increases, thereby augmenting congruency and stability (An *et al.*, 1990; An *et al.*, 1981;

King *et al.*, 1993b; Morrey *et al.*, 1988; Palmer *et al.*, 1982). The elbow flexors and extensors do not confer significant *passive* varus-valgus stability (An *et al.*, 1981; An *et al.*, 1989). However, the superficial muscles of the forearm flexor-pronator group resist *dynamic* valgus forces, particularly the flexor carpi ulnaris (FCU) (An *et al.*, 1981; Lin *et al.*, 2007; Park & Ahmad, 2004; Udall *et al.*, 2009). This has important implications in pitchers, who tend to develop FCU tendinopathy and thus decreased dynamic support. In a cadaveric dissection study, Cohen *et al.* noted that the fascial bands and intermuscular septae of the forearm extensor muscles prevent the forearm from externally rotating away from the humerus when the forearm is supinated (Cohen & Hastings, 1997). This suggests that the forearm extensor-supinator muscles confer dynamic elbow stability against varus and posterolateral rotatory stress. Anconeus also confers dynamic stability during both pronation and supination (Basmajian & Griffin, 1972). Josefsson *et al.* confirmed the important impact of the elbow musculature on dynamic stability by observing that elbow instability following simple elbow dislocation increased when patients were examined under anesthesia, *i.e.* when voluntary muscle tone was decreased (1987b).

1.4 Lateral Collateral Ligament Injury

The elbow is the second most commonly dislocated major joint in the adult population, with an estimated incidence of 5.21 dislocations per 100,000 person-years (Josefsson & Nilsson, 1986; Mehlhoff *et al.*, 1988; Stoneback *et al.*, 2012; Tashjian & Katarincic, 2006). Such dislocations universally cause damage to the LCL and result from high energy mechanisms (Josefsson *et al.*, 1987b). Acute elbow instability is classified into three stages based on the disruption of the Horii circle of soft tissue, proposed by O'Driscoll *et al.*, with injury progression from the lateral to the medial elbow (Figures 1-14 and 1-15; O'Driscoll *et al.*, 2000). The LUCL is disrupted in Stage 1 injuries, causing subluxation and resulting in a condition known as posterolateral rotatory instability (PLRI). This condition is discussed further below (see Section 1.4.1). Stage 2 injuries involve disruption of the remaining LCL structures as well as damage to the anterior and posterior elbow capsule. This can cause incomplete posterolateral dislocation or “perching” where the trochlea appears to rest on the coronoid. Stage 3 injuries involve damage to the MCL and are further divided into three stages. Stage 3A injuries involve disruption of all posterior structures

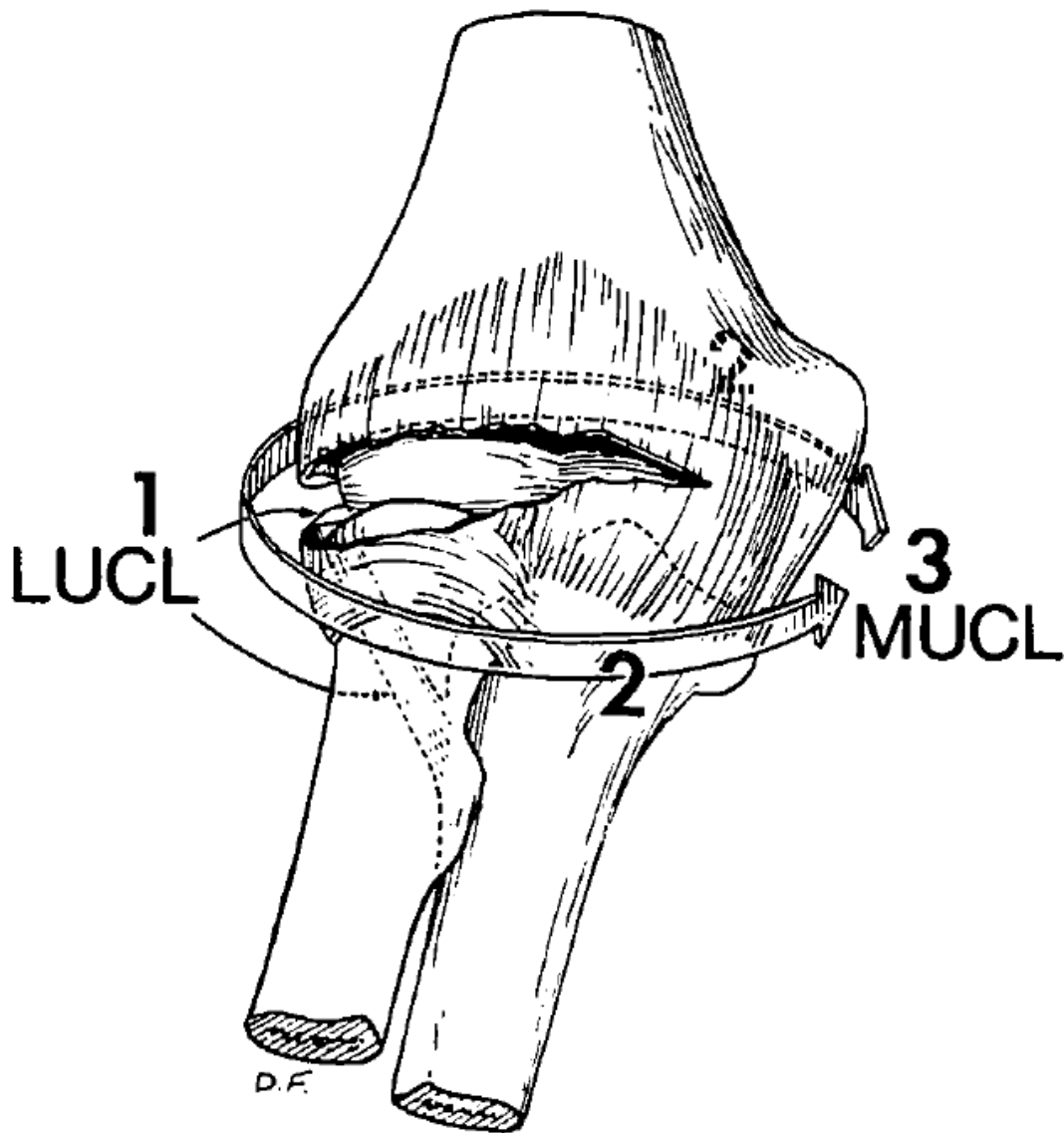


Figure 1-14 - Disruption of the circle of Horii with increasing elbow instability

The Horii circle of soft tissue (double-headed arrows) consists of the elbow capsule and its ligaments. With acute elbow trauma, injury extent progresses from the lateral (left side of image) to the medial (right side of image) side in three stages. Stage 1 injuries involve disruption of the lateral ulnar collateral ligament (LUCL). Stage 2 injuries involve damage to the remainder of the lateral collateral ligament and elbow capsule. Stage 3 injuries involve disruption of part or all of the medial ulnar collateral ligament (also known as the medial collateral ligament (MCL)). (*Reproduced with permission: O'Driscoll, 1992*).

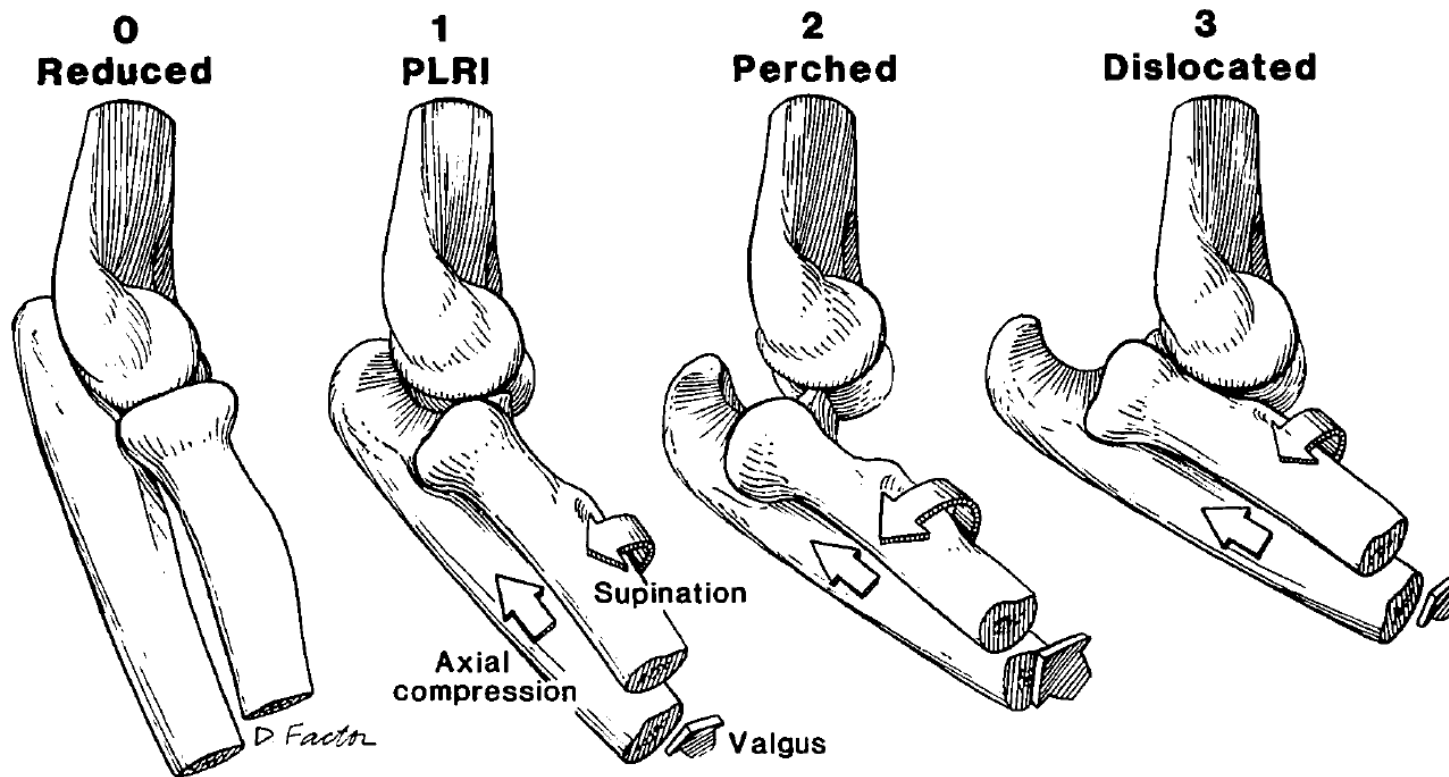


Figure 1-15 - O'Driscoll stages of elbow instability.

Varus elbow instability typically results from an axial compression, supination, and valgus load at the elbow (arrows). In the reduced or native anatomic state (Stage 0, on the left), the distal humerus, proximal ulna, and proximal radius are congruent. Stage 1 injuries can result in recurrent subluxation, known as posterolateral rotatory instability (PLRI). Stage 2 injuries can result in incomplete posterolateral dislocation, or “perching”, where the trochlea appears to rest on the coronoid. Stage 3 injuries result in complete elbow dislocation. (*Reproduced with permission: O'Driscoll, 1992*).

excluding the anterior band of the MCL. Such injuries tend to be associated with fractures of the coronoid process and radial head. Stage 3B injuries involve complete MCL disruption, leading to varus, valgus, and bidirectional rotatory instability. In Stage 3C, the soft tissue trauma is so severe that the elbow can dislocate even when immobilized (O'Driscoll *et al.*, 2000).

Acute isolated LCL injury can arise following traumatic subluxation or dislocation (i.e. from a fall onto an outstretched hand, sports injury, or motor vehicle accident), or iatrogenically from surgical release (Muller *et al.*, 2010; O'Driscoll *et al.*, 2000; Tashjian & Katarincic, 2006). Isolated acute traumatic LCL injuries typically fall into one of six patterns: proximal avulsion (most common), midsubstance rupture (second most common), bony avulsion of the lateral epicondyle, ulnar detachment of the LCL, ulnar bony avulsion, or a combination of the above (McKee *et al.*, 2003). 66% of acute LCL injuries occur in combination with rupture of the common extensor origin. More than half of LCL injuries are associated with rupture of at least the posterolateral part of the elbow capsule off the lateral condyle (McKee *et al.*, 2003). Chronic attritional rupture of the LCL has also been reported, as a consequence of cubitus varus causing recurrent varus loading (O'Driscoll *et al.*, 1991; O'Driscoll *et al.*, 2001), generalized ligamentous laxity (Charalambous & Stanley, 2008), and chronic crutch use (Charalambous & Stanley, 2008; McGuire & Bain, 2013; Singleton & Conway, 2004). It can also arise iatrogenically following radial head resection (Beingessner *et al.*, 2004; Jensen *et al.*, 2005), previous LCL release (Jensen *et al.*, 2005), or corticosteroid injection for lateral epicondylitis (Chanlalit & Limsricharoen, 2013; Kalainov & Cohen, 2005).

1.4.1 *Posterolateral Rotatory Instability (PLRI)*

PLRI is a clinical condition whereby an axial load through the forearm causes external rotatory subluxation of the proximal ulna from the trochlea and posterolateral subluxation of the radial head relative to the capitellum (Figure 1-16) (O'Driscoll *et al.*, 1990; O'Driscoll *et al.*, 1991). Often there is a history of previous trauma or surgery causing damage to the LCL, as outlined above (see Section 1.4). Patients with this condition commonly report clicking, snapping, and functional weakness (Muller *et al.*, 2010).

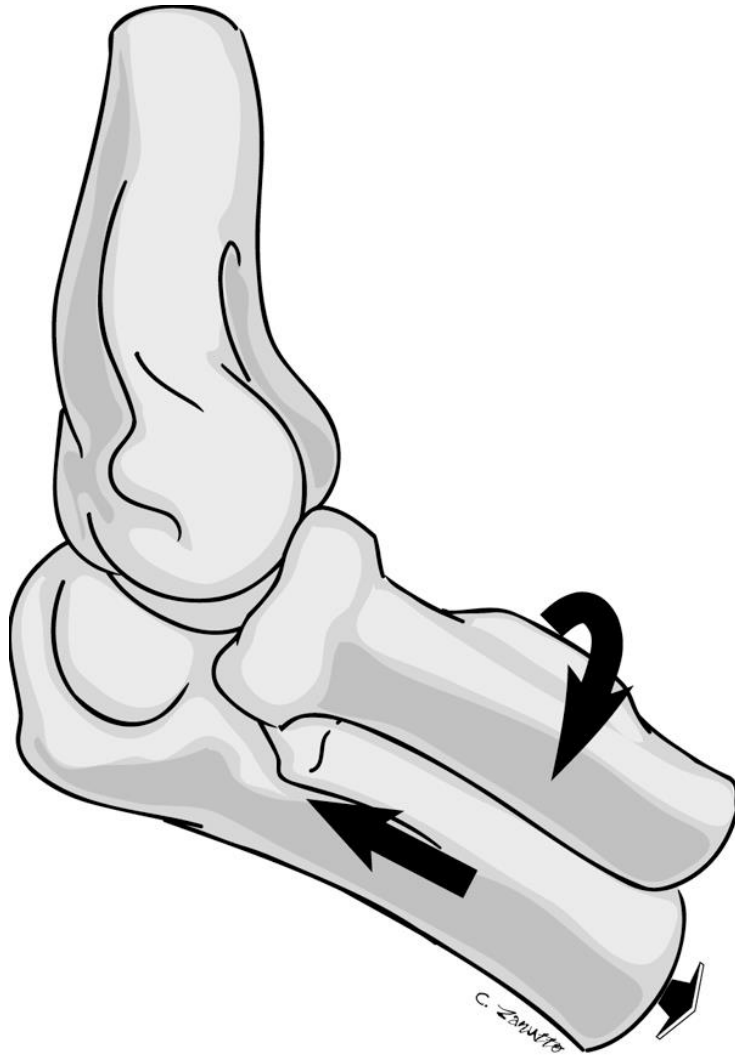


Figure 1-16 - Posterolateral rotatory instability.

When the lateral collateral ligament is disrupted, the elbow is vulnerable to posterolateral rotatory instability (PLRI). In this situation, the radial head subluxates posterolaterally relative to the capitellum, and the ulna rotates externally from the trochlea. This becomes more pronounced when axial compression, supination, and valgus loads are applied (black arrows).

Patients may also note locking, particularly when the elbow is extended and supinated (Reichel *et al.*, 2013). Multiple physical examination maneuvers involving the application of an axial and supination load to the forearm and valgus load at the elbow have been described to elicit this instability, such as the “pivot-shift test”, the “drawer sign”, the “chair sign”, and the “push-up sign” (Reichel *et al.*, 2013).

1.5 Management of Lateral Collateral Ligament Instability

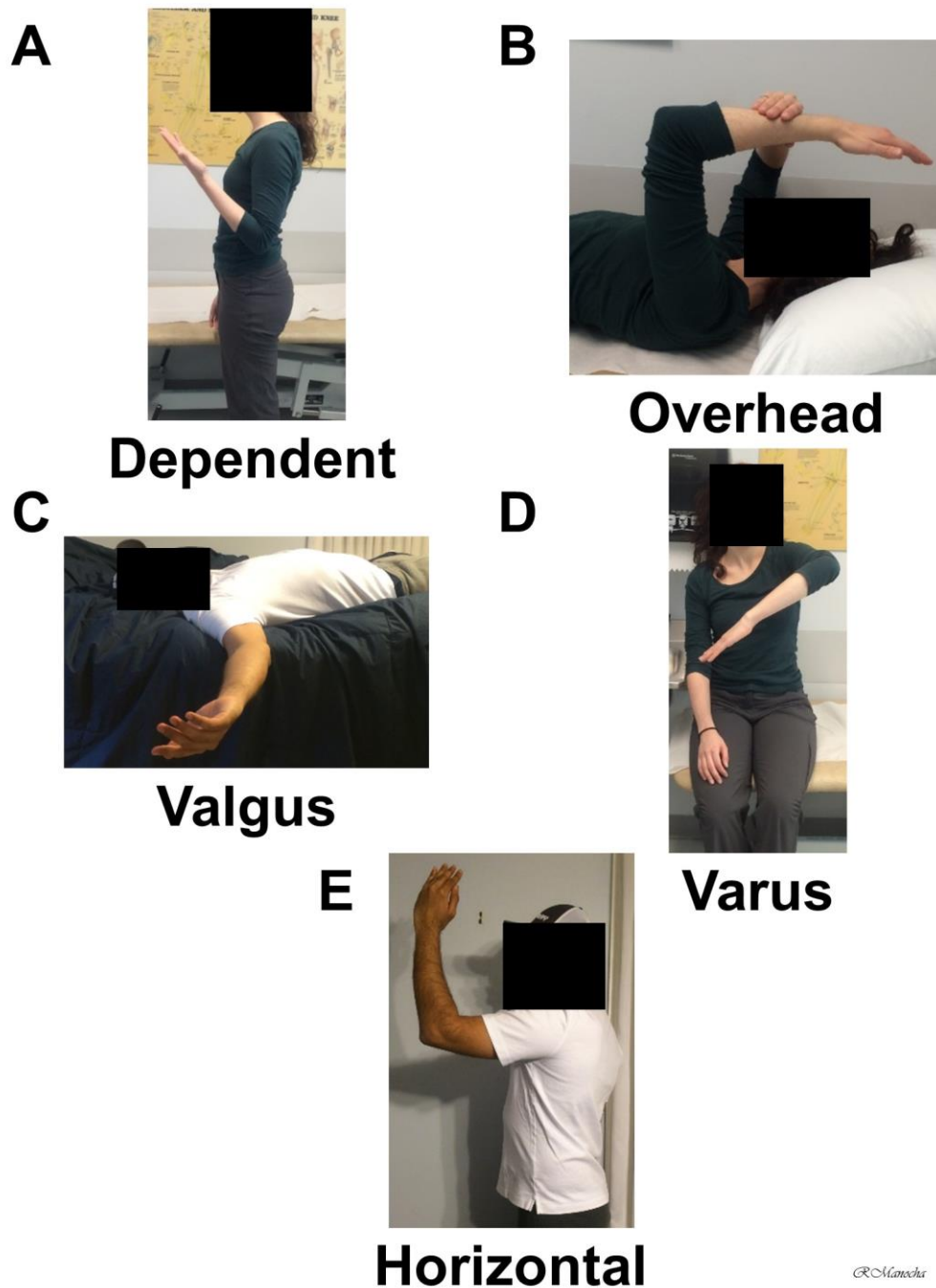
In general, elbow instability can be classified as simple (ligamentous injury without fracture) or complex (ligamentous injury with associated fracture) (Tashjian & Katarincic, 2006). Most simple acute post-traumatic LCL tears are managed non-operatively (Josefsson *et al.*, 1987a; Maripuri *et al.*, 2007; Safran *et al.*, 2005; Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). Rehabilitation protocols typically begin with immobilization and motion restriction, followed by gradual progression of passive-, active-assisted, and active range of motion. Therapy later involves progressive strengthening, and, ultimately, sport-, job-, or other functional-specific activities (Reichel *et al.*, 2013; Wolff & Hotchkiss, 2006). Some surgeons and therapists recommend hinged elbow orthoses (HEOs, colloquially known as braces) for immobilization and motion restriction. However, restricting elbow motion predisposes to stiffness, contracture, and subsequent loss of function (Lansinger *et al.*, 1984; Mehlhoff *et al.*, 1988). The elbow is responsible for allowing the proper placement of the hand in space for ADLs (Szekeres *et al.*, 2008). When the elbow is fused at any flexion angle between 50 and 110°, the shoulder and wrist cannot compensate to allow for completion of functional activities (O'Neill *et al.*, 1992). At a biophysical level, immobilization in mouse hindlimb medial collateral ligament injury models causes ligaments to fail with repetitive low loads (Thornton *et al.*, 2003). Acutely post-injury, however, ligamentous stress increases creep (Thornton *et al.*, 2000) and can increase the risk of repeat subluxation or dislocation (Jockel *et al.*, 2013). In the longer term, this can lead to abnormal joint tracking and post-traumatic arthritis (Josefsson *et al.*, 1984). A small amount of ligamentous stress, however, has also been shown to enhance soft tissue healing (Cyr & Ross, 1998). Thus the rehabilitation of the LCL-deficient elbow involves a balance between encouraging ligament healing and preventing contracture while avoiding worsening instability.

1.5.1 *Muscle Activation*

Passive range of motion (PROM) involves the movement of a joint without autonomous contraction by the patient of the muscles around that joint. This is often achieved by a therapist moving the joint or by the patient using their contralateral extremity to move the affected joint. Active range of motion (AROM) involves a patient actively contracting his or her muscles to move a given joint. Most therapy sessions for lateral elbow instability start with PROM in order to precondition the tissues, followed by AROM later on in the session (Szekeres *et al.*, 2008). In LCL insufficiency, cadaveric studies have found that passive elbow flexion with the forearm supinated in the dependent position (Figure 1-17A) causes instability which can be reduced with simulated (i.e. motion simulator-controlled; described further below, see Section 1.7) active elbow flexion (Dunning *et al.*, 2001b; Duck *et al.*, 2003a). This is likely due to active tensioning of the extensor-supinator muscles providing lateral stability due to their origin on the lateral epicondyle, and contraction of the biceps brachii, brachialis and triceps brachii, which augments the intrinsic constraint of the elbow joint by compressing the articulation together (Olsen *et al.*, 1998; Szekeres *et al.*, 2008).

1.5.2 *Arm Position*

The overhead position has recently become a popular method to rehabilitate elbow LCL injuries (Figure 1-17B) (Szekeres *et al.*, 2008). This is thought to enable the weight of the forearm and the activated biceps brachii, brachialis, and triceps brachii to compress the ulnohumeral joint (Wolff & Hotchkiss, 2006) (Figure 1-18). Although the biceps brachii and brachialis may exert a potentially destabilizing posterior force at the elbow joint, the triceps may counteract this during active extension in the overhead position (Wolff & Hotchkiss, 2006). Lee *et al.* quantified ulnohumeral gapping during passive motion in intact cadaveric elbow specimens, those with a sham “approach only” procedure, and those with LCL sectioning (Lee *et al.*, 2013). They found 104% more gapping with the arm in a dependent position versus in an overhead position and concluded that rehabilitation of the LCL-deficient elbow in the overhead position was safe, whereas loading in the dependent position risked dislocation (Lee *et al.*, 2013). This is the only published study to date that has evaluated the effect of the overhead arm position on elbow kinematics and stability.



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Figure 1-17 - Gravity-loaded humerus positions.

The humerus can be positioned in the gravity-loaded vertical dependent (**A**) or overhead (**B**) positions, in the gravity-loaded valgus (**C**) or varus positions (**D**), or in the horizontal (**E**) position. Typically the dependent, horizontal, and varus positions are seen during activities of daily living. Following LCL injury, the overhead position is employed for exercises and the varus position is avoided.

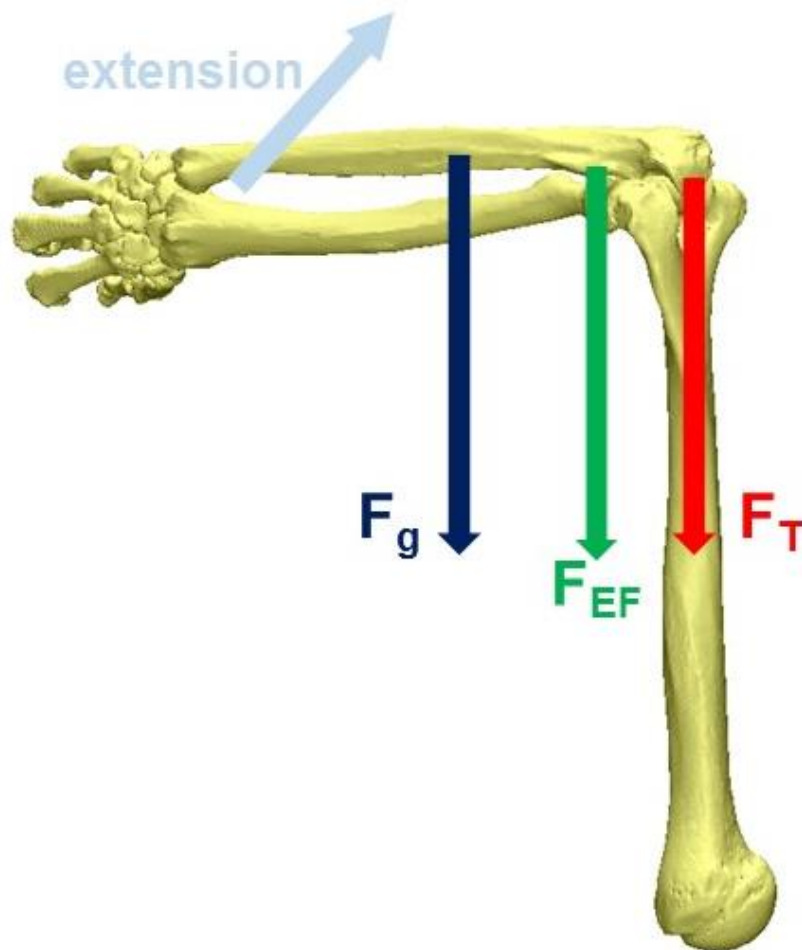


Figure 1-18 - Theoretical elbow joint compressive forces in overhead position.

When the humerus is positioned in the vertical overhead position, the weight of the forearm and hand unit (F_g , dark blue arrow) provides a compressive force at the elbow joint which increases as the elbow moves from 90° to full extension (light blue arrow). When active extension is performed, loading through the triceps muscle (F_T , red arrow) and the elbow flexors biceps brachii and brachialis (F_{EF}) provide an additional compressive force at the elbow joint. For these reasons, clinicians and scientists theorize that overhead arm motion reduces instability following lateral elbow injuries.

No studies have investigated simulated active overhead positioning on elbow kinematics and stability.

Positioning the arm in the gravity-loaded varus position (Figure 1-17D) is typically avoided in the first 6-12 weeks following LCL injuries to avoid putting tensile stresses on the lateral elbow structures (Szekeres *et al.*, 2008). During passive elbow flexion in LCL-deficient cadavers loaded in gravity-loaded varus positions, there were significant increases in maximum varus-valgus laxity, regardless of forearm position (Dunning *et al.*, 2001b). Simulated active elbow flexion and extension have never been studied in LCL-insufficient cadavers in varus orientations because these positions have caused such marked instability that the motion simulators available in the past were not able to reliably initiate and control motion (Alolabi *et al.*, 2012a; Armstrong *et al.*, 2000; Dunning *et al.*, 2001b).

1.5.3 Forearm Position

Cadaveric studies have shown that with the arm oriented in the dependent position, forearm pronation improves the stability of the LCL-deficient elbow relative to forearm supination during both passive and active elbow flexion (Duck *et al.*, 2003a; Dunning *et al.*, 2001b; Fraser *et al.*, 2008). Amongst therapists who deal with LCL injuries, pronation has widely been adopted into rehabilitation regimes (Szekeres *et al.*, 2008).

1.5.4 Orthoses

There is minimal literature on the effectiveness of elbow orthoses in the management of lateral elbow instability (Hijmans *et al.*, Geertzen, 2004). Regardless of whether managed operatively or not, LCL injuries tend to be treated initially with immobilization in a thermoplastic splint with the elbow flexed to 90-120° and forearm pronated (Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). The splint is removed for exercises and personal hygiene but must otherwise be worn continuously for about 4-6 weeks (Szekeres *et al.*, 2008). In cases of significant ligamentous instability, a hinged elbow orthosis (HEO, Figure 1-19) such as a Bledsoe Brace (Bledsoe Brace Systems, Grand Prairie, Texas) or a Mayo Clinic Elbow Brace (Aircast, Summit, New Jersey) is recommended by some authors (Cohen & Hastings, 1998; Morrey, 2000c; Reichel *et al.*, 2013; Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). There is no published data on how frequently such orthoses are used.



Figure 1-19 - Mayo Clinic Elbow Brace.

This device is an example of a prefabricated hinged elbow orthosis (HEO). It has no energy-storing components. It may be used in the first few weeks following elbow LCL injury or surgical repair of such injuries. (*Reproduced with permission: DJO Canada, 2016*).

Initially these devices may be locked at a certain flexion angle and used as a static splint, in a similar manner to the thermoplastic splints previously mentioned (Morrey, 2000c; Wolff & Hotchkiss, 2006). These orthoses may then be adjusted to prevent terminal extension yet allow full flexion (i.e. 40° to 140°) early post-injury or surgery. This extension limit is gradually reduced towards 0° as joint stability improves over 4 to 6 weeks (Wolff & Hotchkiss, 2006). HEOs are worn at all times and often during exercises (Wolff & Hotchkiss, 2006), thus allowing some stress to encourage ligament healing and prevent stiffness and pain (Cyr & Ross, 1998; Morrey, 2000c; Lunsford & DiBello, 2008).

The Mayo Clinic Elbow Brace is prefabricated, with 2 Velcro hook and loop straps at the arm and 2 Velcro hook and loop straps at the forearm to enable proper fit and suspension, and thus adequate mechanical control (Griffin *et al.*, 2008). On the anterior arm, the straps have foam padding to increase skin contact and reduce discomfort. On the posterior undersurface of the most proximal arm strap, there is a C-shaped metal cuff that is adjustable to accommodate for 5 arm widths and which can be secured using an Allen key. Bilateral metal sidebars are aligned axially on the medial and lateral sides of the arm and forearm. There is a mechanical hinge at the elbow flexion-extension axis into which pins can be inserted to limit flexion-extension range of motion. The inner surface of the hinge on both sides of the arm is lined with foam padding. The device has no energy-storing components.

Only one biomechanical study has been published evaluating the effectiveness of HEOs in LCL injury. Lee *et al.* examined seven LCL-deficient cadavers during passive motion when the arm was dependent, and found that ulnohumeral distraction was nearly twice as much in cadavers with a Bledsoe Brace as compared to those that were not braced, although the difference was not statistically significant (2013). This difference was attributed to the mass of the orthosis. No studies have looked at bracing with the arm in any other positions and there are no studies to support the efficacy of these orthoses in terms of secondary injury prevention, enhanced proprioception, or other clinical or functional outcomes.

1.6 Design Principles in Hinged Elbow Orthoses

There are several features that determine how effectively an HEO will provide mechanical stability. These orthoses operate on a four-point pressure system, with the four points on the medial and lateral side being at the level of the arm (provided by the two arm straps) and the level of the forearm (provided by the two forearm straps). This creates a three-point lever system on the medial and lateral aspects of the upper extremity, with the proximal and distal lever arms being on the arm and forearm respectively, and with the orthotic hinge serving as the fulcrum. Longer lever arms theoretically provide more medial-lateral control at the elbow (Lunsford & Contoyannis, 2008). The mechanical control an orthosis will impart is also determined by the surface contact area between the orthosis and the braced limb. Typically, contact area is maximized over areas with minimal soft tissue, as this maximizes mechanical control of the bones beneath the orthosis. Areas with increased soft tissue are subject to the orthosis causing more tissue deflection as opposed to bony control. In the lower extremity, hinged knee orthoses tend to have increased contact at the anterior tibia for this reason (Wolters, 2008). However, in the upper extremity, there is no analogous bony prominence. In this case, wider straps help suspend the orthosis and translate forces of the orthosis over a larger part of the limb to impart control. Alignment of the anatomical joint with the mechanical joint (i.e. the elbow flexion-extension axis with the orthosis' hinge axis) is also important to ensure that motion generated at the arm or forearm does not cause rotation or translational movement outside the flexion-extension axis, as this could risk further subluxation or dislocation with the application of the orthosis (Lunsford & DiBello, 2008).

1.7 Upper Limb Biomechanical Testing

1.7.1 *Joint Motion Simulation Techniques*

In general, a joint's kinematics can be assessed by: observing and quantifying that joint moving naturally in humans (*in vivo*); using a specialized device to move a cadaveric joint (*in vitro*); or using a computer model to simulate how that joint would move (*in silico*) (Ferreira, 2011). There are strengths and limitations to each of these methods.

1.7.1.1 In Vivo

In vivo experiments, which usually involve tracking motion while a human subject is performing a prescribed movement or task, can provide useful clinical and functional information. However, such studies are limited by subject recruitment, the time a subject is willing to spend in the laboratory, and the ability of the subject to perform the desired movement in a repeatable fashion, if necessary. In addition, there is the potential that the novel treatment being investigated, such as a movement protocol or surgical treatment, can harm the subject. Finally, markers must be mounted on the skin since rigid marker mounting is generally considered too invasive for human subjects. Thus *in vivo* joint motion tracking is highly subject to soft tissue artifact (STA) (Akbarshahi *et al.*, 2010; Cappozzo *et al.*, 1996; Heneghan & Balanos, 2010). Humeral internal/external rotation is particularly vulnerable to STA and this is challenging to correct for (Cao *et al.*, 2007; Cutti *et al.*, 2005; Zhang *et al.*, 2011). Some kinematic studies have used fluoroscopy (Jalali *et al.*, 2015; Wu *et al.*, 2010) or four-dimensional computed tomography for joint motion tracking, however, these modalities are associated with high ionizing radiation exposure, which can have damaging effects on deoxyribonucleic acids and potentially predispose to cancer with long-term or repeated use (Brenner & Hall, 2007; “Integration,” 2006).

1.7.1.2 In Vitro

In vitro techniques can address some of the challenges seen with *in vivo* techniques. A device used to move a cadaveric joint for kinematic analyses can result in more repeatable motion patterns for investigation, and multiple investigations can be done with no limitation by patient tolerance. In addition, inserting markers into bone eliminates STA, decreasing the required sample size. If a treatment option is found to cause harm *in vitro*, this can prevent it from being used *in vivo*; similarly, treatments can be optimized prior to being used in patients. Unfortunately, such specialized devices and the cadaveric specimens themselves can be expensive, and testing must be carried out in a designated biohazard facility (Ferreira, 2011). Test duration is limited due to desiccation and biomechanical changes that occur in the soft tissues (King *et al.*, 2000); thus specimens cannot be reused. In addition, the specimens and device hardware and software may be subject to failure. Depending on where the specimens are obtained, there may be a

population skew by age and/or ethnicity. Finally, there may be alterations in fascial plane interactions, cartilage mechanics, and joint loads as compared to natural motion in live subjects. Of course, *in vitro* systems also cannot incorporate features such as cortical control, pain inhibition, and proprioception.

However, *in vitro* systems do allow for the ability to control for various aspects of a system (i.e. distribution of muscle loads, amount of forearm rotation) much better than using human subjects, allowing the investigator to better understand natural joint motion. They also do not have to make the same anatomical assumptions as *in silico* models because the variations in anatomy and ligament and tendon properties that exist between individuals are already incorporated (Ohman *et al.*, 2009). Finally, *in vitro* models allow the incorporation of some clinical variables that are challenging to model *in silico* because of lack of published data, such as mild moments provided by passive range of motion or the torque an orthosis might apply on a specimen.

1.7.1.3 In Silico

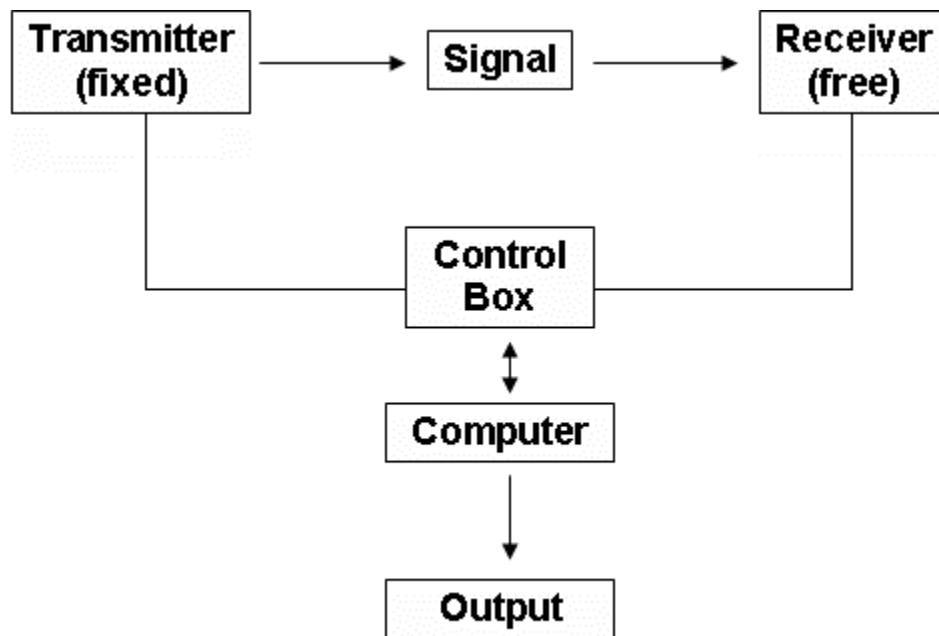
In *in silico* techniques, a live human or cadaveric specimen may be imaged to generate a computer finite element model (FEM) with which different treatment techniques are simulated (Ferreira, 2011; Fisk & Wayne, 2009). The benefits of such models include lower cost, minimal need for subject recruitment, and no surgical safety risk to the investigator. As with *in vitro* techniques, there is no limitation by patient tolerance and novel therapies can be investigated without putting human subjects at risk. Using FEMs, multiple variables can be controlled for and adjusted, and the model can be reused multiple times, unlike *in vitro* specimens. As with *in vitro* techniques, FEMs are limited by the many assumptions that are made in their generation. As outlined earlier in this chapter, elbow motion involves the complex interaction of bones, muscles, ligaments, capsule, and overlying fascia, in the context of the human neuromotor system. In FEMs, because there is little research on the complex dynamic mechanical properties and varying geometries of all of these structures around the elbow, assumptions such as simplifying muscle lines of action (Klein *et al.*, 2007), ignoring viscoelastic and anisotropic effects (Quapp & Weiss, 1998), assuming mechanical properties from other structures (i.e. knee tendon for elbow tendon, or knee tendon for elbow ligament, etc.), and ignoring effects of surrounding soft

tissue are often made. Thus the assumptions made in the model limit its clinical applicability.

1.7.2 *Kinematic Assessment*

Kinematics refers to the study of the motion of a rigid body, without reference to the forces causing the motion. Often motion is described in terms of position (i.e. location of the body in three-dimensional space) and orientation (i.e. angular position of the body in three-dimensional space). This generally results in a six-degree-of-freedom model. Orientation is generally described using Euler angles, i.e. the orientation of the object's frame as a composition of three rotations compared to a fixed reference frame. A downside of this method, however, is gimbal lock, where one degree of freedom is lost when the reference and object frame have two parallel axes. This results in no gimbal available to determine the rotation along the remaining axis (Rab *et al.*, 2002). In healthcare applications, kinematics are often best described in terms of clinically relevant joint motions. In order to do this, a set of universal definitions have been established which align a local bone segment coordinate system with a relevant anatomical or functional axis, such as the bone's long axis or flexion axis (Ferreira, 2011; Rab *et al.*, 2002; Wu *et al.*, 2005). These coordinate systems are known as "joint coordinate systems" (JCS). For the elbow, International Standards suggest JCS for the humerus and ulna, which can then be used to result in an Euler rotation sequence that corresponds to flexion angle, varus angle, and internal rotation of the ulna relative to the humerus (Piazza & Cavanagh, 2000; Wu *et al.*, 2005). Establishing accurate JCSs ensures that misalignment, or "kinematic crosstalk" will not occur with a joint's functional axis (Piazza & Cavanagh, 2000). If misalignment occurs, one joint rotation might be falsely interpreted as another (i.e. flexion interpreted as internal rotation).

Real-time kinematic assessment can be accomplished using a variety of tracking modalities, all of which function according to the same basic principles, illustrated in Figure 1-20 (Manocha, 2008). A transmitter, usually fixed to some location in the operating environment, generates a signal, which can be acoustic, electromagnetic, mechanical, or optical. This signal gets sensed by a receiver, which is generally attached to the object that is being "tracked". Both the transmitter and receiver are connected to a control box, which



R Manocha

Figure 1-20 - Schematic outlining general operation of motion tracking systems.

In general, a transmitter, usually fixed to some location in the operating environment generates a signal (*i.e.* mechanical, optical, electromagnetic). This signal subsequently is sensed by one or more receivers, which are generally attached to the object being “tracked”. A control box integrates the transmitted and received signals and interfaces with a computer to convert the signal into kinematic output (*i.e.* position and/or orientation). (*Reproduced with permission: Manocha 2008*).

processes the relative strengths of the transmitted and received signals through communication with a computer. As a result, the position and/or orientation of the receiver (the “output”) can be determined (Kinzel & Gutowski, 1983; Manocha, 2008).

1.7.2.1 *Optical Tracking*

Devices incorporating optical signals are commonly used in *in vivo* motion analyses (Sardelli *et al.*, 2011; Schnall *et al.*, 2008), but are also used in some *in vitro* studies of motion (Bernas *et al.*, 2009; Hammond *et al.*, 2012). Typically sensors are attached to the limb(s) of interest directly, or otherwise the sensor is affixed to a device attached to the limb surface or to the bone(s) of that limb. Skeletal pins are not practical for *in vivo* motion, although they are commonly used for *in vitro* assessments. Marker movement is then detected either by light reflection from the transmitter to the receiver from retroreflective skin markers, or by videographic analyses of the markers, or a combination of both. Some downsides of this method include the challenge of inserting pins without impinging other structures or motions. In addition, markers attached to wands are likely to impinge on other limb segments and suffer from inertial effects (Rab *et al.*, 2002). Finally, loss of visualization of markers can be common. Imaging techniques such as fluoroscopy have also been used for motion tracking (Lee *et al.*, 2013; Jalali *et al.*, 2015), however these can be expensive and risk exposure to ionizing radiation if used *in vivo*.

1.7.2.2 *Inertial Sensors*

This form of motion tracking involves the use of mechanical sensors such as accelerometers and gyroscopes (Tao *et al.*, 2012). An accelerometer measures change in velocity along an axis, whereas a gyroscope measures change in angular rate of rotation. Such sensors are either attached to various parts of the body or incorporated into garments (Langohr *et al.*, 2016; Liu *et al.*, 2009). Newer “smartphones” contain inertial sensors which can also be attached to limbs for this purpose (Roldan-Jimenez *et al.*, 2015). This technology has become much more affordable and available recently and is well-suited to *in vivo* applications, particularly as they can assess motion outside a controlled laboratory setting (Tao *et al.*, 2012; Patel *et al.*, 2012). In gait analyses, it has been shown that for two-dimensional analyses at slow gait velocities there tends to be good correlation between

inertial sensor data and optical tracking data (Liu *et al.*, 2009; Takeda *et al.*, 2009). However, these devices are prone to STA (described above, see Section 1.7.1.1), and can be subject to error accumulation, particularly with gyroscopes, with higher velocities of motion, and with increased axial rotation (Liu *et al.*, 2009; Tao *et al.*, 2012).

1.7.2.3 *Electromagnetic Tracking*

Most elbow motion simulators (discussed further below, see Section 1.7.3) use electromagnetic tracking systems due to their low cost and ability to function without line-of-sight requirements (An *et al.*, 1988; van Ruijven *et al.*, 2000). With this modality, a series of three orthogonal coils, located in a transmitter, are pulsed in rotation in order to generate a series of radiofrequency (RF) electromagnetic pulses (the signals) (Figure 1-21). Each pulse induces a current in another set of three orthogonal coils located in a receiver. A control box, connected to the transmitter and the receiver, processes the attenuation of the received pulses and from this calculates the position and orientation of the receiver relative to the transmitter. This spatial output can then be used for subsequent real-time motion analysis (An *et al.*, 1988; Koerhuis *et al.*, 2003; van Ruijven *et al.*, 2000). Unfortunately many of these systems rely on alternating current (AC) or steady direct current (DC) signals, which can generate eddy currents in nearby metals, producing secondary magnetic fields and leading to distortions in the transmitted field that is sensed by a receiver, affecting spatial output (McGill *et al.*, 1997; Milne *et al.*, 1996; Raab *et al.*, 1979). The elbow motion simulator used in this thesis and described further below (see Section 1.7.3) relies on a different electromagnetic tracking system (Flock of Birds®, Ascension Technology Corporation, Burlington, VT) which uses pulsed DC signals. These are less susceptible to magnetic field distortions as measurements of the receiver's position and orientation with respect to a transmitter in six degrees of freedom can be obtained once a steady magnetic state has been reached (LaScalza *et al.*, 2003; Milne *et al.*, 1996). The manufacturer's specified static positional accuracy of the device is 0.1 inches root-mean-square (RMS) with a spatial resolution of 0.03 inches. The static angular accuracy is 0.5° RMS with an angular resolution of 0.1° (Ascension, 2004).

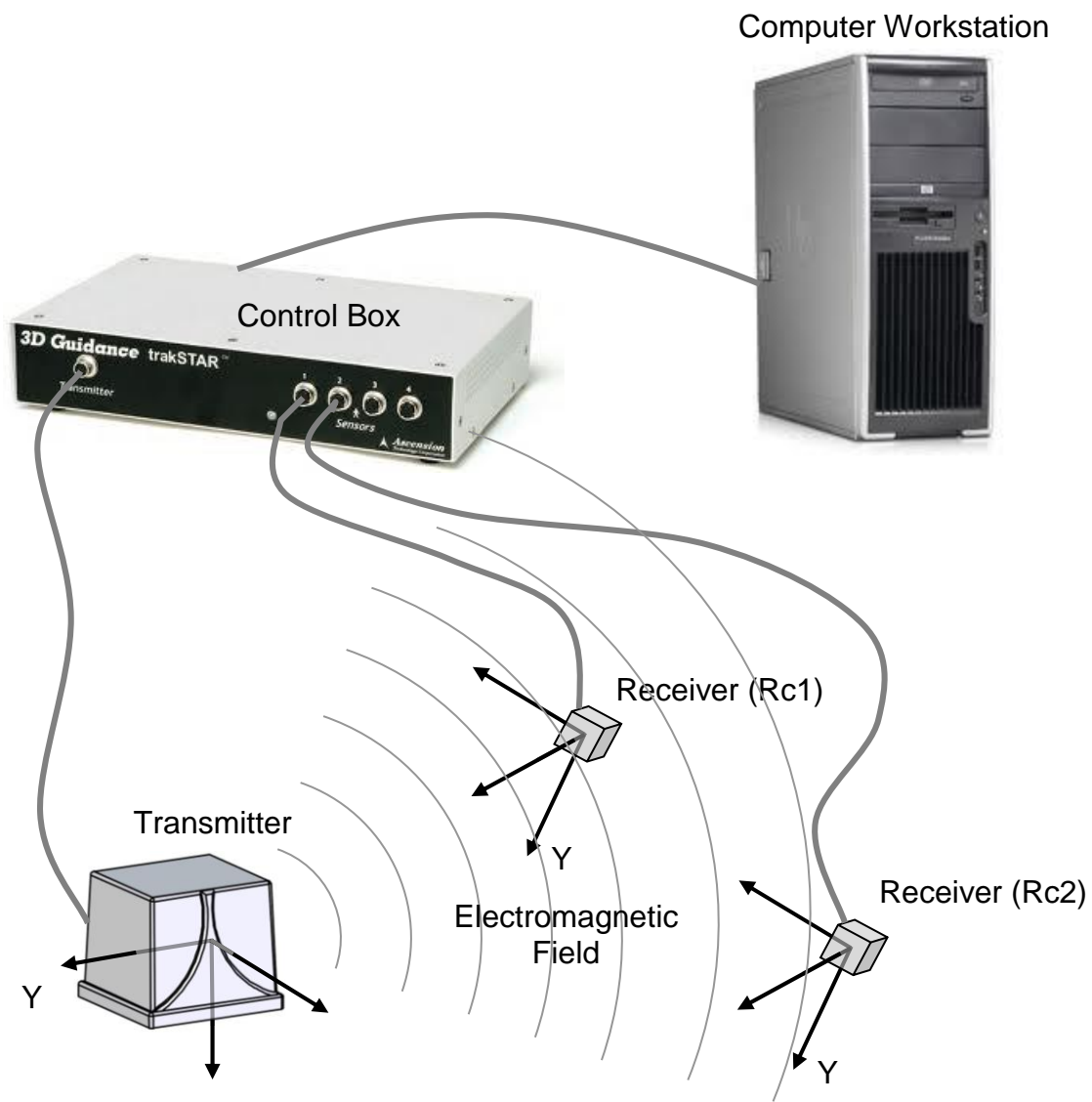


Figure 1-21 - Flock of Birds® electromagnetic tracking system.

In this electromagnetic tracking system (Ascension Technologies, Inc., Burlington, VT), a fixed transmitter emits an electromagnetic field from each of its three orthogonal coils.

Each field induces a current in the antennae of the receivers (Rc1 and Rc2), which are usually fixed to bones of interest. The control box determines the induced currents in the receivers and calculates the attenuation of signal from the transmitter to determine the positions and orientations of the receivers relative to the transmitter. *(Reproduced with permission, Ferreira, 2011).*

1.7.3 In Vitro *Elbow Motion Simulation*

Elbow simulators model joint motion and loading through positioning a cadaveric joint statically or moving it through a range of motion and then measuring the joint's kinematics, contact forces, contact area, or ligament strain (Ferreira, 2011). Cadaveric specimens most closely mimic live human tissues when they are “fresh-frozen” as embalming, decomposition, and dehydration alter tissue mechanics (Fessel *et al.*, 2011; Reilly & Burstein, 1974; Unger *et al.*, 2010; Verstraete *et al.*, 2015; Woo *et al.*, 1986). Most reported systems involve simulated forces with the elbow in static positions or with the elbow being passively flexed or extended by an investigator or device. The latter are known as passive motion simulators and are felt to clinically replicate therapists performing PROM therapy, which has been described earlier in this chapter (see Section 1.5.1). Such devices occasionally have additional simulated muscle forces to enable some joint compression. Multiple studies (Itoi *et al.*, 1994; King *et al.*, 1993a; O'Driscoll *et al.*, 1992; Pomianowski *et al.*, 2001) have used a passive motion simulator developed at the Mayo Clinic in Rochester, Minnesota (Morrey *et al.*, 1991). With this device, the humerus was mounted in a dependent position with static weights with forces of 5% of the maximum potential force applied to the tendons of the biceps brachii, brachialis, and triceps brachii muscles. The investigator then manually performed elbow flexion. The humeral mount could rotate to model gravity-loaded vertical dependent, varus and valgus situations. The use of small “tone loads” with this simulator enabled improved elbow joint contact, likely producing more clinically accurate kinematics.

Active motion simulators enable physiological elbow flexion and extension by using a computer to generate forces through motors and/or actuators connected to tendons (Ferreira, 2011). A novel active elbow motion simulator was developed in the Bioengineering Laboratory of the Hand and Upper Limb Centre (HULC) in London, Ontario, and was first reported in 1997 (Rath). With this device, the mid-shaft of the humerus was rigidly fixed. Stainless steel cables connected the distal tendons of the triceps brachii, biceps brachii, brachialis, brachioradialis, and pronator teres to pneumatic actuators. A computer software program directed electromechanical proportional pressure controllers to provide a desired actuator pressure to produce a proportional force through

each muscle (“load-controlled motion”) (Ferreira, 2011; Rath, 1997). Muscle loads were determined by the maximum voluntary contraction of that muscle crossing the elbow joint *in vivo* based on electromyographic (EMG) analysis (Funk *et al.*, 1987) and the cross-sectional area (CSA) of that muscle (Amis *et al.*, 1979). The humerus could be placed in the dependent, varus, or valgus positions. Simulated active motion could be carried out with good repeatability in the dependent position, where gravity provided a stabilizing vector against elbow flexion while actuators tensioned the biceps brachii, brachialis, and brachioradialis, thus requiring minimal loading through triceps brachii (Dunning *et al.*, 2001a; Johnson *et al.*, 2000). Passive motion could also be assessed in the varus and valgus positions. This simulator was used in multiple investigations (Armstrong *et al.*, 2000; Armstrong *et al.*, 2002; Dunning *et al.*, 2001b, 2001c; Johnson *et al.*, 2000; King *et al.*, 1999; King *et al.*, 2002).

Dunning *et al.* later modified this simulator so that the elbow could be flexed in a “motion-controlled” fashion (2003). In such a system, a “prime mover” of the arm is assigned and the elbow is flexed at a desired rate. The position of the arm is monitored by an electromagnetic tracking system (discussed above, see Section 1.7.2.3) in order to generate a specified excursion rate of the prime mover. The remainder of the tendons are moved in a load-controlled fashion based on computer software that monitors and integrates these inputs and outputs using a custom closed-loop feedback controller. With this simulator, brachialis was considered the prime mover and it was position-controlled using a proportional integral derivative algorithm. Loads were then distributed to the remainder of the muscles (i.e. load-controlled), including triceps, as a ratio of the brachialis load based on the EMG and CSA data used in the previous iteration of the simulator. This motion-controlled simulator was found to produce more reproducible joint velocity and similar or improved repeatability compared to the previous load-controlled version of the simulator in the dependent position (Dunning *et al.*, 2003). It also could simulate active elbow flexion in the varus and valgus positions, however not as reliably as with the arm in the dependent position. It was used in several subsequent investigations (Beingessner *et al.*, 2004; Beingessner *et al.*, 2007).

In 2010, Ferreira *et al.* modified the aforementioned simulator to enable simulated active flexion and extension in the horizontal, varus, and valgus positions (Ferreira *et al.*, 2010). It was more challenging to simulate active motion with the humerus in these positions because the weight of the forearm generates a gravitational moment about the elbow which resists the moments generated by the major elbow flexors and extensors; thus forearm extensors and flexors were used in this iteration of the simulator (see above, Section 1.1.3). The following tendons were incorporated: wrist flexors (flexor carpi ulnaris and radialis), wrist extensors (extensor carpi ulnaris and radialis longus), biceps brachii, brachialis, triceps brachii, brachioradialis, pronator teres, and supinator. Servo-motors with strain gauges on the motor mounts allowed for load-feedback for the brachialis, biceps, and triceps. Load-control outputs were used with pneumatic actuators for the remaining muscles. For each humerus position and forearm position, a certain muscle was designated as the “prime mover” to enable elbow flexion at a given rate. The remainder of the muscles maintained elbow flexion at that rate while maintaining forearm position based on load-control as a function of the load through the prime mover or position-control as a function of flexion angle (Ferreira, 2011). This simulator improved the repeatability of active flexion in the horizontal, varus, and valgus positions compared to the earlier simulator. It has been used in multiple investigations (Alolabi *et al.*, 2012a, 2012b; Ferreira *et al.*, 2015; Sabo *et al.*, 2012a, 2012b). More recently, the simulator has been modified to achieve simulated active and passive motion with the humerus oriented in a vertical overhead position (Kusins *et al.*, 2016). This system is motion-controlled with triceps designated as the prime mover during both flexion and extension.

1.8 Study Rationale

Following ligamentous injury to the elbow with or without surgical repair, it is important to initiate early motion in order to prevent stiffness (Morrey, 2000c). This must be done cautiously as too much motion risks recurrent instability. Most rehabilitation protocols for elbow instability are based on expert opinion (Wilk *et al.*, 1993; Wolff & Hotchkiss, 2006), case series (Rettig *et al.*, 2001; Ross *et al.*, 1999), and modest biomechanical evidence (Alolabi *et al.*, 2012a; Armstrong *et al.*, 2000; Bernas *et al.*, 2009; Dunning *et al.*, 2001b; Fraser *et al.*, 2008; Pichora *et al.*, 2007).

These investigations employ cadaveric specimens in a custom elbow simulator that reproduces *in vivo* forces (see above, HULC simulator, in Section 1.7.3) to study the biomechanical implications of various rehabilitation protocols for lateral elbow instability. Cadaver studies are well-suited for research on elbow rehabilitation since several factors, such as attendance and effort, can be better controlled as compared to clinical studies on patients. As well, cadaver studies can determine possibly deleterious methods of rehabilitation without causing harm to patients with elbow injuries. Such studies may be more repeatable than those involving human participants. Finally, microinstability, not detected by patients or even clinicians, can be measured in the laboratory. This is important as it may compromise ligament healing and lead to degenerative painful arthritis.

In particular, it is important for clinicians to understand whether, as is currently thought, overhead rehabilitation improves stability following lateral elbow injuries. As well, no biomechanical studies have been done on the influence of simulated active elbow extension in the gravity-loaded varus and valgus arm positions. Understanding how this affects kinematics can help determine when such positions can be safely initiated when recovering from an LCL injury. In addition, no studies have looked at the spectrum of LCL injury and its influence on elbow kinematics during AROM. Moreover, HEOs are expensive, but there is little information on whether they are biomechanically effective in the treatment of elbow instability. Understanding how such devices alter kinematics can result in their appropriate prescription. Although this study will focus on LCL injury, it will provide a framework for future studies of MCL and combined MCL and LCL injuries.

1.9 Objectives

The specific objectives of this work are:

1. To compare the stability of the intact elbow to the elbow after:
 - a. Isolated LCL sectioning
 - b. LCL sectioning with and without sectioning of the common extensor origin
2. To better understand the influence of the following in elbow LCL injuries, in order to optimize treatment protocols:
 - a. Arm position (gravity-loaded dependent, overhead, horizontal, and varus)
 - b. Forearm position (full pronation and full supination)

- c. Muscle activation (simulated active and passive motion)
3. To determine the effect of an HEO on an elbow with lateral ligamentous insufficiency

1.10 Hypotheses

The following hypotheses were formulated for the LCL-deficient elbow:

1. In the varus position, instability will increase with increasing lateral soft tissue injury
2. Overhead positioning will minimize instability
3. In the overhead position, pronation will improve stability
4. In the overhead position, active motion will improve stability better than passive motion in both forearm positions
5. A hinged elbow orthosis will not provide additional stability in the dependent, overhead, or horizontal positions
6. A hinged elbow orthosis will reduce instability when the arm is in the varus position
7. While the orthosis is applied, pronation will be more stable than supination
8. While the orthosis is applied, active motion will be more stable than passive motion

1.11 Thesis Overview

Chapter 2 presents the first reported cadaveric study of simulated active motion performed in the overhead and varus positions. Simulated injury to the LCL followed by injury to the common extensor origin is examined with the arm in three positions: dependent, overhead, and varus. In each position, passive and active motion with the forearm in pronation and supination are performed in order to determine the optimal positions for rehabilitation of lateral elbow injuries, depending on the spectrum of lateral injury.

Chapter 3 describes the effectiveness of a hinged elbow orthosis in controlling instability in cadaveric elbows with simulated lateral injuries. The orthosis is evaluated with the arm in four positions (dependent, overhead, horizontal, and varus) during simulated active and passive motion with the forearm in both pronation and supination.

Chapter 4 discusses the impact of Chapters 2 and 3, important conclusions for scientists and clinicians, and directions for future work pertaining to lateral elbow injuries, as well as MCL injuries and elbow dislocations.

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Chapter 2

2 Overhead Rehabilitation in Lateral Elbow Injuries

OVERVIEW: *Following lateral collateral ligament (LCL) injuries, therapists often prescribe active motion exercises with the arm overhead as this is thought to enable gravity and forces through the anterior and posterior arm musculature to compress the elbow joint, improving stability. This effect has yet to be proven biomechanically. This chapter quantifies the effects of muscle activation, arm, and forearm position on elbow stability during simulated rehabilitation exercises following sequential sectioning of the lateral collateral ligament (LCL) and common extensor origin (CEO) of the posterior forearm muscles. Specimens were tested in a custom elbow motion simulator in three arm positions (overhead, dependent, and varus) with the forearm in both pronation and supination. Elbow extension was performed passively by the researcher as well as actively using the simulator. Following combined LCL and CEO injury, overhead positioning enhanced elbow stability relative to the varus and dependent positions ($p < 0.01$ in pronation, $p = 0.04$ in supination). In overhead positioning, forearm pronation improved stability relative to supination ($p = 0.05$). There was no difference in stability between simulated active and passive motion in the pronated overhead position ($p = 0.07$). When the arm was in varus, instability worsened with progressive lateral elbow injury during passive motion ($p = 0.01$ in pronation, $p < 0.01$ in supination). This suggests that rehabilitation with the arm in the overhead position improves elbow stability following lateral soft tissue injuries, and that varus positioning of the arm should be avoided following such injuries.*

Portions of this work were presented at the 2015 Canadian Association of Physical Medicine & Rehabilitation Annual Scientific Meeting and the 2015 American Academy of Physical Medicine & Rehabilitation Annual Assembly.

2.1 Introduction

Acute injury to the elbow lateral collateral ligament (LCL) may occur following trauma causing elbow subluxation, dislocation or fracture-dislocation, such as a fall onto an outstretched hand, motor vehicle accident, or sports injury (O'Driscoll *et al.*, 2000; Tashjian & Katarincic, 2006). Commonly implicated sports include football (Kenter *et al.*, 2000; Muller *et al.*, 2010) and weight-lifting (Kandemir *et al.*, 2002). The common extensor origin (CEO) is injured in 66% of acute traumatic LCL injuries (McKee *et al.*, 2003). These injuries are more likely to cause persistent instability, as the CEO is an important secondary stabilizer of the elbow (Cohen & Hastings, 1997; McKee *et al.*, 2003). LCL insufficiency can also be caused by lateral surgical approaches to the elbow (Morrey & An, 1985). Chronic rupture of the LCL due to recurrent varus tension loading has also been reported. This has been seen in individuals with cubitus varus (O'Driscoll *et al.*, 2001), generalized ligamentous laxity, and following long-term crutch use (Charalambous & Stanley, 2008; Kandemir *et al.*, 2002; McGuire & Bain, 2013; Singleton & Conway, 2004).

Most acute post-traumatic LCL tears without associated fractures are managed non-operatively (Josefsson *et al.*, 1987; Maripuri *et al.*, 2007; Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). Rehabilitation protocols generally begin with immobilization and motion restriction, followed by gradual progression of passive-, active-assisted, and active range of motion (ROM) (Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). Passive ROM involves a patient moving a joint with their other arm or a therapist moving a joint with no assistance from the patient. Active ROM involves a patient actively contracting their muscles to move a given joint. Rehabilitation later progresses to strengthening and, ultimately, sport-, job-, or other functional-specific activities (Reichel *et al.*, 2013; Wolff & Hotchkiss, 2006). The LCL helps prevent external rotatory subluxation of the ulna relative to the humerus and stabilizes the elbow against varus loads (King *et al.*, 2002; McAdams *et al.*, 2005; Morrey & An, 1983). Thus positioning the arm in the gravity-loaded varus position (Figure 1-17D) is typically avoided in the first 6-12 weeks following LCL injuries to avoid putting tensile stresses on lateral elbow structures (Szekeres *et al.*, 2008).

It has been suggested that LCL injuries should be rehabilitated with the arm in a gravity-loaded overhead (Figure 1-17B) position as this is thought to enable gravity and activation of the brachialis, biceps and triceps muscles to cause joint compression and increased congruency, and thus stability (Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006) (Figure 1-18). Lee *et al.* have published the only study to date quantifying elbow kinematics with the arm in the overhead position (2013). Using fluoroscopic analysis to evaluate ulnohumeral distance in cadaveric specimens with sectioned LCLs undergoing passive ROM with the forearm in neutral rotation, they found 104% more displacement with the arm in a dependent (Figure 1-17A) position compared to an overhead position, and concluded that rehabilitation in an overhead position was safe, whereas loading in a gravity-loaded dependent position risked dislocation. Although the overhead position is increasingly used in rehabilitation, no biomechanical studies have assessed the effectiveness of simulated active motion in this position.

Elbow kinematics in the setting of LCL insufficiency have previously been analyzed with the arm in a dependent position. In this position, instability observed with passive flexion was reduced with simulated (i.e. custom motion simulator-controlled) active elbow flexion (Dunning *et al.*, 2001b). Forearm pronation has also previously been shown to improve the stability of the LCL-deficient elbow during active and passive flexion with the arm in the dependent position (Dunning *et al.*, 2001b). While passive motion of the LCL-deficient elbow has been studied with the arm in the varus position (Dunning *et al.*, 2001b), the effect of active motion with this condition has not.

The purpose of this investigation was to quantify elbow stability during simulated rehabilitation exercises with the arm in the overhead, dependent, and varus positions before and after LCL injury with and without concomitant injury to the CEO and lateral elbow capsule. It was hypothesized that following LCL injury:

- 1) Rehabilitation with the arm overhead would minimize elbow instability compared to the dependent and varus positions.
- 2) Active motion would reduce instability compared to passive motion in the overhead position.

- 3) Forearm pronation would reduce instability compared to supination in the overhead position.

2.2 Methods

Seven fresh-frozen cadaveric left upper extremities (mean age \pm standard deviation: 76 ± 10 years; 2 male) amputated at the forequarter level were used. All specimens were scanned using computed tomography to rule out pre-existing arthritis or fracture. Specimens were stored at -20°C and thawed at room temperature ($22 \pm 2^{\circ}\text{C}$) for 18 hours prior to testing and mounted in a custom elbow motion simulator that has been previously described (Dunning *et al.*, 2003; Ferreira, 2010; Johnson *et al.*, 2000; Kusins *et al.*, 2016) (Figure 2-1). The distal tendons of the biceps, brachialis, brachioradialis, pronator teres, triceps, wrist extensors (extensor carpi radialis longus and extensor carpi ulnaris), and wrist flexors (flexor carpi radialis and flexor carpi ulnaris) were sutured with running locking braided Dacron (Gamefish Technologies, Newport Beach, California, USA). Sutures were passed subcutaneously within their respective physiologic compartments to maintain anatomic lines of action of the tendons. In addition, alignment guides were placed at the medial epicondyle for the pronator teres and wrist flexors, at the lateral epicondyle for the wrist extensors, and at the supracondylar ridge for brachioradialis. A custom-machined stainless steel intramedullary humeral mounting rod was inserted into the humeral shaft through the humeral head and cemented with methylmethacrylate. The diameter of the rod was adjusted based on the diameter of the medullary canal of the humerus; the largest rod that could be inserted was used (8mm rod used in 3 specimens, 10mm rod used in 4). This rod was then rigidly mounted into a custom clamp on the base of the elbow motion simulator (Figure 2-2). The rod used to mount the simulator was adjusted based on the arm diameter and upper extremity weight (8mm rod used in 3 specimens, 10mm rod used in 4). The humerus was positioned in neutral ulnohumeral rotation such that when the arm was horizontal and the elbow was flexed to 90° , the forearm was perpendicular to the floor. The sutures for all tendons were then connected via stainless steel cables (0.8mm diameter) to computer-controlled servomotors (for biceps, brachialis, and triceps) and pneumatic actuators (for the remaining tendons). The simulator base could be rotated such that the arm could be positioned in the dependent, overhead, and varus positions.

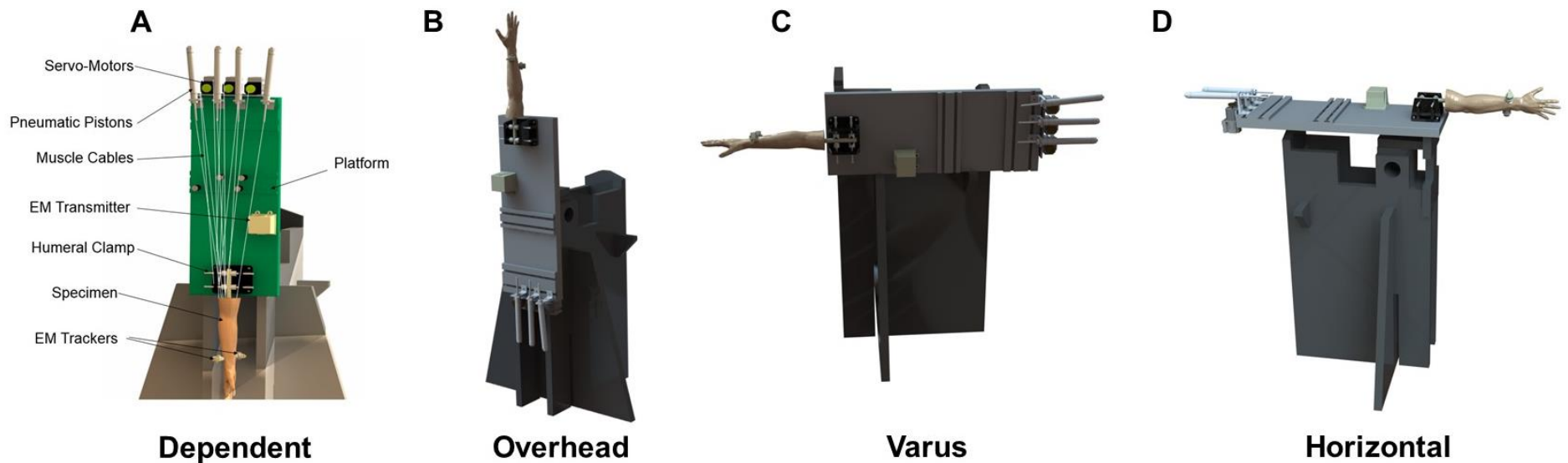


Figure 2-1 - Custom elbow motion simulator in multiple positions.

The parts of the custom simulator are shown in (A), with the humerus in the dependent position. An electromagnetic tracking system, with a transmitter fixed relative to the humerus and a receiver fixed to the ulna, measured ulnohumeral kinematics. Stainless steel cables connected selected tendons of the upper extremity to servo-motors and pneumatic pistons. A computer produced simulated active elbow extension using position feedback. The simulator platform (green) could rotate such that the humerus could be positioned in the overhead (B), varus (C), and horizontal (D) positions (cables, servo-motors, and all actuators not shown). A right upper extremity is shown.

R. Manocha

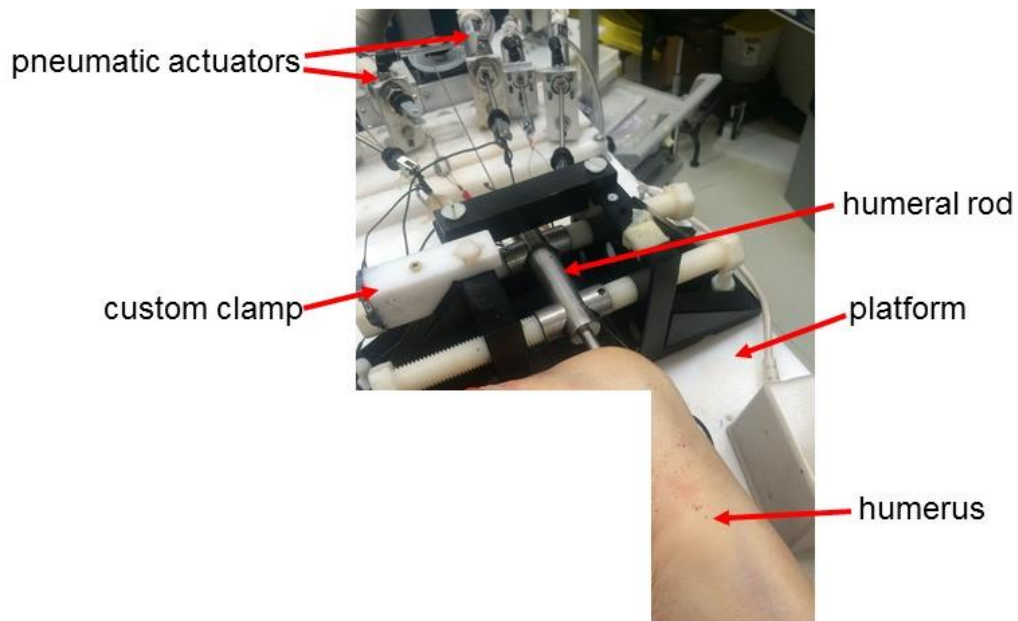


Figure 2-2 - Custom humeral clamp.

A novel humeral mounting system was used in this investigation. A custom-machined stainless steel rod was inserted into the humeral shaft through the humeral head. This rod was then rigidly mounted into the custom clamp which was secured to the simulator platform. Pneumatic actuators are shown for context.

Passive motion was performed by one investigator (RM) manually grasping the wrist and hand to passively rotate the forearm into full pronation or supination until a definite end point of range of motion was reached, and then gently moving the elbow through extension at approximately 10 degrees per second while maintaining the forearm in either full pronation or supination and while avoiding the application of varus or valgus stress. Simulated active motion was achieved as described in previously published studies using a custom-designed LabVIEW program (National Instruments, Austin, Texas, USA) (Ferreira *et al.*, 2010; Kusins *et al.*, 2016). Through sequential timing and loading of each actuator and servomotor, the elbow was actively placed in a starting position, then the desired elbow extension was generated by applying physiologic muscle loads. Simulated active elbow extension was performed at a rate of 10 degrees per second. The following muscles were assumed to be the principle elbow movers: flexors (biceps brachii, brachialis, and brachioradialis) and extensor (triceps brachii). Active forearm rotation was achieved assuming the principle pronator to be pronator teres and the principle supinator to be biceps brachii. During active motion, a 10-N tone load was applied to the wrist extensors and the wrist flexors to stabilize the wrist in a neutral position.

Specimens were examined in the gravity-loaded dependent, overhead, and varus positions. Before testing, in order to minimize viscoelastic effects, five passive then five active preconditioning cycles of elbow flexion and extension through full elbow range of motion with the forearm maintained in both pronation and supination were conducted in all three arm positions. During testing, for each arm position, passive and active elbow extension were performed with the forearm in both pronation and supination. Testing was first conducted with the elbow intact. LCL injury was then simulated by dissecting down to the Kocher interval between anconeus and extensor carpi ulnaris and sectioning the lateral ulnar collateral and the radial collateral ligaments off the lateral epicondyle (“LCL” condition). Complete lateral soft tissue injury was simulated by sectioning the overlying common extensor origin, and the lateral anterior and posterior elbow capsule (“LCL/CEO” condition). The testing sequence was repeated for each injury pattern. During testing, all skin incisions were sutured closed. Specimens were kept moist throughout testing by irrigation with 0.9% normal saline as it is known that mechanical properties of ligaments change with lack of physiologic water content (Thornton *et al.*, 2001).

Ulnohumeral kinematics were recorded using a six degree-of-freedom electromagnetic tracking system (Flock of Birds, Ascension Technologies, Burlington, Vermont, USA) that has previously been shown to have adequate positional and rotational accuracy (Milne *et al.*, 1996). The device's transmitter was rigidly fixed to the base of the simulator such that the receivers would remain within the optimum operating range throughout elbow extension. The first receiver was rigidly fixed to the distal medial ulna, such that the receiver did not limit forearm rotation or cause muscle impingement. Following testing the elbow and wrist were disarticulated and anatomically-derived humeral and ulnar coordinate systems were established from the average of three successive digitizations of bony landmarks using a Delrin stylus attached to a second receiver. The humeral coordinate system was established from: the centre of the humeral shaft; the centre of curvature of the capitellum (using a least-squares sphere-fitting model); and the centre of the trochlear groove (using a least-squares circle-fitting model). The ulnar coordinate system was established from: the centre (using a least-squares circle-fitting model) and plane of the greater sigmoid notch, and the tip of the ulnar styloid (Figure 2-3). The relative motion of the ulna with respect to the humerus was analyzed using the Euler Z-Y-X sequence. Elbow instability was quantified at each elbow extension angle by internal-external rotation of the ulna relative to the humerus.

The effects of active and passive motion, forearm pronation and supination, and arm position on elbow stability for each soft tissue state (intact elbow, LCL injury, combined LCL and CEO injury) were analyzed. A two-way repeated-measures analysis of variance with Greenhouse-Geisser correction (ANOVA) was performed when comparing extension angle and soft tissue state. A three-way ANOVA with Greenhouse-Geisser correction was performed when comparing active and passive motion, with muscle activation (active or passive), soft tissue state, and extension angles as variables. A three-way ANOVA with Greenhouse-Geisser correction was performed when comparing arm position, extension angle, and soft tissue state in the complete injury model. A three-way ANOVA with Greenhouse-Geisser correction was performed when comparing muscle activation, forearm rotation, and extension angle in the complete injury model in the overhead position. For all ANOVAs, statistical significance was set at $\alpha = 0.05$. Post-hoc pairwise comparisons were performed using Bonferroni adjustments.

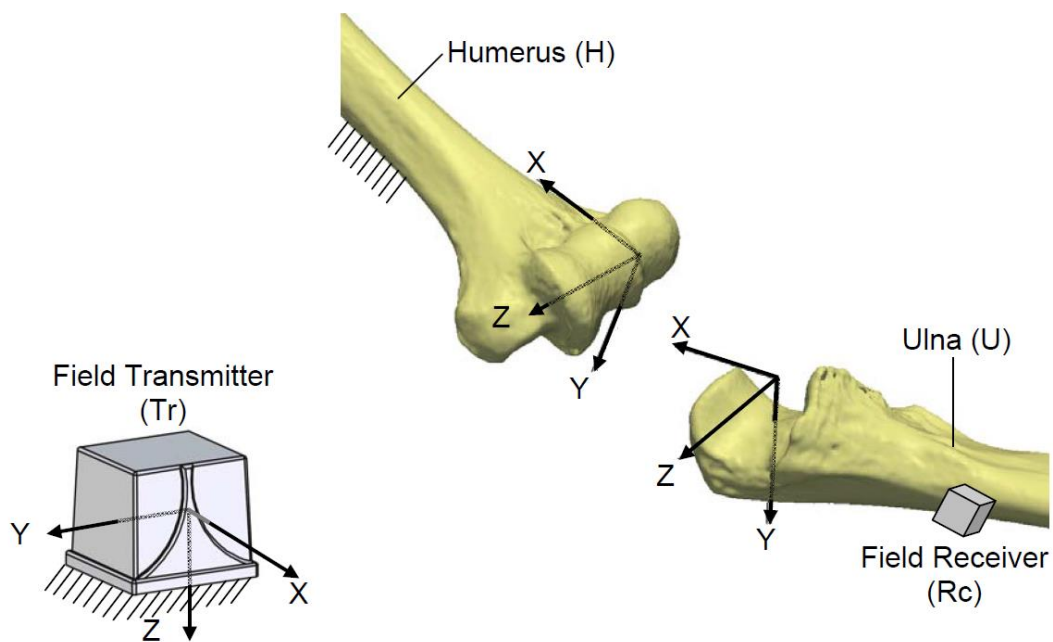


Figure 2-3 - Determination of ulnar and humeral joint coordinate systems.

The transmitter is rigidly fixed to the simulator platform, as shown in Figure 2-1. The humerus is rigidly mounted using the humeral clamp shown in Figure 2-2. The humeral coordinate system is thus derived relative to the transmitter. A receiver is rigidly mounted on the ulna in order to derive the ulnar coordinate system. By convention, the origin of the coordinate system lies at the centre of joint rotation, the x-axis points proximally, the z-axis points medially, and the y-axis points posteriorly. Left upper extremity shown.

(Reproduced with permission, Ferreira, 2011).

2.3 Results

With the arm overhead and forearm pronated, there was no difference in stability by extent of lateral soft tissue injury (active motion, $p = 0.61$; passive motion, $p = 0.19$; Figure 2-4 and Table 2-1). There was also no significant effect of muscle activation (active versus passive ROM) in the overhead position when the forearm was pronated ($p = 0.13$). With combined LCL/CEO injury and forearm pronated, overhead position significantly reduced instability compared to dependent ($p = 0.04$) and varus ($p < 0.01$) positions.

With the arm overhead and forearm supinated, there was no difference in stability by extent of soft tissue injury during active extension ($p = 0.93$; Figure 2-5 and Table 2-2). However, with passive extension, there was significantly increased instability with increased lateral soft tissue injury ($p = 0.01$). Active motion was significantly more stable than passive motion for all 3 arm positions (dependent, $p < 0.01$; overhead, $p = 0.01$; varus, $p = 0.01$) with the forearm supinated. With combined LCL/CEO injury, vertical overhead positioning significantly reduced instability compared to the varus position ($p = 0.01$); however there was no significant difference compared to the vertical dependent position ($p = 0.09$).

In the overhead position, with combined LCL/CEO injury, forearm pronation improved stability relative to supination in both passive ($p = 0.01$) and active ($p < 0.01$) states.

When the arm was varus, instability worsened with progressive lateral elbow injury during both passive ($p = 0.01$ in pronation, $p < 0.01$ in supination) and active motion ($p = 0.04$ in pronation, $p = 0.27$ in supination).

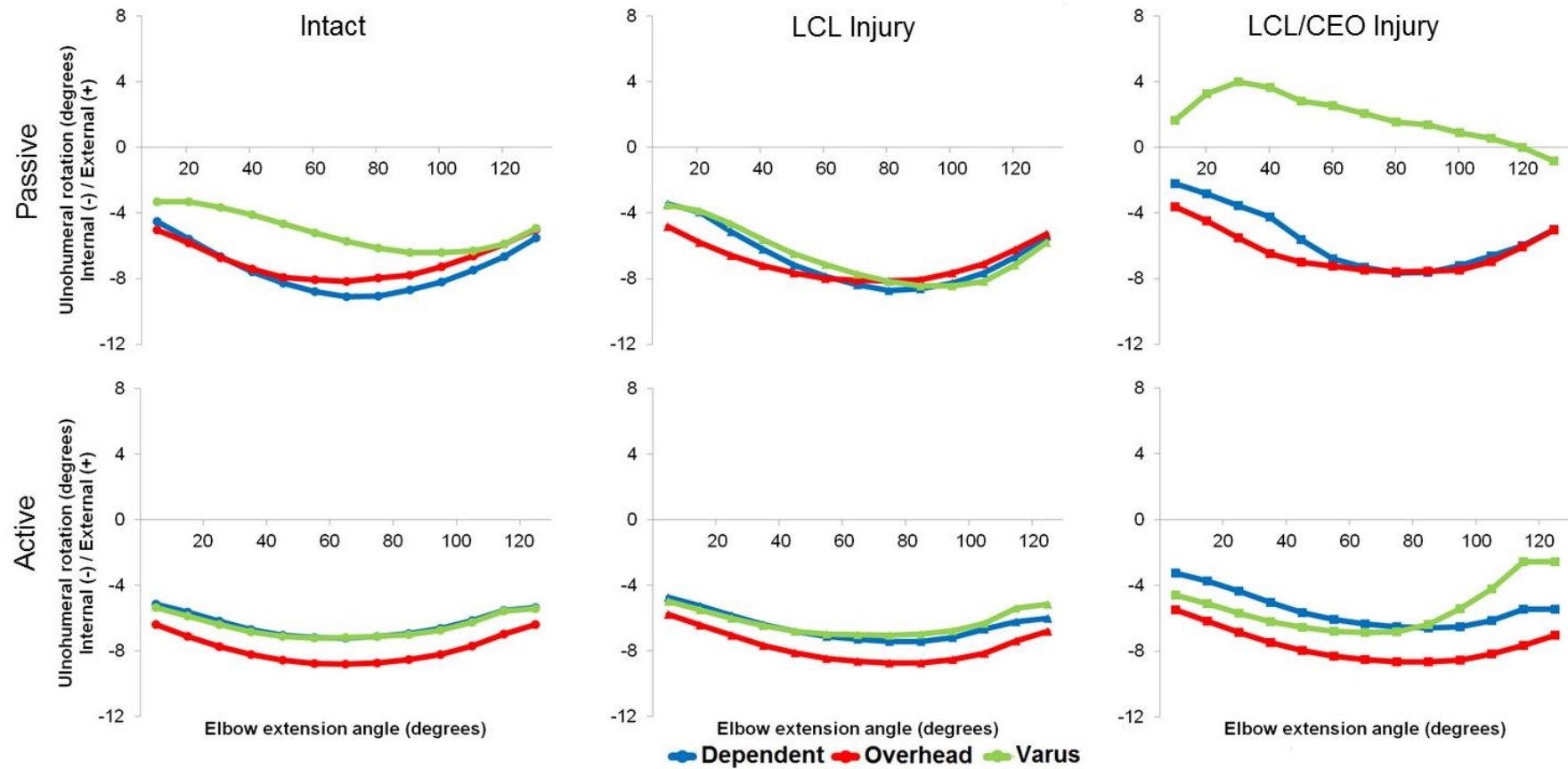


Figure 2-4 - Mean ulnohumeral kinematic profiles during elbow extension with forearm pronated.

Kinematic profiles for passive (top) and simulated active (bottom) elbow extension with forearm pronated are shown for the intact (left), LCL injury (middle), and LCL with CEO injury (right) states. The dependent (blue), overhead (red), and varus (green) humerus positions were examined. Standard deviations were omitted from the graphs for clarity but ranged as follows: active dependent (10.7-13.4°); passive dependent (9.0-13.1°); active overhead (10.8-13.0°); passive overhead (9.9-11.9°); active varus (10.9-13.3°); passive varus (8.8-12.7°).

Table 2-1 - Effect of arm position and muscle activation on elbow stability during extension with forearm pronated.

Arm Position	Muscle Activation	Mean (SD) Ulnohumeral Rotation (degrees)				<i>p</i>	<i>p'</i>
		Intact	LCL Injury	LCL/CEO Injury	Difference		
Dependent	Active	-6.25 (11.57)	-6.04 (11.34)	-5.78 (12.47)	0.47	0.01*	0.68
	Passive	-7.39 (12.22)	-6.73 (10.37)	-5.59 (10.50)	1.81	0.19	
Overhead	Active	-7.86 (11.70)	-7.73 (11.57)	-7.66 (11.63)	0.20	0.61	0.13
	Passive	-6.90 (11.08)	-6.97 (10.68)	-6.34 (10.74)	0.56	0.19	
Varus	Active	-6.47 (11.57)	-6.26 (11.50)	-5.37 (11.64)	1.10	0.04*	0.10
	Passive	-3.48 (11.34)	-2.71 (10.12)	+2.83 (11.39)	6.31	0.01*	

For ulnohumeral rotation, positive values indicate external rotation and negative values indicate internal rotation. “Difference” indicates ulnohumeral rotation for the LCL/CEO state minus that of the intact state. *p*-values describe the significance of ligament state, as the result of a two-way analysis of variance (ANOVA) with ligament state and extension angle as variables. *p'*-values describe the significance of muscle activation, as the result of a three-way ANOVA for muscle activation, ligament state, and extension angle. The asterisk (*) indicates significance ($p < 0.05$). Abbreviations: LCL, lateral collateral ligament; LCL/CEO, lateral collateral ligament and common extensor origin; SD, standard deviation.

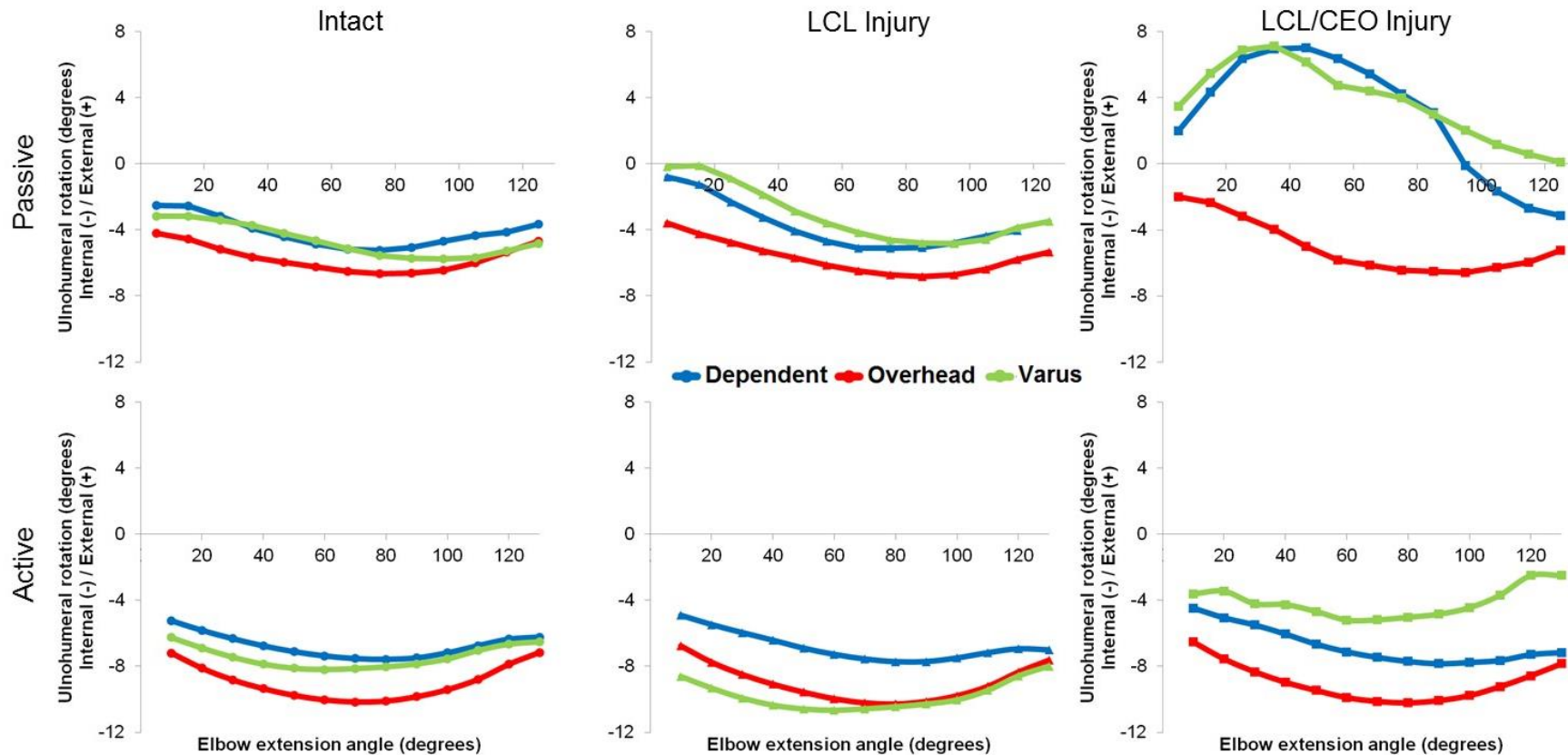


Figure 2-5 - Mean ulnohumeral kinematic profiles during elbow extension with forearm supinated.

Kinematic profiles for passive (top) and simulated active (bottom) elbow extension with forearm supinated are shown for the intact (left), LCL injury (middle), and LCL with CEO injury (right) states. The dependent (blue), overhead (red), and varus (green) humerus positions were examined. Standard deviations were omitted from the graphs for clarity but ranged as follows: active dependent (10.8-11.9°); passive dependent (9.6-18.9°); active overhead (10.8-13.3°); passive overhead (9.9-11.9°); active varus (10.8-11.9°); passive varus (10.3-12.8°).

Table 2-2 - Effect of arm position and muscle activation on elbow stability during extension with forearm supinated.

Arm Position	Muscle Activation	Mean (SD) Ulnohumeral Rotation (degrees)				<i>p</i>	<i>p'</i>
		Intact	LCL Injury	LCL/CEO Injury	Difference		
Dependent	Active	-6.75 (11.53)	-6.81 (11.20)	-6.75 (11.25)	0.00	0.91	<0.01*
	Passive	-4.14 (10.84)	-3.76 (10.61)	2.94 (15.52)	7.08	0.04*	
Overhead	Active	-8.97 (11.99)	-9.02 (11.80)	-8.97 (11.96)	0.00	0.93	0.01*
	Passive	-5.70 (10.62)	-5.69 (10.76)	-5.04 (11.08)	0.66	0.13	
Varus	Active	-10.05 (11.47)	-9.76 (11.20)	-6.27 (11.25)	3.78	0.27	0.01*
	Passive	-4.64 (11.54)	-3.08 (11.67)	3.78 (11.61)	8.42	<0.01*	

For ulnohumeral rotation, positive values indicate external rotation and negative values indicate internal rotation. “Difference” indicates ulnohumeral rotation for the LCL/CEO state minus that of the intact state. *p*-values describe the significance of ligament state, as the result of a two-way analysis of variance (ANOVA) with ligament state and extension angle as variables. *p'*-values describe the significance of muscle activation, as the result of a three-way ANOVA for muscle activation, ligament state, and extension angle. The asterisk (*) indicates significance ($p < 0.05$). Abbreviations: LCL, lateral collateral ligament; LCL/CEO, lateral collateral ligament and common extensor origin; SD, standard deviation.

2.4 Discussion

Previous studies have suggested that active motion and pronation stabilize the LCL-deficient elbow when the arm is in the dependent position (Dunning *et al.*, 2001b; Fraser *et al.*, 2008). This prior work was supported by the results of the current investigation.

Although commonly used in clinical practice, the influence of overhead arm positioning on the stability of the LCL-deficient elbow has not been well-analyzed. To our knowledge, our investigation is the first to look at simulated active motion in the overhead position. This study demonstrates that with combined LCL/CEO injury, during elbow extension with the forearm pronated, overhead positioning reduces elbow instability much more than positioning in the dependent and varus arm positions. With the forearm pronated and the arm overhead, the ulnohumeral kinematics of an elbow with a combined LCL/CEO injury are comparable to those of an intact elbow during both active and passive range of motion. This is likely because of the effect of gravity due to the weight of the forearm and hand unit compressing the elbow joint in this position, increasing bony congruency and thus joint stability (An *et al.*, 1990; Wolff & Hotchkiss, 2006). During active motion with the arm overhead and forearm supinated, there was no difference in ulnohumeral stability based on extent of lateral soft tissue injury, perhaps due to the positive effects of gravity and the force through the activated triceps negating the destabilizing moment caused by forearm supination. However, passive motion in this position created instability that worsened with increasing extent of lateral soft tissue injury. These findings suggest that following LCL and combined LCL/CEO injuries, rehabilitation should be conducted with the arm overhead and forearm pronated.

Given that kinematic pathways between the injured and uninjured elbow are so similar in the overhead position, early motion may be safely initiated in this position following LCL injury or surgical reconstruction. The elbow is particularly prone to stiffness following traumatic injury (Jupiter *et al.*, 2003), thus early range of motion without risking further joint damage or compromising ligament healing can be beneficial in preventing this common complication. Interestingly, active motion was not statistically superior to passive motion in this position, despite evidence that in other arm positions muscular activation increases stability (Dunning *et al.*, 2001b) and the theoretically expected increase in

stability afforded by activated biceps brachii, brachialis, and triceps brachii muscles. This is likely because the stabilizing effects of the overhead position conferred by gravity and forearm pronation outweigh differences due to muscle activation. Most therapy sessions start with passive range of motion in order to precondition the tissues, followed by active range of motion later on in the session (Szekeres *et al.*, 2008), thus this is likely safe to continue doing this with the arm in the overhead position and the forearm in pronation.

This investigation also showed the detrimental effect of placing the arm in a varus position, even during active motion, following LCL injury. Previous work has shown that varus positioning in LCL-deficient cadavers increases elbow instability during passive motion (Dunning *et al.*, 2001b). Most basic activities of daily living (i.e. brushing teeth, dressing, bringing a glass to one's mouth) occur with the elbow in a varus position (Morrey *et al.*, 1981) so it can be a challenging position for patients to avoid. It has also been shown that the average healthy young adult abducts the shoulder to angles greater than 100° approximately 20 times per hour, potentially putting the arm in a varus position (Langohr *et al.*, 2016). This investigation reinforces the importance of reminding patients to restrict motion in the varus position until adequate ligamentous healing has occurred in order to avoid long-term complications such as posterolateral rotatory instability (O'Driscoll *et al.*, 1991; Reichel *et al.*, 2013) or post-traumatic arthritis (Josefsson *et al.*, 1984; Wysocki & Cohen, 2011). This investigation showed that for every condition of muscle activation and forearm rotation, instability in the varus position increased with increasing lateral soft tissue injury. This may further suggest that the timeline for avoiding varus arm positioning should increase based on the extent of injury.

To date, there has been no gold-standard variable for quantifying elbow instability, and there is no value of ulnohumeral rotation that marks instability. As such, we were unable to perform *a priori* power analyses. However, the number of specimens used in this investigation were comparable to that used in similar biomechanical analyses (Fraser *et al.*, 2008; Lee *et al.*, 2013). In addition, in this investigation we simulated LCL with and without CEO injuries. This may not precisely correlate with clinical injuries, however, this was the first investigation to our knowledge to examine the spectrum of lateral soft tissue injuries on elbow stability. In most cadaveric studies of LCL injuries, only the complete

LCL/CEO injury model has been studied (Alolabi *et al.*, 2012; Dunning *et al.*, 2001b; Dunning *et al.*, 2001c; Lee *et al.*, 2013). LCL injuries typically affect those younger than 30 years of age (Stoneback *et al.*, 2012); thus a limitation of many cadaveric studies, ours included, is that specimens of an older age were used. However, at low strain rates, cadaveric tendons and ligaments exhibit no correlation between tensile strength and age (Blevins *et al.*, 1994; Swank *et al.*, 2015; Woo *et al.*, 1991). Finally, cadaveric studies cannot account for some factors that might impact the success of a rehabilitation regime, such as patient motivation, attendance at therapy, and compliance with exercise prescriptions. This study also cannot account for factors that might inhibit range of motion during real-world therapy sessions, such as tactile and visual proprioception, scar tissue formation, and pain (Ervilha *et al.*, 2004; Hodges & Richardson, 1996; Le Pera *et al.*, 2001). However, the results represent a potential worst-case scenario that can help clinicians in providing a reasonable exercise prescription for patients based on biomechanical evidence.

This was the first study to report the effectiveness of an active overhead rehabilitation protocol. The ability to conduct simulated active motion can allow future work in the assessment of the overhead position in conditions of MCL insufficiency, combined MCL-LCL deficiency, as well as ligament injuries combined with fractures. This study also used tone loads in the wrist flexors and extensors in our simulated active motion protocols, something that is not done consistently in cadaveric studies in the literature. The wrist flexors and extensors contribute to elbow stability (King *et al.*, 1993; Park & Ahmad, 2004; Seiber *et al.*, 2009), thus it is likely important to include when simulating active motion. Further research should address how varying wrist flexor and extensor muscle loading affects elbow stability, and how strengthening these muscles could potentially be incorporated into LCL injury rehabilitation protocols.

2.5 Conclusion

The rehabilitation of the LCL-insufficient elbow requires a balance between restricting motion to reduce ligamentous stress to facilitate healing (Jockel *et al.*, 2013), and encouraging motion to reduce stiffness and loss of function (Lansinger *et al.*, 1984; Mehlhoff *et al.*, 1988). Clinicians have recently tried to address this balance by prescribing

range-of-motion exercises with the arm overhead as this is thought to allow gravity to compress the elbow joint, increasing congruency and thus stability. This study is the first to provide a biomechanical basis for this theory. In particular, it suggests that exercises can be safely performed with the arm overhead and the forearm pronated in patients with LCL injuries. Forearm pronation has been previously shown to enhance stability of the LCL-deficient elbow with the arm in the gravity-dependent position (Dunning *et al.*, 2001b); this study demonstrates that this is also true with the arm in the overhead position. Although it was hypothesized that muscle activation would enhance elbow stability in the overhead position, there was no significant difference between active and passive motion in this investigation, suggesting that either can be safely performed. This investigation also illustrates the importance of avoiding varus arm positioning following lateral soft tissue injury in order to enhance ligamentous healing.

2.6 Acknowledgements

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Chapter 3

3 Effectiveness of Bracing in Elbow Lateral Collateral Ligament Injuries

OVERVIEW: *Acute lateral collateral ligament (LCL) injuries are often managed with early immobilization, or protected mobilization, using a hinged elbow orthosis (HEO). There is minimal evidence on how this device affects elbow kinematics or clinical outcomes. This chapter quantifies the effect of an HEO on in vitro elbow stability following LCL injury. Specimens were tested in a custom elbow motion simulator in four arm positions (overhead, dependent, horizontal, and varus) and two forearm positions (pronation and supination) during passive and simulated active elbow extension. The orthosis did not significantly improve elbow stability in any arm position. However there was a trend towards increased instability with the HEO during passive motion in the dependent and horizontal positions. During passive motion when the arm was in the dependent, horizontal, and varus positions, pronation was significantly more stable than supination ($p = 0.02$, $p = 0.04$, and $p = 0.003$, respectively). Active motion was more stable than passive motion when the arm was in the dependent, horizontal, and varus positions. This suggests that an HEO may be beneficial for maintaining the forearm in pronation, and is likely safe to use during active motion. However, an HEO was not effective in preventing elbow instability during passive motion following LCL injury. Caution is required when using an HEO during passive motion in therapy, or when patients are not activating their muscles normally.*

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3.1 Introduction

Injury to the lateral collateral ligament (LCL) is often implicated in cases of elbow instability. Acute LCL injury can arise following a fall onto an outstretched hand, a sports injury, or a motor vehicle accident, leading to a spectrum of dysfunction ranging from posterolateral rotatory instability to frank dislocation (O'Driscoll *et al.*, 2000; Tashjian & Katarincic, 2006). Most acute LCL tears without associated fractures are managed non-operatively (Josefsson *et al.*, 1987; Maripuri *et al.*, 2007; Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). Initially such injuries are treated with immobilization. One method is the thermoplastic splint, which is generally applied with the arm dependent, elbow flexed to 90°, and forearm pronated (Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). The splint is removed for therapy and personal hygiene but is otherwise worn continuously for 4-6 weeks (Szekeres *et al.*, 2008). In cases of more significant instability, a hinged elbow orthosis (HEO) may be used (Cohen & Hastings, 1998; Morrey, 2000a; Reichel *et al.*, 2013; Szekeres *et al.*, 2008; Wolff & Hotchkiss, 2006). Initially these devices may be locked and used as a static splint, in a similar manner as the thermoplastic splints previously mentioned (Morrey, 2000a; Wolff & Hotchkiss, 2006). These devices are typically unlocked to allow motion within a given flexion-extension range early on post-injury. There is no published data on the range typically recommended by clinicians, however terminal extension is typically avoided as the elbow is felt to be more unstable in this position (O'Driscoll *et al.*, 2001). This range is thereafter gradually increased as joint stability improves (Wolff & Hotchkiss, 2006). HEOs are typically worn at all times, including during exercises (Wolff & Hotchkiss, 2006). Early motion within a stable range promotes ligament healing (Cyr & Ross, 1998), prevents stiffness and minimizes muscular deconditioning. There is no published data on how frequently HEOs are used.

Only one biomechanical study has evaluated the effectiveness of HEOs in LCL injury. Lee *et al.* examined seven cadavers with simulated LCL injury during passive motion with the arm dependent and the forearm in neutral rotation, and found that ulnohumeral distraction was nearly twice as much in cadavers with a Bledsoe Brace (Bledsoe Brace Systems, Grand Prairie, Texas) as compared to those that were not braced; although the difference was not statistically significant (Lee *et al.*, 2013). This was postulated to have occurred because the mass of the orthosis increased joint distraction. No reported studies have evaluated bracing

with the arm in any other positions or with active motion, and there are no clinical studies to support the efficacy of HEOs in the context of LCL injury (Hijmans *et al.*, 2004).

The purpose of this investigation was to quantify the effect of an HEO on elbow stability following simulated LCL injury in cadaveric specimens with the humerus and forearm in a variety of clinically relevant positions under both passive and simulated active elbow motion. It was hypothesized that in the setting of LCL injury:

- 1) the HEO would provide no additional stability when the arm is dependent, overhead, or horizontal;
- 2) the HEO would decrease instability when the arm is in varus;
- 3) active motion would be more stable than passive motion when using an HEO;
- 4) pronation would be more stable than supination when using an HEO.

3.2 Methods

Seven fresh-frozen cadaveric left upper extremities (mean age \pm standard deviation: 76 ± 10 years; 2 male) amputated at the forequarter level with no pre-existing pathology were used. Specimens were stored at -20°C and thawed at room temperature ($22 \pm 2^{\circ}\text{C}$) for 18 hours prior to testing. Specimens were mounted in the same custom elbow motion simulator as described in Chapter 2 (Section 2.2). The distal tendons of the biceps, brachialis, brachioradialis, pronator teres, triceps, wrist extensors (extensor carpi radialis longus and extensor carpi ulnaris), and wrist flexors (flexor carpi radialis and flexor carpi ulnaris) were sutured with running locking braided Dacron (Gamefish Technologies, Newport Beach, California, USA) in order to simulate active joint motion. The simulator base was rotated such that the arm could be positioned in the dependent, overhead, horizontal, and varus positions (Figure 2-1). Passive motion was performed by one investigator (RM) manually grasping the wrist and hand to passively rotate the forearm into full pronation or supination until a definite end point of range of motion was reached, and then gently moving the elbow through its arc of flexion and extension at approximately 10° per second while gently maintaining the forearm in either full pronation or supination. Simulated active motion was performed at a rate of 10° per second using a custom-designed

LabVIEW program (National Instruments, Austin, Texas, USA) (Dunning *et al.*, 2001a; Ferreira, 2011; Johnson *et al.*, 2000; Kusins *et al.*, 2016).

Specimens were tested with the arm in the gravity-loaded dependent, overhead, horizontal and varus positions. During testing, for each arm position, passive and active elbow extension was performed with the forearm maintained in both pronation and supination. Testing was first conducted with the elbow intact. LCL injury was then simulated by sectioning the common extensor origin and the lateral ulnar collateral and radial collateral ligaments off the lateral epicondyle, as well as the anterior and posterior lateral elbow capsule off the humerus. The testing sequence was repeated. A left Mayo Clinic Elbow Brace (Aircast, Summit, New Jersey, U.S. Patent #7517329; Figure 1-19) was then applied to the specimen as per the manufacturer's recommendations (Don Joy Global, 2009 & 2011) and testing was repeated. The width of the orthosis was adjusted to ensure good fit to the specimen. In order to eliminate potential motion tracking interference, the metallic loops of the orthosis were replaced with polymer replicas using a three-dimensional printer. The Flock of Birds® (Ascension Technologies, Burlington, Vermont, USA) electromagnetic tracking system was used to record ulnohumeral kinematics in six degrees of freedom. Elbow instability was quantified throughout extension by internal-external rotation of the ulna relative to the humerus.

A two-way repeated measures analysis of variance (ANOVA) with Greenhouse-Geisser correction was performed for each experimental condition, comparing elbow state (intact, LCL injury, LCL injury + HEO) and elbow extension angle. Post-hoc analyses comparing LCL injury to LCL injury with HEO were performed using Bonferroni adjustments. For all tests, statistical significance was set at $\alpha = 0.05$.

3.3 Results

3.3.1 *Dependent Position*

With the arm dependent during passive motion, there was a significant difference in stability between the intact, LCL sectioned and LCL sectioned with HEO elbow states with the forearm in both pronation ($p = 0.03$) and supination ($p = 0.04$) (Figure 3-1 and Tables 3-1 and 3-2). LCL sectioning tended to increase external ulnar rotation relative to the intact

state (pronation: $p = 0.47$; supination, $p = 0.25$). Application of the HEO further increased instability, however this was not statistically significant (pronation, $p = 0.42$; supination, $p = 0.55$). Maximum instability with the HEO occurred at 50° of elbow flexion in the pronated condition and at 40° of elbow flexion in the supinated condition. During passive motion with the HEO and LCL injury, pronation was more stable than supination ($p = 0.02$). During active motion with the arm dependent, the HEO had no significant effect on the stability of the LCL-injured elbow. With the LCL injury and HEO, active motion was more stable than passive motion (pronated, $p = 0.03$; supinated, $p = 0.002$).

3.3.2 *Overhead Position*

With the arm in the overhead position, there was no significant difference in stability of the elbows after LCL sectioning with or without the HEO, regardless of forearm position or muscle activation (Figure 3-2). However, during passive supination, the HEO trended towards reducing instability. This effect was most pronounced at 90° of elbow flexion but did not reach statistical significance. Within the LCL injury with HEO condition, muscle activation had no effect on elbow stability with the forearm in pronation ($p = 0.24$). However, with forearm supination, active motion was more stable than passive motion ($p = 0.02$). During passive motion with the HEO post-LCL injury, forearm rotation had no significant effect ($p = 0.86$).

3.3.3 *Horizontal Position*

With the arm in the horizontal position during passive motion, there was a significant difference in stability between the intact, LCL sectioned, and LCL sectioned with HEO states with the forearm in pronation ($p = 0.01$) but not supination ($p = 0.07$) (Figure 3-3). In pronation, following LCL injury, elbows were no more unstable than the intact state ($p = 1.00$). The braced condition increased instability compared to the unbraced condition, but this was not statistically significant ($p = 0.10$). Instability with the HEO was greatest at 50°. During active motion in the horizontal position, there was no significant effect of LCL sectioning or the HEO with the forearm in both pronation and supination. With the HEO, active motion improved elbow stability relative to passive motion ($p < 0.01$ for both

pronation and supination). When the arm was passively moved with the HEO following LCL injury, pronation was more stable than supination ($p = 0.04$).

3.3.4 *Varus Position*

With the arm in varus during passive motion, there was a significant effect of elbow state in both pronation ($p < 0.01$) and supination ($p < 0.01$) (Figure 3-4). LCL sectioning increased instability ($p < 0.01$ for both pronation and supination). However, adding the HEO did not change elbow stability. During active motion, elbow state had no effect (pronation: $p = 0.11$; supination: $p = 0.28$). With the HEO post-LCL injury, elbows were more stable with active motion than passive motion ($p < 0.01$ for both pronation and supination). During passive motion with the arm in varus while the HEO was applied, pronation was more stable than supination ($p < 0.01$).

3.4 Discussion

During active motion, sectioning the LCL did not worsen instability in any position. Thus, as expected, adding an orthosis during active motion did not alter ulnohumeral kinematics. This supported our hypothesis in the dependent, overhead, and horizontal positions. We had expected the HEO to improve stability in the most provocative varus position but this was not observed. Typically following LCL injury, the arm is braced in the dependent position. These findings suggest that a hinged elbow orthosis can safely be worn following LCL injury during active motion.

During passive motion with the forearm in pronation, elbow state had a significant effect on stability when the arm was in the dependent, horizontal, and varus positions. Within group comparisons, however, only showed a significant increase in instability between the intact state and the LCL sectioned condition with the arm in varus. The addition of the HEO did not improve nor worsen stability in this position. During passive motion in supination, elbow state had a significant effect in the dependent and varus positions. Within group comparisons, however, again only showed a significant increase in instability between the intact state and the LCL sectioned condition with the arm in varus. The addition of the HEO did not improve nor worsen stability in this position. This suggests that varus positioning, with or without a HEO, should be avoided post-LCL injury.

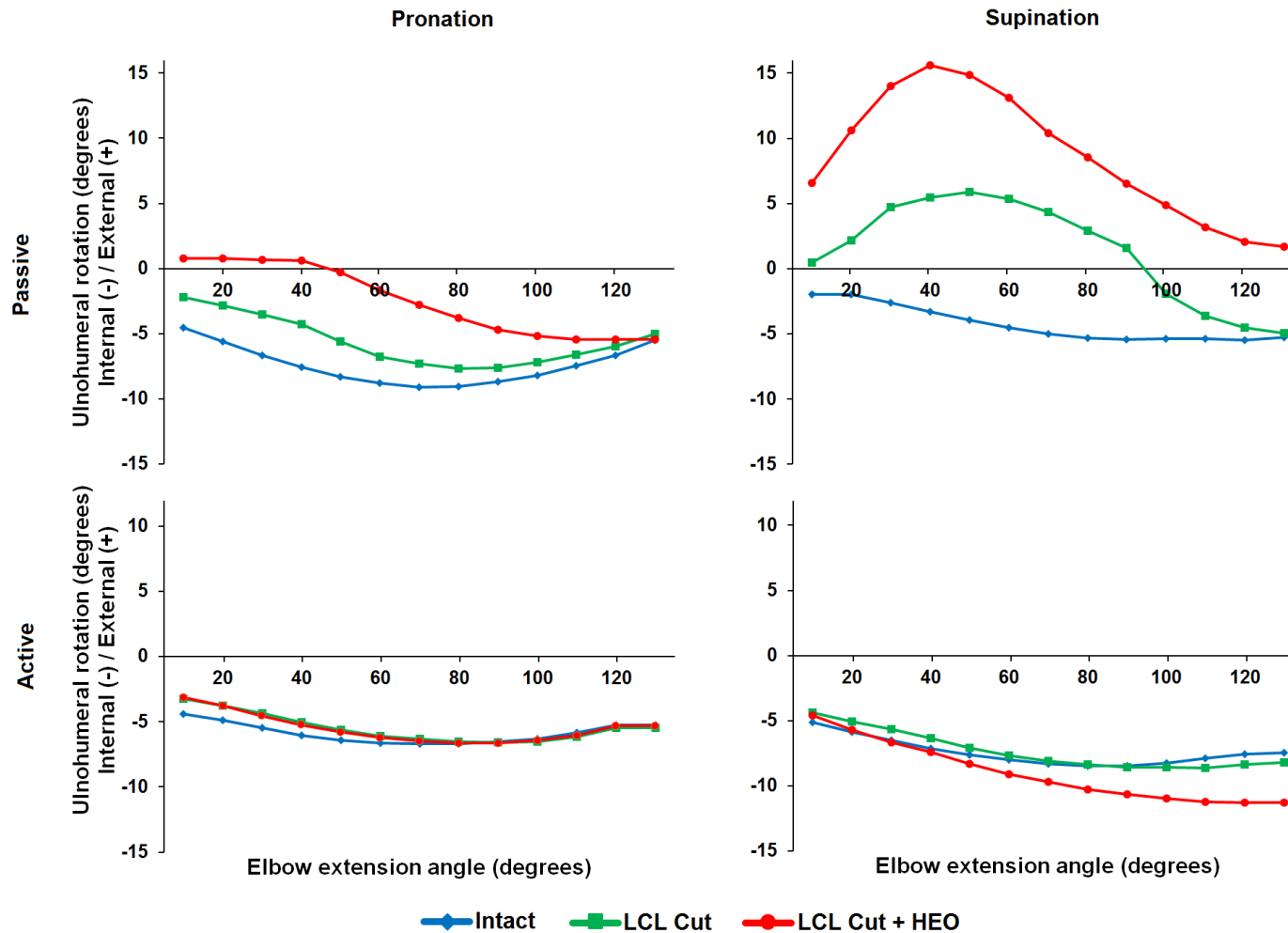


Figure 3-1 - Mean ulnohumeral rotation with arm dependent.

Kinematic profiles for passive (top) and simulated active (bottom) elbow extension are shown for forearm pronation (left) and supination (right). The intact (blue), lateral collateral ligament (LCL) injury (green), and LCL injury with hinged elbow orthosis (HEO; red) states are shown. Standard deviations were omitted from the graphs for clarity but ranged as follows: passive pronated (6.7-13.4°); passive supinated (9.7-18.9°); active pronated (10.6-13.4°); active supinated (11.4-15.5°). During passive motion, there was a significant effect of elbow state (pronation: $p = 0.03$; supination: $p = 0.04$).

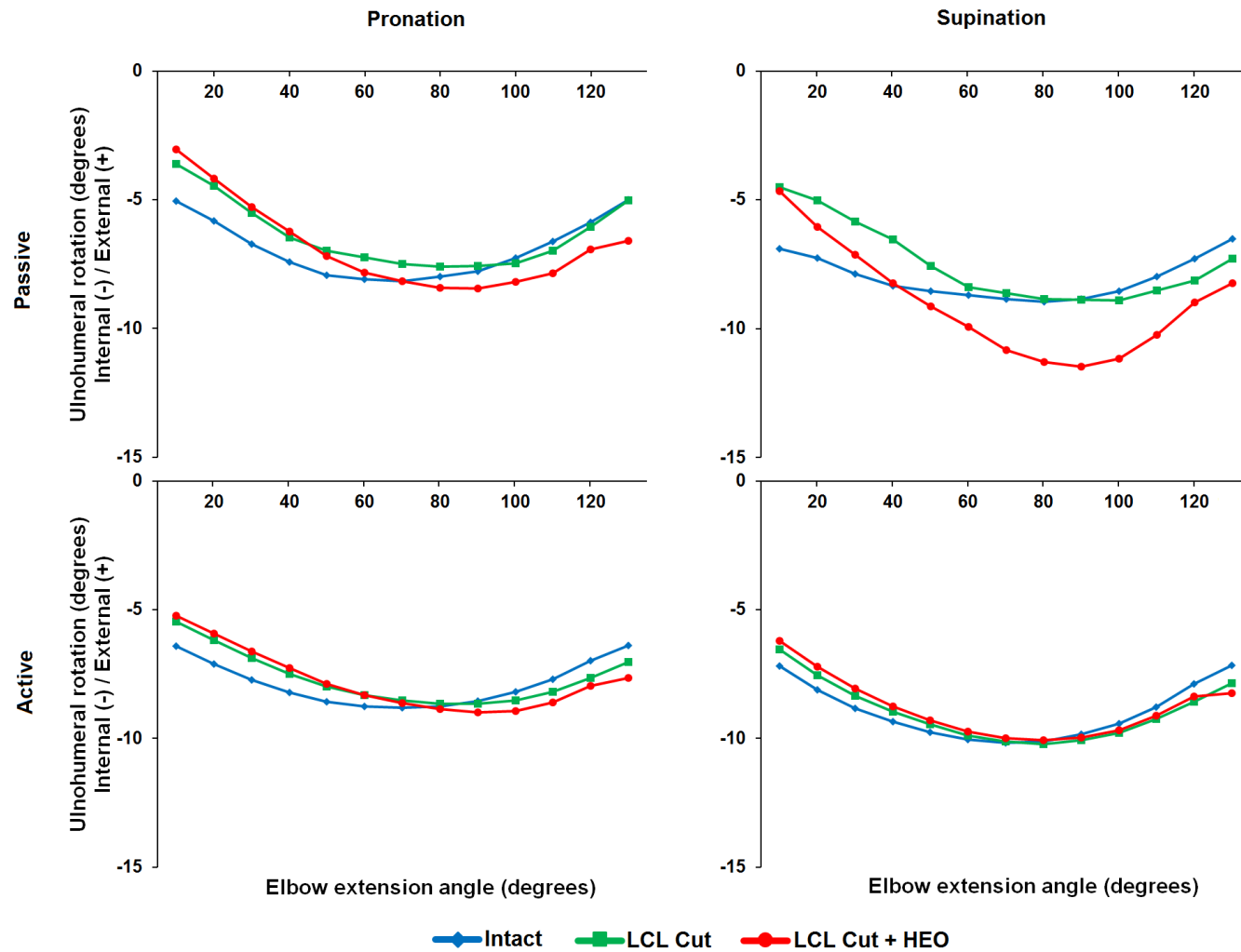


Figure 3-2 - Mean ulnohumeral rotation with arm overhead.

Kinematic profiles for passive (top) and simulated active (bottom) elbow extension are shown for forearm pronation (left) and supination (right). The intact (blue), lateral collateral ligament (LCL) injury (green), and LCL injury with hinged elbow orthosis (HEO; red) states are shown. Standard deviations were omitted from the graphs for clarity but ranged as follows: passive pronated (10.8-14.5°); passive supinated (10.4-12.9°); active pronated (11.8-13.7°); active supinated (11.8-13.9°).

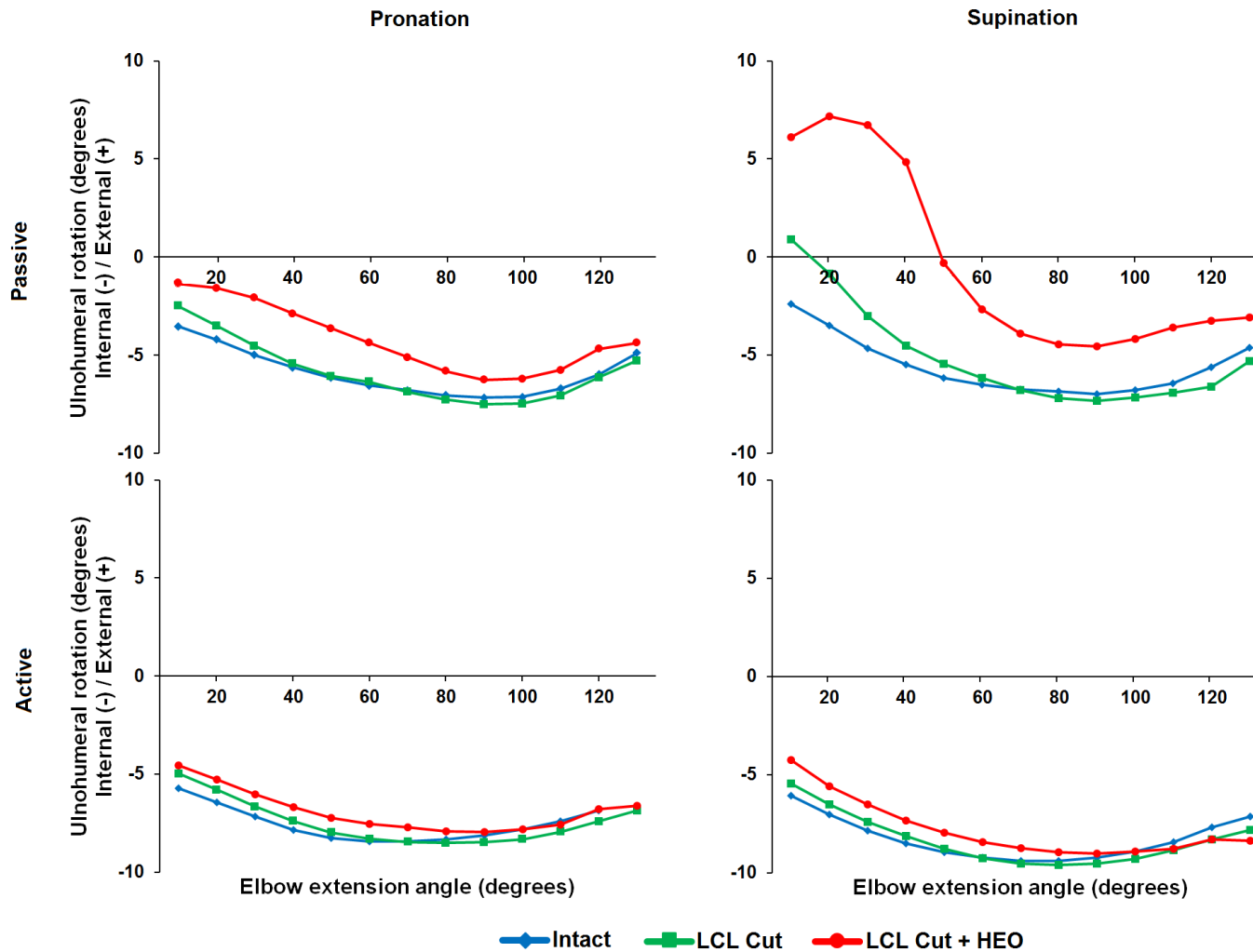


Figure 3-3 - Mean ulnohumeral rotation with arm horizontal.

Kinematic profiles for passive (top) and simulated active (bottom) elbow extension are shown for forearm pronation (left) and supination (right). The intact (blue), lateral collateral ligament (LCL) injury (green), and LCL injury with hinged elbow orthosis (HEO; red) states are shown. Standard deviations were omitted from the graphs for clarity but ranged as follows: passive pronated (10.1-14.7°); passive supinated (8.6-15.4°); active pronated (11.8-13.4°); active supinated (11.5-15.6°). During passive motion with forearm pronated, there was a significant effect of elbow state ($p = 0.01$).

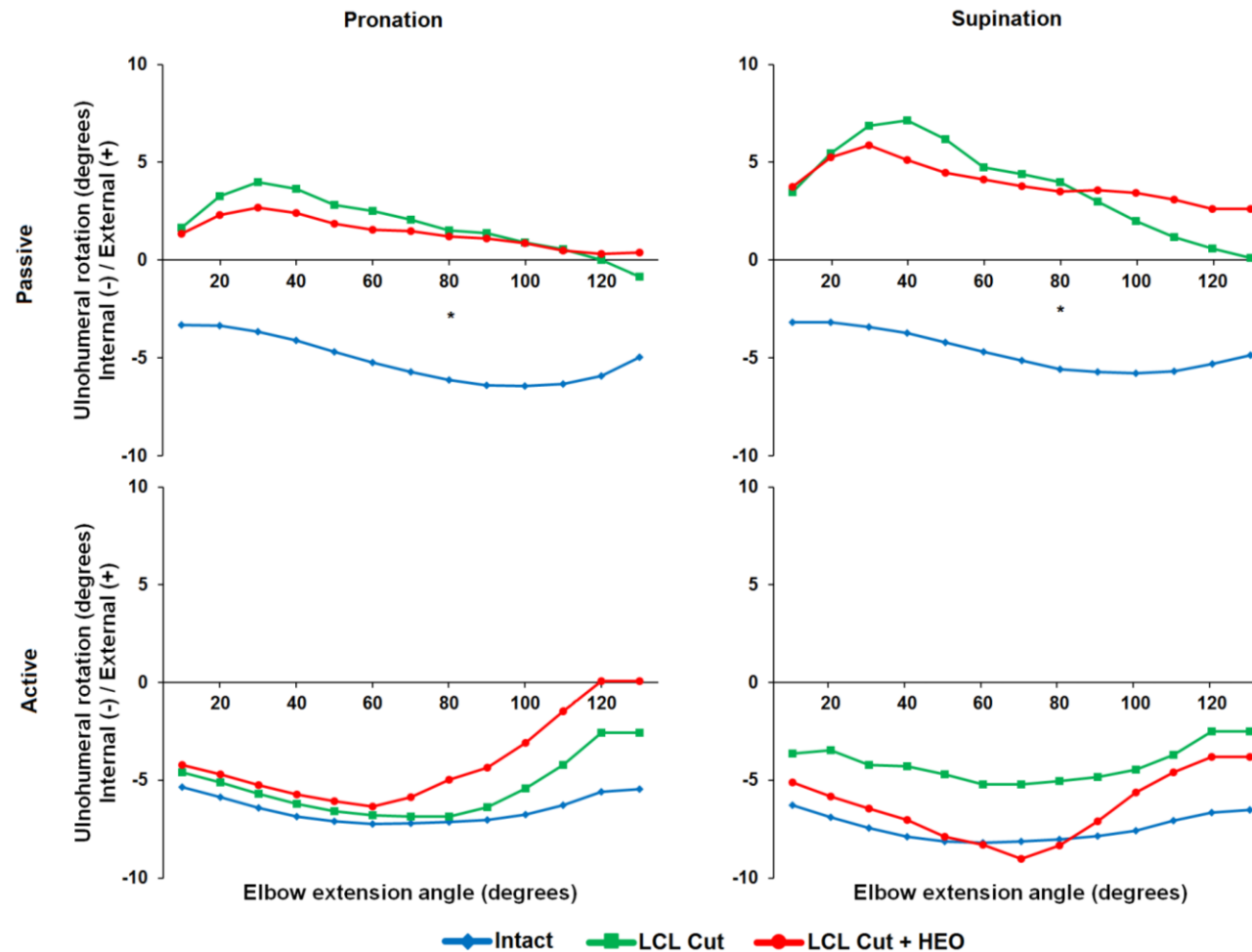


Figure 3-4 - Mean ulnohumeral rotation with arm varus.

Kinematic profiles for passive (top) and simulated active (bottom) elbow extension are shown for forearm pronation (left) and supination (right). The intact (blue), lateral collateral ligament (LCL) injury (green), and LCL injury with hinged elbow orthosis (HEO; red) states are shown. Standard deviations were omitted from the graphs for clarity but ranged as follows: passive pronated (9.6-14.4°); passive supinated (10.9-14.1°); active pronated (11.5-15.1°); active supinated (11.6-15.4°). During passive motion, there was a significant effect of elbow state ($p < 0.01$ in both pronation and supination). LCL sectioning increased instability (*; $p < 0.01$ in pronation and supination). Adding the HEO did not improve or worsen instability ($p = 1.00$ in pronation and supination).

Table 3-1 - Impact of hinged elbow orthosis on elbow stability during extension with forearm pronated.

Arm Position	Muscle Activation	Mean (SD) Ulnohumeral Rotation (degrees)			<i>p</i>
		Intact	LCLI	LCLI + HEO	
Dependent	Active	-5.89 (11.57)	-5.47 (12.47)	-5.53 (11.10)	0.22
	Passive	-7.39 (11.87)	-5.59 (10.37)	-2.45 (9.55)	0.03*
Overhead	Active	-7.86 (12.57)	-7.66 (12.58)	-7.75 (12.74)	0.77
	Passive	-6.90 (11.91)	-6.34 (11.64)	-6.80 (13.18)	0.60
Horizontal	Active	-7.47 (12.36)	-7.46 (12.60)	-6.90 (12.40)	0.07
	Passive	-5.90 (12.00)	-5.84 (13.38)	-4.16 (11.60)	0.01*
Varus	Active	-6.47 (12.50)	-5.37 (12.50)	-3.98 (12.88)	0.11
	Passive	-5.08 (12.23)	1.82 (12.29)	1.39 (12.55)	<0.01*

For ulnohumeral rotation, positive values indicate external rotation and negative values indicate internal rotation. *p*-values describe the significance of elbow state, as the result of a two-way analysis of variance (ANOVA) with elbow state (intact, LCL injury, LCL injury + HEO) and extension angle as variables. The asterisk (*) indicates significance ($p < 0.05$). Abbreviations: HEO, hinged elbow orthosis; LCL, lateral collateral ligament; LCLI, LCL injury; SD, standard deviation.

Table 3-2 - Impact of hinged elbow orthosis on elbow stability during extension with forearm supinated.

Arm Position	Muscle Activation	Mean (SD) Ulnohumeral Rotation (degrees)			<i>p</i>
		Intact	LCL Injury	LCLI + HEO	
Dependent	Active	-7.43 (12.44)	-7.31 (12.23)	-9.01 (13.72)	0.19
	Passive	-4.28 (10.84)	1.37 (15.52)	8.61 (11.39)	0.03*
Overhead	Active	-8.97 (12.91)	-8.97 (12.94)	-8.82 (12.80)	0.89
	Passive	-8.04 (11.38)	-7.47 (11.95)	-9.03 (11.21)	0.15
Horizontal	Active	-8.30 (12.72)	-8.35 (14.25)	-7.79 (12.28)	0.15
	Passive	-5.59 (13.33)	-5.51 (13.36)	-0.39 (10.84)	0.07
Varus	Active	-7.43 (12.65)	-4.13 (14.40)	-6.37 (12.92)	0.28
	Passive	-4.64 (12.47)	3.78 (12.44)	3.94 (12.32)	<0.01*

For ulnohumeral rotation, positive values indicate external rotation and negative values indicate internal rotation. *p*-values describe the significance of elbow state, as the result of a two-way analysis of variance (ANOVA) with elbow state (intact, LCL injury, LCL injury + HEO) and extension angle as variables. The asterisk (*) indicates significance ($p < 0.05$). Abbreviations: HEO, hinged elbow orthosis; LCL, lateral collateral ligament; LCLI, LCL injury; SD, standard deviation.

Table 3-3 - Pairwise comparisons for significant effects of elbow state on ulnohumeral rotation during elbow extension.

Arm Position	Muscle Activation	Pronation			Supination		
		p	p_1	p_2	p	p_1	p_2
Dependent	Passive	0.03*	0.47	0.42	0.04*	0.25	0.55
Horizontal	Passive	0.01*	1.00	0.10	0.07	N/A	N/A
Varus	Passive	< 0.01*	< 0.01*	1.00	< 0.01*	< 0.01*	1.00

p-values describe the significance of elbow state, as the result of a two-way analysis of variance (ANOVA) with elbow state (intact, lateral collateral ligament injury (LCLI), LCLI with hinged elbow orthosis (HEO)) and extension angle as variables. p_1 and p_2 represent the results of pairwise comparisons. p_1 values refer to the difference between intact and LCLI; p_2 values refer to the difference between LCLI and LCLI with HEO.

The asterisk (*) indicates significance ($p < 0.05$).

Interestingly, we found a trend towards increased elbow *instability* with the application of the orthosis to the LCL-injured upper extremity when the arm was passively moved in the dependent and horizontal positions, although this was not statistically significant. Lee *et al.* similarly found that the addition of an HEO following LCL injury with the arm dependent increased ulnohumeral distraction in cadavers undergoing passive elbow flexion (Lee *et al.*, 2013). It is possible that the weight of the HEO (0.47 kg) added an increased gravitational distraction force of 5 N when the arm was loaded in the dependent position, resulting in increased elbow instability. The axial component of such a force would depend on the elbow extension angle. At ranges of elbow flexion less than 90°, axial gravitational forces would tend to be distracting at the elbow joint, whereas at elbow flexion angles greater than 90°, axial forces would tend to have a more compressive component. In this investigation, more instability with the orthosis was seen at elbow flexion angles between 30° and 60° when the arm was dependent, which is consistent with this theory. A trend towards increased instability in the horizontal position occurred particularly between 20° and 60°. These findings suggest that during passive range of motion therapy or when a patient is improperly activating muscles (i.e. due to fatigue, cognitive impairment, altered pain or proprioceptive sensorium, or during sleep), bracing in the horizontal or dependent positions may be harmful by increasing external ulnohumeral rotation. This rotational maltracking may cause pain, impair ligament healing and lead to arthritis. This investigation also suggests that should an HEO be used to manage LCL injuries, it should have an extension block applied to allow motion only at elbow flexion angles greater than 60°, at least early post-injury. This supports clinical experience that the elbow tends to be more unstable at terminal extension (O'Driscoll *et al.*, 2001).

Our hypothesis that the HEO would provide no additional stability when the arm is overhead was confirmed by this investigation. In this position, during passive supination, the orthosis tended to reduce instability, although this was not statistically significant. Previous work has shown that in the dependent position during LCL injury, passive motion is less stable than active motion, and forearm supination is less stable than pronation (Dunning *et al.*, 2001b); thus it is reassuring that an HEO can prevent instability in this situation of forearm supination where the elbow is most at risk for instability. In the overhead position during passive motion with the braced LCL-injured extremity, forearm

rotation had no effect, likely because the compressive gravitational joint force induced by arm position had a much greater effect than destabilizing rotational moments induced by forearm positioning. When the arm was overhead and forearm supinated with the HEO applied, muscle activation provided additional stability. The same effect was not observed with pronation. This is likely because the gravitational moment from the forearm and orthosis weight and the rotational moment conferred from the pronated positioning enabled joint compression that outweighed any further dynamic stability conferred from muscle activation. Clinically, patients often perform exercises with the arm overhead following LCL injury (Szekeres *et al.*, 2008). These results suggest that an HEO is not likely to provide additional benefit during rehabilitation with the arm in this position, except during certain conditions that would not typically be used because they are known to be destabilizing (i.e. passive supination).

Previous work has shown that muscle activation without an orthosis enhances stability during elbow flexion in the LCL-injured elbow when in the dependent position (Dunning *et al.*, 2001b). No studies have looked at the impact of muscle activation on elbow stability following LCL injuries with the addition of an orthosis. In our investigation, when an HEO was applied to an LCL-injured elbow muscle activation enhanced stability when the arm was in the dependent, horizontal, and varus positions. In these positions, as mentioned above, gravitational moments potentially cause increased joint distraction in the LCL-injured elbow. It is likely that the resultant vector of the muscle activation joint reaction forces compressed articular surfaces, augmenting congruency and stability (An *et al.*, 1981; King *et al.*, 1993). As such, it is likely safe to wear an HEO if muscles are being appropriately activated; however, as mentioned earlier, if patient fatigue becomes an issue, it is possible that HEOs may become harmful. We also found that during passive motion, pronation stabilized the LCL-injured elbow more than supination in the dependent, horizontal, and varus positions. As most of the time patients will have their arm in these three positions while performing their activities of daily living (Morrey *et al.*, 1981), an HEO may be beneficial solely to maintain the forearm in pronation.

A limitation of this study was that LCL sectioning only increased instability in the varus condition while the forearm was moved passively. Dunning *et al.* found that LCL

sectioning increased instability in the dependent position during elbow flexion passively and actively, and in the varus position passively (2001b). The entire anterior and posterior elbow capsule was sectioned in that investigation, whereas in the current investigation only half of the lateral capsule was sectioned. The elbow capsule confers significant static elbow stability (King *et al.*, 1993; McKee *et al.*, 2003; Morrey, 2000b; Stroyan & Wilk, 1993), and the lack of instability seen in the dependent position in our study may be related to our decision to section a smaller part of the elbow capsule; however it is likely that this study reflects most clinical capsule injuries associated with LCL tears (McKee *et al.*, 2003). In addition, in this investigation muscle activation was simulated by exerting forces via muscle tendons directly. In reality, when patients contract a muscle, this increases the muscle's diameter (Jones *et al.*, 2008), which would theoretically improve the apposition of the orthosis straps to the skin, improving "fit" and thus the potential of the orthosis to impart some mechanical stability. To account for this, in this *in vitro* investigation, the orthosis was applied tightly, likely tighter than most patients would tolerate with regular use, which should have increased the potential for the orthosis to be effective.

This study also cannot account for some factors that may influence how an orthosis affects ulnohumeral kinematics clinically. It is well-known that ligamentous injury often leads to deficits in proprioception, which is defined as a sensory modality incorporating both joint position sense and joint movement sense (Lephart *et al.*, 1997). This has not been specifically studied in elbow LCL injuries but can be inferred based on studies of other human ligamentous injuries (Barrett, 1991; Corrigan *et al.*, 1992). It has been postulated that the beneficial effect of orthoses in ligamentous injuries may be related to effects on proprioception or neuromuscular control. There have been no studies looking at such effects of an HEO in patients with LCL injuries. However, studies of a variety of hinged knee orthoses and neoprene sleeve-style knee orthoses in the setting of reconstructed or chronic injury to the knee anterior cruciate ligament in humans have suggested that these devices do not significantly improve static (Beynnon *et al.*, 1999) or dynamic (Birmingham *et al.*, 2001) proprioception, muscle contractile forces during isokinetic testing (Wu *et al.*, 2001) or dynamic electromyographic activity of the quadriceps and hamstrings muscle groups, particularly during functional activities (Branch *et al.*, 1989; Ramsey *et al.*, 2003). Other research has suggested these devices may improve gait kinetics

in both reconstructed and ACL-deficient knees (Lu *et al.*, 2006), and static proprioception in ACL-reconstructed knees (Wu *et al.*, 2001). Bracing for ligamentous injury may also have beneficial effects on pain modulation, although again the literature supporting this is conflicting and has not been reported for HEOs in LCL injuries. In general, orthoses may also provide confidence (Birmingham *et al.*, 2008; Zissimopolous *et al.*, 2014) and visible disability (i.e. a patient remembering not to use his or her arm, or a stranger avoiding contact with an injured arm). Again, these factors have not been studied following LCL injury and would be an avenue for future research.

A significant strength of this study is that we preserved the entire length of the humerus as well as the soft tissues under the orthosis, as opposed to potting the mid-shaft of the humerus or denuding the specimen as has been done in other cadaveric bracing studies (Lee *et al.*, 2013; Maurel *et al.*, 2013), which likely helped to ensure sufficient orthosis fit and thus optimize its potential efficacy. To our knowledge, this is also the first reported study to examine the effect of an HEO in the LCL-deficient elbow with the arm in the varus and horizontal positions, and the first to study an HEO during simulated active motion with the arm in multiple positions. Many of the arm positions, forearm rotations, and muscle activations used were physiologic and reflective of activities done by patients in therapy or during daily life.

3.5 Conclusion

In general, there is limited understanding of how orthoses impact elbow biomechanics in the setting of ligamentous injury, and the effects of orthoses are challenging to study with no optimal standard to assess their biomechanical effectiveness. This study attempted to understand how a hinged elbow orthosis affects ulnohumeral kinematics following injury to the lateral collateral ligament of the elbow. This investigation suggests that an HEO may be helpful by keeping the forearm pronated, a position of enhanced stability following LCL injury. It was found that an HEO does not significantly impact elbow stability during simulated active motion when the arm is in a variety of positions. However, during passive motion, use of an HEO may be harmful in arm positions where gravitational forces may increase ulnohumeral distraction, although the effects seen in this study did not reach

statistical significance. In such cases, limiting elbow extension to angles greater than 70° may minimize this risk.

3.6 Acknowledgements

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Chapter 4

4 General Discussion, Conclusions, and Future Directions

***OVERVIEW:** This chapter reviews the objectives and hypotheses stated at the outset of this thesis and discusses the studies performed to address aspects of rehabilitation of lateral elbow injuries. The impact of this work for physicians, therapists, and scientists is reviewed, as well as the strengths and limitations of the investigations performed. Finally, directions for further research in the field of lateral elbow injuries, orthoses, and the application of this methodology to other fields of elbow research is presented.*

4.1 Summary of Hypotheses and Clinical Relevance

This investigation aimed to quantify the effect of several factors employed in the rehabilitation of elbow lateral collateral ligament injuries on elbow stability, using *in vitro* methods. In the opening chapter, three objectives and seven hypotheses were introduced. The subsequent two chapters presented data on the impact of several factors including arm position, forearm position, muscle activation, extent of lateral soft tissue injury, and the presence of a hinged elbow orthosis.

4.1.1 *Instability with Extent of Lateral Soft Tissue Injury*

In Chapter 2, it was shown that ulnohumeral stability worsened with increasing lateral soft tissue injury during active motion in the dependent and varus positions when the forearm was pronated (Objective #1; Hypothesis #1). Instability similarly worsened with further lateral soft tissue injury during passive motion in the dependent position with forearm supinated, and in the varus position during both supination and pronation. In the overhead position, elbow stability did not change significantly with increasing lateral soft tissue injury during active motion.

4.1.2 *Arm and Forearm Position in the Rehabilitation of Elbow Lateral Collateral Ligament Injuries*

It was also shown in Chapter 2 that, following LCL injury, the overhead position is likely best for initiating early active range of motion therapy, in order to maintain elbow stability while preventing the development of elbow stiffness, supporting Hypothesis #2. When the arm was overhead, forearm pronation induced more stability than supination during both active and passive motion, supporting Hypothesis #3. Muscle activation in the overhead position only enhanced stability when the forearm was supinated, partially supporting Hypothesis #4. It was also found that varus positioning should be avoided early post-LCL injury to avoid increased elbow instability (Objective #2).

4.1.3 *Bracing in the Rehabilitation of Elbow Lateral Collateral Ligament Injuries*

In Chapter 3, the presence of a hinged elbow orthosis (HEO) had no significant effect on LCL-injured elbows (Objective #3). This supported Hypothesis #5, but refuted Hypothesis #6. In the dependent and horizontal positions, the addition of the HEO to an LCL-injured specimen tended to increase instability during passive motion, however, this did not reach statistical significance. This suggests that caution should be used when using an HEO during passive ROM in therapy and when patients are not normally activating their muscles, such as during sleep or periods of fatigue. While the LCL-injured elbow was braced, muscle activation enhanced stability when the arm was dependent, horizontal, and varus (Hypothesis #8). It also enhanced stability when the arm was overhead, but in supination only. Within the condition of LCL injury with an HEO, forearm pronation enhanced stability during passive motion when the arm was dependent, horizontal, and varus, but not when the arm was overhead (Hypothesis #7).

4.2 Strengths and Limitations

This body of work has several novel features. It is the first to report on simulated active overhead rehabilitation and quantify the effectiveness of such a motion protocol on elbow stability. It is also the first to investigate simulated active motion during LCL injury with

the arm in varus. Examining the effectiveness of an elbow orthosis during simulated active elbow motion is also unique.

From the standpoint of methodology, we preserved the glenohumeral joint instead of fixing the diaphysis of the humerus. The latter has been done in many cadaveric studies of elbow biomechanics (Bernas *et al.*, 2009; Hammond *et al.*, 2012; Lee *et al.*, 2013). This allowed us to preserve the entire length of the humerus and overlying soft tissues, which was important in ensuring appropriate orthosis fit and in modeling more clinically relevant elbow kinematics. We also looked at multiple arm positions, forearm rotations, and muscle activations that were reflective of activities done by patients in therapy or during daily life. This study also used tone loads in the wrist flexors and extensors during simulated active motion trials. This is not consistently done in the literature, although it is known that these muscle groups contribute to elbow stability (King *et al.*, 1993; Park & Ahmad, 2004; Seiber *et al.*, 2009).

A limitation of this study is that the soft tissue injuries were simulated in both Chapters 2 and 3. Sectioning of the LCL, common extensor tendon origin and elbow capsule may not correlate to clinical injuries. In Chapter 2, however, we presented the first investigation to report on the effect of varying the extent of lateral soft tissue injury on elbow stability, giving further information of clinical relevance to healthcare practitioners regarding treatment protocols. In Chapter 3, complete LCL and CEO sectioning was performed, which is consistent with the majority of clinical LCL injuries (McKee *et al.*, 2003) and is a model which has also been used in multiple prior cadaveric studies of LCL injury (Alolabi *et al.*, 2012; Dunning *et al.*, 2001a; Dunning *et al.*, 2001b; Lee *et al.*, 2013).

With regards to the hinged elbow orthosis, in general it is challenging to perform *in vitro* biomechanical analyses of such devices. In this work, we did not model increases in muscle diameter that occur with muscle activation which may have caused enhanced orthotic “tightness”. We attempted to counteract this by applying the orthosis as tightly as possible, which should have enhanced its potential efficacy.

Finally, it can be challenging to apply *in vitro* work to clinical populations. There are factors that impact the success of any rehabilitation regime that, by design, could not be

incorporated into this investigation, such as patient motivation, attendance at therapy, and compliance with exercise prescriptions. There are also other important factors such as pain, proprioception, ligamentous healing, and scar tissue formation that impact rehabilitation. Some of these factors might also influence the clinical success of an orthosis, although this has not been shown specifically in the literature for elbow LCL injuries.

Despite these limitations, the novel aspects examined in these studies should still help clinicians in providing a reasonable rehabilitation prescription for patients with elbow LCL injuries based on biomechanical evidence.

4.3 Future Directions

4.3.1 *Applying Methodology to Other Clinical Paradigms*

Now that the HULC elbow simulator has been modified to perform simulated active and passive motion in the overhead position, this position should be assessed in the setting of MCL and combined MCL and LCL injuries. Similarly, the ability to simulate active varus and valgus motion should enable the study of valgus motion in MCL injuries and varus and valgus motion in combined MCL and LCL injuries. A similar strategy in investigating extent of medial soft tissue injury could be applied to future investigations of the MCL-deficient elbow. Finally, now that we are familiar with the methodology of using orthoses in cadaveric research, similar studies could be carried out on both MCL-deficient and combined MCL-LCL injuries.

4.3.2 *Expansion of Lateral Collateral Ligament Injury Rehabilitation Research Paradigms*

Further research should be done to investigate other factors involved in rehabilitation of elbow LCL injuries. Firstly, the influence of the forearm extensors on dynamic elbow stability needs to be better elucidated, and studying this can influence how therapists initiate concentric and eccentric strengthening of these muscles in the setting of lateral ligamentous injuries. From an *in vitro* perspective, the tone loads applied through the wrist flexors and extensors in the current simulator could be modified and impacts on elbow stability could be assessed. *In vivo* studies should be carried out with electromyographic

analyses of the forearm extensors in healthy individuals, as has been done in the assessment of the contribution of the wrist flexors to dynamic medial elbow stability (Park & Ahmad, 2004). From the perspective of arm position and forearm rotation, *in vivo* biomechanical analyses may be useful to assess how well current exercises maintain expected positions, and how long patients can sustain repeatable active motion in these positions.

There are several studies which should be done to better understand the role of orthoses for the management of LCL injuries. Other devices could be studied using the same methodology as used in this investigation, such as a locked HEO or custom thermoplastic splint. Custom thermoplastic splints are relatively inexpensive and perhaps could be molded to individual cadavers. It would be helpful to see if customization affects stability differently than a prefabricated HEO. In addition, modifying conditions of the HEO used in this study could also be examined, such as varying strap tightness or brace width. The results of Chapter 3 indicate that the HEO at times tends to worsen instability in the LCL-deficient elbow, potentially because the device itself caused ulnohumeral maltracking. A future avenue for research would be to investigate how varying the varus-valgus angulation of the brace itself, or fixing the forearm rotation provided by the device, affect elbow stability. Ultimately such information could lead to the design of a more biomechanically effective orthosis. Clinically relevant outcomes of HEOs could then be assessed *in vivo*, determining impact on proprioception or pain.

The effects of arm position, forearm position, muscle activation, and presence of an HEO can also be investigated in terms of articular contact or lateral and medial capsule strain to provide more clinical information. Finally, research on the effectiveness of the overhead position and elbow orthoses can be conducted in clinical populations to determine how well these factors reduce risk of development of posterolateral rotatory instability and post-traumatic arthritis following elbow LCL injury.

4.4 Conclusion

This investigation reveals that following elbow LCL injury, active range of motion can be safely initiated early on in the overhead position without risking further instability. This can be helpful to clinicians in preventing the development of elbow stiffness. In addition,

forearm supination and varus positioning of the arm should be avoided early post-injury as these positions risk further posterolateral elbow subluxation.

A hinged elbow orthosis is not helpful in maintaining the biomechanical stability of the elbow following LCL injury. However, it may be helpful solely in keeping the forearm pronated, a position of increased stability, to prevent further subluxation post-injury. There is a risk that such an orthosis will worsen instability during passive motion in the dependent and horizontal positions; thus caution should be used when bracing during passive therapy in these positions or if patients are in states where they may not be activating their muscles normally (i.e. sleep, fatigue, cognitive impairment, altered mental status, altered sensation, etc.). If utilized in these positions, terminal extension should be limited in the HEO to no more than 60°, at least initially.

Despite some limitations of applying this *in vitro* data directly to clinical populations, this thesis provides a biomechanical basis for several important factors that need to be translated to physicians, therapists, and patients in order to improve outcomes amongst those suffering from acute and chronic lateral elbow injuries. There is often limited basic science evidence behind many exercises prescribed in rehabilitation. Cadaveric studies can be useful in determining both safety risks and potential benefits of such exercises in order to better define optimal rehabilitation protocols.

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Appendix A – Glossary

Abduction: The movement of a limb away from a position near the median axis of the body.

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

Active-assisted range of motion (AAROM): The range of motion through which a joint is moved primarily through a patient's efforts to activate adjacent muscles, but accompanied by the aid of an allied healthcare member or the patient's uninjured extremity.

Activities of daily living (ADLs): Functions that an individual must perform for routine self-care; for example: ambulating, bathing, brushing teeth, dressing, feeding, toileting, transferring.

Adduction: The movement of a limb toward a position near the median axis of the body.

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 extensor-supinator muscle group: A group of muscles arising from a common origin located at the lateral epicondyle of the humerus.

Common forearm flexor-pronator muscle group: 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 one or more bone(s) 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.

Creep: The time-dependent deformation of a solid material occurring with the application of a constant stress.

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 no energy-storing components. It consists of 2 Velcro hook and loop straps at the arm and 2 Velcro hook and loop straps at the forearm. A sidebar is aligned axially on the medial and lateral sides of the arm and forearm. There is a hinge at the elbow flexion-extension axis into which pins can be inserted to limit flexion-extension range of motion. This device is often used to reduce instability following ligamentous and/or bony elbow injury.

In silico: Adjective describing the study of a natural process based on computer simulation of that process. In kinematic analyses, this often involves developing a computer model of joint motion and analyzing the impact of altering the model's variables on joint kinematics.

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.

Kinetics: The mechanical study of the forces that result in the motion of points and objects.

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/or 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.

Proprioception: A sensory modality incorporating both joint position sense and joint movement sense.

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, that 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: Movement of a joint that occurs during an *in vitro* study whereby a machine enacts forces on tendon(s) of a cadaver.

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.

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

Ulnohumeral 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 – Appendix to Chapter 3

B.1 Impact of Hinged Elbow Orthosis in the Intact Elbow

Table B-1 is presented to illustrate that the design of the hinged elbow orthosis may have contributed to alterations in elbow kinematics even in the non-injured elbow. This table complements Tables 3-1 and 3-2, and Figures 3-1 to 3-4, in Chapter 3.

B.2 Power for Detecting Differences in Elbow State

Table B-2 is presented to illustrate the results of post-hoc power testing for the ANOVAs performed in Chapter 3. This table complements Tables 3-1 and 3-2 in Chapter 3.

Table B-1 - Impact of hinged elbow orthosis on ulnohumeral rotation in the intact elbow.

		<i>Pronation</i>			<i>Supination</i>		
		Mean Ulnohumeral Rotation (degrees)			Mean Ulnohumeral Rotation (degrees)		
Arm Position	Muscle Activation	Intact	Intact + HEO	<i>p</i>	Intact	Intact + HEO	<i>p</i>
Dependent	Active	-6.37	-6.42	0.87	-6.75	-8.10	0.10
	Passive	-7.39	-5.10	0.04*	-4.14	-4.16	0.98
Overhead	Active	-7.86	-7.80	0.84	-8.97	-8.92	0.80
	Passive	-6.90	-7.11	0.46	-5.70	-6.87	0.02*
Horizontal	Active	-7.47	-6.94	0.11	-8.53	-8.18	0.22
	Passive	-6.32	-5.72	<0.01*	-5.93	-5.24	0.18
Varus	Active	-6.47	-6.06	0.03*	-7.43	-7.16	0.11
	Passive	-5.08	-4.83	0.02*	-4.64	-4.59	0.89

For ulnohumeral rotation, positive values indicate external rotation and negative values indicate internal rotation. *p*-values describe the significance of elbow state, as the result of a two-way analysis of variance (ANOVA) with elbow state (intact, intact + HEO) and extension angle as variables. The asterisk (*) indicates significance ($p < 0.05$). Abbreviations: HEO, hinged elbow orthosis.

Table B-2 - Power analysis for repeated measures ANOVAs in Chapter 3.

Arm Position	Muscle Activation	Power	
		Pronation	Supination
Dependent	Active	0.24	0.24
	Passive	0.64	0.64
Overhead	Active	0.07	0.06
	Passive	0.09	0.33
Horizontal	Active	0.50	0.35
	Passive	0.66	0.46
Varus	Active	0.37	0.19
	Passive	1.00	1.00

Power analyses for two-way analyses of variance (ANOVAs) with elbow state (intact, LCL injury, LCL injury with HEO) and extension angle as variables. Abbreviations: HEO, hinged elbow orthosis; LCL, lateral collateral ligament.

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RE: Permission to Use Copyrighted Material in a Master's Thesis

From: "Van Steenkiste, Francine"
To: Ranita Manocha
CC: Baltrop, Greg; Ure, Alistair
Date: Friday - February 5, 2016 10:21 AM
Subject: RE: Permission to Use Copyrighted Material in a Master's Thesis
Attachments: TEXT.htm; image001.png; Mayo Elbow Brace.jpg; Mime.822

Dear Ranita,

I herewith confirm that it is OK to use the attached image for publication in your master thesis.

Kind regards
Francine

Francine Van Steenkiste
Intl. Clinical Projects & Professional Relations

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Wavre, Belgium / Guildford, UK

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From: Ure, Alistair
Sent: Friday, January 29, 2016 4:46 PM
To: 'Ranita Manocha'; Barltrop, Greg
Cc: Van Steenkiste, Francine; White, Miles
Subject: RE: Permission to Use Copyrighted Material in a Master's Thesis

Hi Ranita –
I have copied Francine Van Steenkiste from our Intl. Clinical Projects & Professional Relations.
Francine will follow up with you shortly regarding your request below.
Best regards,

Alistair Ure
National Market Manager
Bracing and Supports

DJO Canada Inc.
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From: Ranita Manocha
Sent: Friday, January 29, 2016 4:42 PM
To: Ure, Alistair; Barltrop, Greg
Subject: Permission to Use Copyrighted Material in a Master's Thesis

Date: January 29, 2016

Re: Permission to Use Copyrighted Material in a Master's Thesis

Dear Mr. Ure,

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I would like permission to allow inclusion of the photograph you provided of the Mayo Clinic Elbow Brace in my anticipated thesis. The material will be attributed through a citation.

Please confirm in writing or by email that these arrangements meet with your approval.

Yours sincerely,

Ranita Manocha, MD
MSc Candidate, Dept. of Medical Biophysics
Western University

--

Ranita Manocha, MD
Resident, Physical Medicine & Rehabilitation/Clinician Investigator Program
MSc Candidate, Dept. of Medical Biophysics
Schulich School of Medicine & Dentistry, Western University

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Curriculum Vitae

Name: Ranita Harpreet Kaur Manocha

Post-Secondary Education and Degrees: The University of Western Ontario
London, Ontario, Canada
2005-2008 B.A. Cross-Disciplinary Studies
(Western Scholars Distinction)

Harvard University
Cambridge, Massachusetts, United States of America
2006 Course in Psychiatric Anthropology

The University of British Columbia
Vancouver, British Columbia, Canada
2008-2012 M.D.

The University of Western Ontario
London, Ontario, Canada
2014-2016 M.Sc. Medical Biophysics

The University of Western Ontario
London, Ontario, Canada
2012-present Residency, Physical Medicine & Rehabilitation
2015-2016 Clinical Investigator Program
2015-2016 Chief Resident

Academic Honours & Awards: **Western Graduate Research Scholarship** (\$4500/year)
University of Western Ontario, 2014-2016

3rd Place, Abstracts, Resident Research Category (1st Author)
1st Place, Abstracts, Case Reports Category (2nd Author)
Canadian Association of Physical Medicine & Rehabilitation
Annual Scientific Meeting, 2015

1st Place, Abstracts, Systematic Reviews Category (2nd Author)
Canadian Association of Physical Medicine & Rehabilitation
Annual Scientific Meeting, 2014

PSI Foundation Resident Research Grant (\$19,500)
Physician Services Incorporated Foundation, 2014

1st Place, Poster Presentation (1st Author)
Vancouver Coastal Health Research Institute Summer Student
Research Forum, 2010

CIHR Health Professional Student Research Award
Canadian Institutes of Health Research (CIHR), 2010

Florence E. Heighway Summer Research Award
University of British Columbia, 2010

Faculty of Medicine Summer Student Research Program
University of British Columbia, 2010

Oscar E. Forsberg Memorial Scholarship in Medicine
University of British Columbia, 2010

John J. Mason Memorial Scholarship in Medicine
University of British Columbia, 2010

Life Labs Services Prize in Endocrinology (\$1000)
University of British Columbia, 2010

Gwynne-Vaughan Memorial Award in Medicine
University of British Columbia, 2009

NSERC Undergraduate Student Research Award
Natural Sciences & Engineering Research Council of Canada
University of Toronto, 2008
University of Western Ontario, 2007

Department of Medical Biophysics Summer Student Program
University of Toronto, 2008

Richard Konrad Scholarship in Science
Faculty of Science, University of Western Ontario, 2008

Faculty Association Award for Academic Excellence
University of Western Ontario Faculty Association, 2007

Andrew & Sarah Hamilton Scholarship for Academic Excellence
University of Western Ontario, 2006-2008

Continuing Admission Scholarship
University of Western Ontario, 2005-2008

Governor-General's Bronze Academic Medal
South Secondary School, 2005

Local Award Laureate
Canadian Millennium Scholarship Foundation, 2005

National Book Award
University of Toronto, 2004

Peer-Reviewed Publications:

1. MacGillivray M, **Manocha R**, Sawatzky B. The influence of a polymer damper on swing-through crutch gait biomechanics. *Med Eng Phys* 2016;38(3):275-279.
2. **Manocha RH**. On MD/PhD programs and becoming a clinician-scientist. *UBC Med J* 2012;4(1):32.
3. **Manocha RH**. Talking suicide, advocacy, and politics with the Honorable Dr. Fry. *UBC Med J* 2012;3(2):45-46.
4. **Manocha RH**. Playing for fitness: helping seniors stay active. *UBC Med J* 2011;3(1):30-31.
5. **Manocha RH**. One resident, thirty-five cameras: sharing the stories of Inuvik youth. *UBC Med J* 2011; 2(2):26.
6. **Manocha RH**. Vancouver home to Canada's first multidisciplinary vulvodynia program. *UBC Med J* 2010; 2(1):34.
7. **Manocha RH** & Joseph JA. Pulling back the curtains on physicians and the Games. *UBC Med J* 2010;1(2):36-37.
8. Manchanda R, Norman RMG, Malla AK, & **Manocha R**. Antipsychotic use in a first-episode psychosis program. *Int J Psych Clin Prac* 2007;11(2):151-6.

Published Abstracts:

1. **Manocha RH**, Johnson JA, King GJW. Vertical overhead motion in the rehabilitation of lateral elbow injuries: a biomechanical study. *J Rehabil Med* 2016;48:101.
2. Kassam A, **Manocha R**, Sequeira K, Payne M, Batey C, Miller TA. Treatment and rehabilitation in erythromelalgia, a novel approach to a voltage-gated sodium channelopathy. A case of burning red feet. *J Rehabil Med* 2016;48:93.
3. Miller TA, **Manocha RH**, Macaluso S, Sequeira K, Doherty TJ, Ross DC. Scapular winging secondary to spinal accessory nerve palsy following whiplash injury. *Muscle Nerve* 2015;52(S2):S97.
4. **Manocha RH**, Johnson JA, King GJW. Vertical overhead motion in the rehabilitation of lateral elbow injuries: a biomechanical study. *PM&R* 2015;7(9):S189-190.

5. **Manocha RH**, MacGillivray MK, Sawatzky BJ. Aid kinetics during forearm crutch-assisted gait in a transpelvic amputee. *J Rehabil Med* 2014;46:1068.
6. **Manocha RH**, Salter K, Batey C, Macaluso S. Clinical review of acupuncture for non-traumatic shoulder pain. *J Rehabil Med* 2014;46:1065.
7. Batey C, Salter K, **Manocha RH**, Macaluso S. Nutritional supplementation for knee osteoarthritis. *J Rehabil Med* 2014;46:1065.
8. **Manocha RH**, Miller TA, Ross DC, Chinchalkar S. Rehabilitation of bilateral brachial neuritis and radial nerve palsy in hereditary neuropathy: an illustrative case. *J Rehabil Med* 2013;45:1084.
9. **Manocha R**, MacGillivray MK, Sawatzky BJ. The biomechanics of swing-through gait: a proposed comparison of the SideStix™ sports forearm crutch versus standard crutch designs in lower-limb amputees. *UBC Med J* 2011;2(2):S36.

Documentaries:

1. Hughes S, **Manocha R**. The nature of the Coves (Video). Friends of the Cove Subwatershed, London, Ontario, October 2002.

Conference Podium Presentations: (presenter is bolded)

1. **Manocha RHK**, Johnson JA, King GJW. Effectiveness of bracing in elbow lateral collateral ligament injuries: a biomechanical study. *Western University Clinician Scientist Trainee Symposium*, London, Ontario, January 4, 2016.
2. **Manocha RH**, Johnson JA, King GJW. Vertical overhead motion in the rehabilitation of lateral elbow injuries: a biomechanical study. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, Vancouver, May 22, 2015.
3. **Kassam A**, Manocha R, Sequeira K, Payne M, Batey C, Miller TA. Treatment and rehabilitation in erythromelalgia, a novel approach to a voltage-gated sodium channelopathy. A case of burning red feet. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, Vancouver, May 22, 2015.
4. **Batey C**, Salter K, Manocha RH, Macaluso S. Nutritional supplementation for knee osteoarthritis. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, St. John's, June 18, 2014.
5. **MacGillivray MK**, Manocha R, Sawatzky BJ. A kinetic evaluation of a novel forearm crutch design. *BC Orthopaedic Update*, Vancouver, British Columbia, May 2012.
6. **MacGillivray MK**, Manocha R, Sawatzky BJ. The biomechanics of swing-through gait: a comparison of the SideStix™ sports forearm crutch versus a standard crutch

design. 28th *International Seating Symposium*, Vancouver, British Columbia, March 8, 2012.

7. **MacGillivray MK**, Manocha R, Sawatzky, BJ. Kinetic analysis of the SideStix™ Discovery forearm crutches. *International Collaboration on Repair Discoveries Trainee Research Meeting*, Vancouver, British Columbia, May 2011.

Conference Poster Presentations: (presenter is bolded)

1. **Manocha RHK**, Johnson JA, King GJW. Effectiveness of bracing in elbow lateral collateral ligament injuries: a biomechanical study. *Western University Clinician Scientist Trainee Symposium*, London, Ontario, 2016.
2. **Manocha RH**, Johnson JA, King GJW. Bracing in lateral collateral ligament injuries of the elbow: a biomechanical study. *Clinical Investigator Trainee Association of Canada-Canadian Society for Clinical Investigation Annual Scientific Meeting*, Toronto, 2015.
3. **Miller TA**, Manocha RH, Macaluso S, Sequeira K, Doherty TJ, Ross DC. Scapular winging secondary to spinal accessory nerve palsy following whiplash injury. *American Association of Neuromuscular & Electrodiagnostic Medicine Annual Meeting*, Honolulu, 2015.
4. **Manocha RH**, Johnson JA, King GJW. Vertical overhead motion in the rehabilitation of lateral elbow injuries: a biomechanical study. *American Academy of Physical Medicine & Rehabilitation Annual Assembly*, Boston, 2015.
5. **Kassam A**, Manocha R, Sequeira K, Payne M, Batey C, Miller TA. Treatment and rehabilitation in erythromelalgia, a novel approach to a voltage-gated sodium channelopathy. A case of burning red feet. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, Vancouver, May 22, 2015.
6. **Manocha RH**, Johnson JA, King GJW. Vertical overhead motion in the rehabilitation of lateral elbow injuries: a biomechanical study. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, Vancouver, May 22, 2015.
7. **Manocha RH**, MacGillivray MK, Sawatzky BJ. Aid kinetics during forearm crutch-assisted gait in a transpelvic amputee. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, St. John's, June 2014.
8. **Manocha RH**, Salter K, Batey C, Macaluso S. Clinical review of acupuncture for non-traumatic shoulder pain. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, St. John's, June 18, 2014.
9. **Batey C**, Salter K, Manocha RH, Macaluso S. Nutritional supplementation for knee osteoarthritis. *62nd Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, St. John's, June 18, 2014.

10. **Manocha RH**, Miller TA, Ross DC, Chinchalkar S. Rehabilitation of bilateral brachial neuritis and radial nerve palsy in hereditary neuropathy: an illustrative case. *61st Annual Scientific Meeting of the Canadian Association of Physical Medicine & Rehabilitation*, Montreal, June 1, 2013.
11. MacGillivray MK, Manocha R, **Sawatzky BJ**. A kinetic evaluation of a novel forearm crutch with a shock absorption system. *51st Annual Scientific Meeting of the International Spinal Cord Society*, London, United Kingdom, September 2012.
12. **MacGillivray MK**, Manocha R, Sawatzky BJ. A kinetic evaluation of a novel forearm crutch design. *17th Biennial Meeting of the Canadian Society of Biomechanics*, Burnaby, British Columbia, June 2012.
13. **MacGillivray MK**, Manocha R, Sawatzky BJ. A kinetic evaluation of a novel forearm crutch with a shock absorption system. *Interdependence 2012 Global SCI Conference*, Vancouver, British Columbia, May 2012.
14. **Manocha R**, MacGillivray MK, Sawatzky BJ. The biomechanics of swing-through gait: a proposed comparison of the SideStix™ sports forearm crutch versus standard crutch designs in lower-limb amputees. *University of British Columbia Medical Journal-Medical Undergraduate Society Research Forum*, Vancouver, March 2011.
15. **Manocha R**, MacGillivray MK, Sawatzky BJ. The biomechanics of swing-through gait: a proposed comparison of the SideStix™ sports forearm crutch versus standard crutch designs in lower-limb amputees. *International Collaboration on Repair Discoveries (ICORD) 3rd Annual Research Meeting*, Vancouver, February 2011.
16. **Manocha R**, MacGillivray MK, Sawatzky BJ. Biomechanical analysis of swing-through gait: SideStix™ versus standard forearm crutch designs (pilot study). *Vancouver Coastal Health Research Institute Summer Student Research Forum*, Vancouver, August 25, 2010.

Invited Presentations:

Department of Physical Medicine & Rehabilitation, Western University

Annual Research Day

January 11, 2016	Bracing in lateral elbow injuries: a biomechanical study
January 12, 2015	Vertical overhead rehabilitation in lateral elbow injuries
December 2, 2013	Biomechanics of bracing in ulnar collateral ligament tears
April 15, 2013	Proposed Masters during residency: elbow ligament loading with bracing

Journal Club

February 8, 2016	Amputee care and being a “good leader”
June 15, 2015	Technology-enabled aphasia therapy
February 9, 2015	Management of agitation post-brain injury
October 7, 2013	Intravenous ketamine for complex regional pain syndrome
November 26, 2012	Stroke: neuroprotectants and rehabilitation setting

Academic Half Day Lectures

December 7, 2015	Neurogenic Bowel: Structure, Function, and Management
November 9, 2015	Glenohumeral Instability
September 11, 2015	Exercise Terms & the Exercise Prescription
August 17, 2015	Neuroanatomy: Spinal Cord & Spinal Cord syndromes
July 10, 2015	Anatomy: Leg & Ankle
July 10, 2015	Physical Examination: Foot & Ankle
July 6, 2015	Anatomy: Forearm
May 4, 2015	Electrodiagnostic Approach to Lumbosacral Plexopathy
April 27, 2015	Electrodiagnosis of Ulnar Neuropathy at the Elbow
March 16, 2015	Medical Complications of Spinal Cord Injury
August 24, 2014	All About Parasport
August 8, 2014	Anatomy: Ventricles & Meninges
July 7, 2014	Anatomy: Arm & Elbow
July 7, 2014	Physical Examination: Thoracic Outlet Syndrome
April 28, 2014	Pharmacological Management of Spasticity
July 29, 2013	Patellofemoral Pain Syndrome
July 22, 2013	Anatomy: Pelvis & Hip
July 15, 2013	Anatomy: Shoulder & Neck
July 15, 2013	Physical Examination: Triangulofibrocartilage Complex Tear
July 8, 2013	Kienbock's Disease
August 20, 2012	Mood Disorders: Diagnosis & Management
August 13, 2012	Anatomy: Skull
August 10, 2012	Achilles Tendinopathy & Rupture
July 30, 2012	Anatomy: Lumbosacral Plexus
July 16, 2012	Acromioclavicular Joint Sprains
July 9, 2012	Anatomy: Brachial Plexus

Department of Medical Biophysics, Western University

Graduate Seminars

December 17, 2015	Elbow lateral collateral ligament injuries: A biomechanical evaluation of the effectiveness of bracing
February 12, 2015	Vertical overhead rehabilitation in the management of lateral elbow injuries

Schulich School of Medicine & Dentistry, Western University

Year 4 Undergraduate Medicine Communications Course Small Group Facilitator

March 12, 2014	Communicating in Teams, Breaking Bad News
March 11, 2014	Giving & Receiving Feedback

Year 2 Undergraduate Medicine Musculoskeletal Course

Clinical Methods

February 28, 2014	Patellofemoral Pain Syndrome (Knee Examination)
February 26, 2014	Patellofemoral Pain Syndrome (Knee Examination)
February 26, 2014	Ankylosing Spondylitis (Spine Examination)

Anatomy Lab

February 3, 2015	Anterior and Medial Thigh
February 4, 2013	Thigh & Gluteal Region
January 28, 2013	Anterior Forearm

Parkwood Hospital, London, Ontario

“Amp Up Your Knowledge”: Amputee Rehabilitation Program In-Service Education Sessions

April 1, 2015	Acute wheelchair-related injuries amongst lower extremity amputees
March 4, 2015	Anticipated outcomes of individuals with transtibial amputations and contralateral limb dysfunction
February 4, 2015	Cognition and lower extremity amputations

Others:

1. Cassidy C, **Manocha R**, Payne M. What is Physiatry? *Western University Physical Medicine & Rehabilitation Interest Group*, London, Ontario, February 18, 2015.
2. **Manocha R**, Woodward E. “Abilities in Focus”: Sport & Disability Advocacy Through Photographic Research. *University of British Columbia Physical Medicine & Rehabilitation Academic Half-Day*, Vancouver, British Columbia, November 16, 2011.
3. Simonett G, **Manocha R**, Tawse H, Bowie K. All about wheelchair sports. *University of British Columbia Physical Medicine & Rehabilitation Academic Half-Day*, Vancouver, British Columbia, July 6, 2011.
4. **Woodward E**, Manocha R. Abilities in focus: photo novella as a way to explore disabled athletes’ experiences in sport. *University of British Columbia Island Medical Program Class of 2012*, Victoria, British Columbia, November 7, 2010.
5. **Manocha R** and the Rural Education Action Program. Becoming a rural healthcare provider. Presented to Glenmerry Elementary School grade seven students, Trail, British Columbia, June 21, 2010.
6. **Manocha R**. The Alternative Spring Break program: how service learning fits into the classroom. *University of Western Ontario Department of Statistical & Actuarial Sciences Research Group Retreat*, London, Ontario, May 1, 2008.
7. **Manocha R**. Creating change at home. *2007 Activate! Youth Sport Leadership Conference*, Esteem Team Canada, Ottawa, Ontario, May 17, 2007.

8. Khan A, **Manocha R**, & Wei L. So we served in the Dominican – now what?
University of Western Ontario Alternative Spring Break Showcase, London, Ontario,
April 2007.

Interviews:

1. All you ever wanted to know about the elbow with Ranita Manocha [Podcast].
Gradcast, for the Western University Society of Graduate Students. January 20, 2016,
London, Ontario, Canada.
2. The #AAPMR2015 Experience [YouTube Video]. American Academy of Physical
Medicine & Rehabilitation. October 3, 2015, Boston, Massachusetts, United States of
America. Available at: <https://youtu.be/ITALsGFksG0>.
3. #AAPMR2015 Connections [YouTube Video]. American Academy of Physical
Medicine & Rehabilitation. October 3, 2015, Boston, Massachusetts, United States of
America. Available at: <https://youtu.be/pNL6gaCPJIQ>.
4. Breakfast Television [Television]. The New PL. The Canadian Association for Girls in
Science. November 7, 2005, London, Ontario, Canada.

Reviewer Activities:

- 2014 Reviewer, Musculoskeletal Abstracts, American Academy of Physical Medicine
& Rehabilitation 2014 Annual Assembly (18 abstracts)
- 2013 Reviewer, Musculoskeletal Abstracts, American Academy of Physical Medicine
& Rehabilitation 2013 Annual Assembly (14 abstracts)
- 2010 Reviewer, University of British Columbia Medical Journal (2 articles)

Supervision:

- | | |
|---------------------|--|
| January 2016 | Erica Yang, Co-Op High School Student
Oakridge Secondary School, London, Ontario |
| January - June 2015 | Nick Asapu, Co-Op High School Student
London Central Secondary School, London, Ontario |
| May - August 2014 | Jennifer Dowling-Medley, Undergraduate Biomedical Engineering
Co-op Student (Year 4), University of Guelph |
| May - August 2014 | Rafael Gomes Pereira, Undergraduate Mechanical Engineering
International Exchange Student (Year 4), Universidade Federal de
Pernambuco |
| June - July 2014 | Allison Pellar, Masters of Engineering Science Student, UWO |

Certifications:

ACLS Provider (2015)

Workplace Hazardous Materials Information System (WHMIS) Basic Training (2015)

Classifier, Canadian Wheelchair Basketball Association (2011)

Tri-Council Policy Statement Ethical Conduct for Research Involving Humans (2008)

Professional Memberships:

2015-present Clinical Investigator Trainee Association of Canada

2012-present Ontario Medical Association

2012-present College of Physicians & Surgeons of Ontario

2011-present American Academy of Physical Medicine & Rehabilitation

2009-present Canadian Association of Physical Medicine & Rehabilitation

2008-present Canadian Medical Association

2008-present British Columbia Medical Association

2008-2012 College of Physicians & Surgeons of British Columbia

Non-Academic Honours **Rick Hansen 25th Anniversary Relay Medal-Bearer**
2012

2010 Olympic Winter Games Torchbearer
2009

Letter of Accomplishment in Community Leadership & Service Learning
President of the University of Western Ontario, 2007

Honour W Award for Outstanding Student Leadership
University Students' Council, University of Western Ontario, 2006

Novice Champion
Skate Canada National Synchronized Skating Championships
2001