The Design and Validation of a Computational Rigid Body Model of the Elbow.

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THE DESIGN AND VALIDATION OF A COMPUTATIONAL RIGID

BODY MODEL OF THE ELBOW

A Thesis submitted in partial fulfillment of the requirements for the degree of Master of
Science in Biomedical Engineering at Virginia Commonwealth University

by

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B.S., University of Virginia, 2004

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# Table of Contents

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acknowledgement</td>
<td>iii</td>
</tr>
<tr>
<td>List of Tables</td>
<td>viii</td>
</tr>
<tr>
<td>List of Figures</td>
<td>ix</td>
</tr>
<tr>
<td>Abstract</td>
<td>xii</td>
</tr>
<tr>
<td>1. INTRODUCTION</td>
<td>1</td>
</tr>
<tr>
<td>1.1. OVERVIEW OF COMPUTATIONAL MODELING</td>
<td>1</td>
</tr>
<tr>
<td>1.2. RIGID BODY MODEL SIMULATION</td>
<td>3</td>
</tr>
<tr>
<td>1.3. COMPUTIONAL SOFTWARE PACKAGES</td>
<td>5</td>
</tr>
<tr>
<td>1.4. RIGID BODY MODELING OF THE ELBOW</td>
<td>7</td>
</tr>
<tr>
<td>1.5. OBJECTIVES</td>
<td>8</td>
</tr>
<tr>
<td>2. ELBOW ANATOMY &amp; PHYSIOLOGY</td>
<td>9</td>
</tr>
<tr>
<td>2.1. BONY ANATOMY</td>
<td>9</td>
</tr>
<tr>
<td>2.2. SOFT-TISSUE ANATOMY</td>
<td>14</td>
</tr>
<tr>
<td>2.2.1. Capsuloligamentous Constraints</td>
<td>14</td>
</tr>
<tr>
<td>2.2.2. Distal Ligament Constraints</td>
<td>17</td>
</tr>
<tr>
<td>2.2.3. Muscle Constraints</td>
<td>20</td>
</tr>
<tr>
<td>2.3. JOINT DEFINITIONS</td>
<td>22</td>
</tr>
<tr>
<td>3. THREE-DIMENSIONAL BODY ACQUISITION</td>
<td>27</td>
</tr>
<tr>
<td>3.1. OVERVIEW</td>
<td>27</td>
</tr>
<tr>
<td>3.2. COMPUTED TOPOGRAPHY (CT)</td>
<td>27</td>
</tr>
<tr>
<td>Section</td>
<td>Title</td>
</tr>
<tr>
<td>---------</td>
<td>-------</td>
</tr>
<tr>
<td>3.3</td>
<td>MASK CREATION AND THRESHOLDING</td>
</tr>
<tr>
<td>3.4</td>
<td>PREPROCESSING</td>
</tr>
<tr>
<td>3.4.1</td>
<td>Cropping &amp; Boolean operations</td>
</tr>
<tr>
<td>3.4.2</td>
<td>Morphology Operations</td>
</tr>
<tr>
<td>3.4.3</td>
<td>Boundary Detection &amp; Slice Interpolation</td>
</tr>
<tr>
<td>3.5</td>
<td>STEREOLITHOGRAPHY (STL) FILES</td>
</tr>
<tr>
<td>3.5.1</td>
<td>Smoothing</td>
</tr>
<tr>
<td>3.5.2</td>
<td>Triangle Reduction</td>
</tr>
<tr>
<td>3.6</td>
<td>STL REGISTRATION</td>
</tr>
<tr>
<td>3.7</td>
<td>CHARACTERIZING THE MODEL</td>
</tr>
<tr>
<td>4</td>
<td>THREE-DIMENSIONAL KINEMATIC SIMULATIONS</td>
</tr>
<tr>
<td>4.1</td>
<td>OVERVIEW</td>
</tr>
<tr>
<td>4.2</td>
<td>ORIGINS AND INSERTIONS</td>
</tr>
<tr>
<td>4.3</td>
<td>JOINT APPROXIMATIONS</td>
</tr>
<tr>
<td>4.3.1</td>
<td>Flexion/Extension</td>
</tr>
<tr>
<td>4.3.2</td>
<td>Pronation/Supination</td>
</tr>
<tr>
<td>4.4</td>
<td>COSMOSMOTION</td>
</tr>
<tr>
<td>4.4.1</td>
<td>Solver Parameters</td>
</tr>
<tr>
<td>4.4.2</td>
<td>Contact parameters</td>
</tr>
<tr>
<td>4.5</td>
<td>MUSCLE FORCES</td>
</tr>
<tr>
<td>4.6</td>
<td>LIGAMENT CONSTRAINTS</td>
</tr>
<tr>
<td>4.6.1</td>
<td>Ligament Arrangement</td>
</tr>
</tbody>
</table>
List of Tables

Table 4.5-1: Muscle force magnitudes and lines of action..................................................66
Table 4.6-1: Ligament mechanical properties.....................................................................72
# List of Figures

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1-1</td>
<td>Bony anatomy of the distal humerus</td>
<td>11</td>
</tr>
<tr>
<td>2.1-2</td>
<td>Bony anatomy of the proximal radius and ulna</td>
<td>13</td>
</tr>
<tr>
<td>2.2-1</td>
<td>Medial and lateral ligament complexes of the elbow</td>
<td>16</td>
</tr>
<tr>
<td>2.2-2</td>
<td>Sagittal view of the interosseous membrane (IOM) of a forearm</td>
<td>19</td>
</tr>
<tr>
<td>2.2-3</td>
<td>Musculature of the upper arm</td>
<td>20</td>
</tr>
<tr>
<td>2.3-1</td>
<td>Anterior view of the EJC with overlaid average SDA</td>
<td>25</td>
</tr>
<tr>
<td>3.2-1</td>
<td>Isometric view of a 3-D reconstruction of the cadaveric arm</td>
<td>28</td>
</tr>
<tr>
<td>3.3-1</td>
<td>Stack of axial DICOM images arranged within MIMICS</td>
<td>30</td>
</tr>
<tr>
<td>3.4-1</td>
<td>Axial mask slice showing noise outside the distal ulnohumeral joint</td>
<td>33</td>
</tr>
<tr>
<td>3.4-2</td>
<td>Bony mask of the radial head before and after mask manipulation</td>
<td>35</td>
</tr>
<tr>
<td>3.4-3</td>
<td>Axial slice of the proximal ulna showing polylines</td>
<td>36</td>
</tr>
<tr>
<td>3.5-1</td>
<td>MIMICS Remesher interface</td>
<td>38</td>
</tr>
<tr>
<td>3.5-2</td>
<td>An STL triangulation of the distal humerus</td>
<td>40</td>
</tr>
<tr>
<td>3.6-1</td>
<td>The UH joint shown before and after STL registration</td>
<td>44</td>
</tr>
<tr>
<td>3.6-2</td>
<td>Hybrid elbow model showing low and high resolution models</td>
<td>45</td>
</tr>
<tr>
<td>3.7-1</td>
<td>Sphere fits of the humeral head and radial head</td>
<td>46</td>
</tr>
<tr>
<td>4.3-1</td>
<td>Oblique view of the distal humerus with overlaid UH SDA</td>
<td>56</td>
</tr>
<tr>
<td>4.3-2</td>
<td>Medial view of UH in both the 30° and 90° positions</td>
<td>57</td>
</tr>
<tr>
<td>4.3-3</td>
<td>Anterior view of the forearm showing pro/supination ROM</td>
<td>58</td>
</tr>
<tr>
<td>4.4-1</td>
<td>COSMOSMotion interface showing RBM solver parameters</td>
<td>62</td>
</tr>
</tbody>
</table>
Figure 4.4-2: COSMOSMotion diagram showing the penalty regularization force........64
Figure 4.4-3: COSMOSMotion interface showing 3-D contact parameters...............65
Figure 4.6-1: Sagittal view of the ligaments of the LCLC and MCLC of the model ........68
Figure 4.6-2: Sagittal view of the ligaments of the forearm..................................70
Figure 5.2-1: Isometric views of the cadaveric and the model setups ......................80
Figure 5.2-2: Materials testing equipment used in the cadaveric and model setups.........81
Figure 5.2-3: View showing the humerus and global coordinate systems..................85
Figure 5.2-4: Anterior aspect of the proximal ulna with variable CP resections...........87
Figure 5.3-1: View of the distal humerus represented by STL vs. IGES files...............89
Figure 5.4-1: Typical Load vs. Time plot for the model during varus excursion (Hull) ....94
Figure 5.4-2: Constraining load at 1cm varus vs. CP resection at 30° flexion............95
Figure 5.4-3: Constraining load at 1cm varus vs. CP resection at 45° flexion.............95
Figure 5.4-4: Constraining load at 1cm varus vs. CP resection at 60° flexion............96
Figure 5.4-5: Constraining load at 1cm varus vs. CP resection at 90° flexion.............96
Figure 5.4-6: Constraining load at 1cm varus vs. CP resection at 120° flexion.........97
Figure 5.4-7: Constraining load at 1cm varus across flexion angles vs. CP resection.....99
Figure 5.4-8: UH contact force vs. CP resection at 0cm varus.................................101
Figure 5.4-9: RC contact force vs. CP resection at 0cm varus ................................101
Figure 5.4-10: Relative percentages of EJC contact force vs. CP resection ..........103
Figure 5.4-11: Magnitude of forearm ligament tension vs. flexion angle .................105
Figure 5.4-12: Tension developed in the IOM-CB/AB vs. CP resection ..................106
Figure 5.4-13: Tension developed in the IOM-DOB vs. CP resection .....................107
Figure 6.2-1: 3-D recreation of the Biomet ExploR radial head implant
Figure 6.2-2: Lateral view of the LUCL-r insertion on the UH joint in 90° flexion
Figure 6.2-3: Proximal radius showing radial neck axis and RH implant alignment
Figure 6.2-4: Initial MTA load in the SDA position vs. the pre-valgus position
Figure 6.3-1: Typical Load vs. Time plot for the model during varus excursion (Fern)
Figure 6.3-2: Initial constraining load at 0cm varus for 30° and intact CP process
Figure 6.3-3: Initial constraining load at 0cm varus for 90° and intact CP process
Figure 6.3-4: The change in varus load of the model at 30° vs. % CP resection
Figure 6.3-5: The change in varus load of the model at 90° vs. % CP resection
Figure 6.3-6: Constraining load at 1.5cm varus for the LrRr repair
Figure 6.3-7: Constraining load at 1.5cm varus for the LrRx repair
Figure 6.3-8: Constraining load at 1.5cm varus for the LxRr repair
Figure 6.3-9: Constraining load at 1.5cm varus for the LxRx repair
Figure 6.3-10: Lateral view of the LxRxC100% injury state in 30° flexion
Figure 6.3-11: Change in model load across repair groups and CP resections
Figure 6.3-12: The relative force contribution of the UH and RC articulations at 30°
Figure 6.3-13: The relative force contribution of the UH and RC articulations at 90°
Figure 6.3-14: LUCL-r tension vs. CP resection in the LrRx and LrRr states
Figure 6.4-1: Posterolateral view of the LrRr injury state in 90° of flexion
The use of computational modeling is an effective and inexpensive way to predict the response of complex systems to various perturbations. However, not until the early 1990s had this technology been used to predict the behavior of physiological systems, specifically the human skeletal system. To that end, a computational model of the human elbow joint was developed using computed topography (CT) scans of cadaveric donor tissue, as well as the commercially available software package SolidWorks™. The kinematic function of the joint model was then defined through 3D reconstructions of the
osteoarticular surfaces and various soft-tissue constraints. The model was validated against cadaveric experiments performed by Hull et al and Fern et al that measured the significance of coronoid process fractures, lateral ulnar collateral ligament ruptures, and radial head resection in elbow joint resistance to varus displacement of the forearm. Kinematic simulations showed that the computational model was able to mimic the physiological movements of the joint throughout various ranges of motion including flexion/extension and pronation/supination. Quantitatively, the model was able to accurately reproduce the trends, as well as the magnitudes, of varus resistance observed in the cadaveric specimens. Additionally, magnitudes of ligament tension and joint contact force predicted by the model were able to further elucidate the complex soft-tissue and osseous contributions to varus elbow stability.
1. INTRODUCTION

1.1. OVERVIEW OF COMPUTATIONAL MODELING

Normal function of the musculoskeletal system requires the interplay of numerous physiological entities. Chief among these are the three-dimensional articular surface contacts and soft-tissue constraints. Computational modeling has become an increasingly prevalent tool for characterizing this complex system. Subjects of interest within such models include muscular forces, ligament tensions, soft-tissue in situ strains, articular contact forces, and relative spatial positions of the bones in a given joint. Once validated against real-world experiments, these models can be used to investigate new surgical techniques, uncommon or difficult to repair pathologies and traumas, as well as to elucidate joint parameters not easily measured experimentally. Furthermore, these models allow for quantitative study of vast numbers of physiologic parameters simultaneously, without the need for large numbers of samples or trials, and at a fraction of the cost of experimental models.

Traditionally, computational modeling has manifested itself in two forms; continuum based finite element analysis (FEA) or multipart rigid body modeling (RBM). Though recently, some effort has been made to create models that incorporate both methodologies, significant limitations remain. [24, 29, 56, 61] Therefore, a majority of
investigators still choose one technique over the other based on the nature of the information they desire.

FEA is an extremely valuable tool for investigating the stress and strain distributions of a body under a set of static loading conditions. Under this approach, the body in question is divided into ever smaller conjoined polygonal elements, each of which is comprised of at least three nodes or vertices. The experimentally derived material properties are then assigned at each node, and boundary conditions set. In the study of biomechanics, FEA algorithms have been used to predict the stress maxima and distribution of musculoskeletal joints under a variety of loading scenarios. [24, 33] Additionally, as it is widely accepted that the site of interface in prosthetic implants is the most apt to fail, many studies have been conducted to evaluate the feasibility, longevity, and physiologic consequences of various orthopaedic implant designs. [16, 24, 27, 28, 51, 95, 107, 121, 126] However, due to the complexity of FEA models and the many inherent unknowns, the processing time required to solve very fine meshes can be prohibitive. For this reason, their application has been limited in areas where the model bodies are not static, as this requires that the model be solved over the time domain. Such is the challenge when attempting to use FEA in the study of kinematics and dynamic biomechanical function.
1.2. RIGID BODY MODEL SIMULATION

RBM has proved itself an effective strategy for studying the forces imparted by bodies in motion, and offers many advantages over finite element analysis alone. The rigid body modeling approach assumes that each body is inelastic and incompressible, which by definition precludes the possibility of deformation in the model. Such assertions are appropriate when studying materials with high stiffnesses, and observable stresses that would result in <<1% total strain. [40] Under these circumstances, strains are considered infinitesimal, and therefore, the deformed case is identical to the undeformed. Additionally, the algorithms used in RBM are able to resolve a model in the time domain much faster than those used in FEA. [61]

Recently, RBM has become more prevalent in the study of musculoskeletal kinematics and joint function. [22, 30, 36, 37, 42, 43, 46, 47, 51, 62-65, 67, 68, 85, 87, 98, 107, 111] While many models in the literature use computed topography (CT), laser, or magnetic resonance imaging (MRI), to ensure the accurate reproduction of the articular surfaces, most also constrain joint movement to a small number of degrees of freedom (DOF). For example, the elbow and knee joints are often approximated as simple hinges, with only a single rotational DOF. While this approach can still yield valuable information concerning approximate ranges of motion, joint moments, and muscle forces when solving the driven, or inverse, scenario, the omission of geometry defined movements limits the models’ usefulness in predicting joint function under non-idealized motions. The alternative to idealized mechanical joints is to allow totally unconstrained motion with no predefined limits on degrees of freedom, wherein motion is defined solely by contact
between the articular surfaces and representative soft-tissue constraints. Though less common, this approach allows for the more realistic and experimentally relevant prediction of joint mechanical function based on the orientation and magnitude of soft-tissue forces as well as external perturbations. [37, 56, 61, 67, 124]

Wismans et al and Hirokawa et al were among the first to model a musculoskeletal joint whose motion was dictated by articular surface geometry without idealized restrictions on joint DOF. In these biomechanical knee studies, force vectors spanning the joint were used to represented ligament and muscle soft tissues while reconstructed articular surfaces dictated motion. Bony contacts were assumed to be rigid and constant, and cartilaginous soft tissues were not considered. [52, 124] Kwak et al expanded on these concepts by allowing overlap between bodies while still mandating that body rigidity was preserved. This overlap allowed for the prediction of contact area and contact force to be expressed as a function of overlapped volume and contact stiffness. Furthermore, this methodology did not require continuous surface contact as had been stipulated previously. Instead, a vector normal to any portion of the surface provided a line equation that could be used to extrapolate distance, contact, and overlap between any two bodies. [61] Numerous models have followed that use this methodology for deducing joint contact, either implemented through custom coding or in bundled software packages.
1.3. COMPUTATIONAL SOFTWARE PACKAGES

In addition to custom in-house coding, there are several rigid body modeling
software packages that are utilized in the investigation of kinematic behavior of
musculoskeletal joints. Chief among these are SIMM (Software for Interactive
Musculoskeletal Modeling, Musculographics Inc., USA), VIMS (Virtual Interactive
Musculoskeletal System, BioMed Central Ltd., USA), MSC ADAMS (Automatic
Dynamic Analysis of Mechanical Systems, MSC Software Corp, USA), and more recently
AnyBody (AnyBody Technology A/S, Denmark). While each of these is capable of
solving the constitutive equations required in RBM, SIMM, VIMS and AnyBody are either
significantly limited or unavailable in a commercial form. [25, 26] Furthermore, the
SIMM and AnyBody software packages do not allow fully dynamic joint motion in their
native states. Instead, function is idealized as simple mechanical joints with few DOFs,
thereby limiting the predictive power of the model when attempting to solve the forward
dynamics case. [11, 25] VIMS, by contrast, does support full geometry based forward
dynamics simulation; however, this functionality requires additional proprietary coding
that is not commercially available at this time. [26]

The ADAMS solver offers several advantages over alternative modeling packages.
In terms of joint constraints, the ADAMS package does not require that the motion
between two bodies be prescribed by any sort of idealized mechanism. Instead, the RBM
algorithms instead utilize user-defined contact parameters to dictate the geometric
interferences between two parts. Similar to the architecture described by Kwak et al,
ADAMS is able to track the volumetric overlap of two rigid bodies and then express the
contact force and orientation as a function of the material stiffness and degree of penetration. [1, 3, 61] This distinction enhances the predictive power of the force driven or forward dynamics scenario by allowing model structure to dictate function without introducing any motion limiting artifacts that could be caused by improperly restraining model bodies. [61, 66]

A second advantage of the ADAMS solver is its wide commercial application and relative availability. Since its introduction in 1977, ADAMS has become prevalent in many areas of engineering modeling, including aerospace, automotive and machine design. [66] The package is versatile and can accommodate many different engineering formats and offers well documented support literature and an extended support community. It is capable of importing and manipulating many standard computer-aided-designs (CAD) drawing files, such as Parasolid, IGES, DXF, and STL, which consequently, negates the need for specialty or obscure modeling packages to create the initial rigid body solids. Likewise, the ADAMS package supports many standardized output formats for animation, image creation or FEA boundary condition export. Finally, because the ADAMS solver is available as a native application programming interface (API) within other commercially available CAD programs, such as part of the COSMOSMotion add-in (Structural Research and Analysis Corp, Santa Monica, CA) to the Solidworks design software (SolidWorks Corp, Concord, MA), solid body creation and rigid body simulation can be performed within a single software architecture. This simplification allows for quick results implementation and design alteration without the need for clumsy conversion, compression, or import/export functions. [65, 66]
1.4. RIGID BODY MODELING OF THE ELBOW

Computational models of the elbow have been used extensively to study the biomechanical function of the joint. [22, 42, 43, 46, 47, 53, 63, 85, 111] Predominately these studies have focused on the inverse dynamics scenario, wherein a particular flexion/extension or pronation/supination angle is dictated and the requisite soft-tissue stabilizing force magnitude, orientation, or moment arm is extrapolated. However, to date only the work by Fisk et al has allowed osteoarticular surface contact to dictate joint function. [37] By contrast, every other study has imposed small DOF movements about fixed joint centers and neglected the influence of ligaments on kinematic function. Such approximations do not permit the accurate study of many relevant joint parameters including joint contact area and contact force, ligament function, such as in situ strain, and investigations of normal physiologic rotations and translations outside of flexion/extension, such as varus/valgus tilt and ulnohumeral joint laxity. Therefore, while many elements of elbow function are well understood, there is currently a paucity of data that accurately reflects the biomechanical function of the native joint. Consequently, until these omissions are addressed, the robustness of computational elbow models will be insufficient for implementation in broader clinical applications.
1.5. OBJECTIVES

It is the objective of this thesis to develop and validate a computational model of the elbow for the purposes of elucidating biomechanical function of the in vivo joint. Joint kinematics will be defined by osteoarticular and soft-tissue constraints derived from computed topography of native bony surface geometry as well as reported physiological and cadaveric parameters. Creation of the solid model will be facilitated through the use of the commercially available CAD program Solidworks and rigid body model simulation will be implemented through the MSC ADAMS modeling package native to the COSMOSMotion add-in. Motion will not be limited by or prescribed according to idealized small DOF joints. Finally, the model will be validated against two cadaveric studies concerning the varus stability of the intact and disrupted elbow. [34, 55] This work expands on the computational elbow model developed by J.P. Fisk and J.S. Wayne, 2009 wherein cadaveric CT scans were used to create a 3-D surface model of the elbow for the purposes of investigating range of motion and osteoarticular stability of the elbow joint. [37]
2. ELBOW ANATOMY & PHYSIOLOGY

2.1. BONY ANATOMY

The elbow is a synovial joint located approximately at the midpoint of the arm, and consists of the three long bones of the upper extremity, the humerus, radius, and ulna. The humerus is positioned superior to the radius and ulna, which each lay roughly parallel to one another just distal to the humerus. The elbow is considered diarthrodial in its range of motion and works in concert with the shoulder and wrist to position the hand in three-dimensional space. Due to the extremely precise fit between the concavities of the proximal ulna and radius and the convexities of the distal humerus, the elbow has been described as one of the most congruous articulations in the body. [78] Consequently, a complete understanding of the osteology of each of the three bones is paramount in attempting to describe the overall function of the elbow joint complex (EJC).

The proximal humerus is situated lateral to the scapula and articulates with the thorax at the glenohumeral joint of the shoulder, the most rostral aspect of the bone is roughly spherical at the humeral head. This structure then transitions from roughly spherical to cylindrical via the surgical neck of the humerus which emerges caudal and slightly lateral to the center of the humeral head. Continuing distal beyond the surgical neck, the bone reshapes into the cylindrical diaphysis, a segment which accounts for the
majority of the length of the upper arm as well as the attachment site for most of the upper arm’s musculature. CT and MRI scans of this structure are characterized by a nearly circular outline of the highly attenuated thick cortical shell encasing the medullar canal of the bone. Beyond the diaphysis and beginning just proximal to the supracondylar ridges, the distal humerus becomes pyramidal in appearance by splaying in the mediolateral plane while retreating on the posterior aspect. The terminant of each of the medial and lateral aspects of the splay are designated as the medial and lateral epicondyles, while the void created on the posterior aspect is referred to as the olecranon fossa. [88]

The most distal portions of the humerus have an approximately 30° anterior tilt relative to the long axis and terminate in the two major articular surfaces of the elbow joint, shown in Figure 2.1-1. [78] The larger of these two structures is the pulley-shaped trochlea and it constitutes the bulk of the ulnohumeral articular surface. The structure is positioned on the medial half of the distal humerus such that the central axis of the feature is oriented approximately mediolateral. Anteriorly, the structure emerges from the coronoid fossa and presents roughly 300° of hyaline covered articular surface before becoming continuous with the olecranon fossa on the posterior aspect. [8, 78] The larger medial rim and smaller lateral rim of the trochlea create a border around the central trochlear sulcus which wraps helically from the posterior fossa to the anterior fossa. On the lateral half of the distal humerus, the smaller half-spheroid capitulum is responsible for all of the radiohumeral articulation in the elbow. The structure becomes distinct just distal to the radial fossa and sweeps anteriorly and inferiorly before terminating at the trochlear tubercle on the inferior aspect of the humerus and becoming continuous with the lateral
trochleocapitellar groove adjacent to the medial ridge of the trochlea on the medial and inferior surface of the humerus. The capitulum does not present as much articular surface as the trochlea; however, it is covered in the same hyaline cartilage for articulation with the radial head.

The proximal ulna articulates with the medial aspect of the distal humerus and is the largest articulating surface in the elbow. (Figure 2.1-2) Beginning at the most superior portion of the ulna, the olecranon protrudes anteriorly and arcs slightly inferior creating a beak-like structure that defines the superior limit of the trochlear notch. Distal to the olecranon, the proximal anterior surface of the ulna presents two deeply inset parallel concave grooves which begin from either side of the anterior most point of the olecranon and sweep roughly 190° distally to the medial and lateral sides of the tip of the coronoid process. A portion of fatty tissue running transverse to the grooves divides the sigmoid into its anterior coronoid portion and the posterior olecranon portion. [78] Of these two

Figure 2.1-1: Bony anatomy of the distal humerus. [8]
grooves, the medial is deeper, while the lateral groove is shallower and merges with the approximately 70° ellipsoidal concavity of the radial notch on the anterolateral aspect of the bone. [8, 78] Collectively, these indentations form the characteristic “C”-shaped greater sigmoid notch of the ulna whose orientation approximately 30° posterior to the diaphysis of the bone complements the 30° anterior tilt of the distal humerus, providing an articulating surface to receive the medial and lateral rims of the trochlea with the raised guiding central ridge of the sigmoid riding in the trochlear sulcus of the humerus. [78] Both the greater sigmoid notch and the radial notch are lined in hyaline cartilage.

The radius is the third and final bone to have articulations in the elbow joint. (Figure 2.1-2) On the proximal end of the bone, the radial head resembles a broad ellipsoidal cylinder perched on the more slender radial neck, with the center line of the two forming an angle of 12-17° with the diaphysis of the bone.[78, 116] The major axis of the ellipsoidal radial head runs posteromedially to anterodistally with the forearm in neutral rotation resulting in a radial head that is torqued about the radial neck by approximately 55° relative to the distal portions of the bone. [116] In this orientation the long arc of the medial side of the radial head articulates with the radial notch of the proximal ulna allowing the rotation necessary for forearm pronation/supination, but limiting axial translation. [116] Additionally, the most superior end of the bone is capped with a shallow depression of approximately 2.4mm that abuts the capitulum of the humerus allowing rotation and pivoting, but limiting axial translation. [59] Hyaline cartilage covers the entirety of this radiocapitular articular surface as well as the posteromedial 240° of the circumference of the radioulnar articulating margin. [78]
Though not technically part of the elbow, the distal radioulnar articulations are important in understanding proximal radioulnar positions. The diaphyses of the radius and ulna emerge distal to the radial and ulnar tubercle, respectively, and extend roughly parallel to one another down the length of the fully supinated forearm. Both bones are triangular in cross section with each having an anterior, posterior, and interosseous border clearly demarcated along the length of the diaphysis. The interosseous borders of each bone are also parallel in full supination and provide the attachment site for the interosseous membrane of the forearm. Furthermore, the interosseous border and membrane serve to demarcate the anterior surfaces of the forearm long bones from the posterior surfaces for the purpose of Pronator/Supinator muscle attachments.
2.2. SOFT-TISSUE ANATOMY

Normal biomechanical function of the intact elbow requires the presence of significant soft-tissue constraints, working in concert with the osteoarticular congruities within the joint. Morphologically, these tissues can be separated into the passively stabilizing fibrous capsuloligamentous constraints and much bulkier active muscular constraints.

2.2.1. Capsuloligamentous Constraints

The entire elbow joint complex is enveloped in a thin fibrous capsule that inserts proximal to the radial and coronoid fossae and just distal to the coronoid and radial articular margins on the anterior surfaces of the elbow (see Figure 2.2-1). On the posterior aspect of the joint, the capsule attaches to the margins of the olecranon fossa and the superoposterior surface of the proximal ulna. Medial and lateral limits of the capsule trace the articular margins of the trochlea-sigmoid notch and the radiocapitular articular margins, respectively. The anterior and posterior portions of the capsule are generally thin mechanically weak structures, while the medial and lateral aspects have substantial thickenings that form the collateral ligament complexes. [78]

The medial collateral ligament complex (MCL), shown in Figure 2.2-1, has three distinctly oriented portions termed the anterior, posterior, and transverse bundles. Despite its relative prominence on the medial aspect of the joint, the transverse bundle (ligament of Cooper) does not have an observed kinematic function. [14, 78] The anterior portion is the
most discrete of the three ligaments; indeed, there is debate as to whether it in fact represents an extracapsular constraint. [14, 78, 79] The structure has a broad origin on the anteroinferior surface of the medial epicondyles, distinct from the medial lip of the trochlea, and inserts along the anteromedial border of the coronoid process at the sublime tubercle. [8] Most researchers maintain that the primary function of the anterior bundle is to maintain elbow stability in extension, asserting that as the flexion angle increases, its fibers become lax. [8, 14, 78-80, 83] However, Fuss et al has showed that a portion of the bundle maintains taut throughout the entire range of motion, terming this the “guiding bundle.” [41] Additionally, the anterior bundle has been shown to impart significant elbow stability under valgus displacements. [78]

The posterior bundle shares its origin with that of the anterior portion, but is significantly broader, as it inserts across the proximal half of the trochlear notch’s medial margin. The bundle is generally lax at flexion angles less than 90° and develops tension in a sequential anterior to posterior progression with further increasing flexion. [41, 99] Similar to the anterior bundle, the posterior bundle has been shown to significantly resist valgus displacements of the forearm. [78]
Figure 2.2-1: Lateral ligament complex and capsule (left), medial ligament complex (right). [76]

The lateral collateral ligament complex (LCL) is generally less defined in its emergence from the joint capsule. The LCL has four distinct bundles, the radial collateral ligament (RCL), the lateral ulnar collateral ligament (LUCL), the non-collateral annular ligament, and the intermittent accessory collateral ligament. [72, 78] The annular ligament is a strong fibrous band of radially arranged tissue that originates on the anterior margin of the radial notch, encircles the articular margin of the radial head, and inserts back on the ulna at the posterior margin of the radial notch. This semi-circular band “contributes about four-fifths of the fibro-osseous ring” that keeps the articulating surfaces of the radial head seated in the radial notch. [78] Recently, Bozkurt et al identified two obliquely oriented bands inserting at the superior and inferior limits of the annular ligament that presumably further anchor the radial head. [20] When present, Martin defined the accessory lateral...
collateral ligament as those fibers emerging distinct from the anterolateral portion of the annular ligament with discrete insertions on the supinator crest of the ulna. [72] The RCL originates from the lateral epicondyle and spreads in a fan-like manner across the radiocapitular joint and inserts indistinguishably on the annular ligament. [78] In contrast to the ligaments of the MCL, the RCL is taut throughout the entire range of flexion/extension suggesting that it lies on, or very near, the axis of rotation of the joint. The LUCL also originates from the lateral epicondyle, but inserts distal to the annular ligament on the supinator crest of the ulna. The primary function of the ligament is widely accepted to be varus stabilization of the lateral joint; however, some have noted that its blending with the anterolateral aspect of the annular ligament likely also acts as a “posterior buttress” to the radial head, thereby preventing subluxation. [76, 78, 79, 99] The quadrate ligament and oblique cord represent two extracapsular constraints of questionable kinematic importance. The quadrate originates just distal to the margin of the radial notch and extends anterodistally to its insertion just below the articular margin of the radial head. Spinner and Kaplan demonstrated that the anterior fibers of the ligament provide supplemental stabilization to the radioulnar joint during full supination, while the less substantial posterior fibers supported the joint in full pronation. [108]

2.2.2. Distal Ligament Constraints

Though not technically part of the elbow joint complex, a number of fibrous constraints in the forearm and at the distal radioulnar joint have important stabilizing
actions that influence overall elbow function. Originating just distal to the joint capsule, the oblique cord is a thin fibrous band running from the distolateral edge of the coronoid process to the distal medial aspect of the radial tubercle. [73, 108, 113] Though the structure has been suggested as a proximal radial stabilizer in full supination, its inconsistent presence and sparse fiber arrangement has led most to conclude that its physiological significance is limited. [78]

The interosseous membrane (IOM), shown in Figure 2.2-2, is a fibrous membrane distal to the oblique cord that runs obliquely from the interosseous border of the radius distomedially to insertions along the interosseous border of the ulna. [106] Along the proximal quarter of the IOM, the central band (CB) emerges as a distinct thickening of the membrane approximately mid-diaphysis of the forearm bones. The average width of this band is approximately 1.1cm measured transverse the fiber alignment and oriented at an average angle of 21° with the ulnar diaphysis [54] Additionally, there are a variable number of accessory bands (AB) present just distal to the CB that blend into the membranous tissue extending to the DRUJ and with orientations similar to those of the CB. Distal to the ABs, the distal oblique bundle (DOB) is a variably present isometric stabilizer of the DRUJ, similar in thickness to the CB but much less wide. Its fibers originate from the distal 1/6th of the ulnar diaphysis and run distally and obliquely to their insertion on the inferior margin of the sigmoid notch of the distal radius. [77, 89, 101] Several distinct portions of the IOM, most importantly the CB, have been demonstrated to have functional significance, chiefly in limiting proximal axial migration of the radius as well as providing a stabilizing force to the distal radioulnar joint (DRUJ) during full
supination. [75, 77, 89, 94, 106] Additionally, the DOB is thought to provide secondary rotary support to the DRUJ as well as limit distal axial translation of the radius relative to the ulna in the approximately 40% of the population in which is observed. [58, 77, 89, 120]

Figure 2.2-2: Sagittal view of the interosseous membrane (IOM) of a forearm in neutral rotation. [89]

The triangular fibrocartilage complex (TFCC) is a complex arrangement of fibrous connective tissue and meniscus-like cartilaginous soft-tissue that anchor the DRUJ to the radiocarpal and ulnocarpal joints of the wrist. [122] Integral to this complex are the distal radioulnar ligaments (DRULs) that emerge as distinct longitudinally arrayed fibers about the circumference of the fibrocartilage disc. The fan-shaped ligament has two functional bands on the dorsal and palmer aspects of the complex that span the DRUJ. The dorsal ligament originates at the dorsal rim of the radius and runs anteriorly and obliquely to insert at the ulnar fovea at the margin of the styloid process. The palmer ligament attaches at the palmer rim of the distal radius and runs dorsally and obliquely to a common insertion with the dorsal radioulnar ligament. [15, 86] In concert with the meniscus
homologue of the TFCC these structures anchor the DRUJ, providing rotary support to the joint as the radius pivots about the ulnar during pronation/supination. [86, 102, 104]

2.2.3. Muscle Constraints

There are four groups of muscles that cross the elbow joint: the anterior bicep group, the posterior tricep group, the lateral extensor-supinator group, and the medial flexor-pronator group. [8] Of these, Morrey and An identified the biceps brachii, brachialis of the anterior bicep group, and the triceps brachii of the posterior tricep group as among the largest active stabilizers of the joint. (Figure 2.2-3) [9, 78, 80]

Figure 2.2-3: Musculature of the upper arm (left), stabilizing component of the triceps (TR), brachialis (BR), and biceps (BC) (right). [5, 78]
The biceps brachii is an elbow flexor with a large cross-sectional area. The muscle has two distinct origins, the long head at the supraglenoid tubercle of the scapula, and the short head at the coracoid process of the scapula. The line of action of the muscle runs along the anterior surface of the humerus from the bicipital groove (long head) and medial aspect of the glenoid distally toward the major insertion on the anterior aspect of the radial tubercle. In addition to its osseous insertions, the bicipital aponeurosis extends distomedially, wrapping around median nerve and brachial artery to insert onto the deep fasciae of the forearm. [78, 88] Though large in its cross-sectional area, the relatively close proximity of the muscle insertion with the flexion/extension axis of the elbow gives the muscle diminished mechanical advantage in flexion. Additionally, the muscle is a powerful supinator when the forearm is in the pronated position. [9, 78]

The brachialis muscle is another large muscle of the anterior bicep group with the largest cross-sectional area of any flexor. The muscle originates along the anterior surface of the humerus distal to the deltoid and coracobrachialis insertions and terminates onto the distal coronoid process and ulnar tuberosity. Though very large, the brachialis also has a line of action with poor mechanical advantage. [9, 78, 88]

The triceps brachii is a broad muscle with three distinct origins occupying the entire posterior musculature of the arm and is the predominate extensor of the forearm. The most proximal of the three bundles is the long head which originates on the infraglenoid tuberosity of the scapula, followed distally by the lateral head originating from the proximal lateral intramuscular septum of the humerus, and further distally by the medial head originating along a wide surface on the distal half of the posteromedial aspect
of the humerus. Approximately mid-substance, the three bands become continuous and extend distally to their insertion along the entire proximal tip of the olecranon via the triceps tendon. [78, 88]

2.3. JOINT DEFINITIONS

The articulations between the humerus, radius, and ulna create four discrete joints and two distinct rotational movements in the upper extremity. Within the elbow itself, the highly congruous ulnohumeral articulation is responsible for forearm flexion/extension and accounts for the majority of articular contact surface area within the joint. The radiohumeral articulation, which shares the same rotational axis as the ulnohumeral articulation, acts as a secondary stabilizer to this flexion/extension moment as well as provides the majority of axial compression resistance to the joint. [50, 70, 78] The proximal and distal radioulnar articulations allow for the pivoting pronation/supination movement of the forearm about an axis oblique to the diaphyses of both the radius and ulna.

Traditionally classified as a hinge (ginglymus) joint with a single rotational degree of freedom, the predominant motion allowed by the ulnohumeral articulation is flexion/extension of the forearm. When viewing the upper extremity hanging in a neutral position from the sagittal aspect, this motion would have the wrist scribing the arc of a semi-circle anterior to the center line of the body, with the elbow at the approximate center. Average range of motion for this joint is approximately 150° from full extension,
with the forearm roughly parallel to the humerus, to full flexion. [6, 49, 78] The limiting factor in passive extension of the elbow is the impact of the anterior tip of the olecranon within the olecranon fossa, though active extension is likely constrained by anterior capsuloligamentous tautness preceding such an abutment. Conversely, passive flexion is limited by contact between the radial head and radial fossa and/or contact between the coronoid process and coronoid fossa; the relative contributions of which are dependent upon the degree of pronation/supination of the forearm. Again, active flexion is constrained by soft-tissue tension prior to osseous contact. [78]

Qualitatively, the axis of this rotation has been described as a line passing through the medial and lateral epicondyles. [19] However, though useful in predicting movements of the forearm and the gross position of the wrist and hand, this definition does not allow for any out of plane rotation or translation of the ulna and by definition precludes the possibility of a non-static, instantaneous axis that is dependent on joint angle. Furthermore, the single DOF joint approximation is inconsistent with the flexion angle dependent alignment of the ulna, which rotates from relative eversion (valgus) in full extension, to inversion (varus) in full flexion. [10, 69] Clinically, insufficiently precise approximations of the flexion/extension axis have been cited as a likely contributor to loosening failures of total joint arthroplasties and external fixators of the elbow. [17-19, 69, 78, 83, 84, 110] Consequently, when attempting to characterize, and especially when attempting to quantitatively predict elbow motion, a more precise and dynamic model of elbow joint behavior must be used.
Accurate kinematic measurements of both the active and passive flexion/extension axes are abundant in the literature, with widespread agreement that the axis is not static, but instead is flexion angle dependent. [17-19, 31, 78, 83, 109, 110, 112, 118] Tanaka et al were among the first to use three dimensional electromagnetic tracking to dynamically describe passive elbow flexion and extension. Presented in terms of Eulerian angular notation, they showed that as the elbow was moved through flexion the instantaneous axis of rotation would translate as well as pivot in its projection through the distal humerus. [112] Bottlang et al expanded on this work by quantifying forearm rotation in accordance with the screw displacement algorithms put forth by Beggs wherein a body may rotate about, whilst simultaneously translating along, any given line in space. [13, 17] This augmented methodology confirmed the earlier findings by Tanaka et al concerning the continuously dynamic nature of the flexion/extension axis, but also reported anteroposterior and mediolateral translations of the ulna not previously quantified. As described by Bottlang, “the axode… of the screw displacement axis traces the surface of a double quasi-conic frustum of a nominally elliptical cross section.”
Figure 2.3-1: Anterior view of the EJC with overlaid average SDA orientation (left).

An isometric view of the elliptical frustum formed by the migration of the SDA throughout flexion. [17-19]

This complex movement was simplified by averaging the continuous axis data across the 10-130° flexion range observed, thus yielding an average screw displacement axis (SDA) that was oblique to both the humeral and ulnar long axes and passed through the approximate centers of the capitulum and trochlea in agreement with many in the literature. [17, 19, 78, 83] Though still an approximation, this two DOF model of flexion has been widely accepted as an accurate predictor of elbow flexion/extension, varus/valgus tilt, and joint laxity. This translational and rotational axis is shown in Figure 2.3-1.

The second major rotation of the forearm is pronation/supination. The two articulations which constitute this joint are categorized as pivot (trochoid) joints. In this arrangement, the proximal and distal radioulnar articulations allow axial rotation and
limited axial translation but prohibit any rotations transverse to the axis of rotation. Axial translation is further constrained by the close proximity of the radial head and capitulum. Kinematic data suggests the radius rotates about an axis that originates proximally at the center of the capitulum, coincident on the ulnohumeral flexion/extension axis, and travels obliquely in the mediodistal direction, passing through the center of the radial head and terminating at the center of the distal ulnar fovea. [78, 83, 109, 118, 119]

This orientation causes the axis of rotation to lie approximately orthogonal to the ulnohumeral flexion/extension axis, thus forearm pronation/supination is generally considered to be flexion angle independent. [78, 83, 109, 118] Furthermore, because the forearm axis of rotation is coincident upon points which are fixed relative to the ulna, forearm pronation/supination is defined as exclusive of any ulnar component and containing radial rotation only.
3. THREE-DIMENSIONAL BODY ACQUISITION

3.1. OVERVIEW

The goal of creating an elbow joint model was to accurately predict both soft tissue and bony surface contributions of the humerus, ulna, and radius, to elbow stability throughout the range of normal motion. To accomplish this, the high fidelity acquisition of the 3-D geometry of the elbow joint was paramount. To this end, computed tomography (CT) was selected as the most reasonable option for transcribing the bony geometry of an anatomic specimen into a representative stack of 2-D images from which the third dimensional component could be interpolated. Once the slice data was acquired, it was transformed into solid body information through the meshing and triangulation of the corresponding voxelated information.

3.2. COMPUTED TOPOGRAPHY (CT)

A single fresh frozen right upper extremity, disarticulated at the shoulder, was harvested from a 91 year old female cadaveric specimen. The appendage was free from any obvious morphological or pathological deformities and passive range of motion tests revealed normal flexion/extension and pronation/supination movements. The elbow to
outstretched middle fingertip distance was 428.83mm which placed the specimen in the 25th% of female upper extremities. [4, 48]

The appendage was secured with wire tires and wooden dowels to a custom wooden mounting jig created by Fisk et al (2009). [37] The jig consisted of two plywood swing arms, one for the upper arm and the second for the forearm and hand, affixed by a single nylon bolt at the joint. The appendage was positioned onto the apparatus, shown in Figure 3.2-1, such that the forearm was maintained in approximately neutral forearm rotation and with the axis of the nylon bolt traveling through the approximate elbow flexion/extension axis. This axis was visualized as the line connecting the medial epicondyle (EM) and the lateral epicondyle (EL) through the distal humerus. Angle markings on the base of the apparatus allowed the flexion angle to be adjusted.

Figure 3.2-1: Isometric view of a 3-D reconstruction of the cadaveric arm with the volar aspect facing up, positioned in the custom scanning apparatus.
The specimen and apparatus were then placed into a SOMATOM Sensation 64 helical scanner (Siemens AG, Forchheim, Germany) with the long axis of the humerus positioned roughly normal to the transverse plane of the scanner and three separate CT scans were taken. The first scan captured the entire appendage flexed at approximately 30° with a slice thickness of 2mm. The second scan was taken without any repositioning of the specimen, but focused only on a section of the arm from approximately 10cm proximal to, and 10cm distal to the elbow joint itself. Additionally, this scan utilized a higher resolution 0.4mm slice thickness in order to better capture the osteoarticular surface details of the joint complex. The third and final scan was taken of the elbow joint at 0.4mm resolution at approximately 90° of flexion with the long axis of the humerus approximately 45° to the transverse plane of the scanner.

3.3. MASK CREATION AND THRESHOLDING

The output of the CT scans was in the form of three stacks of transversely arrayed panes, with each pane comprised of a single 512x512 pixel image. Though the orientation and/or slice thickness varied between the three scans, the method of interpolation was the same for each. The commercially available medical imaging software MIMICS (Materialise's Interactive Medical Imaging Control System, Materialise, Ann Arbor MI) was used to transform the arrayed 2-D DICOM (Digital Imaging and Communications in Medicine) images generated by the CT scanner into a cohesive 3-D triangulated body. The general procedure for this transformation was as follows. The MIMICS program was used
to import and sequentially arrange the stacks of DICOM images based on the orientation of the scanner and order of the image capture. Once imported, the sequential panes were aligned such that successive identically positioned 2-D pixels became the top and bottom faces of 3-D voxels, with slice thickness dictating the axial edge size of each enclosed cuboid space. Sets of arranged images sharing a common threshold value were then grouped into “masks” within MIMICS and edited together. (Figure 3.3-1) A binary thresholding operation was then used to assign either an active (bright) or inactive (dark) state to each pixel on each image thereby allowing only regions of interest, e.g. bone, to be selected.

Figure 3.3-1: Stack of axial DICOM images arranged within MIMICS showing a mask of the distal humerus (yellow).

The value of the global threshold was chosen according to the Hounsfield unit (HU) that would allow the most bony material to be detected without including a prohibitive number of soft tissue or noise related voxels. Because the Hounsfield scale is a standardization of
the relative linear attenuation coefficient of a radiographed body compared to that of water, the initial threshold value used was approximately 400, as that is the generally accepted value for cortical bone. However, due to the much lower density and correspondingly lower linear attenuation coefficient value of the trabecular bone found in the epiphysis of the bones, this value was adjusted downward to 225. This adjustment represented a compromise between low HU values that would detect all bony regions while including dramatic noise and soft tissue interference and higher HU values that may not have captured the full expanse of the trabecular bone articular surfaces.

3.4. PREPROCESSING

The majority of musculature and capsuloligamentous material are known to attenuate x-rays below the threshold set and the use of low attenuating materials in the positioning apparatus and support structure helped minimize the interference from non-anatomic sources, e.g. the wooden positioning jig, wooden support dowels, zip ties, etc. However, many images from each scan still contained a large number of pixels that attenuated at or above the 225 HU value selected but did not represent osseous material within the specimen. In general, these false positives were removed though a series of preprocessing operations within the MIMICS package designed to eliminate soft-tissue and apparatus noise while preserving the fidelity of the osseous surfaces.
3.4.1. Cropping & Boolean operations

Following the creation of a whole specimen mask, the first method employed to eliminate soft-tissue and apparatus noise was multiple slice cropping. This procedure utilized the traditional cropping tool and the multiple-slice edit tool in concert to ensure that only those areas within or immediately adjacent to bone were selected. First, the cropping tool was used to deselect those active voxels that were outside of the smallest cuboid containing the entire bony geometry of the arm. Though a very rough approximation, this step was able to eliminate all noise originating from the positioning apparatus as well as those structures distal to the DRUJ. A typical result of this operation is shown in Figure 3.4-1. Second, the multiple-slice editing tool was used to deselect those highlighted voxels which were not immediately adjacent to the osseous surfaces as with those in the skin. Finally, manual slice by slice cropping was used in those areas where voxel noise was immediately adjacent to the bony surface or when separating one bony surface from another.
Once a noise-free mask had been created that encompassed all three long bones of the appendage, it was advantageous to separate each into its own mask. This not only simplified mask manipulation, but more importantly, ensured that once interpolated into 3D bodies, the three bones would be able to translate and rotate with respect to one another. To obviate the need for simply repeating the cropping function described above, MIMICS provides a Boolean Operations tool that allows one mask to be added or subtracted from another. Therefore, it was possible to coarsely crop out the entirety of any one of the three long bones, e.g. the radius, and that mask could be subtracted from the original whole arm mask to yield the resulting, noise-free mask containing only the bone of interest.
3.4.2. Morphology Operations

Within MIMICS, a mask would often contain jagged edges around the bony surface due to surface porosity or very thin cortical bone; this was especially problematic in those regions at the epiphyses where bone density is lowest. Here, the global threshold applied to the mask would capture the majority of the surface, but would contain discontinuities that were inconsistent with the bony outlines visible in the slice data. While these edges do not present a problem when isolated within the medullar portions of the bone, the presence of these discontinuities would threaten the fidelity of any surface reconstructions for the bone. Therefore, a combination of manual and automated morphology operations were used to compensate for these inadequately defined surfaces.

The coarse “dilate” and more refined “close” commands were used to ensure continuity of the articular surfaces. Both of these functions project active pixels or voxels onto adjacent inactive space according to a set of user-defined conditions. Specifically, the dilate command enlarges a mask by highlighting up to eight inactive pixels in 2-D, or twenty six in 3-D, that are adjacent to a selected pixel. This operation has the effect of filling out holes in the articular surfaces where the cortical shell is defined only by a thin band of highlighted voxels, though it was used sporadically to limit the introduction of non-physiologic artifacts in areas without surface discontinuities. The close function operates in the same manner as the dilate function, but will not swell surfaces without discontinuities. This discrimination therefore, preserves areas like the diaphysis that already show good surface resolution. Figure 3.4-2:A shows the original mask with initially detected cortical bone highlighted in teal and gaps in the outer edge of the surface.
shown in red circles. Figure 3.4-2:B and Figure 3.4-2:C show that same portion of the mask following application of the close and dilate morphology operations, respectively, with successfully closed gaps shown in green circles. While Figure 3.4-2:B demonstrates the “close” function’s ability to fill in small surface discontinuities, larger imperfections persist. Additionally, while Figure 3.4-2:C shows that the “dilate” function is able to close all surface gaps; it does so at the expense of enhanced mask noise outside of the bone, shown in yellow.

Figure 3.4-2: Original bony mask of the radial head (teal) showing surface gaps in red circles (left). Mask following use of "close" function, showing new pixels highlighted in purple and successful (green) and unsuccessful (red) closing of surface gaps (middle).

3.4.3. Boundary Detection & Slice Interpolation

Once each mask accurately reflected the surface geometry of the bone throughout its length, another MIMICS function was used to create a set of contours on each slice called polylines. These lines would outline any region with active voxels to create closed
loops within each slice. Additionally, this approach required that only the outermost surface, the cortical shell, of the bone needed to be defined, since any inactive voids in the mask resulting from very low trabecular bone density or medullar space would still be encompassed within a closed loop. These voids could then be filled using the fill polylines command. For example, throughout the diaphyses of the various masks, the cortical shell would be highlighted but those voxels representing the medullar canal would be inactive. Thus, by encircling the cortical shell and filling those closed spaces, each mask could be made solid without having to manually selecting those low attenuation areas on each successive slice as shown in Figure 3.4-3. Additionally, the fill polylines function offered a final check against discontinuity before slice interpolation since any slice that did not have a closed loop encircling the cortical shell would not fill and would therefore appear hollow.

Figure 3.4-3: Axial slice of the proximal ulna with polyline (purple) detection of its cortical edge (left). Polylines calculated for all axial slices of ulna (middle). View of interpolated surface (yellow) between polylines (right).
With the bones of the appendage each defined by a separate continuous mask, solid bodies could be created by interpolating the voxel space between adjacent slices. To ensure that no extra-osseous material was created, the shell reduction function was used. This confined solid body creation to the single (user-defined) largest volumetric body, discarding any small voxelar artifacts still present in the mask.

3.5. STEREOLITHOGRAPHY (STL) FILES

The solid bodies created by MIMICS were output in a format known as stereolithography (STL). This file type has a number of benefits that make it ideal for computer modeling. Chiefly, the STL format is near universally compatible with commercial 3-D design software packages. Additionally, it is able to be read in ASCII or binary, and the files themselves are unambiguous and can be relatively small. MIMICS created these bodies by detecting the boundaries within each slice and then creating a linear interpolation between successive slices in a given mask. The combined surfaces are then represented through triangulation. Numerically, each individual triangular facet is described by just twelve Cartesian coordinates, three for each vertex of the triangle and a fourth set of three describing the end of a unit vector normal to the surface, pointing outwards. [7]

Initially, the STL bodies output by MIMICS contained many tens of thousands of triangular facets for each bone. While these are the most faithful 3-D constructions given the original voxelated masks, using models with so many facets is often unnecessary for
RBM and will drastically slow down, if not prevent, certain 3-D software packages from running normally. For this reason, a remeshing routine was used within the native MIMICS environment to refine the mesh and selectively remove or combine many of the triangles within the STL. The result was a final 3-D body was reasonably smooth, free from any abrupt non-physiologic surfaces, and comprised of a computationally manageable number of triangles.

To evaluate the effectiveness of the surface modifications performed, MIMICS allowed that the relative quality of the constituent triangles be measured quantitatively and presented as a histogram within the remesher subroutine, shown in Figure 3.5-1.

![Figure 3.5-1: MIMICS Remesher interface showing R-in/R-out shape measure histogram.](image)

This graph diagrammatically represents the uniformity of the triangular mesh according to a parameter known as the “R-in/R-out Normalized ratio.” Here, twice the radius of an inscribing circle is divided by the radius of an ascribing circle. That value is
then normalized to two. This ratio therefore is a dimensionless value between 0 and 1, with one representing an equilateral (and equiangular) triangle. Triangles with small R-in/R-out ratios were usually elongated slivers with highly acute/obtuse inner angles. The user-defined parameters and error thresholds used in the smoothing and triangle reduction operations were chosen to maximize this ratio, maintaining the maximum number of surface triangles while not exceeding the computational limits of the hardware or software available. [7]

3.5.1. Smoothing

The first surface modification performed involved removing some of the surface cavitations and small contour discontinuities that resulted from bone density fluctuations and surface porosity observed at the mostly trabecular epiphyses. This was done by smoothing the surface within the MIMICS remesher subroutine. The smoothing algorithms follow a first order Laplacian approximation that is fundamentally different from the methods used by MIMICS to create the triangular mesh. Within this approach, each triangular vertex position is evaluated according to the weighted contributions of its neighboring triangles. Thus, in contrast to the triangulation process, each vertex is not considered fixed. Specifically, 

\[(E_1, E_2, \ldots, E_k)\] are triangles sharing a common vertex \(v^*\) with all other vertices \((v_1, v_2, \ldots, v_k)\).

\[\sum_i W_i (x_i - x) = 0\]  

(1)

39
\[ \sum W_i (y_i - y) = 0 \]  
\[ \sum W_i (z_i - z) = 0 \]  
\[ v^* = (v_1 + \ldots + v_k)/k \]

Where \( W1 \) is the weighted position function of a given triangle and \((x_i, y_i, z_i)\) are the Cartesian coordinates of the \( i \)th vertex of the adjoining facets excluding \( v^* \). [35] Within MIMICS, the weights assigned to adjoining triangles are given according to the Smoothing-Factor ratio. When this ratio is very low, the initial position of \( v^* \) is chiefly responsible for its new position; when the ratio is high, approaching 1, the weights of all adjacent triangles are considered equally. [7] Because this method of smoothing calculates a translation for each \( v^* \) at every vertex in the model, multiple iterations were necessary to reach convergence. For the models used, this was generally less than fifteen.

![Figure 3.5-2: An STL triangulation of the distal humerus (left) and the effect of triangle reduction (middle) and smoothing (right).](image-url)
3.5.2. **Triangle Reduction**

The second surface modification technique was triangle reduction. Through trial and error, it was discovered that approximately ten thousand triangles was the usable computational limit of the Solidworks CAD software package when importing binary STL files. Thus, following the MIMICS smoothing routine, a triangle reduction algorithm was used to shrink the initial mesh size of as many as 100,000 triangles, down to a manageable number. Accomplishing this task involved two separate phases. First, a shape-quality threshold was applied, such that all triangles below the user-defined value were eliminated. This function has the effect of removing highly acute sliver triangles, resulting in a more uniform mesh. The value of this threshold ranged between 0.15 and 0.3 for the models in this work. Depending on the initial quality of the mesh, this function would have an insignificant to minor effect on the total number of triangles but would greatly improve the resulting quality of the mesh. [7]

The second step in the triangle reduction was the iterative evaluation of each triangle to triangle edge angle within the mesh. Here a flip threshold angle was assigned that defined the minimum angle uniqueness to preserve shape geometry. Thus, any two triangles that meet with an angle below this value could be approximated as a single facet in one plane; any two triangles that meet at an angle above this value were identified as representing a unique curvature and thus preserved to keep the original surface contours. Additionally, any two triangular surfaces meeting at angle below the threshold angle were
bound by a geometrical error tolerance that set the maximum total translation of the resulting new surface from the original. This value was set at the default level of 0.05 to ensure the total 3-D body volume was unaffected by the triangle reduction protocol. [7]

With a limit of approximately 10,000 triangles imposed for any single STL file to be imported into Solidworks, the individual bony masks of the 30° high resolution scan, which focused on the EJC, were cropped just outside of the joint itself. For the distal humerus, the mask was cropped just superior to the supracondylar ridges parallel to the CT slice orientation. For the proximal radius and ulna, each mask was cropped just distal to the radial and ulnar tuberosities, respectively. This was also done along a plane parallel to the CT slice orientation. The cropping operation ensured that the articular surfaces of each bone were represented by the largest number of triangular facets possible and that the number of triangles per unit surface area was highest in those areas where 3-D bony contact would dictate RBM function.

3.6. STL REGISTRATION

The solid body creation process resulted in three sets of elbow models, each with three bones resulting in nine distinct STL bodies being exported from MIMICS. For the purposes of RBM, the ideal reconstruction of the joint involved the use of both 30° flexion scans to create a hybrid assembly representing the complete surface geometry of the three long bones of the arm. Specifically, this entailed creating an assembly of the humerus, radius, and ulna STL files derived from the lower resolution 2.0mm slice thickness scan
and merging it with the STL files from the higher resolution 0.4mm slice thickness scan that focused specifically on the elbow joint complex. As a result, the hybrid model created accurately represented the size and shape of each of the long bones of the arm with very high detail of the articular surfaces within the elbow joint.

To create the hybrid model described the commercially available CAD modeling software SolidWorks 2007 SP4 was used to import and manually manipulate the STL files output by MIMICS. First, each whole bone STL from the lower resolution scan was one by one uploaded into Solidworks. During this process, SolidWorks would parse each file, assigning a 3-D point to the three vertices described by the STL format, and assemble the collection of triangular facets into a single cohesive body. Next, the three high resolution 30° STLs were loaded into the low resolution 30° MIMICS environment. Using the STL Registration feature, each high resolution STL was mapped onto its corresponding low resolution full bone counterpart through the use of the Global Registration algorithm. This operation sequentially calculated the distance between each common vertex in the two models and adjusted the position of the low resolution STLs to minimize those point-to-point distances. The final translation and rotations applied by MIMICS were output in a 4x3 transformation matrix text file. Visually, the success of this registration could be assessed by the completely aligned overlap of the surface features between the two low and high resolution STLs (Figure 3.6-1). Quantitatively the degree to which the point-to-point distances were minimized was described by the sum of least squares residual error measurement in MIMICS given in equation 5 below.
Next, the high resolution 30° STLs were each loaded into the SolidWorks, parsed, and saved as “part” files. Each low resolution part was matched with its high resolution analog and both were loaded into a hybrid bone assembly. Since the global registration had transformed the high resolution part into the identical place and orientation of the corresponding low resolution part, these hybrid bone assemblies were exactly matched in 3-D space. Since the low resolution STLs represented the entire bony geometry of the humerus, radius, and ulna, each hybrid bone assembly had a low and a high resolution surface represented in the EJC. As we were only interested in the high resolution geometry, the low resolution bone in each hybrid assembly was manually cropped at the level that the high resolution bone began. For the humerus, this was at the level of the
supracondylar ridges; for the radius and ulna, this joint was located just distal to the respective tuberosities of the proximal bones.

Finally, each of these new hybrid bone assembly files was brought together into a higher level assembly file shown in Figure 3.6-2. Herein, the relative positions of the three bones were kept consistent from the CT scan to the 3-D model by keeping the bones defined in terms of the CT scanner’s reference frames. An interference algorithm available within the SolidWorks package confirmed that there were no overlaps or non-physiologic interferences present in the assembly file following construction of the model.

Figure 3.6-2: Hybrid elbow model showing the 0.4mm resolution EJC STLs (dark colors) incorporated into the 2mm resolution whole bone STLs (light colors).
3.7. CHARACTERIZING THE MODEL

The ability to quantitatively and reproducibly identify certain skeletal features was important in the creation of a computational model. To accomplish this, the MedCad module within MIMICS was utilized as a commercially available alternative to custom in-house coding. This interface allowed spheres, cylinders, lines, and planes to be fit onto the polylines representing the bony contours on each slice. These geometry fits were employed to capture the 3-D best fit line for the long axis of each bone based on its diaphysis, the central axis of the proximal ulna superior to the ulnar tuberosity, the proximal radius superior to the radial tuberosity, the spherical centers of the radial and humeral heads, and the best fit plane describing the central ridge of the trochlear notch of the ulna (Figure 3.7-1).

Figure 3.7-1: Sphere fits of the humeral head (left) and radial head (right), shown with their respective best-fit long axes overlaid.
The appropriate geometric expression of the fitted object was output by MIMICS to a text file. Numerically, 3-D lines were reported as two (x,y,z) Cartesian coordinates in space with a third point relaying a unit directional vector originating at point one, spheres were represented by a 3-D coordinate at the center of the sphere and a second point representing the tip of a vector with length equal to the radius of the fit sphere. Best fit planes were fully described by the coordinates of a single point on a plane and a second point at the terminant of the unit vector normal to it. Goodness of fit was quantified for each body accompanied the geometric expression and was denoted by the deviation value (D) that represented the minimum sum of squares between any point in the voxelar mask to the line or plane being fit:

\[
D = \sum_{i=1}^{N} \frac{d(x_i, \bar{x})^2}{N}
\]

(5)

In addition to geometry fitting, each bone was given a unique coordinate system and set of corresponding reference frames. These landmarks were based on prominent anatomic landmarks, mathematically fit 3-D objects, and an external SolidWorks API designed to calculate and mark the center of gravity (COG) of any part. The designation of unique references frames was used extensively to characterize joint motion, orient testing apparatus, and define cut planes for radial and ulnar ostectomies. Where applicable, the reference planes and coordinate systems were defined in accordance with the recommendations made by Wu et al (2004) on behalf of the International Society of Biomechanics (ISB). [125]
The humerus coordinate system had its origin \((Oh)\) at the sphere-fit center of the proximal humeral head \((GH)\). The \(y\)-axis \((Yh)\) was defined as the line connecting the midpoint of the line between the medial and lateral epicondyles \((EM, EL)\) pointing superiorly. The \(x\)-axis \((Xh)\) was the line normal to a plane containing the \(GH, EL,\) and \(EM\) pointing anteriorly. The \(z\)-axis \((Zh)\) is the common line perpendicular to both \(Xh\) and \(Yh\) pointing to the right, (laterally for this right upper extremity). The anatomic planes of reference were also defined consistent with this coordinate system. Herein, the anteroposterior \((AP)\) plane contains \(Yh\) and \(Xh\), the mediolateral \((ML)\) plane contains \(Yh\) and \(Zh\), and the transverse \((TR1)\) plane contains \(Xh\) and \(Zh\). Additionally, it was useful to maintain a second definition for the transverse plane based on the 3-D line approximation of the humeral diaphysis. This definition placed the transverse plane normal to this line and containing the humerus COG as identified by a third party SolidWorks API.

The ISB coordinate system was also defined for the ulna; however, due to the paucity of easily defined bony landmarks in the forearm, this definition required the forearm be in neutral rotation and \(90^\circ\) flexion. The origin \((Ou)\) was assigned as coincident with the tip of the ulnar styloid \((US)\). The \(y\)-axis \((Yu)\) pointed proximally from \(Ou\) toward the midpoint of the line connecting \(EM\) and \(EL\). The \(x\)-axis \((Xu)\) was represented as a vector normal to the plane containing \(EL, EM,\) and \(US\) pointing anteriorly. The \(z\)-axis \((Zu)\) was thus described by the common line perpendicular to both \(Yu\) and \(Xu\) pointing to the right (laterally). [125]

Anatomic reference planes dictated by the ISB coordinate system for the ulna were also defined but had limited application due to the constraint that neutral forearm rotation
and 90° flexion be maintained. Instead, a second series of reference planes was defined based on the 3-D line fit of the ulnar diaphysis and the ulnohumeral SDA described in sections 3.7 and 4.3.1. The AP plane for the ulna was defined as containing both the long axis of the ulna and the SDA of the ulnohumeral joint. The ML plane represented a 90° rotation of the AP plane about the ulnar long axis. The transverse plane was defined as normal to this long axis and containing the ulna COG. This alternate definition allowed the position of the ulna to be uniquely described in space at any flexion angle and irrespective of radial movement.

The ISB coordinate system for the radius was applied, but like the restrictive definition given for the ulna, reference planes based on the radius coordinate system are constrained to instances of neutral forearm rotation and 90° flexion. The origin (Or) of the radius coordinate system is coincident with the apex of the radial styloid (RS) and has a y-axis (Yr) that projects proximally toward EL. The x-axis (Xr) was defined as the vector normal to a plane containing RS, US, and EL, and finally, the z-axis (Zr) was defined as the common line perpendicular to Yr and Xr. Reference planes were assigned in accordance with ISB recommendations but again had limited use. However, an alternate definition of the transverse plane was utilized that described the feature as the plane normal to the 3-D line fit of the long axis of the bone also containing the radius COG.
4. THREE-DIMENSIONAL KINEMATIC SIMULATIONS

4.1. OVERVIEW

Following the acquisition and reconstruction of each of the long bones of the arm, the bodies were assembled, positioned, and constrained in 3-D space. SolidWorks was used to orient the bones of the model such that the original specimen flexion angle, pronation angle, and joint articular space were faithfully represented. Furthermore, SolidWorks was used to identify certain skeletal and soft-tissue features on each of the bones in order that they could be consistently identified and quantified. The SolidWorks add-in COSMOSMotion was used to impose force and displacement constraints on the model as well as to define the boundary conditions for inter-body interactions. These parameters include constant magnitude force vectors to represent muscles, tension only spring elements to represent ligamentous/capsuloligamentous soft-tissues, and inter-body contact calculation and restoration through the product of material stiffness and normal penetration distance. At the core of the COSMOSMotion package, the ADAMS solver provided the computational algorithms used to iteratively solve the constitutive equations of the model under various load and displacement perturbations. Thus, a RBM model was created wherein the reconstructed solid body elbow assembly was constrained only by muscle contraction, ligament stiffness, and articular surface contacts.
4.2. ORIGINS AND INSERTIONS

The solid body creation process described in Chapter 3 yielded triangulated surfaces for each bony part comprised of many thousands of triangular facets. The vertices and edge midpoints of these facets provided an abundance of easily identifiable 3-D points, fixed in their relative position on each bone, that could be used to represent the origins and insertions of the various soft-tissues applied to the model. The locations selected for these landmarks were made qualitatively using anatomic atlases and published cadaveric studies. When an area of soft-tissue attachment was diffuse across a large segment of bone, e.g. the insertions of the triceps tendon onto the superior olecranon, multiple points were used to represent the structure. More discrete attachments were characterized by the qualitative center of the origin or insertion described in the literature. Thus, lines of action for all soft-tissues were represented as a straight line vector between the origin and insertion(s). This methodology was therefore able to constrain the bones within the assembly in a physiologically relevant way through the use of relatively few fiber analogues.

4.3. JOINT APPROXIMATIONS

In order to accurately measure certain physiologic parameters such as flexion-dependent ligament lengths and joint angles, idealized joint movements were defined for both the ulnohumeral flexion/extension axis as well as the radioulnar pronation/supination
axis. In both instances, only osseous constraints were considered in determining the axes of rotation and therefore the resulting approximations reflect only passive elbow kinematics. Furthermore, each joint approximation was made independently of the other, thereby allowing any physiologically reasonable combination of flexion and pronation to the modeled. This separation is generally supported in the literature as most have observed the radioulnar axis of rotation to intersect the ulnohumeral flexion/extension axis at the approximate center of the capitulum. [50, 54, 78, 83, 118] Though useful for characterizing the model, no assumptions regarding joint behavior were made during any RBM simulation. Instead, idealized motions were used to position the joint prior to simulation with soft-tissue constraints and osseous surface interactions dictating the kinematic function of the joint for all time t >0.

4.3.1. Flexion/Extension

As suggested in section 2.3, the flexion/extension axis of the forearm was approximated by a screw displacement axis qualitatively described as passing through the centers of the capitulum, trochlea, and greater sigmoid notch. Quantitatively, the orientation of the axis was extrapolated from relative position data of three discrete points on the proximal ulna as represented in the 30° and 90° scans of the original specimen. As it is generally accepted that flexion/extension is independent of forearm rotation, the relative position of the radius was not considered. [78, 83]
In order to infer the rotational and translational movement of the ulna relative to the humerus in each scan, a hybridized assembly was created using the distal humerus as a common frame of reference in each scan. To accomplish this, a sub assembly of the distal humerus and proximal ulna was created within SolidWorks for both the 30° and 90° scan STLs. Then, using the MIMICS STL registration feature described in section 3.6, the precise position of each facet in the 90° humerus was transformed onto the 30° humerus. Following this step, the two humeri showed complete interference with one another, indicating that they were occupying the identical position and orientation within 3-D space. Next, the same transformation applied to the 90° humerus was applied to the 90° proximal ulna. The final result of these two transformations was a common humerus position and two distinct ulna positions, the first in 30° flexion and the second in 90° flexion.

Once positioned, the MedCad module within MIMICS was used to characterize various aspects of the humerus and ulna geometries for each orientation (see section 3.7). First, the centerline for the humerus and each ulna was fit as well as a point defining the most proximal tip of the coronoid process. Next, the intersection of a line perpendicular to the ulnar centerlines and coincident with the tip of the coronoid process was marked on each ulna. Finally, a point was marked 20mm superior to the intersecting point described above along the centerline of the ulna. Thus, each ulna was given three common non-collinear points that could be used as inputs for extrapolating the helical transformation of the ulna from one flexion angle to another.

The location of the screw displacement axis was calculated using the approach outlined by Beggs (1983) for extrapolating the helical transformation of a body given only
the starting and ending positions of three points on that body. [13] Herein, Pi is transformed into Pi’ where i=1,2,3, through a rotation of angle $\sigma$ and translation of magnitude $s$ along the SDA. Mathematically,

$$
\begin{bmatrix}
1 \\
P'_1 \\
P'_2 \\
P'_3
\end{bmatrix} = 
\begin{bmatrix}
1 & 0 & 0 & 0 \\
S_{21} & S_{22} & S_{23} & S_{24} \\
S_{31} & S_{32} & S_{33} & S_{34} \\
S_{41} & S_{42} & S_{43} & S_{44}
\end{bmatrix}
\begin{bmatrix}
1 \\
P_1 \\
P_2 \\
P_3
\end{bmatrix}
$$

where,

$$
S_{21} = |\hat{i} \cdot \hat{s}| - o_x (S_{22} - 1) - o_y S_{23} - o_z S_{24}
$$

$$
S_{22} = \cos(\sigma) + \text{versin}(\sigma) \left(\hat{i} \cdot \hat{s}\right)^2
$$

$$
S_{23} = -\sin(\sigma) \left(\hat{k} \cdot \hat{s}\right) + \text{versin}(\sigma) \left(\hat{i} \cdot \hat{s}\right) \left(\hat{j} \cdot \hat{s}\right)
$$

$$
S_{24} = \sin(\sigma) \left(\hat{j} \cdot \hat{s}\right) + \text{versin}(\sigma) \left(\hat{k} \cdot \hat{s}\right) \left(\hat{i} \cdot \hat{s}\right)
$$

$$
S_{31} = |\hat{j} \cdot \hat{s}| - o_x S_{32} - o_y (S_{33} - 1) - o_z S_{34}
$$

$$
S_{32} = \sin(\sigma) \left(\hat{k} \cdot \hat{s}\right) + \text{versin}(\sigma) \left(\hat{i} \cdot \hat{s}\right) \left(\hat{j} \cdot \hat{s}\right)
$$

$$
S_{33} = \cos(\sigma) + \text{versin}(\sigma) \left(\hat{j} \cdot \hat{s}\right)^2
$$

$$
S_{34} = -\sin(\sigma) \left(\hat{i} \cdot \hat{s}\right) + \text{versin}(\sigma) \left(\hat{j} \cdot \hat{s}\right) \left(\hat{k} \cdot \hat{s}\right)
$$

$$
S_{41} = |\hat{k} \cdot \hat{s}| - o_x S_{42} - o_y S_{43} - o_z (S_{44} - 1)
$$

$$
S_{42} = -\sin(\sigma) \left(\hat{j} \cdot \hat{s}\right) + \text{versin}(\sigma) \left(\hat{k} \cdot \hat{s}\right) \left(\hat{i} \cdot \hat{s}\right)
$$

$$
S_{43} = \sin(\sigma) \left(\hat{i} \cdot \hat{s}\right) + \text{versin}(\sigma) \left(\hat{j} \cdot \hat{s}\right) \left(\hat{k} \cdot \hat{s}\right)
$$

$$
S_{44} = \cos(\sigma) + \text{versin}(\sigma) \left(\hat{k} \cdot \hat{s}\right)^2
$$
Here, \( \hat{s} \) is a unit vector originating at \( o_j \) for \( j=x,y,z \) and versin(\( \theta \))=1-cos(\( \theta \)). [13]

The application of this methodology was accomplished through a MATLAB script developed by Fisk et al (2009) wherein the three discrete ulna points from the 90° and 30° orientations were used as inputs for Pi’ and Pi, respectively. [37] The outputs from that routine are given below.

\[
[S] = \begin{bmatrix}
1 & 0 & 0 & 0 \\
-370.9916 & 0.4768 & 0 & -0.8790 \\
-16.6958 & -0.1034 & 0.9931 & -0.0561 \\
-344.5736 & 0.8729 & 0.1176 & 0.4735
\end{bmatrix},
\]

(6)

\[
\hat{s} = \begin{bmatrix}
0.0985 \\
-0.9934 \\
-0.0586
\end{bmatrix}, \quad |\hat{s}| = 0.2566 \text{mm of total translation},
\]

(7)

\[
o = \begin{bmatrix}
0 \\
1040.4 \\
-422.1
\end{bmatrix}, \quad \sigma = 1.0796 \text{mm} = 61.857°.
\]

(8)

Graphically, the extrapolated SDA is shown below in Figure 4.3-1.
Figure 4.3-1: Oblique view of the distal humerus with ulnohumeral SDA shown passing through the approximate centers of the trochlea and capitulum.

The SDA forms an angle of 84.96° with the long axis fit of the humeral diaphysis, compared to 83.19° for the line connecting EM and EL. By inspection, the axis travels through the approximate center of the trochlea and capitulum, as viewed from a sagittal projection of the bone. Additionally, the calculated rotation $\sigma$ from the 30° scan to 90° was 61.857° in good agreement with the approximate value determined during specimen scanning.

Ulnohumeral flexion/extension was quantified as the angle formed between the ulnar and humeral geometry fit long axes as projected onto a plane normal to the SDA. In accordance with ISB recommendations as well as generally accepted convention, flexion was defined as a positive rotation about the SDA. [125] Consequently, the original scan
orientation was determined to be $+36.63^\circ$ flexion as opposed to the approximately $30^\circ$ intended in the scan.

**Figure 4.3-2:** Medial view of ulnohumeral flexion angles mapped between the 30° and 90° CT scans. Note the actual difference in flexion is 61.89° vs. the intended 60°.

Though not characterized by any idealized motions, measurements of carrying angle were made with model at various flexion angles. This value was defined as the angular displacement between the 2-D frontal plane projections of the ulnar and humeral geometry fit long axes. Lateral ulnar deviation was defined as a positive carrying angle. The magnitude of this value ranged from 7.15° at full extension to 0.25° at 145° flexion and is within the ranges reported in the literature. [10, 12, 78]
4.3.2. Pronation/Supination

The forearm pronation/supination axis was defined as the line originating at the center of the proximal radial head and projecting obliquely toward the center of the distal ulnar fovea. (Figure 4.3-3) The location of the radial head center was determined through a sphere fit characterization of the feature as described in section 3.7 while the center of the ulnar fovea was determined by inspection.

Figure 4.3-3: Anterior view of the forearm in 30° flexion with -80° of supination (light yellow) and +80° of pronation (dark yellow).

Neutral forearm rotation was defined according to the ISB recommendations of Wu et al (2005) as occurring when a line connecting the radial and ulnar styloids is parallel to
the sagittal plane of the body. Pronation was defined as a positive rotation about this axis resulting in the palmar aspects of the distal radius being directed medial and inferior. \[125\] Therefore, the specimen was in +14.13° of pronation in the scanned position.

4.4. COSMOSMOTION

The COSMOSMotion rigid body simulation package is seamlessly incorporated into SolidWorks. As a consequence, all aspects of the original bony model were easily transported to and from the COSMOS rigid body simulator without the loss of pertinent position, orientation, or geometric characterization data. However, in order that COSMOS recognize certain geometric constraints at the surfaces of bones and that it arrives at a convergent solution in a reasonable amount of time, it was necessary to define a number of parameters within the COSMOSMotion module. The most fundamental of these parameters was the identification of the bodies within the model which were allowed to move and those which were grounded. This designation as a “moving part” was given to both the radius and ulna assemblies such that each bone was unconstrained in all six DOFs, and that only soft-tissue elements and osteoarticular geometry could influence motion. In contrast, the humerus assembly was designated as a “grounded part.” This meant that the humerus part had zero unconstrained degrees of freedom and was defined as at rest at all times during the RBM. In contrast to the work by Fisk et al (2009), gravitational forces were not neglected in the model. \[37\] This force was applied at the centroid of each bone and oriented relative to the grounded humerus bone.
Additional functionality included the ability to constrain specific motions or DOFs in the assembly. This was done through the application of “joints” that impose mechanical limitations on assorted combinations of translations and rotations. For instance, a translational joint removes all three rotational DOFs and two translational DOFs from a body, leaving only axial translation as an allowed motion. Though critical to the replication of aspects of real world experiments, joints were not used to define any physiologic movement within the elbow itself. Instead, these mechanical constraints were only implemented as a means of applying external perturbations to the model. These elements will be discussed further in the validation of the computational model (see section 5.2.2, 6.2.2).

Following the completion of each RBM simulation, COSMOSMotion offered a number of export and analysis options for the resultant kinematic data. Chief among these are the spreadsheet export, MSC ADAMS solver file export, FEA meshing output, and animation viewing options. These outputs allowed for both qualitative and quantitative analysis of results, as well as a means of presenting derived kinematic data in universally accepted formats.

4.4.1. Solver Parameters

Embedded within COSMOSMotion, the MSC ADAMS software package is responsible for modeling the mixed dynamics behavior of the elbow joint assembly. This collection of algorithms iteratively applies the equations of motion to all parts within the
model not explicitly defined as motionless. While there are a number of integrators available within the ADAMS package, the “Gear Stiff Integrator” (GSTIFF) was chosen since it is widely accepted as a robust and streamlined methodology for solving “stiff” ordinary differential equations. Such equations are said to be “stiff” because the integration error is not significant compared to the large variability in the resultant magnitude possible when the integration of high frequency oscillatory systems is performed over insufficiently small time steps. [21, 44, 45] The GSTIFF algorithm addresses these problems by applying backward differentiation formulas that iteratively predict the error of the system for a given time step, compare it to a user defined threshold for that error, and then lower, or correct, the value of the time step until convergence is reached and the error threshold is not violated. [1] Thus, this “predict, compare, correct” methodology allows the model to be solved quickly with large time steps that are dynamically and automatically adjusted downward in order to capture very short duration events such as intermittent rigid body contact. [1]

In addition to the choice of an integrator, Figure 4.4-1 shows a number of other user-defined parameters available within the COSMOS environment. The values of these parameters were assigned based on ADAMS solver recommendations for capturing short time events as well as trial and error assessments of model stability and total computational time. [1, 2] Here, the accuracy threshold and minimum time step values were set to the lowest possible level while the Jacobian pattern was set at its maximum. The combined effect of these changes ensured that the GSTIFF integrator would be able to capture even the briefest of events in the model, accrue the least amount of error provided by the
interface, and accurately capture all rigid body contact by recalculating the Jacobian matrix at every time step. Additionally, adaptivity was set to zero thereby ensuring that all rigid body contact would be incorporated into subsequent time steps of the integrator. The maximum number of iterations was upwardly adjusted from the default setting of 25 to 50 ensuring that the GSTIFF algorithm would have sufficient opportunities to meet the convergence criteria specified above.

Figure 4.4-1: COSMOSMotion interface showing user-defined RBM solver parameters.

4.4.2. Contact parameters

In order to faithfully simulate the complex motion of the native elbow joint, no approximations of physiologic joint movement were enforced during the RBM
simulations. Only, soft-tissue elements and surface to surface contacts were considered. Therefore, it was necessary to define the nature of the contacts possible between the bones of the arm. This was done for each of the three long bones of the arm as well as the annular ligament body using the 3-D contacts feature within COSMOS. This process causes SolidWorks to check for overlap of the cuboid spaces bounding each selected body at each time step of a simulation. Then, in the event of overlap, the software performs a finer interference calculation wherein the entire precise triangulated geometry of the bone is considered. [2] If an overlap is still detected, SolidWorks calculates the total volume of overlap on each bone as well as the penetration distance normal to the surface of the body. The ADAMS solver then incorporates this SolidWorks information and acts to counter the overlap through the process of penalty regularization. This involves applying a force originating at the centroid of the overlapped volume that projects outwardly against both bodies in contact (depicted in Figure 4.4-2). The magnitude of this force is calculated as the product between the material stiffness of the body in N/mm and the penetration distance normal to the surface. [1, 2] The ADAMS solver also incorporates a viscous element to the penalty regularization force in order to maintain the stability of the system. The degree of damping is variable dependent upon the rate of interference (dg/dt) between the two bodies and is bounded by the user-defined limit of damping, \( c_{\text{max}} \). [1, 3]

\[
F_n = k * g^e + \text{Step}\left(\frac{dg}{dt}, 0, 0, c_{\text{max}}, d_{\text{max}}\right)
\]

(9)

Where:

\( F_n \) = separating force normal to the overlapping surfaces
k = material stiffness (N/mm)
g = penetration distance
e = exponent
c_{\text{max}} = maximum damping force
d_{\text{max}} = penetration distance at maximum damping
dg/dt = penetration velocity at the point of contact

Figure 4.4-2: COSMOSMotion diagram depicting the origin and orientation of the penalty regularization force vector calculated in response to a detected 3-D contact.

There are a number of user-definable parameters available to better characterize the nature of 3D contacts (Figure 4.4-3). The most important of these is the stiffness value assigned to the material. A value of 8,000 N/mm was chosen in order to ensure that material overlap was kept to a minimum while not causing the integration difficulties possible as material stiffness tends larger. Geometric accuracy was set to its maximum value to ensure faithful detection of each facet in the triangulated bodies. Finally, friction
between articulating bodies was neglected due to the exceptionally low frictional coefficients observed within synovial joints. [60, 96, 97]

![Figure 4.4-3: COSMOSMotion interface showing user-defined 3-D contact parameters.](image)

**Figure 4.4-3: COSMOSMotion interface showing user-defined 3-D contact parameters.**

### 4.5. MUSCLE FORCES

Active muscle contraction was modeled by straight line, constant magnitude vectors anchored at discrete point insertions and directed towards the respective muscle origins defined by the methodology of section 4.2. In agreement with An et al (1991), only the contributions of the biceps brachii, brachialis, and triceps brachii were considered in this model as they are widely recognized as the predominate muscular stabilizers of the elbow joint. [9] Due to limitations in the COSMOSMotion modeling package, the muscles within this model were unable to account for muscle vectors that wrap bony surfaces or function along non-linear lines of action. Therefore, the origins for the biceps and triceps muscles were defined as coinciding with the point closest to the anatomical origin but lying on the physiological line of action for the tissue. For instance, the long head of the biceps originates on the supraglenoid tubercle of the scapula yet the force of the muscle is
redirected along the intertubercular sulcus. Therefore, the biceps origin established in this computational model was represented by a point at the superior limit of this groove.

Furthermore, the complex bicipital aponeurosis was not modeled in the current iteration of the rigid body elbow assembly; instead the biceps was defined as having a singular insertion on the radial tuberosity. The values of the muscular soft-tissues modeled are given in Table 4.5-1.

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Muscle Name</th>
<th>Origin</th>
<th>Insertion</th>
<th>Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>BC</td>
<td>Biceps brachii</td>
<td>Intertubercular sulcus*</td>
<td>Radial Tuberosity†</td>
<td>19.6</td>
</tr>
<tr>
<td>BR</td>
<td>Brachialis</td>
<td>Anterior surface of distal humeral shaft</td>
<td>Ulnar Tuberosity</td>
<td>19.6</td>
</tr>
<tr>
<td>TR-L</td>
<td>Triceps brachii, Lateral Insertion</td>
<td>Posterior surface of proximal humeral shaft**</td>
<td>Olecranon, Lateral Margin</td>
<td>13.067</td>
</tr>
<tr>
<td>TR-C</td>
<td>Triceps brachii, Central Insertion</td>
<td>Posterior surface of proximal humeral shaft**</td>
<td>Olecranon, Central</td>
<td>13.067</td>
</tr>
<tr>
<td>TR-M</td>
<td>Triceps brachii, Medial Insertion</td>
<td>Posterior surface of proximal humeral shaft**</td>
<td>Olecranon, Medial Margin</td>
<td>13.067</td>
</tr>
</tbody>
</table>

* = Origin of Bicipetal long head line of action  
** = Only lateral head of Triceps brachii modeled  
† = Bicipetal aponeurosis not modeled

Table 4.5-1: Muscle force magnitudes and lines of action.

Passive muscle contributions were generally ignored within the model since quantification of results was only done after all forces had reached equilibrium. One exception to this was the use of a 15 N/sec*mm translational damper acting collinear with
the biceps. The inclusion of this time dependent component helped to temper the un-
physiologic rapid acceleration and resulting force ringing associated with step jumps in
force on a muscle responsible for combined flexion and supination moments.

4.6. LIGAMENT CONSTRAINTS

In contrast to the constant magnitude vectors used to represent musculature within
the model, ligamentous soft-tissues were represented as damped linear tension only forces
with constant stiffness and dynamic length-dependent tension. Orientation and attachment
of fibers within the model were based on anatomic atlases and cadaveric studies.
Mechanical properties of the structures were modeled after published cadaveric
experiments investigating stress free lengths, stiffness or relative stiffness, and kinematic
stabilizing effects.

4.6.1. Ligament Arrangement

Within the elbow joint complex, the ligaments of the LCLC and MCLC were
modeled as discrete straight line fibers passing directly between origin and insertion. On
the medial aspect, the MCL was represented by three elements, two for the posterior fan-
shaped MCL, and one for the anterior MCL. (Figure 4.6-1) The posterior fibers originated
on the inferior and posterior aspects of the medial condylar surface and inserted on the
articular margin of the greater sigmoid notch of the ulna, superior to the central non-
articulating strip. The origin of the anterior band of the MCL was placed at the
approximate isometric center of the humerus on the anterior and inferior surface of the
medial condyle. The location of this point was chosen coincident with the center of the
sagittal projection of the medial lip of the trochlea in agreement with the work by Fuss et al

![Image](image1.png)

**Figure 4.6-1:** Sagittal view of the ligaments of the LCLC (left) and the MCLC (right).

The LUCL and RCL components of the LCLC were represented by one and three
tension elements, respectively. (Figure 4.6-1) The origin of the LUCL and RCL fibers
was placed just inferior to the lateral epicondyle. The single LUCL insertion was placed
on the proximal aspect of the tubercle on the supinator crest of the ulna. [79, 90] The RCL
was represented by three elements that project from this common origin to insertions on
the posterior, central, and anterior surfaces of the annular ligament body described below.

The annular ligament was not able to be modeled by simple straight line elements
since the native structure does not function in simple linear tension between two separate
bones. Instead the endogenous structure originates and inserts on the ulna, creating a looping projection that wraps about the articular margin of the radial head. Therefore, in order to create a tension only element analogous to the annular ligament without unphysiologically constraining the pronation/supination movement of the proximal radioulnar joint, additional solid geometries were incorporated. To this end, a SolidWorks part created by Fisk et al (2009) was used to approximate the curvature of the native annular ligament and span the ulnar origin and insertion while accommodating the radial head. The annular ligament body (ALB) was created by sweeping a sketch of an idealized annular ligament cross-section 225° about a central axis. [37] The interior four corners were then used as insertion points for the four ALB ligaments originating on the medial and lateral margins of the radial notch of the ulna. In this way the effective lines of action for the annular ligament elements were made to assert force along non-linear paths. Additionally, three RCL insertions were defined on the outer medial, central, and posterior portions of the surface in order to approximate the diffuse fan-shaped projections of the ligament.

The part was then initially positioned in the model by defining the ALB central axis as collinear with the radial head central axis defined using the MedCad module of MIMICS. However, this positioning definition was not used to approximate ligament function and was disabled prior to all RBM simulations. Physiologic function was instead dictated through a 3-D contact constraint (see section 4.4.2) defined between the inner surface of the ALB and the radial head. This prevented the ALB from passing through the radial head when pulled into tension, allowing the tensile forces produced by the ALB
ligaments to impart motion restrictions on the radius by way of the incorporated solid surface. Thus, the proximal radius was constrained within the radial notch of the ulna without limiting pronation/supination motions of the forearm.

The interosseous membrane of the forearm was modeled as five bands spanning the distal two thirds of the radius and ulna. (Figure 4.6-2) The central band of the IOM was represented as two elements originating mid-substance on the interosseous border of the radius and extending distally to insertions on the interosseous border of the ulna. The perpendicular width of the band bracketed by these two elements was approximately 12.25mm, within the range of findings regarding the average width of the CB of the IOM presented in the literature. [77, 89, 106] Additionally, the CB ligaments were prescribed to meet the ulna long axis at an angle of 20° in agreement with published cadaveric studies. [106]

Figure 4.6-2: Sagittal view of the ligaments of the forearm.

Accessory bands of the IOM are present in variable number and location in the population. [75, 77, 89, 106] Therefore, only the most constantly observed AB fibers were modeled with a pair of elements at the proximal and distal margins of the interosseous borders of
each bone. While the proximal AB fiber was constrained to 20° just as those fibers in the CB, the distal AB insertion on the radius was chosen based on anatomic landmarks and therefore creates a more obtuse angle with the long axis of the ulna. The final band of the IOM to be modeled was the DOB. This fiber was positioned along the interosseous border of the forearm such that it inserted at a point located at a distance 85% the length of the ulna and 90.1% the length of the radius, in agreement with published cadaveric data. [77, 89] These insertions caused the fiber to run obliquely and distally from the ulna to the radius at an angle of approximately 30° relative to the long axis of the ulna.

Lastly, the ligaments of DRUJ were modeled using two tension only elements. These fibers represented the dorsal and palmar aspects of the TFCC and originated at the base of the styloid, within the fovea of the ulna, and projected to insertions on the radial sigmoid notch.

4.6.2. Mechanical Properties

Ligament mechanical properties were modeled after published cadaveric data regarding stiffness, in situ strain, and stress free length. Stiffness values were applied in terms of N/mm of elongation. Modulus was not used since the straight line approximations of the ligament elements did not incorporate physiologic soft-tissue cross-sectional areas. Ligament strains and stiffnesses are given in Table 4.6-1.
Table 4.6-1: Ligament mechanical properties.

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Ligament Name</th>
<th>Stiffness (N/mm)</th>
<th>In Situ Strain (%)</th>
<th>Tension-Free Flexion Angle (α)*</th>
</tr>
</thead>
<tbody>
<tr>
<td>A-MCL</td>
<td>Medial Collateral, Anterior bundle</td>
<td>72.3</td>
<td>n/a</td>
<td>22°</td>
</tr>
<tr>
<td>P-MCL-A</td>
<td>Medial Collateral, Posterior bundle, Anterior</td>
<td>26.1</td>
<td>n/a</td>
<td>80°</td>
</tr>
<tr>
<td>P-MCL-P</td>
<td>Medial Collateral, Posterior bundle, Posterior</td>
<td>26.1</td>
<td>n/a</td>
<td>110°</td>
</tr>
<tr>
<td>RCL - A</td>
<td>Radial Collateral, Anterior</td>
<td>15.5</td>
<td>n/a</td>
<td>43°</td>
</tr>
<tr>
<td>RCL - C</td>
<td>Radial Collateral, Central</td>
<td>15.5</td>
<td>n/a</td>
<td>30°</td>
</tr>
<tr>
<td>RCL - M</td>
<td>Radial Collateral, Posterior</td>
<td>15.5</td>
<td>n/a</td>
<td>85°</td>
</tr>
<tr>
<td>LUC</td>
<td>Lateral Ulnar Collateral</td>
<td>57.0</td>
<td>n/a</td>
<td>107°</td>
</tr>
<tr>
<td>AL - A,D</td>
<td>Annular, AnteroDistal</td>
<td>28.5</td>
<td>2.0</td>
<td>n/a</td>
</tr>
<tr>
<td>AL - A,P</td>
<td>Annular, AnteroProximal</td>
<td>28.5</td>
<td>2.0</td>
<td>n/a</td>
</tr>
<tr>
<td>AL - P,D</td>
<td>Annular, PosteroDistal</td>
<td>28.5</td>
<td>2.0</td>
<td>n/a</td>
</tr>
<tr>
<td>AL - P,P</td>
<td>Annular, PosteroProximal</td>
<td>28.5</td>
<td>2.0</td>
<td>n/a</td>
</tr>
<tr>
<td>IOM - A,D</td>
<td>Interosseous Mem., Accessory Distal</td>
<td>18.9</td>
<td>0.6</td>
<td>n/a</td>
</tr>
<tr>
<td>IOM - A,P</td>
<td>Interosseous Mem., Accessory Proximal</td>
<td>18.9</td>
<td>0.6</td>
<td>n/a</td>
</tr>
<tr>
<td>IOM - C,D</td>
<td>Interosseous Mem., Central Distal</td>
<td>65.0</td>
<td>0.5</td>
<td>n/a</td>
</tr>
<tr>
<td>IOM - C,P</td>
<td>Interosseous Mem., Central Proximal</td>
<td>65.0</td>
<td>0.8</td>
<td>n/a</td>
</tr>
<tr>
<td>IOM - DOB</td>
<td>Interosseous Mem., Distal Oblique Bundle</td>
<td>65.0</td>
<td>0.8</td>
<td>n/a</td>
</tr>
<tr>
<td>DRUL-D</td>
<td>Distal RadioUlnar, Dorsal</td>
<td>13.2</td>
<td>2.0</td>
<td>n/a</td>
</tr>
<tr>
<td>DRUL-P</td>
<td>Distal RadioUlnar, Palmar</td>
<td>11.0</td>
<td>2.0</td>
<td>n/a</td>
</tr>
</tbody>
</table>

* = Angles derived from Regan et al (1991)

Stiffness values as well as flexion angle dependent stress free lengths for the LUCL, RCL, and posterior portions of the MCL were taken from Regan et al (1991). As there is some ambiguity in the literature regarding the stress free length of the anterior MCL, or guiding bundle, values for this ligament were modeled as constantly taught with

Stiffnesses for the CB of the IOM were obtained from Gabriel et al (2003) in agreement with Pfaeffle et al (1996). Stiffnesses of the AB of the IOM were inferred from a publication by Hotchkiss et al (1989) that showed that the CB is responsible for 71% of IOM tension, with the AB fibers responsible for the remaining 29%. The stiffness value for the DOB of the IOM was inferred from an anatomical study conducted by Noda et al (2009) wherein the average DOB width was reported as approximately one half that of the CB, while the average thicknesses for the two structures were equivalent.[89] Thus, the DOB was modeled using one half the stiffness of the CB as reported by Pfaeffle et al.

Since the CB, AB, and DOB have been shown to be isometric through forearm pronation/supination, all ligament lengths were arbitrarily defined in neutral forearm rotation. [77, 89] Distal radioulnar ligament stiffnesses were consistent with those published by Schuind et al (1991, 1995). [103, 105]

In situ strains applied to isometric ligaments were taken from published values when possible. This group included the fibers used to model the central and accessory bands of the IOM. The strain within the DOB was assumed to be equivalent to that found in the CB. Ligaments that are physiologically isometric but that do not have published in situ strain values were modeled as having 2.0% strain consistent with Liacourus et al (2007). [67] This approximation applied to the fibers of the ALB as well as the elements of the DRULs. Additionally, those ligaments that did not have an observable in situ strain, i.e. those whose length was flexion angle dependent, were pretensioned with 0.5-1.0%
strain. This was done in order to minimize instability within the model associated with oscillating step jumps in tension.

4.6.3. FORTRAN Expressions

COSMOSMotion contains inherent functionalities that allow it track a host of pertinent RBM variables including displacements, velocities, and relative positions. However, the program does not contain a ready-made means of modeling the behavior of ligamentous soft-tissue. Specifically, the ability to define tension only elements with user specified stiffnesses and intercepts is absent. However, by combining multiple COSMOSMotion functions in a single parametric expression, the tension in each ligament can be expressed in terms of the element displacement and its user defined stress free length. This relationship was input into COSMOSMotion through the use of the FORTRAN computing language which is readily understood by the SolidWorks software package.

In general, ligaments were modeled as straight line elements initially tensioned through either the application of an in situ strain or with a defined in situ stress free length. Those ligaments approximated by elements with a defined in situ strain were the simplest and most common soft-tissue constraint used in the model. Herein, the origin and insertions for a given tissue were identified according to section 4.2, and COSMOSMotion result expressions were used to track the relative distance between the two points in SolidWorks 3-D space. A straightforward logic function then applied the desired tension
to the element based on the distance, stiffness, and strain of the ligament in the model. The
generalized form of this equation is given in equation 10.

\[
T = \text{IF}(\text{ActualLength}, L) \cdot 0.0, -S \cdot (\text{DM}(P_1, P_2) - L_0) - 0.1 \cdot V(R(P_1, P_2))
\]

(10)

This expression states that the tensile force in the ligament (T) is defined in terms of the
absolute magnitude distance between origin and insertion points (DM (P1,P2)) minus the
predefined stress-free length (L0) for lengths below, equal to, and above the stress free
length of the element.

Therefore,

\[
T = \begin{cases} 
0 & L - L_0 < 0 \\
0 & L - L_0 = 0 \\
- S \cdot L & \text{for ligament lengths } L - L_0 > 0 
\end{cases}
\]

(11)

Where, S is the published stiffness of the material in N/mm and L is the actual distance
between insertion and origin as defined by

\[
L = \text{DM}(P_1, P_2) - L_0
\]

(12)

The stress free value denoted by L0 was calculated for ligament structures with in situ
strains. These structures included the interosseous membrane, the annular ligament, and
the distal radioulnar ligaments. The magnitude of this value was derived using published
strain values applied to the Lagrangian deformation of the ligament element. Where,

\[
L_0 = \frac{L}{1 + \varepsilon}
\]

(13)

L is the actual distance separating the origin and insertion points in space and \( \varepsilon \) is the in
situ strain of the material from published cadaveric data.
For ligaments with an observed in situ stress free length, L0 was measured directly as the strain line distance between origin and insertion with the ligament in its stress free orientation. Ligaments defined in this manner include those of the LCLC and the MCLC. For example, Regan et al (1991) observed that the LUCL is lax through flexion angles to \( \sim 107^\circ \) for an elbow with no varus or valgus tilt. Thus, the L0 for the LUCL was measured with the model positioned in 107° of flexion. [99] In order to avoid the ringing associated with a lax ligament element suddenly becoming taut, the origin to insertion straight line distance was measured only after the model had equilibrated in its tensioned state. This configuration was then iteratively solved with incremental increases in stress free length or in situ strain in order to ensure that the model was evenly constrained by all elements.

A final component in the FORTRAN expression defining ligament function was the inclusion of a damping element resisting the rapid development of ligament tension. This force was applied through the relative velocity (VR) term within the expression given in equation 10. Herein, the velocity was tracked between P1 and P2 and a resistance to motion applied with a damping coefficient equivalent to 0.1 N*s/mm. Though this term does lend viscoelastic behavior to each ligament element, it was not intended to approximate physiologic time dependent behavior. Furthermore, the VR term does not alter the linearity of the FORTRAN expression given. Therefore, since the RBM results reported reflect the equilibrium state, the final magnitude of tension developed within each ligament was equivalent regardless of the value any damping coefficient.
5. CORONOID PROCESS STUDY

5.1. OVERVIEW

The triangulated rigid body elbow model was first validated against a cadaveric experiment conducted by Hull et al (2005) entitled the “Role of the coronoid process in varus osteoarticular stability of the elbow.” [55] Herein, intact cadaveric elbow joints with incremental levels of coronoid resection and a range of flexion angles were subjected to a varus displacement. The resistance to that displacement was quantified and reported as a percentage of the intact case. In this way, Hull et al demonstrated diminished varus stability for elbows with traumatic deficiencies in the coronoid process, with greater than 50% resections resulting in statistically significant decreases in stability throughout the range of 30° to 120° elbow flexion.

5.1.1. Hull et al Experimental Method

The protocol used in this cadaveric experiment involved eleven fresh frozen cadaveric upper extremities which were disarticulated at the glenohumeral and radiocarpal joints; mean age of the specimens was 66.2 years. Muscular and cutaneous soft-tissue was dissected away from the specimen in addition to the medial and lateral ligament complexes as well as portions of the anterior and posterior capsular tissues. All other soft-tissues
were preserved, including the annular ligament, the interosseous membrane of the forearm, and the TFCC, as well as the tendinous insertions of the biceps, brachialis, and triceps muscles. The humeri were transected 12cm proximal to the medial epicondyle and potted in polyvinyl chloride (PVC) pipe secured in poly-methylmethacrylate (PMMC). The potted humeri were then attached via a custom clamping mechanism to an angular positioning platform that oriented the humeral shaft horizontally with the medial epicondyle oriented vertically. A goniometer was used to align the humeral and ulnar shafts in 30°, 45°, 60°, 90°, and 120° of flexion. An aluminum bracket was mounted to the medial aspect of the distal ulnar shaft approximately 18cm from the medial epicondyle. A shoulder bolt was passed through this bracket such that it was oriented parallel to the ulnar long axis. The entire apparatus was then positioned in a biaxial servo-hydraulic materials testing machine (MTA) (Instron model 1321; Instron Corporation, Canton, MA) by means of a spherical rod end bearing attached to the shoulder bolt approximately 19.5cm from the medial epicondyle. This connection allowed unconstrained axial rotation associated with pronation/supination as well as distraction/compression of the ulnohumeral joint space while preventing changes in flexion angle and controlling varus/valgus extension of the forearm at the rod end bearing. An S-beam load transducer (Omega Engineering Inc, Stamford, CT) was positioned between the rod and MTA actuator in order to capture the force induced by displacement. [55]

At each flexion angle, the varus/valgus tilt of the specimen was adjusted until the load at the transducer read zero, which was defined as the neutral position of the arm, i.e. 0cm of varus displacement. Monofilament lines were used to route the biceps, brachialis,
and triceps lines of action from their insertions through eye bolts on the PVC pipe at 4, 2, and 2cm from the center, respectively. These distances were chosen so as to accurately and reproducibly replicate the various moment arms of the elbow musculature. Muscle tension was achieved by means of a 4kg hanging weight for the triceps and a 2kg hanging weight for the biceps and brachialis each attached to the appropriate muscle monofilament line. The MTA was then displaced to 1cm at a rate of 15mm/min and the equilibrated resisting load was recorded. Subsequent tests involved incremental resection of the coronoid process at 25%, 33%, 40%, 50%, 75%, and 100% of the total coronoid height as measured by digital calipers from the deepest point of the greater sigmoid notch to the tip of the coronoid.

Incremental resections of the coronoid process were created by guiding a rotary reciprocating saw blade along the anterior face of a custom cutting jig. This jig was comprised of a small cuboid block with holes threaded for bolts attaching it to the U-shaped void in a second larger block. In this way, the smaller block served as a cutting deck with variable height adjustment relative to the fixed lower block. This lower block was placed adjacent and aligned perpendicular to the coronoid process by means of two threaded guide pins inserted along the proximal ulnar shaft on the medial aspect of the bone. Digital calipers were used to measure the total height of the coronoid process from the tip of the bone to the deepest point in the greater sigmoid notch in order that the anterior surface of the jig’s lower block could be set flush with this lower limit of the feature. The total height of the cutting jig’s adjustable anterior face was then modified to attain the desired coronoid resection.
5.2. METHODS

5.2.1. Overview

The experimental procedure described by Hull et al was replicated in the 3-D space of the SolidWorks model (seen in Figure 5.2-1). Only those soft-tissues retained in the cadaveric study were enabled in the computational model. Constant magnitude force values of 19.6, 19.6, and 39.2 N were assigned to represent the tension of the 2, 2, and 4 kg weights freely hung on the biceps, brachialis, and triceps monofilament lines, respectively.

Constraining load was recorded for all seven coronoid process resection states at each of the five flexion angles ranging from 30° to 120°. This resulted in a total of 35 simulations for the model, just as in the cadaveric experiment. Since all data reported in Hull et al
regarding constraining load reflected the equilibrium state, time dependent aspects of the cadaveric experimental protocol were not replicated. Specifically, Hull et al applied 1cm of varus displacement at a rate of 15mm/min whereas the computational model was displaced at a rate of 1cm/sec. This rate was chosen through trial and error as that which would not disrupt model stability but could limit overall simulation time to a minimum while still preserving the validity of the solution reported. [55]

5.2.2. Materials Testing Apparatus

The Instron materials testing apparatus and positioning assembly was replicated within the model through three simple parts created in SolidWorks, shown in Figure 5.2-2.

Figure 5.2-2: Materials testing equipment utilized in the cadaver experiment (left) and the 3-D MTA used in the model.
The first was a reproduction of the bracket and shoulder bolt assembly affixed to the distal ulnar shaft and was referred to within the model as the MTA Bracket. This part consisted of two L-shaped ends facing one another separated by a cylinder 6.35mm in diameter and 33.3mm long. This part was created by J.P. Fisk and its use is reflected in the data presented by Fisk and Wayne. [37] The MTA Bracket was positioned such that the central axis of the shoulder bolt was parallel with the diaphysis long axis of the ulna at a distance of 0.75 inches from the surface of the ulna. Additionally, the underside of the bracket was constrained to a position parallel with the mediolateral plane of the ulna. In this way, the MTA bracket was fully defined with respect to the ulna. In order to simplify model computation these joints were not actively solved within COSMOSMotion, instead, the part was defined as grounded in reference to the ulna.

The second part, termed the MTA, was a simplified cylinder drawn to a point at one end which was used to represent the load transducer and rod end bearing actuator. The point of this cylindrical rod was affixed to the central axis of the shoulder bolt through the use of the in-line joint primitive which enforced a point-line coincidence between the two parts. The long axis of the rod was defined as being oriented vertically with respect to the MTA Base (described below) at a straight-line distance of 19.5cm from the medial epicondyle. Thus the assembled MTA was unconstrained in all three degrees of rotational freedom allowing pronation/supination movements as well varus/valgus pitching of the cylindrical rod relative to the shoulder bolt long axis. Furthermore, the joint allowed unconstrained axial distraction/compression along the ulnar long axis. Anteroposterior translation was not permitted and thus fixed the flexion angle of the model for a given
simulation. Lastly, mediolateral translation was dictated by a mathematic expression defining the 1cm step and hold in displacement described by Hull et al. [37, 55].

Initially, a more realistic rod end bearing assembly was created that more closely mimicked the spherical joint used in the cadaveric experiment, however, its inclusion unnecessarily increased the model complexity while offering no demonstrable advantage over the idealized joint described above. As a result, the simplified joint primitive was chosen for use in the computational simulations to minimize unnecessary complexity within the model.

Finally, a third part was created within SolidWorks that resembled the PVC pipe used to pot the humerus. The piece was created by extruding a short cylindrical segment perched approximately 75mm atop a square base plate. This part was referred to as the MTA Base and was analogous in function to the pipe, clamp, and positioning apparatus described by Hull et al. In addition to providing clear visualization of the eye bolts and PVC pipe used to pot the transected humerus and orient the muscle lines of action, the MTA Base stem was also used to define the direction of gravity vectors.

5.2.3. Data Acquisition

The mathematic function defining the translational movement of the MTA is given in equation 14. As interpreted by COSMOSMotion, the function prescribes a static hold of the initial position for a period of six seconds after which time a displacement of 1cm/s
was applied in the $+z$ direction (see section 5.2.4) for a period of one second, followed by a second hold for six seconds.

\[ z = \text{STEP} \left( \text{TIME}, 6, 0, 7, 10 \right) \]  

(14)

Here “STEP” is an embedded COSMOSMotion function that approximates the slope between an initial and final value according to a cubic polynomial line regression. The general formula is given by

\[ y = \text{STEP} \left( x, x_0, h_0, x_1, h_1 \right) \]  

(15)

where $x$ is the independent variable being controlled, in this case the time lapse of the simulation (TIME), $x_0$ is the value at which $x$ begins, $h_0$ is the initial value of the ramp, $x_1$ is the end value of $x$, and $h_1$ is the final value of the ramp. The observed result of this function is the dependent variable $y$. Mathematically, this expression is applied according to the following system of equations. [2]

\[ a = h_1 - h_0 \]  

(16)

\[ \Delta = \frac{(x - x_0)}{(x_1 - x_0)} \]  

(17)

\[ \text{STEP} = \begin{cases} 
   h_0 & x \leq x_0 \\
   h_0 + a \cdot \Delta^2 (3 - 2\Delta) & x_0 < x < x_1 \\
   h_1 & x \geq x_1 
\end{cases} \]  

(18)

The force associated with the above ramp in displacement was recorded from the MTA at a rate of 1 point per frame with approximately 100 frames per second. This yielded a “Force (N) x Time (sec)” plot with roughly 1300 points. The actual number of frames could be larger if the GSTIFF integrator had difficulty with the default length of the
initial time step and as a result reduced its value, (see section 4.4.1). These data were exported from COSMOSMotion to the Excel spreadsheet program (Microsoft Corporation, Redmond, WA) for all statistical analysis.

5.2.4. Assigning Coordinate Systems

In order to orient the model in a way that was physiologically relevant to the cadaveric experiment being replicated, the global coordinate system within SolidWorks was transformed from its original arbitrary position in space. A new origin and axis orientation was chosen that would afford the computational model the same x, y, and z directions used by Hull et al. (Figure 5.2-3)

Figure 5.2-3: View showing the orientation of the ISB humerus coordinate system relative to the orientation of the global coordinate system and gravity.
Specifically, the +z direction was oriented vertically, the +y direction directed proximally toward the humeral head, and the +x direction as the common line perpendicular to z and y pointing posteriorly. To accomplish this, the proximal most center of the transected and potted humerus was chosen as a point both easily identifiable in both experiments as well as one that could be defined relative to the grounded humerus part. This point was defined as the global model origin. The global +y and +z were then defined as collinear with the yh and parallel to the zh axes, respectively, as suggested by the ISB and described in section 3.7. [125] Relative positions of the radius and ulna to the humerus were maintained by defining a coincidence between the anatomic planes of each bone of the forearm and those of the humerus. As a result, any transformation applied to the humerus would be automatically extended to the radius and ulna, thus maintaining the scan orientation of the model during positioning.

5.2.5. Coronoid Process Resection

The incremental resections of the coronoid process were achieved by cutting the model geometry described in chapter 3 along a plane perpendicular to the central ridge of the greater sigmoid notch. This best fit plane was defined by the MedCad module of the MIMICS software package as described in section 3.7. The plane perpendicular to the central ridge was therefore defined as a 90° rotation of the MedCad fit plane about the ulnar long axis.
The coronoid cut plane, shown in Figure 5.2-4 was represented by an anterior offset of this perpendicular plane coincident with the lower margin of the non-articular strip of the trochlear notch. This location reflected a complete (100%) resection of the feature. Furthermore, based on this definition, the total height of the coronoid process in the computational model was measured to be 17.13mm. Additional cut planes were defined at larger anterior offsets in order to represent 75, 50, 40, 33, and 25% resections.

5.3. ALTERATIONS TO THE WORK BY FISK AND WAYNE

The study presented by Hull et al was also chosen by Fisk and Wayne (2009) as a means for validating a rigid body model of the elbow. Similar to the model formulation described in Chapters 3 & 4, Fisk and Wayne used the ADAMS solver package native to the COSMOSMotion module within SolidWorks to create a model with dynamic tension-only spring elements representing ligamentous soft-tissue and constrained only by muscle
force vectors, external varus perturbations, and osteoarticular contact forces. This study reported good correlation with the cadaveric results reported by Hull et al with regard to trends observed in initial constraining load, as well as resistance to load at 1cm of varus across flexion angles and coronoid resection levels. Additionally, the computational model was able to quantify tension in each soft-tissue element as well as rigid body contact within the joint. As a consequence, Fisk and Wayne were able to show general agreement between the model and cadaveric behavior suggesting that beyond 50% coronoid resection varus elbow stability is dramatically diminished, especially at lower flexion angles. [37, 55]

While similar in its formulation, the model described above differs in a number of important ways from the work of Fisk and Wayne. Foremost is the use of triangulated surfaces to represent bony geometry in the present model versus the Initial Graphics Exchange Specification (IGES) contour files used in the study by Fisk and Wayne. While accurate in their representation of the bony surfaces along the diaphysis of the bones, these IGES files were unable to automatically represent closed contours such as would be present at the articular epiphyses of a bone. Furthermore, in areas with multiple parallel protuberances or “branches,” the IGES format was unable to automatically interpolate a surface represented by divergent polylines. [37]
Therefore, each branch needed to be masked and isolated separately, then manually stitched together to form a cohesive surface. Thus, the creation of a solid body based on IGES contour lines required a great deal of manual surface editing to approximate the native bony surfaces (see Figure 5.3-1); a step that not only introduced additional unknowns into the creation of the surfaces, but was also a laborious and time consuming task. In contrast, the use of binary STL files provided a generally straightforward process by which a masked MIMICS file could be triangulated according to automatic detection of surface boundaries while still providing pertinent user-definable options to minimize post-processing. Though not without limitations, namely in the total number of facets able to be imported into SolidWorks, this alternative surface representation greatly reduced the
amount of manual editing necessary for creating a high fidelity bony model from a CT
scan data set while still maintaining wide format compatibility across commercial CAD
software packages.

A second concern addressed in the present iteration of the computational elbow
model is the inclusion of gravitational force vectors. [37] Though relatively simple to
implement, the application of gravity imparted a force that was unbalanced within the
assembly when applied in the absence of full ligament and muscular tension. Therefore
instability in the model would often arise if the sudden application of the gravitational
force caused the bones of the elbow to shift in their initial position prior to the full
tensioning of various soft-tissue constraints. This was especially true of the triceps and
brachialis force vectors and annular ligament fibers. Indeed, though Fisk and Wayne
showed good correlation between the model and cadaveric experiments in the force trends
of the initial and final constraining loads expressed as percent of intact, they noted that the
computational model often produced generally smaller magnitudes of forces compared to
the cadaveric data. J.P. Fisk addressed this point by including a pilot simulation that did
include gravitational effects on the forearm wherein the model was fixed in 90° flexion
with an intact coronoid, then displaced to 1 cm of varus excursion by the MTA actuator.
He noted that when gravitational forces were considered, the mean varus constraining
force increased from 4.41 N to 7.53 N as compared to the 7.71 N reported by Hull et al.
[37, 38, 55] In contrast to work described by J.P. Fisk, the computational model outlined
in Chapters 3 & 4 does consider the effects of gravity. Vectors representing this force
were applied at the centroid of each of the bones of the forearm and oriented relative to the
stationary MTA base and thus depict a more realistic recreation of the cadaveric experimental setup. The inclusion of this force imparted a larger magnitude varus resistance at both the initial 0cm varus excursion as well as at 1cm displacement.

A third difference involved the iterative pretensioning of the all of the ligaments of the forearm. This was done to ensure that any unphysiologic articular space was removed between the proximal and distal radioulnar joints prior to assigning in situ strain values. This step was necessary because the in situ strain values reported in the literature for each of the ALS, IOM, and DRULs reflect a radius and ulna in constant articular contact throughout the range of pronation/supination. Thus, it was important that any voids between the bones caused by the lack of articular cartilage or as an artifact of the triangulation of the bony surface did not cause the ligaments’ stress free length to be overestimated. The procedure through which this pretensioning was accomplished involved initially arranging the forearm in -80° of supination. Then the literature reported values for in situ strain in each ligament element were assigned and the model was allowed to run without any other force or perturbation present. Any movement of the radius relative to the ulna was tracked by means of the radial coordinate system defined in section 3.7, and the resulting distance between the origin and insertion of a given ligament element was used to calculate the new stress free length of the element for the following iteration. This process was stopped once both proximal and distal radioulnar contact forces were non-zero and the tension values reported in each of the ligament elements corresponded to the appropriate tension that would be developed given the known stiffness and in situ strains.
A final difference of note between the model formulated by Fisk and Wayne, and the one described here was the application of a user defined coordinate system to orient the model. [37] Though the previous model used a similar methodology for identifying landmarks on each bone, the coordinate systems based on those landmarks could only be considered local to the specific part file, in this case the single bone from which they were derived. That is to say that the global coordinate system was not manipulated in such a way as to reflect design intent. A consequence of this was the necessity that all data exporting made regarding the load resisting varus displacement observed at the MTA had to be made relative to the that part’s local coordinate system. While this is a valid approach, it makes comparisons of forces from separate parts difficult and introduced additional steps in the data acquisition process. By contrast, the present model uses the coordinate system of the MTA base to orient the entire elbow model and MTA relative to a single grounded part. In terms of design, the positions of all parts could be defined in terms of the global coordinate system and all forces can be reported in both their magnitude as well as resolved xyz components in a coordinate system that is common throughout the model.
5.4. RESULTS

The varus restraining load measured at the MTA during its +z translation was tracked during the 13 second simulation setup described in section 5.2.3. Though the magnitude of this load varied across flexion angles as well as across coronoid resection levels, certain behaviors were consistent across testing configurations. Initially during the first few simulation frames, the MTA magnitude varied greatly under constant 0cm displacement due to short-lived high frequency oscillations within the model. These were typically followed by a damped decay in force magnitude to a steady-state magnitude by second 2. This steady-state would continue until second 6, at which time the prescribed varus translation of 10mm/sec for one second caused the force to steadily raise at the MTA reaching its maximum force at 7 seconds. Following this, the system showed a damped decay to the steady-state 1cm varus constraining load by second 8-9. A typical curve is given in Figure 5.4-1.
Figure 5.4-1: Typical example of the MTA constraining load measured in the model from 0-1cm varus excursion. Ramp in displacement occurred from second 6-7.

Just as was reported in the cadaveric experiment conducted by Hull et al, comparisons of coronoid resection states within flexion a given flexion angle were expressed as a percent of the intact (0%) resection state. Thus, by definition, 0% resection was plotted at 100% of the intact case. Results for the MTA constraining load at 1cm for both the cadaveric experiment as well as the computational model are given for each of the five flexion angles in Figure 5.4-2 - Figure 5.4-6. Plot points for the cadaveric experiment represent an average across multiple specimens and are thus presented with error bars representing a single standard deviation.
Figure 5.4-2: Constraining load at 1cm varus vs. coronoid resection at 30° flexion.

Figure 5.4-3: Constraining load at 1cm varus vs. coronoid resection at 45° flexion.
Figure 5.4-4: Constraining load at 1cm varus vs. coronoid resection at 60° flexion.

Figure 5.4-5: Constraining load at 1cm varus vs. coronoid resection at 90° flexion.
Constraining load values predicted by the model correlate very well with those reported by Hull et al for all flexion angles and resections at or below 50%, with every value falling well within a single standard deviation. Across these resection levels, the constraining load measured within the model was relatively constant or decreasing by less than 10% of the intact case at angles up to 90° and decreasing by approximately 18% at 120°. At 75% resection, both the model and the experimental data showed a decrease in constraining load yet the model underestimated this reduction relative to the cadaveric data for all flexion angles, resulting in constraining load percentages more than double the value of the standard deviation for 45° and 60°, but less than 10% more than the standard deviation for 90°. Values for 30° and 120° were within a single standard deviation. For
the 100% resection case, both the model and experimental data again showed a further decrease in constraining load as a percent of the intact case. These predicted values were within a single standard deviation for flexion angles of 45°, 60°, and 90° and were just below and above for angles 30° and 120°, respectively.

Shown below in Figure 5.4-7 are the percent constraining load values for each coronoid resection value average across all five flexion angles. Here, it can be seen that the model was able to predict the percent constraining load well within a single standard deviation across all values of coronoid resection with a great deal of accuracy with the exception of the 75% cut case wherein the model generally underestimated the loss in varus constraining load at just over a one standard deviation increase over the experimental case. The average magnitude of the constraining load at 1cm varus was generally higher than those reported by Hull across all resection values by an average of 4.98N up to 50%, and 1.1N and 0.7N at 75% and 100% resections, respectively.
Figure 5.4-7: Constraining load at 1cm varus averaged across all flexion angles vs. coronoid resection for the cadaver study (blue) and model (pink).

In addition to the MTA varus constraining force magnitude, many other simulation parameters were tracked and plotted in order to fully explain the behavior of the model during RBM simulations. These included soft-tissue and contact constraints not measured experimentally, such as ligament force magnitude for all tension-only soft-tissue elements and magnitude of rigid body contact forces at the ulnohumeral, proximal and distal radioulnar, and radiocapitular joints.

Shown in Figure 5.4-8, the ulnohumeral contact force magnitude was plotted for each of the 35 testing configurations prescribed by Hull et al. These data show the initial force magnitude at 0cm varus and within resection levels up to 50% to only slightly
decrease as the elbow moves through higher degrees of flexion. This trend supports the widely held description of the intact ulnohumeral articulation as providing elbow stability throughout the entire range of flexion. However, at resection levels of 75% and 100%, the contact force was significantly diminished for lower flexion angles from a minimum at 30° to near normal values at 90°. This suggests that the portion of the ulna responsible for most of the ulnohumeral contact at lower flexion angles is located near the base of the coronoid process, whereas at high angles of flexion, the site of predominate ulnohumeral contact is outside of the coronoid. Contact force magnitude values for the radiocapitular joint interactions given in Figure 5.4-9 support these assertions about the site of ulnohumeral contact. Herein, there was zero force developed in the radiocapitular joint of the intact elbow at any flexion angle; however, at coronoid resection values greater than 50%, this contact force increased dramatically, with 100% resection causing more radiocapitular contact than the 75% case for all contacting flexion angles except 45°.
Figure 5.4-8: Ulnohumeral contact force developed within the model vs. coronoid resection level at 0cm varus excursion for each flexion angle.

Figure 5.4-9: Radiocapitular contact force developed within the model vs. coronoid resection level at 0cm varus excursion for each flexion angle.
Lastly, each of the EJC contact forces was plotted as a percent of the total EJC forces as a confirmation of the model’s predictions that with increased levels of coronoid disruption, it is only the location of the contact forces that are changing and not the total sum of elbow forces. This plot, given in Figure 5.4-10, shows that prior to varus displacement; ulnohumeral articulations were the predominant stabilizing contact within the joint and were responsible for approximately 90% of the EJC total, while the radiocapitular and proximal radioulnar were responsible for approximately 10% and 0%, respectively, at coronoid resections through 50%. At coronoid resections of 75%, the ulnohumeral contribution to overall contact force decreased from 90% to less than 70% of the total force developed in the joint. Concurrently, there was an increase of approximately 20% in the contribution of the radiocapitular articulating force and no change in the proximal radioulnar contribution. At full resection of the coronoid, ulnohumeral contact contribution increased slightly to 75% of the EJC contact total, representing about a 4.1% increase over the 75% coronoid resection state; however, at this same resection level, the radiocapitular component of the total EJC force also increased by 7.2%, with the 12% increase in humeral articular contact being offset by a zeroing of contact force in the proximal radioulnar joint. Finally, Figure 5.4-10 also shows the relative force contributions of each these three EJC articulations after 1cm of varus excursion. As expected, through 50% coronoid resection, there was no change in force distribution from the 0cm varus displacement line. At 75% coronoid resection, wherein the radiocapitular articulation force accounts for a larger percentage of overall EJC contact force, the application of a varus displacement transferred stabilizing load away from the
radial head and back to the ulna. At 100% coronoid resection, ulnohumeral and radiocapitular contact forces remain constant during varus displacement.

Figure 5.4-10: Relative percentages of total joint contact force vs. coronoid process resection level. Contributions from ulnohumeral (UH, pink), radiocapitular (RH, red) and ulnoradial (UR, blue) articulations are shown as solid and dashed lines for 0cm and 1cm varus, respectively.

The force developed in each tension-only ligament element was also tracked for each of the 35 testing scenarios. The total ligament tension was calculated as the sum of its constituent fiber forces, thus the total annular ligament tension was the sum of its four fibers, the DRUL force was the sum of its dorsal and palmer fibers and the total IOM
tension was the sum of the two CB, two AB bands, and single DOB band. However, because of the near orthogonal line of action of the IOM-DOB relative to that of the IOM-CB and -AB, additional calculations were made that further differentiated the IOM into two groups. The proximal group consisted of the IOM-CB and IOM-AB whose fibers run obliquely from the proximal radius to the distal ulna, and the distal group consisting only of the IOM-DOB whose single element runs obliquely from the proximal ulna to the distal radius. The total tension developed in each ligament structure averaged across all coronoid resection values, both before and after 1cm varus displacement, is shown in Figure 5.4-11. Here, both the ALS and DRUL ligaments showed an increase in their tension prior to varus displacement as the flexion angle increased. The IOM also showed an increase in total force with increasing flexion; however, maximum tension was developed at 90° flexion, beyond which the IOM developed decreased tension equivalent to the 45° flexion state. For 60° flexion and above, the IOM and DRUL showed increases in tension after varus displacement, the ALS ligament demonstrated this same increase in tension due to varus displacement for flexion angles of 90° and 120°.
Figure 5.4-11: Average magnitude of ligament tension in the IOM (blue), ALS (green), and DRULs (orange) vs. flexion angle. Note that the dashed and solid plots indicate tension at 0cm and 1cm of varus, respectively.

There was a clear inverse relationship between the tension developed by the proximal group of IOM fibers, including the –CB and –AB bands, and those of the IOM-DOB as the level of coronoid process resection increased. As shown in Figure 5.4-12 and Figure 5.4-13, for flexion angles at and below 60° and coronoid resections through 50%, the fibers of the IOM-CB and IOM-AB were tensed at approximately 24N. Beyond 50% coronoid deficiency, there was a sharp decrease in the tension developed by these structures with lowest flexion angles developing lowest final tension. At 90° of flexion and coronoid resections up to 75%, the initial tension developed by the IOM-CB and IOM-
AB was approximately 19N. At full coronoid resection, the total ligament tension decreased significantly but still remaining higher than any of the lower flexion angles. At 120°, the ligament developed only approximately 12N and was generally insensitive to coronoid deficiency through 75% resection, at 100% resection; there was a small increase in total tension.

Figure 5.4-12: Magnitude of tension developed in the IOM-CB/AB vs. coronoid resection level.

For the IOM-DOB band of the IOM, the trends observed in the –CB and –AB tension were generally reversed. At coronoid resections through 50%, the fibers of the IOM-DOB developed increasing tension ranging from 7.8, 10.1, 12.1, 15.1, and 17.6N for each flexion angle from 30° to 120°. Beyond 50% coronoid deficiency, there was a sharp
increase in the tension developed by this structure for all flexion angles with the 60° angle resulting in the highest tension at 100% resection. The increase in tension from 50% to 100% resection appeared linear for each flexion angle with the slope growing less steep as the flexion angle increased, with the exception of the 45° and 60° slopes which were reversed.

Figure 5.4-13: Magnitude of tension developed in the IOM-DOB vs. coronoid resection level.

The total elapsed solve time for each approximately 1300 frame rigid body simulation was recorded based on the elapsed CPU clock time rounded down to the nearest minute. There was a trend of increasing solve times across flexion angles as well as across coronoid resections beyond 50%. Times for all 35 simulations ranged between 5 and 31
minutes, with setups involving 50% or less coronoid resection averaging less than 6.5
minutes, and simulations of 75% and 100% resection each averaging less than 14 minutes.

5.5. DISCUSSION

The musculoskeletal rigid body model of the elbow developed in Chapters 3 & 4 was able to predict with great accuracy the constraining load of a varus loaded elbow when subjected to the testing parameters outlined in the cadaveric experiment conducted by Hull et al. [55] Following 1cm of varus displacement at the distal ulna, the constraining load of the MTA was observed to drop off significantly for those elbows with more than 50% resection of the coronoid process of the ulna in agreement with cadaveric results. Additionally, the computational model was able to show that with increasing elbow flexion, there was decreased sensitivity to coronoid resections beyond 50% as was described by Hull et al. Finally, the average magnitudes of varus constraining loads for each resection level were predicted to within a single standard deviation of the cadaveric data. For the 100% coronoid resection level the average constraining load was predicted to be diminished to 21.5% of the intact (0%) coronoid case as compared to 23.2% reported in the cadaveric study.

In addition to predictions concerning the constraining load measured by the MTA, the computational model was able to further quantify observations by Hull et al regarding mode of failure in elbows with large coronoid deficiencies of 75% and 100%. Herein, the
authors described the gradual migration of load from the ulnohumeral to radiocapitular joints following muscle loading but prior to varus displacement. The computational model was able to show this transition as a percent of the total EJC contact force for all flexion angles below 120°. With respect to relative joint contact forces, the computational model was able to predict values of ulnohumeral contact force percent of total close to those values predicted by Markolf et al [71] wherein it was demonstrated that the ulnohumeral articulation joint is responsible for 93% of the axial load bearing of an intact elbow in varus alignment. In addition to the relative percentages, the model predicted total joint contact magnitudes that were physiologically reasonable (Figure 5.4-8 & Figure 5.4-9) given the 80N collective muscle load prescribed by Hull et al, though no direct comparison to the cadaveric data experiment was possible. As for the predominant location of contact within the ulnohumeral articulation, the model showed a high sensitivity to coronoid resections of 75% and higher wherein ulnohumeral contact force would drop off significantly in these corresponding injury states. Thus it can be inferred that the location of primary ulnohumeral contact within the model was at the base of the coronoid process within the sigmoid notch. This location is coincident with those generally accepted in the literature. [50, 71, 78, 81, 92]

The tension-only force expressions used to represent ligament tension within the model provided a means of quantifying the soft-tissue contribution to varus stability through load transfer between the radius and ulna. During low flexion angle simulations with coronoid resections of 75% and 100%, ulnohumeral contact force diminished drastically while radiocapitular force increased sharply. This pattern of loading suggests
that at low flexion angles and with a highly disrupted coronoid, the normally compressive component of the triceps muscle becomes distracting. Thus, in the absence of the full coronoid process, radiocapitular abutment and forearm ligament tension provide the majority of the joint resistance proximal migration of the ulna relative to the radius. Of the ligament structures modeled, the IOM provided the most force transmission between bones as well as the primary stabilizer of proximal migration of the ulna relative to the radius in simulations with large coronoid resections, though the DRULs also saw a significantly increased tension for these cases. It can be seen in Figure 5.4-12 and Figure 5.4-13 that at coronoid resection levels above 50%, there was a slackening of the IOM-CB and -AB fibers accompanied by a tensioning of the IOM-DOB. Given that the IOM-CB and –AB have lines of action oriented from the proximal radius to the distal ulna, these fibers were best able to resist the proximal migration of the radius relative to the ulna such as might be seen for elbows in varus alignment when the radius is axially compressed. They were however, inefficient stabilizers of proximal ulnar migration relative to the radius as was demonstrated within the model. Under conditions wherein the coronoid is heavily disrupted and thus the trochlea is not anteriorly buttressed, proximal migration of the ulna relative to the stationary radius caused a decrease in strain within the proximal bands of the IOM. In contrast, the proximal ulna to distal radius orientation of the IOM-DOB caused this element to develop increased tension and thus allowed it to most efficiently limit just such a dislocation. Therefore, within the model, as the level of coronoid resection increased for elbows at or below 90° of flexion, the predominant soft-tissue stabilizer limiting proximal migration of the ulna was the IOM-DOB. This prediction was in
agreement with the conclusions of Schneiderman et al (1993) and Wantanabe et al (2005) that concluded that fibrous portions of the distal IOM, directed oblique to those of the central band, acted as secondary stabilizers limiting proximal movement of the ulna relative to the radius in forearms with Galeazzi fractures of the radius. [101, 120]

These results are generally in good agreement with the simulations performed by Fisk and Wayne (2009) regarding the varus constraining load recorded for the various injury states. Furthermore, this model showed a very similar trend for increasing proximal migration of the ulna relative to the radius as well as increased radiocapitular contact force for elbows with more than 50% coronoid deficiencies. However, it was noted that the predominant soft-tissue elements resisting this proximal migration were not consistent between the models. While Fisk and Wayne saw a marked increase in ALS tension with increased coronoid resection with no significant change in IOM or DRUL tension, and thus identified the ALS as the chief soft-tissue constraint in the disrupted coronoid elbow, the current model developed significant tension within the IOM and DRULs, and only minor increases in ALS strain for the same simulations. Likely causes for this discrepancy are the inclusion of the IOM-DOB element within the current model. Though not significantly different in terms of stiffness or strain from any of the structures modeled by Fisk and Wayne, the IOM-DOB is oriented in a manner that was not represented in any previous work. Furthermore, the iterative pretensioning of the ligaments of the forearm described in section 5.3 was not employed previously. This step caused the current model to consistently develop significantly tension in all of the bands of the IOM as well as those of
the DRUL whereas Fisk and Wayne often reported seeing no tension in the distal portions of the IOM and smaller magnitude tensions in the DRULs.

Possible sources of error within the model include the use of an idealized SDA joint based on passive joint motion to predict elbow location. While this method is supported in the literature, extrapolating the average SDA based only on two scans could have introduced position errors if the cadaveric specimen was not positioned precisely in neutral carrying angle when scanned. Another possible source of error is the use of computed tomography to create the 3-D surface geometry of the bones. Though superior to MRI for detection of bony surfaces, the use of this technology prevented the detection of articular surface cartilage within the joint. While this soft-tissue is generally less than 1mm thick, it is possible that fluctuations in the surface thickness could have altered the position and orientation of the average SDA calculated.
6. TERRIBLE TRIAD STUDY

6.1. OVERVIEW

Following validation against the cadaveric experiment conducted by Hull et al, the computational elbow model was subsequently applied to a more complex study conducted by Fern et al (2009). [34] This second study entitled “Complex varus elbow instability: A terrible triad model” used the catastrophic, but less common, terrible triad injury model to investigate the interplay between soft-tissue and osseous constraints in an elbow under varus load. Specifically, the injury involves radial head fractures resulting in complete resection, complete ruptures of the LUCL and variable levels of coronoid process fracture. Accordingly, each cadaveric specimen was subjected to a varus excursion at the distal forearm under the conditions of an intact, resected, and repaired radial head, an intact, resected, and repaired LUCL, and seven levels of coronoid process resection ranging between intact and 100%. The force resisting this excursion was measured for each injury state at both 30° and 90° flexion.

Thus, Fern et al were able to show that neither radial head resection nor LUCL rupture alone reduced varus elbow stability when the coronoid process is preserved. Also, under conditions of deficient radial head and LUCL but intact coronoid, they observed that the varus stability of the joint was diminished from the intact state. At coronoid resections
beyond 50%, neither LUCL repair nor radial head prosthesis alone conferred additional varus stability to the joint. At coronoid resections equal to or above 67%, varus stability was diminished regardless of LUCL and radial head repair. These results support the findings by Hull et al that the coronoid is the predominant osseous stabilizer of an elbow loaded in varus. [55] Additionally, they suggest that the LUCL is the primary varus stabilizer of the joint for coronoid injuries involving less than 50% resection. At coronoid resections beyond 50%, Fern et al showed that the varus stability of the joint will be reduced regardless of repair to LUCL and radial head structures. Finally, these data show that isolated resection or repair of the radial head does not alter varus stability in the otherwise intact joint. [34]

6.1.1. Fern et al Experimental Method

The experimental setup described by Fern et al utilized ten fresh frozen cadaveric arms with a mean age of 58 years, which were disarticulated at the glenohumeral and radiocarpal joints, preserving the ligamentous structures of the TFCC. Muscular, ligamentous, and cutaneous soft-tissues were preserved around the elbow joint with the exception of the biceps, brachialis, and triceps muscles which were excised at their respective musculotendinous junctions. The humerus of each specimen was transected mid-diaphysis, approximately 12cm proximal to the medial epicondyle and potted in a 1.5in-diameter polyvinyl chloride (PVC) pipe using poly-methylmethacrylate (PMMC).
Prior to placing the specimen into the materials testing machine apparatus, the appropriate injury state was replicated in the joint. First, the ulnohumeral articulation was accessed through a medial incision which allowed visualization of the coronoid process. The cutting deck of the custom jig employed by Hull et al (described in section 5.2.5), was aligned with the tip of the coronoid process and secured to the medial ulna by means of two threaded guide pins passed into the bone. [55] Because capsuloligamentous soft-tissues were not removed from around the joint, the deepest point of the greater sigmoid notch was not able to be visualized. Therefore, Fern et al employed a small Kirschner wire passed into the medial ulnohumeral articular space in order to gauge the depth of the feature. Incremental resections of the coronoid process were made at 25, 40, 50, 67, 75, and 100% of the total height using a rotary reciprocating saw.

Lateral structures of joint were accessed through the Kocher approach at the posterolateral aspect of the elbow. [34, 123] For both of the injury states involving the LUCL, the ligament was cut at its origin on the humerus and stitched to a length of braided microfiber suture. The ruptured LUCL state (Lx) was simulated by leaving the ligament unrepaired and therefore under no tension. For the repaired state (Lr) this suture was passed through a small diameter hole drilled through the mediodistal humeral head, beginning at the isometric point of the joint on the lateral side of the capitulum. The specimen was manipulated into 90° of flexion after which the ligament was tensioned to 25N using a static weight freely hung from the suture at its point of exit from the posterolateral aspect of the lateral epicondyle. [34, 57] The suture was clamped in this position following tensioning and the weight removed prior to testing.
To induce traumas to the radial head, the structure was first accessed through the surgical technique described above. For both the excised (Rx) and repaired states (Rr), the Biomet ExploR (Biomet, Warsaw, IN) modular radial head prosthetic was employed. The modular design of this device was chosen for its ability to fit a large range of radii as well as the ability to separate the articular surface of the implant without removing the stem component once it was driven into the medullar cavity of the bone. Three radial head implant diameters (20, 22, 24mm) were available for each of five radial head heights (10, 12, 14, 16, 18mm). Implant stems were also provided in five sizes (5×22, 6×24, 7×26, 8×28, 9×30mm). Thus, there were 75 total combinations of implant head and stem from which to choose the appropriate size. Once an implant size was selected, the specimen’s radial head was excised transverse to the axis of the radial neck at a point proximal to the radial tuberosity as designated by the Biomet surgical guide for the given implant size. [74] Following excision, the appropriately sized stem component of the implant was pressed into the rasped medullar canal of the bone. Afterwards, the head component was attached to the stem and affixed with a set screw on the anterolateral margin. For the excised radial head state (Rx) all of the above steps were performed; however, the head component of the implant was left unattached. As a consequence of the surgical approach used, all tests reflecting the Rx or Rr states required incising the annular ligament. This structure was not repaired and was thus only intact for those tests involving the intact radial head (Ri).

Once the appropriate injury state had been created, the PVC pipe was clamped to a custom angular positioning apparatus such that the humeral shaft lay horizontal with the medial epicondyle oriented vertically. Flexion angles of 30° and 90° were identified
through the use of a goniometer. Heavy monofilament lines were routed through eyebolts on the PVC pipe at 2, 2, and 4cm from the pipe center axis to the tendinous insertions of the biceps, brachialis, and triceps, respectively. Tensions of 20, 20, and 40N were drawn on the three muscles by means of hanging weights. The mediodistal margin of ulna was attached to a biaxial servo-hydraulic materials testing apparatus (MTA) (Instron model 1321; Instron Corporation, Canton, MA) via a load cell and spherical rod end bearing slid along a lubricated shoulder bolt affixed to the bone as described by Hull et al. (See section 5.1.1) [55] To ensure that each joint was subjected to the same varus moment, the spherical rod was positioned at a straight-line distance of 19.5cm from the medial epicondyle.

Biomechanical testing of the joint consisted of two stages. Initially, the tendon loads were applied and the MTA was adjusted until the load cell read 0N. This position was marked and designated as 0cm varus extension. Second, the MTA was cycled from 0 to 1.5cm of varus at a rate of 15mm/min for 5 cycles. Following completion of testing, resistance to varus extension was recorded at the end of the 4th cycle (specimen at 0cm varus) and at the end of the 5th cycle (specimen at 1.5cm of varus). The sequential testing of excised/repaired LUCL and excised/repaired radial head at each of the prescribed coronoid resection values resulted in a total of 64 tests per specimen.
6.2. METHODS

6.2.1. Overview

Just as with the cadaveric study by Hull et al, the experiment conducted by Fern et al was recreated within the 3-D SolidWorks environment. Each of the injury states described in the terrible triad study was replicated at both 30° and 90° of flexion. Active muscle loads, as well as capsular and distal ligamentous constraints used in the study were likewise represented in the computer model. Passive muscle contributions and other soft-tissue constraints were not included. Rigid body simulations were performed for each of the intact, excised, and repaired states for the LUCL and radial head as well as the seven coronoid resection levels, yielding 62 of the 64 testing scenarios described by Fern et al. The duplicate tests of the fully intact state performed at 30° and 90° with and without sham surgeries were not replicated as discussed below in section 6.2.5. Additionally, intact LUCL (Li) and intact radial head (Ri) injury states with variable coronoid resections were modeled, though such tests were not physically possible in the study by Fern et al. [34]

6.2.2. Materials Testing Apparatus

The MTA base, bracket, and rod parts described in section 5.2.2 were again utilized to maintain the position and orientation of the model in 3-D space, as well as to apply the varus displacement prescribed under Fern et al. Likewise, because the study performed by Fern et al used the same specimen orientation as that presented by Hull et al, the computational model was assigned the same global coordinate system used previously.
Herein, the long axis of the humerus, as represented by a line connecting the sphere fit center of the humeral head and the midpoint of a line between the medial and lateral epicondyles, was oriented horizontally with the proximal portions of the bone pointing in the +y direction. Also the MTA and MTA Base vertical axes were made parallel with the +z direction of the assembly as defined in section 5.2.4.

6.2.3. Data Acquisition

The translational motion of the MTA was again governed by a COSMOSMotion expression whose form was nearly identical to that described in section 5.2.3, differing only in the total magnitude of the varus displacement. This value was set to 1.5cm in agreement with the study conducted by Fern et al thus yielding the revised function given below in equation 19.

\[ z = \text{STEP} \left( \text{TIME}, 6, 0, 7, 15 \right) \]  

The application of this function resulted in a 13 second total simulation time wherein the translation of the MTA was held static at 0cm varus for six seconds, followed by a +z displacement (see section 5.2.3) of 1.5cm/s for one second, and ended with a second static hold for six seconds. Though both the 3-D contact parameters and ligament expressions used within the COSMOSMotion modeling package display time-dependent behaviors, these were not intended to accurately represent native viscoelastic properties of the various soft-tissues in the arm. Accordingly, the cyclic application of varus
displacement and relatively slow rate of MTA movement prescribed by Fern et al were not replicated within the model.

The force resisting this varus displacement was calculated for each of the approximately 100 frames/second of the simulation, resulting in roughly 1300 data points. These data were exported to Excel where the average equilibrium values for the initial constraining load, load at 1.5cm varus, and change in load from 0 to 1.5cm varus were calculated for each of the tested injury states.

6.2.4. Biomet ExploR© Prosthesis Creation

The 3-D reconstruction of the ExploR prosthesis used in the study by Fern et al was created within SolidWorks in agreement with size and material specifications gleaned from Biomet product releases and surgical technique guides. [74] Additionally, fitment guidelines provided within the surgical guide were used to decide what implant head diameter and stem size were appropriate for the size of the model radius. Given that the native radial head of the computational model specimen was approximately 22mm on its elliptical major axis, the 22mm diameter by 12mm tall radial head implant was chosen to represent in the model. Likewise, based on the overall length and diameter of the radius within the model, the 5mm wide by 22mm long implant stem was chosen as the most appropriate size to model.
The implant itself consisted of three separate parts joined together in an assembly as shown in Figure 6.2-1. The first part created was the implant stem which was comprised of a 22mm cylindrical shaft drafted inward at approximately 5°, terminating in a half-sphere cap. The base of the stem was affixed to the center of the stem flange, a flat circular disc 2mm thick and 17.05mm in diameter oriented such that the long axis of the stem was normal to the bottom face of the disc. Secured to the opposite face of the flange was the T-shaped channel lock onto which the implant head was attached.

The implant head was the second part created. Unlike the stem however, this part was comprised of many complex curves that could not be easily extrapolated from the product descriptions available from Biomet. To overcome this, an image of the entire ExploR assembly, taken from the sagittal aspect was uploaded into Adobe Photoshop 7.0 photo editing software (Adobe Systems, San Jose, CA). Once imported, the photo was scaled according to the known stem flange thickness of 2mm. [74] Then, using the outline
of the radial head as a template, a freeform 3-D spline was fit to the photo and saved as a “work path.” This outline was then imported into SolidWorks where it was incorporated into a profile that was revolved 360° about its central axis, thus recreating the 22mm radius implant’s outer shape. The depth of the radial head’s articular surface was created by cutting a paraboloid indentation into the superior surface of the implant with a total depth at center of 2.4mm. This depth value was arrived at through depth measurements of the physical ExploR implant as well as published anatomical studies on the average size of the feature. [59, 74, 116] Finally, a T-shaped channel was cut from the medial aspect of the structure such that it accepted the lock extruded on the surface of the stem flange. The last component of the assembly created was a surgical set screw used to affix the head portion of the implant to the stem. Though not necessary from a rigid body modeling perspective, this part was useful for orienting the implant assembly in a manner consistent with its clinical application. As such, the implant screw and screw recess were positioned facing outward in an anterolateral direction where they would not interfere with the radioulnar articular margin.

6.2.5. Terrible Triad Modeling

The terrible triad injury model described by Fern et al involved the partial or complete disruption of the coronoid process, the LUCL, and the radial head. Traumas to the coronoid process were modeled by cutting along a plane oriented normal to the feature as described in section 5.2.5. The level of resection was calculated as a percent of the total
intact coronoid height, measured from the tip of the bony process to the deepest recess of the greater sigmoid notch. For the ulna used within the model, the height of the coronoid process was determined to be 17.24mm.

In the cadaveric study, ruptures of the LUCL were simulated by incising the ligament at its humeral origin and leaving the structure loose around the capsule. In the computational model, ligamentous soft-tissue was represented by mathematical expressions of force only and therefore did not have any solid geometry associated with them. As a consequence, the Lx injury state was able to be replicated by deactivating the FORTRAN expression governing the intact structure’s behavior. The LUCL repair state was created in the cadaveric specimens by rerouting the ligament’s line of action through the isometric point of the joint, as projected onto the lateral aspect of the capitulum, shown in Figure 6.2-2.
Figure 6.2-2: Lateral view of the ulnohumeral joint in 90° flexion with inset detail of the circle-fit center of the capitulum used for the LUCL-r state.

Once secured, the LUCL was pretensioned to 25N at 90° of flexion using a hanging weight. This scenario was replicated in SolidWorks using many of the same steps. The ulna was rotated about the ulnohumeral SDA (defined in section 4.3.1) with the radius fixed relative to it, until the model was positioned in 90° flexion. Then, to visualize the isometric point of the joint, the ulnohumeral articulation was isolated and viewed from the sagittal aspect of the humerus. A circle was scribed onto this plane whose perimeter was coradial with anterior circular margin of the capitulum. The center of that circle was then projected onto the surface of the humerus to designate the “isometric point centered on the lateral aspect of the capitellum [sic]” described by Fern et al. [34] Pretensioning of the
ligament was accomplished by modifying the strain value in the FORTRAN expression of the repaired LUCL such that the total extension was equivalent to 25N at the beginning of the simulation. Specifically, since the LUCL was modeled as having a stiffness of 57.0 N/mm (as given in Table 4.6-1), a shortening of 0.4386mm applied to the resting length was used to create a tension-only force of 25N. The general format of the expression is given below in equation 20.

\[
T_{LUCL-r} = IF(DM(P_1, P_2) - (L - 0.4386) : 0, 0, -57 \times (DM(P_1, P_2) - (L - 0.4386)) - 0.1 \times VR(P_1, P_2))
\]

Where \( T_{LUCL-r} \) is the tensile force in the ligament, \( DM(P_1,P_2) \) is the absolute magnitude distance between origin and insertion points, \( L \) is the actual distance between origin and insertion at \( t=0 \), and \( VR(P_1,P_2) \) is the relative velocity between the two points.

Therefore at \( t = 0 \), the actual distance of the ligament is governed by equation 21,

\[
DM(P_1, P_2) - L = 0
\]

but the tension calculated for the ligament is equal to the expression shown in equation 22.

\[
T_{LUCL-r} = -57 \times DM(P_1, P_2) - (L - 0.4386)
\]

Radial head injuries modeled for this study included the fully excised radial head (Rx) and the full radial head arthroplasty (Rr). Both states required that the native radial head be resected within the computer model in a manner consistent with the procedure employed by Fern et al. For most radial head resections and arthroplasties, including those used for the Biomet ExploR implant design, the radius is transected along a plane normal to the long axis of the radial neck and proximal to the radial tuberosity. Clinically,
selection of the correct cut angle is critical to ensure proper implant orientation and to minimize the possibility of implant impingement with the capitulum. [23] Additionally, some have noted changes in position and kinematic function of the radius following implantation of poorly oriented or improperly sized radial head arthroplasties. [23, 114, 117] However, though recognized as a critical aspect to successful implant placement, there is debate in the literature regarding the precise definition of the radial neck. [59, 116] For the injury states created within the model, a modified definition based on the anthropometric observations by Van Riet et al (2004) was used to identify this feature as accurately as possible.

To identify the central axis of the radial neck, a plane normal to the radioulnar axis of rotation (defined in section 3.7) was first drawn at a level coplanar with the most superior aspect of the radial tuberosity, identified by inspection. A circular sketch was then used to inscribe the outer contour of the radial cortex on this plane, with its center marking the center of the medullar space of the radial neck. Following this, a line was drawn from this center point to the sphere fit center of the radial head; this second line represented the central axis of the radial neck. (Figure 6.2-3) As quantified by the methods put forth by Van Riet, the angle formed between this radial neck axis and a line drawn from the center of the crinis radii to the center of the radial head was determined to be 5.47°, within the range reported. [115, 116]
Figure 6.2-3: Anteromedial view of the proximal radius showing the end-to-end long axis vs. the axis of the neck (left), a plane normal to the axis of the radial neck used to resect the head (middle), and the completed arthroplasty of the radial head using the ExploR implant system (right).

Finally, the amount of bone resected was defined as the distance between a plane normal to the axis of the radial neck and the most superior point on the native radial head. Given the size of the implant assembly created, the ideal resection level should have been 14mm. However, as Fern et al noted that the cadaveric specimens were consistently pushed into further varus following the arthroplasty, it was decided that the total resection distance should be 13.00mm, which represented the idealized 14mm value suggested by Biomet, offset by one millimeter. Once the radial head had been resected, the ExploR implant was positioned in the model by defining the collinearity between the long axis of the implant stem and the radial neck as well as a coincidence between the underside of the stem flange and the resection plane.
6.2.6. Setting Model Valgus and Forearm Rotation

Once the appropriate injury state was replicated and the model was set into the MTA testing apparatus, the neutral 0cm varus state needed to be defined. While the scan orientation could have been arbitrarily selected as the neutral carrying angle position, this assumption would have led to unphysiologic representations of the articular space within the joint in that position. This error arises from the differences in the orientation of the specimens used by Fern et al and that of the specimen that was scanned for model creation. Fern et al positioned their specimens such that the medial epicondyle was directed vertically and the supinated forearm was subjected to a valgus moment created by the hanging weight of the forearm, whereas in the dissimilar position of the scanned cadaveric tissue, the forearm was in approximately neutral rotation and supported by the scanning base platform, see Figure 3.2-1.

To account for this discrepancy, the intact model with activated muscle forces and ligament tension was initially tested with no constraint on the translational movement of the MTA. Therefore, while still fixed in flexion, the forearm was free to pronate/supinate and rotate in varus/valgus tilt in what Morrey and An called “gravity valgus.” [81] When the model had reached an equilibrium state, the positions of the radius and ulna, defined according to ISB recommendations, were each marked in 3-D space. In order to verify that the model had not been pulled into an unphysiologic valgus position, a second simulation was run wherein the radius and ulna were moved to these new starting positions, but the
translational motion constraints of the MTA were enforced. Under these conditions, the MTA registered no force as shown in Figure 6.2-4, indicating that the valgus moment imparted by gravity was equally balanced by the muscular and ligamentous tension around the joint. This result is in agreement with the methods described by Fern et al wherein the intact specimens were initially positioned in the testing apparatus and the position of the servo-hydraulic actuator adjusted until the load cell registered zero. Thus, before each model simulation, the initial position radius and ulna within the model were defined according to these the steady-state valgus positions.

![Graph](image)

**Figure 6.2-4:** MTA constraining load at 0cm varus for the model in the SDA predicted position (blue line) and in the gravity valgus position (pink line).
In addition to carrying angle, it was beneficial to preset the approximate level of forearm rotation in the model. Though not specifically quantified in the study presented by Fern et al, the authors noted that the specimens each rotated themselves into full supination during initial muscular tensioning. Pilot testing within the model indicated that when the radius was constrained only by radioulnar and radiocapitular 3-D contact, muscular forces, and ligament tension, the forearm within the model would fully supinate to a value of approximately -100° according to the definition set forth in section 4.3.2. Additionally, it was noted that the final resting position of the radius was not affected by the initial level of forearm rotation, whether it began in the cadaveric scan orientation, neutral orientation, or any other value of supination. However, though the final position of the bone was not dependent on the starting angle of forearm rotation, the total time required for the simulation to reach equilibrium was increased. Therefore, the radius was positioned into -80° of forearm supination prior to testing by rotating the bone about the radioulnar axis of rotation. This step ensured that the radius was able to reach its steady-state position in a short amount of time and that the entire simulation could be run faster.

6.3. RESULTS

The load resisting 1.5cm of varus excursion was recorded at the MTA for the 13 second simulation setup described in section 6.2.3. The protocol outlined by Fern et al (see section 6.2.5) involved testing of the elbow joint in the completely intact state, 4 intact coronoid but either disrupted LUCL or RH states, and 28 separate injury states involving
both LUCL and RH repair and variable levels of coronoid process resection, at both 30° and 90° of flexion. Thus, the complete testing protocol yielded a total of 66 simulations. Though each simulation generated a unique set of data, the characteristic shape of the ramped step in force shown in Figure 6.3-1 was routinely observed. Because this curve represents a ramp step in varus from the free-hanging gravity valgus position described in section 6.2.6, the equilibrium load prior to varus excursion was noted to be near zero for the intact case. Furthermore, for all but the most disrupted trauma state (LxRx), the magnitudes observed at the equilibrium position after varus excursion were generally of a much higher magnitude than those observed in the Hull study.

![Figure 6.3-1: Characteristic MTA Load vs. Time plot for the model during the 1.5cm varus excursion prescribed by Fern et al. Ramp in varus displacement occurred from second 6 to 7.](image-url)
Results from both the computational model, as well as the experimental data derived from Fern et al for the initial intact and four partially disrupted states for 30° and 90° of flexion are given in Figure 6.3-2 and Figure 6.3-3, respectively. It can be seen in Figure 6.3-2, that for the 30° case, the model predicted initial values of varus constraining load similar to those reported by Fern et al with overall magnitudes within a single standard deviation for each case except LrRi. For the 90° case (Figure 6.3-3), the model predicted the same sign of the resisting load as in the 30° case, though with generally larger magnitudes. The exception to this trend was the LiRx case wherein the model predicted the specimen would be acting to accommodate a varus displacement for the 90° simulation versus resisting this displacement for the 30° simulation. With regards to the intact case (LiRi) at 30° and 90°, the model recorded a near zero force while the experimental data was slightly negative. This discrepancy arises due to the perfectly elastic nature of the computational model, wherein, regardless of cycles, the force output measured at the MTA will be identical. By contrast, the experimental data reflects the hysteresis effects of soft-tissue constraints within the elbow that upon the application of four cycles of 0-1.5cm varus excursion have diminished the tension developed in the lateral soft-tissues of the arm. A second portion of the plots of particular note is the negative initial constraining load recorded for the LiRr case. Just as within the Fern study, the inclusion of a radial head implant has slightly lengthened the radius and thus the radiocapitular contact is pushing the forearm into varus. Though the cadaveric experiment did not quantify this degree of radial lengthening, the plots below suggest that because the model predicts a larger negative magnitude for this varus pressure, the degree of
lengthening in the cadaveric experiment was likely less than the 1mm value described in section 6.2.4. Finally, it was observed that the model predicted the LrRi repair case at 90° to have an increased varus constraining load versus the 30° case, whereas the cadaveric experiment reported a decrease in this load. This discrepancy stems from differences in the manner in which the model and the cadaveric joints strained the LUCL-r. In the cadaveric elbow, a shift in position of the radius from 30° to 90° caused the RH to impart less force on the ligament and thus less tension to be developed in the repaired LUCL. However, in the model, this same RH shift did not allow a decrease in the strain of the ligament element. Thus, the LUCL-r tension within the model was only sensitive to strain caused by ulnohumeral position, whereas the cadaveric elbows allowed for the ligament to be strained through relative changes in both ulnohumeral and radiohumeral position.

![Figure 6.3-2: Initial constraining load magnitude at 0cm varus for 30° flexion and intact coronoid process for both the model (purple) and the cadaveric (blue) data.](image-url)
Figure 6.3-3: Initial constraining load magnitude at 0cm varus for 90° flexion and intact coronoid process for both the model (purple) and the cadaveric (blue) data.

The change in constraining load following 1.5cm of varus excursion was also tracked for the 28 terrible triad injury specimens for each flexion angle. These results are given in Figure 6.3-4 and Figure 6.3-5. From Figure 6.3-4 it can be seen that within each coronoid resection level, additional instability results from having an unrepaired LUCL or RH and that repair of the soft-tissue structure restores greater varus stability than repair of the RH alone. Repair of both the LUCL and RH offers the most stability, while repair of neither offers the least. Similar results are seen at 90° (Figure 6.3-5) wherein loss of either the LUCL or RH confers greater varus instability to the joint at a given coronoid resection level. When analyzed within repair group, but across coronoid resection level, Figure 6.3-4 shows that there is a reduction in stability at greater than 67% coronoid resection for...
all four injury groups at 30°, while Figure 6.3-5 shows that at 90°, the model only predicted additional instability for the LrRx and LxRx cases. This later observation suggests that at 90° a greater percentage of the elbow’s varus stability is conferred by osseous constraints than at 30°.

The magnitude of the remaining varus constraining load after 1.5 cm excursion for the intact case is overlaid in red for both the 30° and 90° plots. For the 30° plot, the intact case was found to undergo a 32.54 N increase in load versus 24.8 ± 13.8 N for the experimental study. For the 90° plot, the intact case underwent an increase of 29.51 N versus 32.9 ± 15.2 N for the experimental case.
Figure 6.3-4: The change in load from 0-1.5cm varus displacement of the model at 30° flexion vs. percent coronoid resection. The intact joint value is marked in red.

Figure 6.3-5: The change in load from 0-1.5cm varus displacement of the model at 90° flexion vs. percent of coronoid resection. The intact joint value is marked in red.
In order to more directly compare the experimental values reported by Fern et al, and those predicted by the model, each of the data points from Figure 6.3-4 and Figure 6.3-5 were plotted as the change in load from 0-1.5cm given in terms of the percent of the intact coronoid case within the given injury state. Thus by definition, the 0% coronoid resection level point for each of the four injury groups is 100%. Likewise, the data from the corresponding Fern et al specimens were plotted alongside the model data in the same manner. The results for the LrRrC0-100% simulations for 30° and 90° are given in Figure 6.3-6. The results for the LrRxC0-100% simulations for 30° and 90° are given in Figure 6.3-7. The results for the LxRrC0-100% simulations for 30° and 90° are given in Figure 6.3-8. Finally, the results for the LxRxC0-100% simulations for 30° and 90° are given in Figure 6.3-9. Error bars given in each plot represent a single standard deviation, as reported by Fern et al.
Figure 6.3-6: Remaining constraining load at 1.5cm varus (as percent of intact coronoid) for the LrRr repair case at 30° flexion (left) and 90° flexion (right). Note: the cadaveric data is shown in blue and the computation model data in pink.

Figure 6.3-7: Remaining constraining load at 1.5cm varus (as percent of intact coronoid) for the LrRx repair case at 30° flexion (left) and 90° flexion (right). Note: the cadaveric data is shown in blue and the computation model data in pink.
Figure 6.3-8: Remaining constraining load at 1.5cm varus (as percent of intact coronoid) for the LxRr repair case at 30° flexion (left) and 90° flexion (right). Note: the cadaveric data is shown in blue and the computation model data in pink.

Figure 6.3-9: Remaining constraining load at 1.5cm varus (as percent of intact coronoid) for the LxRx repair case at 30° flexion (left) and 90° flexion (right). Note: the cadaveric data is shown in blue and the computation model data in pink.
The fully repaired LUCL and RH case (LrRr) shown in Figure 6.3-6 indicates that, for the 30° flexion angle, the model predicted a percentage loss in varus stability within a single standard deviation for all coronoid resection levels. However, while the cadaveric data shows very little change in constraining load until 50% resection, with a steady decline thereafter, the model plot indicates that there is an initial drop after only 25% resection, followed by relative stability until 75% resection has been reached. The 90° plot shows a similar experimental trend to the 30° plot; however, the model predictions for this flexion angle indicate relative insensitivity to resection of the coronoid, with only modest decreases for even 75% and 100% resection levels.

The fully repaired LUCL and unrepaired RH case (LrRx) shown in Figure 6.3-7 shown very similar trends to Figure 6.3-6 in experimental and model data for both 30° and 90°. Herein, the 30° is again within a single standard deviation of the experimental values for all coronoid resection levels, while the 90° plot indicates that the model was less sensitive to disruptions to the coronoid than were the cadaveric specimens. Only slight differences exist for the 30° plot wherein the model predicts a step-wise reduction in constraining load but beginning at a resection level of 40% instead of 25%.

The resected LUCL and fully repaired RH case (LxRr) shown in Figure 6.3-8 indicates that the model is able to accurately predict the corresponding loss in varus stability following coronoid resection for the 30° well within the standard error reported by Fern. Likewise, the 90° plot shows that the model predicts this behavior less well; though the LxRr injury group does show a larger percentage drop in constraining load at both 75% and 100% in closer agreement with the experimental data.
The most unstable injury state, a resected LUCL and an unrepaired RH (LxRx), is shown in Figure 6.3-9. For tests in this group, the 30° model is no longer able to predict the steady decrease in varus stability following coronoid resection. Indeed, there was a spike in stability observed at 100% that was not observed experimentally for any of the repair groups. Additionally, this increase in constraining load at 100% coronoid resection coincided with the resected radial head stem coming into contact with the capitulum. This was the only simulation to predict such interference and it was not a reported occurrence in the cadaveric study. (Figure 6.3-10) The data for the LxRx case at 90° shows plots very similar to those of the LxRr case wherein the model predicts a modest decrease in stability for coronoid resections of 75% and 100%.

![Figure 6.3-10: Lateral view of the LxRxC100% injury state in 30° flexion showing large proximal migration of the ulna causing abutment of the radial head prosthesis stem with the lateral edge of the capitulum.](image)
Finally, when the change in varus constraining force was compared across all injury states and irrespective of coronoid resection level (Figure 6.3-11), a clear pattern of increasing stability was observed for those groups with greater repair. Specifically, it was noted that repair of the radial head alone conferred additional varus stability onto the coronoid deficient elbow at both 30° and 90° of flexion. While significant, this increase in stability was less than that conferred by LUCL repair alone and both LUCL repair alone and RH repair alone were inferior to the simultaneous (LrRr) repair state. These trends were consistent with those reported by Fern et al, though the model generally overestimated the varus constraining contribution of a repaired LUCL while slightly under-representing the contribution of a repaired RH.
Figure 6.3-11: Change in constraining load from 0-1.5cm varus for each repair group and averaged across all coronoid resection levels. Note the values predicted by the model are shown in the purple and blue bars, while those measured experimentally are shown as purple and blue line plots.

In addition to varus constraining load, the computational model was also able to predict values for articular contact force in the radiocapitular and ulnohumeral joints for each separate injury state. Magnitudes for this force ranged from 41.9 to 181.7N for the 30° injury states and from 155.5 to 318.5N for the 90° injury states. Shown in Figure 6.3-12 and Figure 6.3-13 is a plot of the relative contribution that each articulation has to the overall contact force developed in the elbow. At 90°, Figure 6.3-13 the osseous stability of the elbow is dominated by the congruous ulnohumeral articulation for both the LxRr and LrRr injury states. At levels of coronoid resection above 50% there is a slight
transfer of load from the ulnohumeral articulation to that of the radial head and capitulum. This transfer becomes significantly larger for the 100% resected coronoid wherein ulnohumeral contact percentage is diminished to approximately 72.0 and 74.2% for the LxRr and LrRr cases, respectively. Also shown is a corresponding rise in the contribution of radiocapitular force to the overall joint force across the seven levels of coronoid disruption. The plot of the 30° flexion angle injury states Figure 6.3-12 indicates a more even distribution of contact force across the joint, with ulnohumeral and radiocapitular forces shared 47/53 for the LxRr injury state and 55/45 for that of the LrRr. As in the 90° case, at coronoid resections above 50%, there is a trend increasing contact force being shifted from the ulnohumeral contact to that of the radiocapitular articulation. At full coronoid resection, this trend was observed to have transferred approximately 8.5 and 5.3% of the joint force from the ulna to the radius for the LxRr and LrRr cases, respectively. Lastly, unlike the 90° case, the LxRr and LrRr injury states have 0% joint contact distributions which are significantly different from one another. Herein, the reason for the discrepancy is likely due to the stabilizing effect of the repaired LUCL on the ulnohumeral articulation. Such a ligament force would create larger 3-D contact force magnitudes with the LrRr case, thus biasing the joint force distribution towards the ulnohumeral articulation.

Though not plotted below, ulnohumeral and radiocapitular contact force was also tracked for the LxRx and LrRx injury states in both 30° and 90° of flexion. However, because these injury states did not include a radial head, the entirety of the elbow contact
force was assumed, and likewise observed to be dependent only on the ulnohumeral contact force.

![Graph](image_url)

**Figure 6.3-12: The relative contribution of the ulnohumeral and radiocapitular articulations on total elbow joint contact force for the model at 30° flexion.**
The tension within the repaired LUCL was tracked for each pertinent injury state at 30° and 90° of flexion. For this element, it was observed that at 30° the model was very sensitive to increased coronoid resection beginning at the 25% level. Further resections through 50% caused very little change in overall tension of the element, followed by a marked decrease in tension for the 75% and 100% cases. The 90° injury cases were, by contrast, much less sensitive to coronoid resection. For these simulations, the model predicted no significant difference between the repaired and resected RH states nor did it predict any change in LUCL-r tension for disrupted coronoid states through resections of
67%. At resections above 67%, the ligament element showed only modest decreases in tension.

![Graph showing LUCL-r tension vs. percent coronoid resection](image)

Figure 6.3-14: LUCL-r tension vs. percent coronoid resection in both the LrRx and LrRr injury states for the model at 30° and 90° flexion.

6.4. DISCUSSION

The rigid body elbow model developed in chapters 3 and 4 was used in conjunction with a 3-D representation of the Biomet ExploR implant assembly to predict the varus constraining load for an elbow following the disruption of osseous and ligamentous tissues categorized by the terrible triad injury. In accordance with the experimental protocol set forth by Fern et al, 66 separate simulations representing intact, resected, and repaired states of the radial head, coronoid process, and lateral ulnar collateral ligament were modeled at
both 30° and 90° of flexion. [34] For each injury state, the computational model was positioned into the appropriate flexion angle as well as the gravity valgus position determined in section 6.2.6. The model was then subjected to a 13 second rigid body simulation wherein the varus alignment of the joint was constrained at 0cm distal ulnar excursion for 6 seconds, followed by a 1 second ramp in varus displacement at a rate of 15mm/sec, and concluding with an additional 6 second hold at the 1.5cm level of displacement. During each of these simulations, the varus constraining load measured at the distal ulna was recorded as well as a host of other physiologic parameters.

For the intact and partially disrupted elbows, the model was able to predict initial constraining loads within, or just outside of a single standard deviation of those values reported by Fern et al. (Figure 6.3-2 and Figure 6.3-3) [34] This included accurate predictions regarding the inadvertent lengthening of the radius when inserting the radial head implant. The model was further able to capture the change in varus resistance magnitudes for the various injury groups reported in the cadaveric study. (Figure 6.3-4 & Figure 6.3-5) In general, the 30° simulations showed a marked decrease in change in varus constraining load for increasing levels of LUCL and RH disruption as well as for coronoid resections above 67%. Furthermore, the data plots produced for these simulations correlated very well with those reported by Fern et al, mimicking the overall shape, as well as the magnitudes of the cadaveric data well within the single standard deviations observed experimentally. As was observed in Figure 6.3-12, as well as in section 5.4, the model’s sensitivity to coronoid height suggests that the primary site of ulnohumeral contact was located at the base of the coronoid for elbows in low angles of flexion in agreement with
Morrey and An (2005). [81] For the simulations conducted at 90° of flexion, the model predicted trends for increasing LUCL and RH deficiency that were in also agreement with the cadaveric data. However, the model showed less sensitivity to coronoid resections than did the experimental elbows. The magnitudes of varus constraining force were still very close to the cadaveric data at low levels of coronoid resection but began to diverge from the cadaveric data for coronoid deficiencies of greater than 50%. These data reflect the model’s ability to describe the trend of decreasing varus stability with increasing coronoid resection as well as identify the LUCL and RH as the primary and secondary varus stabilizers in agreement with the conclusions reported by Fern et al.

The most likely cause of the discrepancies seen for the 90° injury states was the inability of COSMOSMotion to model ligament elements with curved lines of action. A specific consequence of this limitation was that the LUCL line of action was modeled as the straight line vector between origin and insertion. In situ, this ligament is continuous with the annular ligament and wraps the lateral aspect of the radial head in a manner that allows the ligament to be tensioned by means of the radial head pressing against the fibers orthogonal to the their normal longitudinal alignment. [57] Thus, in the native joint, the intact RH is vital for redirecting the line of action of the LUCL, thereby creating tension in the structure under varus loading. Within the computational model, the LUCL was insensitive to the presence of the RH in terms of the tension it develops, and was thus shorter in length due to its straight line path from origin to insertion. At 90°, these factors caused the LUCL ligament element to act along a line that passes through the radial head.
prosthesis, as shown in Figure 6.4-1, and in so doing overestimated the total strain in the
element during varus excursion.

Figure 6.4-1: Posterolateral view of the LrRr injury state in 90° of flexion. Note the
line LUCL-r element (dark blue) passes through the repaired RH.

Tension developed in the LUCL-r element for the LrRx and LrRr repair cases at
30° and 90° was shown to follow the same trend as the overall varus constraining load of
the joint thus supporting the generally supported notion that the LUCL in the primary soft-
tissue varus stabilizer of the elbow. [78, 82, 93, 100] (Figure 6.3-4, Figure 6.3-5 & Figure
6.3-14) For these repair cases, the 30° injury cases showed a greater decrease in varus
constraining load with increasing coronoid resection than those at 90°. This observation
further reinforces the assertion that an unphysiologic increase in LUCL tension is
responsible for the model’s diminished sensitivity to coronoid resection at 90°.
In addition to varus constraining load, the model was also able to measure the joint contact forces for each simulation. In general it was observed that there were lower total joint contact forces for injury cases at 30°, as well as a more even sharing between radiocapitular and ulnohumeral contact forces for those injury states with a repaired RH. At the 90° injury states, it was observed that the overall magnitude of the joint contact forces increased as well as the proportion of that force borne by the ulnohumeral articulation. These predictions are in agreement with those in the literature that suggest that increasing flexion angle causes more of the joint’s stability to arise from osseous interactions, specifically the ulnohumeral articulation. [57, 78, 81, 82]

Generalized sources of error for this computational model are similar to those reported in section 5.5, while sources unique to this experimental setup are given below. The first source is the possible unphysiologic tension in the MCLC or LCLC due to gravity valgus pre-positioning of the model. Though the experimental procedure used to attain this zero force initial state was faithfully reproduced within the model, the cadaveric elbows were tested with musculature and capsule intact. These additional soft-tissue constraints, which were not modeled computationally, could have caused the forearm to hang in more or less valgus than what was observed for the model. If this were the case, then it is possible that the magnitudes of the varus constraining loads could be affected, though the overall trends would not have been drastically altered by this initial offset. A second potential source of error derives from the positioning of the Biomet ExploR radial head prosthesis. While the method described in section 6.2.5 for quantifying the axis of the radial neck was supported in the literature and the 3-D representation of the ExploR
implant was consistent with Biomet’s fitment guidelines, clinical placement of the prosthesis is heavily dependent on the judgment and experience of the surgeon. [116] Thus, alternative placements of the implant are possible and would affect the radiocapitular contact force as well as the initial valgus position predicted by the model. Additionally, the radial head offset value of 1mm was somewhat arbitrarily chosen as the maximum radial lengthening error in the placement of the implant. While the model data suggests that this offset was greater than what was attained experimentally, the precise degree of lengthening in the cadaver study was not quantified
7. CONCLUSION

The computational model developed within this thesis represents a continuation and adaptation of the work by Fisk and Wayne (2009), with the continued objective of creating a high fidelity rigid body model of the human elbow joint wherein function is dictated only by soft-tissue muscle and ligamentous constraints and osteoarticular surface interactions. [37] In the creation of this model, the commercially available medical imaging software MIMICS was used to arrange and manipulate high resolution computed topography scans of a cadaveric right upper extremity according to the linear attenuation of the cadaveric tissue represented within each slice image. In this manner, the bony surfaces of the humerus, ulna, and radius were successfully separated from the soft-tissue of the arm. From this, three dimensional triangulations of these bones were created and transferred into the SolidWorks computer design software by means of the STL file format. Within the SolidWorks environment, force, position, and contact constraints were created that accurately approximated the native anatomy and function of the cadaveric arm throughout the normal physiologic range of forearm flexion/extension, pronation/supination, and varus/valgus tilt. Finally, the ADAMS rigid body motion solver native to SolidWorks’ COSMOSMotion add-in was used to analyze the behavior of the muscle and ligamentous soft-tissue, osseous interaction, and prescribed external perturbation in a complete and quantifiable manner.
Validation of the model involved the simulation of a pair of cadaveric elbow studies by Hull et al (2005) and Fern et al (2009), wherein the complex mechanism of varus elbow stability was investigated. [34, 55] Under the testing conditions prescribed for each experiment, the model was able to accurately replicate the trends, and in many instances the magnitudes, observed in the MTS constraining load resisting varus for a variety of injury states and in numerous levels of flexion/extension, pronation/supination, and varus/valgus tilt for each of the respective studies. Furthermore, the model was able to predict physiologically reasonable magnitudes for the joint contact force within the elbow, the location of osteoarticular interface, and the tension developed within the ligaments of the elbow joint and forearm. Additionally, the methods employed herein offered an improvement over those put forth by Fisk and Wayne in terms of the manual manipulation required in generating the bony 3-D anatomy of the arm, as well as the time required to solve the rigid body simulations for the various position and injury states prescribed by Hull et al.

Though the model was able to make accurate predictions concerning a number of physiologically relevant parameters, certain assumptions regarding the material behavior of articular cartilage, bone, and ligamentous soft-tissue were necessary. With regards to the contact constraints enforced within the model, all interactions were defined as rigid body. Thus, the physiologic deformation and stiffness contributions of articular cartilage were not modeled. While this approximation is valid for cases wherein the overall deformation for any body within the joint is very small, it may not be an appropriate assumption in situations wherein very high stresses would be expected to cause deformation within a
cadaveric model, as those alterations in geometry would not be captured in the computational model. Thus, future elbow models could potentially incorporate parallel finite element analysis studies in order to more completely describe the contact stresses within the joints.

Additionally, ligament function within the model was governed by linear tension only force expressions without true viscoelastic behavior though more elaborate definitions are enforceable using the ADAMS solver package. These assumptions were made because precise experimental investigations into the stress/strain and viscoelastic behavior for the ligaments of the elbow are generally lacking in the literature. However, there is support for the notion that ligaments are linearly elastic within their physiologic range and all results reported for the model represent the equilibrium state, thereby mitigating the impact of these approximations. Furthermore, the in situ strains present in a number of the ligaments of the elbow have not been thoroughly investigated. As the mechanical behavior of these structures is further elucidated, future ligament expressions could be created that more accurately reflect this complex behavior.

A final concern relating to the application of soft-tissue forces within the model was the requirement that they be straight line elements between their origin and insertions. While this definition does not cause conflict for ligaments loaded only longitudinally, ligaments with physiologic action orthogonal to the longitudinal direction, such as the annular and lateral ulnar collateral ligaments, potentially require additional solid bodies to be incorporated into the model to ensure a physiologic line of action. Future improvements in the model’s ligament orientations could focus on more completely
describing the secondary stabilization effects of these structures without introducing additional rigid body elements by allowing non-linear force vectors to wrap bony geometries instead of passing through them.

The rigid body computational model of the upper arm outlined in this thesis was able to predict the physiologic function of both the intact and deficient elbow joint under a variety of loading scenarios and at a number of flexion angles with a high degree of accuracy. Furthermore, the model was able to readily quantify parameters such as ligament tension, contact force, and contact area that are not easily measured experimentally. This ability allowed the model to elucidate the causes of the varus instability observed clinically in a depth and detail that were not possible to within the context of the two cadaver studies presented. Additionally, the computational model was able to clarify these behaviors while utilizing only commercially available software and at a much lower cost than cadaveric experiments. Future applications for this type of modeling are numerous and could impact fields ranging from the academic to the clinical. Examples include anything from pilot investigations of possible cadaveric experimental protocols to patient-specific biomechanical preoperative decision making to custom arthroplastic implant design and positioning. Thus, rigid body modeling of complex physiologic structures will likely be used as an ever more prevalent tool in the investigation of musculoskeletal kinematics.
Literature Cited
Literature Cited

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2. COSMOSMotion 2007 Help Pages. 2007, Structural Research and Analysis Corporation, MSC Software Inc.: Santa Monica, CA.


APPENDIX A

List of Abbreviations

General Abbreviations

2-D  two dimensional
3-D  three dimensional
ADAMS Automatic Dynamic Analysis of Mechanical Systems
API Application Programming Interface
ASCII American Standard Code for Information Interchange
CAD Computer Aided Design
CT  Computed Topography
DICOM Digital Imaging and Communications in Medicine
DOF Degrees of Freedom
DRUJ Distal Radioulnar Joint
DRULs Distal Radioulnar Ligaments
DXF Drawing Exchange Format
FEA Finite Element Analysis
HU Hounsfield Unit
IGES Initial Graphics Exchange Specification
ISB International Society of Biomechanics
MIMICS Materialise's Interactive Medical Imaging Control System
MRI Magnetic Resonance Imaging
MSC MacNeal-Schwendler Corporation
PMMC poly-methylmethacrylate
PVC polyvinyl chloride
RBM Rigid Body Modeling
SDA Screw Displacement Axis
SIMM Software for Interactive Musculoskeletal Modeling
STL Stereolithography
VIMS Virtual Interactive Musculoskeletal System
ISB Abbreviations

Oh Origin of Humerus Coordinate System
Yh Y-axis of Humerus Coordinate System
Xu X-axis of Humerus Coordinate System
Zh Z-axis of Humerus Coordinate System

Ou Origin of Ulna Coordinate System
Yu Y-axis of Ulna Coordinate System
Xu X-axis of Ulna Coordinate System
Zu Z-axis of Ulna Coordinate System

Or Origin of Radius Coordinate System
Yr Y-axis of Radius Coordinate System
Xr X-axis of Radius Coordinate System
Zr Z-axis of Radius Coordinate System

AP Anteroposterior
ML Mediolateral
TR Transverse

Anatomic Abbreviations

COG Center of Gravity

EM Medial Epicondyle of Humerus
EL Lateral Epicondyle of Humerus
GH Center of Proximal Humeral Head (sphere-fit)
CP Coronoid Process of Ulna
US Ulnar Styloid
RH Radial Head (Proximal)
RS Radial Styloid
ALB Annular Ligament Body

IOM Interosseous Membrane
CB Central Band of IOM
AB Accessory Band of IOM
DOB Distal Oblique Bundle
RCL Radial Collateral Ligaments
MCL  Medial Collateral Ligaments
LCL  Lateral Collateral Ligaments
LUCL  Lateral Ulnar Collateral Ligament
DRUL  Distal RadioUlnar Ligaments
TFCC  Triangular Fibrocartilage Complex

EJC  Elbow Joint Complex
UH  Ulnohumeral Joint
RC  Radiocapitular Joint
UR  Proximal RadioUlnar Joint

Fern et al Abbreviations

Ci  Coronoid Intact
C#  e.g. 25% or 50% Coronoid Resection
Li  LUCL Intact
Lx  LUCL Excised
Lr  LUCL Repaired
Ri  Radial Head Intact
Rx  Radial Head Excised
Rr  Radial Head Repaired
VITA

Edward Meade Spratley was born and raised in Central Virginia, just outside of the City of Richmond. He attended Powhatan High School and upon graduation in May of 2000, he was accepted and enrolled in the University of Virginia. In 2004, Meade graduated with a Bachelor of Science degree with a concentration in chemistry. For the following three years, Meade lived alternately in Vail, CO and Tucson, AZ traveling and pursuing his two passions, rock climbing and alpine skiing. In the spring of 2007, he returned to the Richmond area and was enrolled in the Master of Science program at Virginia Commonwealth University. In the spring of 2009, Meade was presented with the Phi Kappa Phi Student Scholarship award and was also later inducted into the VCU chapter of Alpha Eta Mu Beta honor fraternity for Biomedical Engineers. Following graduation, Meade will continue on at Virginia Commonwealth in pursuit of his Doctoral degree. Portions of this work were presented at the summer 2009 meeting of the Virginia Academy of Science where it earned the Best Student Paper award in the General and Biomedical Engineering section.