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MECHANICAL STRUCTURES RESISTING ANTERIOR INSTABILITY IN A COMPUTATIONAL GLENOHUMERAL JOINT MODEL

Kevin Elmore
Virginia Commonwealth University

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MECHANICAL STRUCTURES RESISTING ANTERIOR INSTABILITY IN A COMPUTATIONAL GLENOHUMERAL JOINT MODEL

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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ABSTRACT

MECHANICAL STRUCTURES RESISTING ANTERIOR INSTABILITY IN A COMPUTATIONAL GLENOHUMERAL JOINT MODEL
By Kevin A. Elmore, B.S.

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Virginia Commonwealth University, 2009
Major Director: Dr. Jennifer S. Wayne
Departments of Biomedical Engineering and Orthopaedic Surgery

The glenohumeral joint is the most dislocated joint in the body due to the lack of bony constraints and dependence on soft tissue, primarily muscles and ligaments, to stabilize the joint. The goal of this study was to develop a computational model of the glenohumeral joint whereby joint behavior was dictated by articular contact, ligamentous constraints, muscle loading, and external perturbations. Validation of this computational model was achieved by comparing predicted results from the model to the results of a cadaveric experiment in which the relative contribution of muscles and ligaments to anterior joint stability was examined. The results showed the subscapularis to be critical to stabilization in both neutral and external rotations, the biceps stabilized the joint in neutral but not external rotation, and the inferior glenohumeral ligament resisted anterior displacement only in external rotation. Knowledge gained from this model could assist in pre-operative planning or the design of orthopedic implants. Use of this model as a companion to cadaveric testing could save valuable time and resources.
1 INTRODUCTION

1.1 Modeling of Musculoskeletal Systems

A musculoskeletal model is a virtual representation of body movements designed to scientifically study the biomechanics of motion. Models vary widely in complexity and can range from a two dimensional idealized joint that reacts to applied muscle forces, to an anatomically correct 3D representation of a joint complete with muscle loads and ligamentous restraints. The information to develop these models comes from a long history of anatomic studies of the human body. More recently, mechanical testing of cadavers has allowed for the development of models that can realistically and accurately describe the complexities of biomechanics.

![Musculoskeletal models of a shoulder joint, in 3D (left) and 2D (right). Muscles are modeled as deformable bodies that connect the origin to the insertion in the 3D model, while the forces in the 2D model are approximated by moving the insertion point (i.e. the middle deltoid) to the humeral head [1].](image)

The usefulness of models in the medical field lies in their ability to produce an abundance of information in a relatively short amount of time. Unlike a cadaver study, a
study based on a musculoskeletal model can be easily repeated with a slight change in experimental methodology. However, a cadaver study would require an exhaustive and expensive process to recreate the previous experimental conditions. Data collection can be problematic in a cadaver model, requiring a separate measuring device for each parameter of interest. Simultaneously capturing this data can be challenging throughout the experiment. Computational models can collect vast amounts of parallel data for many variables, including those that may be difficult to measure experimentally. For instance, joint contact areas are particularly challenging to quantify in vivo, though can be easily assessed in a computational model. With validation of a computational model, the sole dependency on repetitive cadaveric testing can be alleviated.

There are many clinical applications for musculoskeletal models. For instance, a model that could predict the function of a joint after alteration of a muscle insertion site would be an invaluable tool for pre-operative planning. The surgeon could vary the insertion site within the model to quantify muscle moment arm data, the effect on surrounding muscles, and resulting efficiency of the joint. This same information could be used post-operatively to coordinate a rehabilitation regimen for that patient specific surgery. Furthermore, measurements attained from models could help a surgeon choose the correct size, type, and placement of a prosthetic implant and determine the effects of these decisions before performing any operation. Surgeons could also use these models to assess the outcome of a specific osteotomy or ligament shortening procedure. With the advent of powerful computational technologies, highly detailed musculoskeletal models make such surgical decisions much more predictable.
1.2 Computer Methods in Musculoskeletal Models

Most computerized musculoskeletal models fall into one of two categories: rigid body models or deformable models (also known as Finite Element (FE) models). These categories do not have distinct dividing lines, and features of any one model may allow it to span these categorical distinctions. Rigid body models consist of a system of solid bodies that react to external perturbations, though do not account for deformations of these solid bodies [2]. Applying this method to musculoskeletal models, the bones are modeled as the solid bodies and muscles as force producing elements that connect the origin to the insertion of the muscle. Since the bodies are rigid, contact between the surfaces of the bodies affects the overall kinematics of the joint.

There are some limitations to rigid body modeling. One of the difficulties of rigid body modeling has involved ensuring that the line of action of a muscle is physiologically correct. Often, a straight line from the origin to the insertion is sufficient for modeling purposes, but a change in the position of the bones may interfere with this line of action. “Wrapping” of the muscle around bone geometry thus requires special attention to retain the correct moment arm, and is sometimes accomplished by creating points through which the line of action is redirected (Figure 1-2) [3].
Figure 1-2. Wrapping of muscle around bone can be accomplished by intermediate points. Instead of connecting the insertion (b1, on humeral head) to the origin (b4, on the glenoid), a point (b3) is used to correct the line of action to a more accurate direction [4].

Also, many models of this nature oversimplify the joint to the point that realistic movement of the joint is not possible. For instance, the shoulder joint has been modeled as a ball and socket joint [5] or as a three-hinge joint [6], but neither of these approximations allow for translational movement of the humeral head. Thus, stability studies can not be performed on these particular models. Finally, rigid body models cannot quantify stress or strain developed in a body due to an applied force, and therefore are not useful in determining how prosthetic implants might react with the surrounding bone. With advances in rigid body modeling software, some of these limitations have recently been overcome.

A few musculoskeletal models exist today, though they have been designed for various needs and have their own limitations. Software for Interactive Musculoskeletal Modeling (SIMM) is a 3D analysis tool for studying biomechanical systems [7]. The program integrates representations of bones, muscles, and ligaments and can be configured to produce joint movements based on muscle forces. The system relies on computed muscle moment arms about the joint to produce a rotational movement [3].
Unfortunately, the joints are simplified to only a few degrees of freedom so that they function as a ball and socket or hinge type joint. Users can study the effects of changing muscle insertions, forces, and bone geometry, but further dynamic analysis is restricted due to the joint definitions.

Other software packages exist that are more versatile. The Virtual Interactive Musculoskeletal System (VIMS) software program also consists of 3D representations of bones, muscles, and ligaments, but can also incorporate any solid body [8]. The main advantages that VIMS has over SIMS include the ability to not only study joint motions but all associated static, kinematic, kinetic, and stress concentrations under a wide variety of loading and boundary conditions. This program has been used to develop an analysis of the throwing motion for the shoulder to study the moment of arms of the shoulder muscles under different arm positions [9]. In this study, VIMS was used to elucidate which muscles were most effective to producing the desired throwing motion. Unfortunately, these advanced abilities are only available with proprietary software packages that expand on VIMS basic architecture [8].

Unlike rigid body modeling that sacrifices accuracy for simplicity, deformable models (or FE models) utilize material properties of biologic structures to provide a more realistic representation of a joint. These material properties are generally non-linear elastic or viscoelastic. The models have the ability to assess plastic deformation, creep, and fatigue failure behaviors [2]. Applications of FE models include prosthetic design and implantation with special emphasis on fixation methods. Quantifying the differences in stress on either the prosthetic or the surrounding bone tissue due to a change in
positioning can provide invaluable data for designing and implanting such devices. Additionally, the problem with muscle wrapping for rigid body modeling can be resolved simply in FE models by utilizing deformable muscle bodies that can flex around bony geometry (Figure 1-1) [1,10]. Because of the complexities of deformable modeling, usually some concessions to accuracy have to be made. Instead of using all non-linear elastic or viscoelastic material behavior, many models use idealized material properties. Additionally, boundary conditions may be restrained to simplify the model. Another method of simplification may be to replace individual muscle forces with one force, representative of the total joint reaction force [11,12].

1.3 Modeling of the Glenohumeral Joint

The glenohumeral joint of the shoulder allows the most range of motion of any joint in the body. The lack of bony constraints in this ball and socket type joint allows movement that is predominately restricted by the surrounding soft tissue. Unfortunately, the mechanics of this joint make it highly susceptible to injury. Over 1.7% of people experience a shoulder dislocation in their lifetime [13]. After the initial dislocation, many people experience joint laxity and an increased chance of re-dislocation. According to one study, 20 to 30% of the population experiences some shoulder pain, while 8.8% experience functional impairment [14,15]. Furthermore, many clinical issues dealing with the shoulder have proven to be more persistent and challenging to overcome than other joints in the body. More needs to be understood about the shoulder to overcome these challenges, and computational modeling can potentially fill this void and provide a deeper understanding of shoulder pathology and treatment [2].
In the past, simple models have accurately been able to describe motions at the knee and hip joints, as the motions here can be approximated effectively with two dimensional models. The shoulder model, however, requires three dimensional modeling to take into account the entire range of motion. Only recently has the software and computational power existed to handle such complexities. One potential application of computational modeling deals with prosthetic design and implantation. The number of shoulder joint replacement procedures has tripled in the past 10 years, with over 29,000 partial and total replacements were performed in 2004 in the US [16]. The common complications involve loosening of the glenoid prosthetic and joint instability following surgery. Finite element approaches could be used to determine alternate methods of fixation, and rigid body modeling can further investigate joint conformity, muscle loading, or the influence of pins or screws to return stability to the implant.

To accomplish these goals, improvements to the current computational models of the shoulder must be made. Some models currently are based on assumptions of a ball and socket joint for the shoulder [7,17]. With this limitation, translations of the humeral head are restricted negating the usefulness in stability testing. Others models take into account only the muscular forces present at the shoulder joint. As the shoulder relies heavily on ligamentous restraints under extreme motions, injury studies will suffer from inaccuracies [1,10]. Still others determine the tension and strain within ligaments under various positions and loadings, though neglect muscle forces [4,18]. Currently, there are no existing computational models of the glenohumeral joint that can simultaneously
predict the biomechanics of the shoulder while incorporating anatomically correct bone contact, muscle forces, and ligament restraints.

1.4 Objective

The goal of this work was to create a computational model of the glenohumeral joint whereby joint behavior was dictated by articular contact, ligamentous constraints, muscle loading, and external perturbations. Three-dimensional anatomy of the bones of the shoulder was obtained from computer tomographic scans of a human upper extremity. Glenohumeral ligament insertions were located on the bones and their function reproduced by linear springs while select muscles were represented as constant-magnitude force vectors along their physiologic line of action. A commercial rigid body dynamics program was then implemented to simulate joint function, with validation accomplished by a comparison of model predictions to results obtained in a published experimental study.
2 ANATOMY BACKGROUND

2.1 Bone Anatomy

The scapula, clavicle, and the humerus are the three main bones that comprise the shoulder joint (Figure 2-1). The scapula is the main stabilizing structure of the shoulder and serves as the point of contact of the humerus [19]. Its name comes from the Latin word for shovel due to its resemblance to the blade of a shovel (Figure 2-2). The scapula is roughly triangular in shape, though the lateral portion of the bone splits into two processes: the acromion and the coracoid processes, which serve as attachment points for the acromioclavicular ligament and the coracohumeral ligament (CHL) [19]. The spine of the scapula runs along the medial-lateral direction, and serves as the anterior border for the supraspinatus muscle.

Figure 2-1. The shoulder is composed of the scapula, clavicle, and humerus. The bones articulate through the interactions at the glenohumeral and acromioclavicular joints. The acromion is the most lateral bony process on the scapula. [20]
The other supporting structure of the shoulder is the clavicle, also known as the collar bone, and is located on the anterior aspect of the thorax. The clavicle acts to support the scapula and also the arm by directing force to the axial skeleton. The medial end of the clavicle connects to the sternum at the sternoclavicular joint, while the lateral end meets with the acromion of the scapula, termed the acromioclavicular joint, constituted by the acromioclavicular ligament [19]. The anterior face of the lateral portion of the clavicle serves as the origin of the anterior deltoid. The midsection of the clavicle is also connected to the scapula at the coracoid process via the coracoclavicular ligament.

The humerus contacts the scapula at the glenohumeral joint, composed of the convex head of the humerus and the concave glenoid cavity, located on the lateral aspect of the scapula (Figure 2-3). Surrounding the glenoid cavity is a cartilaginous ring called the glenoid labrum, which acts to deepen the socket [22]. The humerus is the largest bone in the arm and has two tubercles on the proximal shaft. The greater tubercle lies more lateral than the humeral head and serves as the connection point for the
supraspinatus muscle [19]. More inferiorly, also on the greater tubercle, are other attachment sites for the infraspinatus and teres minor tendons. The lesser tubercle, on the anterior aspect of the humerus, provides a location to insert the subscapularis. The bicipital groove is situated between the two tubercles, and directs the long head of the biceps tendon toward its origin on the rim of the glenoid. Another bony prominence, the deltoid tuberosity, located at about half of the length of the humerus on the lateral side, allows for the connection of the deltoid muscle. Motion at the glenohumeral joint is composed of rotations of both the scapula and the humerus. The scapula rides along the posterior aspect of the thoracic cage as the humerus rises to create abduction. This simultaneous movement is termed scapulothoracic rhythm. For each 3° of abduction, 1° is due to rotation at the glenohumeral joint, while the remaining 2° is due to humeral rotation. Therefore, 90° of abduction of the arm involves a 30° upward rotation of the scapula [23].

The distal humerus ends in two articular surfaces that serve as articulating surfaces for the radius and ulna. The medial and lateral epicondyles are the attachment sites of some of the ligaments that stabilize the elbow.
2.2 Musculature

There are seven major muscles that are relevant to movement and stabilization of the glenohumeral joint. The rotator cuff muscles (subscapularis, supraspinatus, infraspinatus, and teres minor) are the underlying muscles that surround the glenohumeral joint (Figure 2-4). These muscles, in addition to helping move and suspend the arm, have the important role of centering the humeral head within the glenoid cavity. The deltoid, divided into three regions, helps to abduct and flex the humerus.

![Figure 2-4. Muscles of the rotator cuff [25].](image)

The subscapularis is located on the anterior aspect of the glenohumeral joint (Figure 2-4). It originates on the costal surface of the scapula and extends around the edge of the glenoid cavity, to insert on the lesser tubercle of the humeral head. The subscapularis assists in internal rotation of the arm, while also acting to protect the humeral head from anterior dislocation from the glenoid cavity. At 0° of abduction, the subscapularis is tight over the humeral head in external rotation only [26]. At 45° and 90° of abduction, the subscapularis is loose in internal rotation, tight in neutral rotation,
and very tight in external rotation [26]. Turkel’s findings state that the subscapularis does not limit anterior dislocation at higher degrees of abduction, as the muscle fibers rise too far superiorly to cover the midportion of the joint [26].

The supraspinatus lies on top of the scapula and aids in abducting the arm, especially below 30 degrees of abduction [27] (Figure 2-4). The origin lies in the supraspinatus fossa, located above the scapular spine. From here, the muscle traverses the superior aspect of the scapula, under the acromion process. Then it passes above the glenohumeral ligament and inserts on the most anterior aspect of the greater tubercle on the humeral head [28,29]. Mochizuki documents that in 21% of the population, the supraspinatus is a split tendon that spans the bicipital groove to the lesser tubercle [29]. This muscle counteracts the force of gravity by actively restraining the humeral head from subluxing inferiorly [30]. A deficiency in the supraspinatus tends to cause upward migration of the humeral head during abduction of the humerus [1].

The infraspinatus lies on the posterior aspect of the scapula (Figure 2-4). It attaches to the infraspinous fossa laterally on the scapula, and traces medially across the scapula towards its insertion on the greater tubercle of the humerus [29]. This muscle aids in external rotation and adduction of the arm [27].

Another external rotator, though less structurally significant than the infraspinatus, is the teres minor muscle [27] (Figure 2-4). This muscle attaches to the lateral border of the scapula, and inserts on the inferior aspect of the greater tubercle on the humeral head. Sometimes this muscle becomes bundled with the fibers of the
infraspinatus, and thus is occasionally combined with the infraspinatus as a single unit during cadaveric stability experiments [31].

The deltoid muscle lies above the glenohumeral joint. It has three defined regions: anterior, middle, and posterior (Figure 2-5). The medial attachment sites for this muscle extend from the medial portions of the clavicle, around the acromion, and then inferiorly and medially along the scapular spine [19]. As the deltoid extends laterally, it envelops the superior region above the glenohumeral joint, finally anchoring on the lateral aspect of the medial humerus. The separate regions of the deltoid allow for increased muscular division. The anterior region of the deltoid assists in abduction, especially when the arm is externally rotated [27]. Additionally, this group of fibers performs forward flexion and transverse flexion motions. When the arm is in internal rotation, the middle bundle conducts abduction, while also assisting in transverse flexion [27]. The middle group of fibers performs transverse adduction when in external rotation. Finally, the posterior region’s primary motion is transverse extension.

Figure 2-5. Lateral view of deltoid. Proximal attachments include clavicle, acromion, and spine of scapula. The muscle covers the glenohumeral joint and attached at the deltoid tuberosity on the humerus [32].
2.3 Ligamentous Restraints

Beneath the rotator cuff muscles lies the glenohumeral joint capsule. This structure encircles the articular surface of the humeral head as well as the glenoid fossa. The capsule is a continuous structure of ligamentous material, but has been described as having distinct thickened regions with varying mechanical properties [33-35]. The ligaments of the capsule are responsible for limiting extreme rotations of the humerus with respect to the scapula, as well as resisting dislocation of the head of the humerus from the glenoid socket. To allow the shoulder the range of movement it has, the capsule is quite lax in the neutral position. As the humerus is rotated around its axis or the arm is moved, certain ligaments become taut, resisting motion. The capsule has been traditionally divided into four regions, some of which include sub regions (Figure 2-6). These are the superior glenohumeral ligament, the middle glenohumeral ligament, the inferior glenohumeral ligament, and the posterior capsule. Although each of these regions adds to joint stability, their importance depends strongly on the position of the humerus.
The superior glenohumeral ligament (SGHL) has two origins, one at the superior tip of the labrum (adjacent to the biceps tendon) and also at the base of the coracoid process [26]. There is some variability here, as Boardman et al. reports that in some specimens, the SGHL arose entirely from the superior glenoid labrum [36]. This ligament extends over the glenohumeral joint and inserts superior to the lesser tuberosity on the humeral head (Figure 2-7). The SGHL’s insertion is intimately bundled with that of the coracohumeral ligament. This union is sometimes referred to as the rotator interval capsule, named because it lies in the region between the subscapularis and supraspinatus tendons [37]. This ligament has been found to limit external rotation of the adducted arm [37-40]. Additionally, this ligament has been found to resist inferior dislocation of the...
humerus, especially when the muscles supporting the humerus are paralyzed [30]. Over a
short period of paralysis, the fibers crossing the superior portion of the joint (SGHL and
coracohumeral ligament) becomes stretched in this condition, leading to joint laxity [41].

![Figure 2-7](image.png)

Figure 2-7. Anterior view of glenohumeral ligaments at 0°, 45°, and 90° abduction (posterior view inset). As abduction increases, the SGHL and MGHL become lax, while the inferior regions become taut. AB and PB are the anterior and posterior bands of the IGHL [40]. Note the coracoid and acromion processes have been removed for clarity.

The middle glenohumeral ligament (MGHL) traverses the anterior aspect of the
glenohumeral joint. The origin of the MGHL lies just anterior and inferior to the origin
of the SGHL, and extends antero-inferiorly towards its insertion on the medial aspect of
the lesser tuberosity of the humerus. Again, the position of the humerus has a significant
impact on the role the MGHL plays towards glenohumeral stability. At zero degrees of
abduction, only in external rotation, the MGHL is tight. The MGHL becomes taut and
helps to prevent anterior dislocation when the arm is in external rotation at 45 degrees of
abduction [26]. At 90 degrees of abduction and internal rotation, the MGHL/SGHL
complex restricts anterior dislocation equally with the superior band of the inferior
glenohumeral ligament [42]. Of all the capsular ligaments, the MGHL has the most
variability in structure and attachment site; some individuals lack the MGHL entirely
[43]. This deficiency would be accounted for by variations in the surrounding structures.
The inferior glenohumeral ligament (IGHL) plays an important role in stabilizing the glenohumeral joint. Many studies have demonstrated the importance of the IGHL in restraining anterior instability [26,34,38,42-45]. This portion of the capsule encompasses nearly half of the circumference of the joint [44]. It originates from the anterior aspect to the postero-inferior margins of the glenoid labrum. On the humerus, it inserts on the inferior aspects of the anatomical and surgical necks, along the articular border. The IGHL has been described as having three separate regions: the anterior band of the IGHL (AB-IGHL), the axillary pouch of the IGHL, and the posterior band of the IGHL (PB-IGHL) [46]. To better describe the three regions of the IGHL complex, Bigliani et al. defined a coordinate system for consistent dissection during their study [44].

Figure 2-8. Ligament insertions on humerus with respect to bony landmarks, anterior aspect (left) and medial aspect (right). Here the AB-IGHL is labeled as the superior band of the IGHL [26].

The anterior band of the IGHL (also known as the superior band of the IGHL [26]) originates from the antero-inferior aspect of the glenoid fossa, and can be intimately attached to either the glenoid labrum, medial to the glenoid rim, or a combination of both [47]. It inserts on the humerus inferior and posterior to the MGHL, along the anatomical
neck of the humerus. The effectiveness of this region of the ligament depends entirely on the position that the arm is in. At zero degrees of abduction, the anterior band is only taut when in external rotation. When the arm is at 45 degrees of abduction, the ligament becomes strained with the humerus in neutral rotation or external rotation. At 90 degrees, the AB-IGHL is tight across the lower one-half of the joint in neutral rotation [26,42]. The AB-IGHL moves cranially when the arm is placed in external rotation to traverse the mid-portion of the anterior surface of the joint. In this position, the AB-IGHL is the primary static restraint to anterior dislocation [26,42].

Figure 2-9. The IGHL complex cradles the humeral head in neutral rotation at 90° abduction (a). Internal rotation rotates the IGHL complex towards the posterior, decreasing its effectiveness in anterior stability (c). As the arm is externally rotated, the IGHL is brought over the anterior portion of the humeral head and is much more efficient in anterior stability (d) [34].

The axillary pouch and posterior band of the IGHL (PB-IGHL) compose the remainder of this ligament complex. Fibers of the axillary pouch run caudally and curve posteriorly to their attachment on the anatomical neck of the humerus [26], and work like
a hammock to suspend the humeral head. The fibers in the PB-IGHL curve anteriorly and cranially towards their attachment site on the surgical neck of the humerus. With the arm at 0° of abduction, this portion of the IGHL was seen to be lax in both neutral and external rotations [26] (Figure 2-6). At 45° of abduction in neutral rotation, the axillary pouch begins to tighten, and continues to tighten with either external rotation or greater abduction. At 90° of abduction, the axillary pouch is tight, but becomes tighter with either external or internal rotation (Figure 2-9).

The posterior capsule (PC) extends across the rest of the glenohumeral joint capsule. It extends from the superior margin of the posterior axillary pouch of the IGHL to the posterior margin of the biceps insertion, located at the apex of the glenoid labrum (Figure 2-10) [48]. The posterior capsule has previously been described [49] as “thin, translucent, and relatively featureless” [48]. Bey et al., after dividing the PC into three equal portions, concluded that the thickness of the superior region was statistically similar to the AB-IGHL (same as the SB-IGHL) [48]. Therefore, this ligament may be more clinically significant than previously thought. The PC serves to prevent posterior dislocation of the humeral head, as well as limiting internal rotation of the humerus with the arm above 40 degrees of abduction in the scapular plane [50,51]. Blasier et al. determined that the posterior capsule acts as a secondary restraint to anterior dislocation when the arm is in external rotation and 90 degrees of abduction [42].
The coracohumeral ligament (CHL) is not considered a part of the glenohumeral joint capsule as it extends from the lateral border of the coracoid process (Figure 2-11). It does, however, merge with the SGHL and the supraspinatus to insert superior to the bicipital groove, and just anterior to the greater tubercle [37]. This robust ligament functions to restrain inferior translation of the humeral head, as well as to check external rotation of the adducted arm [38,37,39,40,53]. Edelson et al. agreed that the ligament suspends the humerus, but went further to state that the CHL was taut in flexion and external rotation, and that the ligament stiffened when moving the humeral head anterior or posterior in the sagittal plane [53].
Figure 2-11. Anterior view of the glenohumeral joint. The CHL extends from the base of the coracoid process, extending over the humeral head to merge with the SGHL and split over the bicipital groove, anchoring the biceps tendon. The ligaments are shown as separate structures, without the capsule, for clarity [52].
3 DEVELOPMENT OF COMPUTER MODEL

3.1 Computed Tomography Images

The three bones of the shoulder joint that were used in the computational model were derived from computed tomography (CT) scans acquired from the National Library of Medicine’s Visible Human Project (VHP). The images were obtained from the cadaver of a 39 year old male, and were selected for the VHP to become the basis of a “digital image library of volumetric data representing complete, normal adult male” anatomy [54]. These same images have been used for many clinical medicine and biomedical research projects since their availability in November 1994.

A CT scan is actually a compilation of a series of two-dimensional X-ray slices that are taken around a central axis of rotation. The X-rays are then processed by a computer and can be converted into a three dimensional image. The collection of CT slices consists of 1,871 cross sectional images at 1 mm intervals. Each 2-D image has a resolution of 512 x 512 pixels, with each pixel having 12 bits of data. Each pixel thus has 4,096 (2^12) variations for grey tone (to represent density). Each cross-sectional image shows a snapshot of the density of different regions within that particular slice. A slice of a CT scan of the abdomen can be seen in Figure 3-1 which shows the spine and pelvis bone easily.
The first step in the process of creating 3D bones from 2D CT images involves importing the images to a software program capable of handling the image data. Various software packages exist to display CT images, but one of the more well known programs is Mimics, available from Materialise N.V. [56]. Mimics allows users to import and stack the CT data, perform any editing of the images that may be desired, transform the pixels into voxels (a 3D pixel), and then assemble these voxels into a 3D solid body.
model. Only the regions of the scan data pertaining to the shoulder and upper extremity were imported into Mimics for compiling and editing.

3.2 Editing of Computed Tomography Images

Mimics operates on the principle of “masking” the data to eliminate portions not relevant to the desired goal. Within one set of scan data there is a multitude of information that may not be needed and can essentially be hidden from view by applying what Mimics calls “Masks.” Each mask created represents a different portion of data (copied from the original), and can be edited separately. As the editing process evolves, the masks begin to represent only those portions of the data set that are important to the user. At the conclusion of the editing process, three masks for the three separate bones will result, with the remaining data in the image set hidden from view.

The most dramatic editing effect applied to the CT images is the application of “thresholding”, and is typically the first step when creating a mask. This process selectively removes data from the CT image according to the radiodensity (determined by the level of grey scale) of each pixel, and thus permits the user to filter out data that corresponds to less dense material. Within Mimics, the user can select from pre-defined threshold values that correspond to Hounsfield units of radiodensity. By selecting the preset threshold for bone, 226 to 2603 Hounsfield units, only the pixels that were picked up as regions of bone will be displayed.

Working with the images requires a large amount of processing power. This issue can be alleviated by reducing the working range of the images. The “Crop” function in Mimics can be used to section off areas of the images that are unnecessary to the project.
The CT scans of the cadaver from the VHP were taken in approximately an anatomic position, thus all data superior to the scapula and clavicle, and all data inferior to the distal end of the humerus was removed with the Crop function. Further, only the left shoulder was used in the study, so everything medial to the scapula was also removed. The section of image data left included the three bones of interest and could then be further subdivided into separate masks for each bone.

This small region of data for each bone could then be visualized by creating a preliminary 3D model of the remaining data. The 2D image data for each bone was assembled into a 3D body, by selecting “Create 3D Body from Mask.” This function takes all of the voxels, assembles them together, and creates a body constructed of nodes (points or vertexes) connected by straight lines (Figure 3-3). The nodes and lines together create the surface mesh which is usually composed of triangles, but could be connected using other shapes such as tetrahedrons. The body shown is a virtual resemblance of the scapula, but is far from usable in a computational model. Further editing is required to remove any noise in the data due to occasional poor data collection or scan resolution issues.
Acquiring CT images is a delicate process and is susceptible to any number of problems that will result in poor image quality. Since the images from the VHP were taken at 1 mm intervals in 1994 and included a significant amount of noise, further editing of the bones was required to achieve the desired surface smoothness, especially the articulating surfaces. There are many techniques to remove noise from the image data, but these should be used in sequence of how precisely the techniques can remove the noise.

The following techniques are performed in either 2D or 3D. It is important to remember that Mimics makes changes in 2D while editing in 3D, and vice versa. The first technique, “Edit Mask in 3D,” is the most powerful and it removes the highest number of bad pixels at once. Mimics can easily render a mask in 3D, and with this tool the user can select regions of connected voxel data that may be separate from the bone.
This tool efficiently removes noise data that is not connected to the surface of the bone. The second technique, called “Morphology Operations,” works to expand and contract connected voxel data. The “Dilate” function bridges gaps and holes by expanding voxels, then reduces the voxels with the “Erode” function. Another technique involves returning to the threshold value for Hounsfield units discussed previously, and altering the value so only denser pixel data is displayed.

The last and most laborious technique requires the user to view every 2D image slice for each bone. The noisy pixels can easily be seen as outliers around the body of the bone. As the user scrolls through the images, the outline of the bone is visible, along with any holes that are missing in the outer surface border of the bone.

Figure 3-4. Image slice through humeral head, before (left) and after (right) repairs. Notice missing pixels (noise) in the outline of the humeral head. Noise is present as can be seen by the green specks outside the humeral head. The glenoid on the scapula can also be seen in this image.

The “Edit Mask” feature allows users to draw in missing pixels or delete extra pixels from the image data. A “Multiple Slice Edit” tool is available to make this process more efficient. Using all these methods of editing the image data to remove noise will result in a more defined surface of the bony anatomy.
When all the surface editing was completed, Mimics was used to perform a cavity fill to create a solid body. The surface data was corrected in the previous step, but the assembled 3D bodies were still mostly hollow. To fill the interior space, the “noise reduced” 3D solid body was then transformed into “Polylines,” meaning circumferential lines were computed that contoured the surface of the 3D body (Figure 3-5). Once these lines were calculated, a “Polyline Cavity Fill” function was performed to fill the voids within the bone.

Figure 3-5. Humeral head showing concentric polylines which were used to fill the interior space with solid material.

3.3 Remeshing the 3D Body

With the interior regions of the 3D bones solid and the exterior surfaces devoid of noise and gaps, the next process involved a separate module within Mimics called “Remesh” (part of the FEA module). Earlier in the image editing phase, Mimics created
a mesh whenever 2D images were transformed into 3D objects. The mesh created here
was largely irregular because it was based only on the pixel and voxel data. An
optimized mesh consists of similarly shaped and sized triangles, as well as no sharp
extruding edges (Figure 3-6). A regularly and consistently generated mesh performs
better in computational simulations because it can reduce discrepancies in the data and
computation time.

![Irregular mesh (left) and optimized mesh (right). Equilateral triangles are much more apparent in the optimized form.](image)

Within Mimics Remesh are a multitude of tools available to tailor the mesh to the
desired characteristics. The number of triangles must be reduced to fewer than 10,000 as
SolidWorks, the computational modeling program we use in the next step, has a solid
body import limitation of 10,000 triangles. The outer surface must be as smooth as
possible, with special attention paid to the articulating surfaces of the humerus and
scapula. The quality of the triangles must also be as high as possible. A perfect mesh
would consist of all equilateral triangles, but in a complex solid body this is impossible.

Mimics Remesh uses various measurements of shape quality for determining the
quality of the triangles. For instance, an equilateral triangle would have a shape quality
ratio of 1 if the length of the shortest side was divided by the length of the longest side, since all sides are equal length. The number would be less in any other triangle, and thus triangles with a lower shape quality would be less ideal. A histogram showing the number of present triangles with varying shape qualities helps the user to determine how to direct the formation of the mesh.

The first step to acquiring the desired mesh quality was to apply the “Auto Remesh” tool. The settings for this tool used R-in/R-out as the shape measure, with the inspection measure being the largest angle of the triangle, and an area growth ring of 1. The R-in/R-out is a ratio of the radius of an inscribed circle to the radius of an ascribed circle around the three points of the triangle. The R-in value (the smaller radius) is normalized by a factor of two so that the shape measure is 1 for a perfectly equilateral triangle. The shape quality threshold was set at 0.3, with a maximum geometrical error of 0.075. The Auto Remesher chose those triangles that were below the shape quality threshold and reconstructed them to be more equilateral (above a 0.3 threshold).

A “Smoothing” function was also applied to remove variations in surface elevation. The function looks at one node with respect to the surrounding nodes, and relocated that node so that it is more in line with the other nodes. The smoothing factor is variable to permit the new node location to rely either more on the old node location, or on the location of the other surrounding nodes. The smoothing factor used was 0.7, with 13 iterations, meaning the new node placement was calculated for each node in the mesh 13 times. Afterward, “Triangle Reduction” was performed with default settings to reduce the overall number of triangles in the body. The settings used were Flip Threshold Angle
of 15, Geometric Error of 0.02, and 12 iterations. Following this, the “Triangle Reduction” process was performed again, but with varying values for the geometric error with a target of achieving just under 10,000 triangles. Each time the geometric error was changed, the previous attempt was deleted so there were no consecutive reductions. A final “Smoothing” was performed at 0.7 smoothing factor for 5 iterations.

Before completing the remeshing, a final tool was utilized called “Auto-Fix” that runs a consistency check over the mesh. Occasionally, malformations of triangles and inconsistencies in the surface geometry lead to a non-unified mesh. This will present problems later when importing into Solidworks, as that program will recognize a single 3D object as two separate objects. The Auto-Fix tool reviews each triangle and checks that the mesh has been properly formed. This process was similar for all three bones in the project.

When the remeshing process was complete, all three bones were digitally composed of fewer than 10,000 triangles, the triangles that composed the mesh were of optimum quality, and the surface of the bones was as smooth as possible. Exiting Mimics Remesh will return the 3D objects to Mimics, and allow them to be exported as a STL (Stereolithography) file. This file format is used by a wide variety of engineering applications, and simply provides the data to reconstruct the triangles in 3D space.

3.4 Import to SolidWorks/COSMOSMotion

Once the 3D body is in the STL file format, SolidWorks can read the file, rebuild the triangles, and display the 3D geometry of the meshed bone. Each bone was imported separately as a solid body and saved as its own SolidWorks part file. Then, an
assembly was created that brings the three bones together. Within an assembly, the part files are free to move with respect to each other until the user defines their positional relationships. This can be done by “Mating” two components, for instance making two parts’ origins coincident, or two parts’ faces congruent. To assemble the bones back into their initial positions during the CT scan, it was a matter of making each of the part files’ origins coincident, then mating their reference planes.

COSMOSMotion is an add-on package for SolidWorks that enables users to simulate motion by applying forces, springs, dampers, and displacement. The simulation is highly customizable, and allows users control over all the force elements within the system. Further, the program can simultaneously store data for any number of variables throughout the simulation such as linear displacements, applied force, reaction force, and contact area. This data can then be exported and saved for later analysis. The current
study utilized COSMOSMotion by simulating action-only forces as muscle forces, ligaments as tension-only springs, dampers to portray viscoelastic effects, linear motors to apply external perturbations, and a force to represent gravity.

3.5 Ligament Structures

3.5.1 Origins and Insertions

Origins and insertions of the ligaments were mapped onto the humerus and scapula using anatomical textbooks, literature, and electronic sources [3,19,26,33-36,44,46,58-62]. Since the glenohumeral joint involves so much variability in anatomy, multiple sources were used as a basis for determining the attachment site for the ligament structures. In an effort to make the results more repeatable, an article by Bigliani served as the foundation for representing the origins and insertions [44]. In order to consistently dissect the cadavers in the study, Bigliani chose a coordinate system that divided the IGHL into three representative regions by means of degrees or sectors of a 360° circle (Figure 3-8). From the picture, it was assumed that the circle was inscribed on a plane that was parallel to the face of the glenoid rim. The long axis of the glenoid cavity served as the vertical axis. At the center of the circle lies the geometric center of the glenoid cavity. On the humerus, it was assumed that the long axis of the humerus served as the vertical axis, and the horizontal direction was parallel with a line drawn between the two distal epicondyles of the humerus. The center of the circle lies at the geometric center of a sphere fit to the humeral head. These assumptions were used as a guiding tool to choose where the insertions and origins of the ligaments should attach.
Figure 3-8. Coordinate system from Bigliani to determine regions of IGHL [44]. Insertions on scapula for IGHL posterior axillary pouch (also called PB-IGHL): 210° to 270°, axillary pouch: 270° to 330°, superior band (also called AB-IGHL): 330° to 30°. The coordinates are shifted clockwise (for left shoulder) on the humeral surface between 20° and 30°. For example, IGHL superior band on humerus would be from 310° to 10°.

In the computational model, a “compass” was drawn on a plane parallel to the glenoid rim, with the axes lined up as described above (Figure 3-8). A splined loop was drawn around the rim of the glenoid to guide the positions of the ligament origins. The first ligament to be mapped was the IGHL and its attachment points were placed approximately 2 mm outside of the splined loop. Considering the variability within the population, the ligaments may attach either on the glenoid rim itself, to the glenoid labrum, or medial to the labrum attachment directly to the scapula [43,46,47,63]. The value of 2 mm was chosen for consistency purposes.

The region of origin for the IGHL on the scapula covers fully half of the entire circumference of the circle, from 210° to 30°. Once the IGHL had been positioned, its origins were used to approximate the insertions for the remainder of the joint capsule. The remaining portions were divided up using a few more assumptions. First, the posterior capsule extended from the posterior edge of the posterior axillary pouch all the
way to the apex of the glenoid cavity, where the biceps tendon typically inserts. This would translate into a region that covers 210° to 90° on the scapula on Bigliani’s coordinate system. The SGHL insertion on the scapula was set at 75°, while the MGHL insertion was at 45°. The locations of the ligament insertions on the humeral head were rotated 20° clockwise (for a left shoulder), and were positioned with respect to a spline loop drawn around the edge of the articular surface.

Since the IGHL encompassed such a large area of the joint, it was deemed necessary to simulate the function of the IGHL by several ligament linear springs. Each of the three regions of the IGHL was split into two separate ligament forces, equally spaced within the sector. Instead of only three vectors describing the IGHL, a total of six were used to fill in the large gaps. First, this makes the model more representative of the actual anatomy of the body. Second, since the IGHL has been known to be an important stabilizer in anterior displacement, the division of the ligaments allowed greater detail in

Figure 3-9. Compass drawn on scapula (left) and humerus (right) to describe ligament origins. The splined loops represent the rim of the glenoid of the scapula and the margin of the articular surface on the humeral head. See Appendix B for specific degrees pointing towards attachment site.
the data to compare the different regions of the IGHL to each other. The posterior capsule region has been characterized by three separate regions by Bey et al, and thus was modeled using three different spring elements. The remaining ligaments, the SGHL, MGHL, and the CHL, were each modeled with only one spring element.

3.5.2 Ligament Modeling

Ligaments were modeled as tension-only springs in order to more closely approximate the material properties of ligament tissue. Ligaments function like rubber bands, in that they provide resistance when pulled upon, however they produce no restraint to compression. Therefore, a simple spring element could not be used in our model to represent a spring, as it would produce an opposing force to compression. When a ligament shortens in length in the body, it simply deforms or folds over itself. In order to model a ligament as a tension-only spring, we must first know the material properties of the ligament.

The two relevant material properties that are required for modeling a ligament are the stress free length (SFL) and the stiffness. Most ligaments in the human body are under a small amount of strain, usually about 2%. This small stretch in the ligament is what provides many joints their inherent stability, as the ligament is normally compressing the joint. When the ligament is excised from the body, it shrinks to its original length. This excised length is the measurement when the ligament is not tense, the stress free length. For each ligament in the glenohumeral joint to be modeled accurately, the stress free length for each ligament must be known.
Research of the literature provided most of the necessary SFLs, though some had to be approximated. Additionally, there were some inconsistencies in the data that are reported in the table below as well.

<table>
<thead>
<tr>
<th>Ligament</th>
<th>Author</th>
<th>Free Length (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>CHL</td>
<td>Average of Bigliani IGHLs</td>
<td>40.7*</td>
</tr>
<tr>
<td>SGHL</td>
<td>Novotny, 2000 [4]</td>
<td>40.7* (Average of Bigliani IGHL?)</td>
</tr>
<tr>
<td></td>
<td>Debski, 1999 [18] (ten cadavers)</td>
<td>44.1 ± 2.9</td>
</tr>
<tr>
<td>MGHL</td>
<td>Novotny, 2000 [4]</td>
<td>40.7* (Average of Bigliani IGHL?)</td>
</tr>
<tr>
<td></td>
<td>Debski, 1999 [18] (ten cadavers)</td>
<td>49.4 ± 8.1</td>
</tr>
<tr>
<td>IGHL – Superior**</td>
<td>L.U. Bigliani [44] (sixteen shoulders from 9 male 7 female)</td>
<td>41.3 ± 4.5*</td>
</tr>
<tr>
<td></td>
<td>Ticker [58] (eight shoulders, eight from 4 male 4 female)</td>
<td>44.4 ± 6.3</td>
</tr>
<tr>
<td></td>
<td>Debski, 1999 [18]</td>
<td>50.1 ± 4.9</td>
</tr>
<tr>
<td>IGHL – Ant. Pouch**</td>
<td>L.U. Bigliani [44]</td>
<td>39.8 ± 5.6*</td>
</tr>
<tr>
<td></td>
<td>Ticker [58]</td>
<td>42.5 ± 4.7</td>
</tr>
<tr>
<td></td>
<td>McMahon [64] (eleven cadavers)</td>
<td>37.1 ± 0.9</td>
</tr>
<tr>
<td></td>
<td>Debski, 1999 [18]</td>
<td>50.5 ± 7.8</td>
</tr>
<tr>
<td>IGHL – Post. Pouch**</td>
<td>L.U. Bigliani [44]</td>
<td>41.0 ± 4.2*</td>
</tr>
<tr>
<td></td>
<td>Ticker [58]</td>
<td>43.4 ± 4.1</td>
</tr>
<tr>
<td>PC - Inferior</td>
<td>Average of Bigliani IGHLs</td>
<td>40.7*</td>
</tr>
<tr>
<td>PC - Middle</td>
<td>Average of Bigliani IGHLs</td>
<td>40.7*</td>
</tr>
<tr>
<td>PC - Superior</td>
<td>Average of Bigliani IGHLs</td>
<td>40.7*</td>
</tr>
</tbody>
</table>

Table 3-1. Overview of ligament lengths found in the literature, as well as those values which were chosen for the computational model (*). **Debski reports 50.1 ± 4.9 mm for anterior band IGHL, 50.5 ± 7.8 mm for posterior band IGHL but does not report superior band IGHL (ten cadaveric specimens). Also, he used ligaments lengths from Bigliani instead of his own for the IGHLs.

Anatomical variations are commonly found in the shoulder leading to large values for standard deviations, as can be seen for most of the values in Table 3-1. Debski, Bigliani, McMahon, and Ticker published their findings based on cadaveric studies [18,44,58,64]. Curiously, Debski chose to use Bigliani’s values for the AB-IGHL (aka superior band of IGHL) and the PB-IGHL (aka posterior axillary pouch of IGHL) in his analytical model though he measured the lengths of those ligaments from the cadaver
study [18]. The computational model here uses Bigliani’s lengths for the three regions of the IGHL, and averages the lengths of these three to determine approximated values for the SGHL and MGHL. Novotny used the same method in his computational model [4], though Debski’s actual lengths were available for the SGHL and MGHL at the time [18]. Novotny’s decision to approximate all ligaments of the shoulder as being near the same length was reflected in this model. Therefore, the CHL and all three regions of the PC were approximated as 40.7 mm for lack of evidence in the literature.

The next important material property for the ligaments in the model is the stiffness value. Some papers report the stiffness value for a ligament; others report a modulus. These measurements differ as the stiffness already factors in the cross sectional area of the ligament, and can be used directly in the spring equation. The modulus must be converted to a stiffness value by factoring in the geometric measurements of the ligament according to the equation below.

\[
Stiffness = \frac{Modulus \times Width \times Thickness}{Length}
\]

Properties for calculating the stiffness from the modulus can be found in Appendix A. The stiffness values used for all the ligaments are presented below in Table 3-2. The values for the CHL and SGHL were the only stiffness values reported in the literature, the others had to be computed from the modulus with either known geometry or approximated ligament geometry.
Table 3-2. Ligament stiffnesses used for tension-only spring definition. Mean value was utilized in all cases. *Approximated geometry to calculate stiffness, see Appendix A.

Tension-only springs were not a built-in feature in COSMOSMotion and had to be custom programmed as an action/reaction force to get the desired mechanical behavior of a ligament. Each instance of a ligament required its own custom code that included the stiffness and stress free length. The action/reaction force was programmed as an “IF” statement in FORTRAN code. The code takes the instantaneous length of the ligament and compares it to the stress free length. If the current ligament length is longer than the stress free length, the spring stiffness is applied in relation to the difference in length from the stress free length. If the ligament is shorter than the stress free length, there is no tension developed. The code is written in COSMOSMotion as it appears below:

\[
\text{If((Ligament Length)-SFL} < 0, 0, \text{Stiffness} \times (\text{Ligament Length}-\text{SFL}) \]

In SolidWorks 2008, a result plot must first be created that actively measures the distance between the insertion and origin point during the simulation. This result plot can then be called into the FORTRAN code as the ligament length so that the tension produced changes dynamically with the ligament’s length. Defining the ligaments in this
manner most accurately describes the physiologic function of the ligaments as a mechanical structure.

3.5.3 Force Line of Action

Unfortunately, COSMOSMotion does not have the ability to direct an action/reaction force around an object. Instead, a force can only act along a straight line from its origin to its insertion, leading to potential problems with the force being directed through the bone of the humeral head, as opposed to around it. This presents problems such as incorrect ligament lengths (that are used to compute the tension generated) and incorrect moment arms of muscles or ligaments. This issue is common in computational models of the shoulder joint and is generally referred to as “wrapping” of either muscles or ligaments.

To resolve this issue, two techniques were attempted. The first involved a series of beads connected by action/reaction forces. These acted like a pearl necklace rolling over the head of the humerus (Figure 3-10). Each ligament whose line of action passed through the bony anatomy would require a separate chain of beads.
This theory was plausible, though required an excessive amount of computing power (even for only one chain of beads) due mainly to the fact that there were more parts in contact with each other. Contact is the most intensive part of the modeling process, and thus requires the most computation time. Removing all but one bead seemed to provide good results, though once two beads were present in the model (one bead for each ligament), the computation time increased dramatically. It seemed that the beads had too many degrees of freedom, allowing them to bounce off the surface of the humerus, and then spring back once the tension-only spring expression was activated.

The second technique used swinging arms for each ligament that required it. These arms were curved to follow the contours of the humeral head. Their degrees of freedom simply allow them to rotate about the ligament insertion on the humerus, with the tip of the cone being parallel with a line drawn from the insertion point to the...
geometric center of the humeral head. Seven swinging arms were used in all, two for each of the three regions of the IGHL and one for the MGHL.

![Image of GHJ showing swinging arms for the IGHL ligament](image)

**Figure 3-11. Inferior aspect of GHJ showing swinging arms for the IGHL ligament. Anterior direction is to the left. The blue arrows are the tension-only springs representing the ligaments. IGHL has been broken into three regions: anterior band (AB), axillary pouch (AP), and posterior band (PB). Humerus is in 90° abduction in scapular plane with external rotation.**

This technique worked well, and was used in the final version of the model. It did add some complexity to the model, as the expressions earlier defined for the tension-only ligament forces had to be updated to take into account the fixed length of the swinging arm. The swinging arms were used for only some of the simulated ligaments: the MGHL and the six regions of the IGHL. The SGHL and the three regions of the PC simply used a straight line distance to compute the ligament stiffness Figure 3-12. This approximated the path of the ligament, but for these ligaments it was determined that the straight line path between the endpoints of the ligament did not pass through a significant portion of
3D bodies representing the bones. Additionally, the ligament length during both rotations at all ranges of displacement was sufficiently below the stress free length so the ligament would not be generating a tension in any case. Since the length of these ligaments were far below their stress free lengths, swinging arms for these ligaments would have extended computation time while not providing more accurate results.

![Figure 3-12. Anterior (left) and posterior (right) views of simulated ligaments. In the anterior view, the ligaments shown with blue arrows are (from superior to inferior): SGHL, MGHL, IGHL AB 1 and 2. In the posterior view (from inferior to superior): IGHL AP 1 and 2, IGHL PB 1 and 2, and the posterior capsule (inferior, middle, and superior).](image)

3.6 Muscular Simulation

As described earlier, there are two functional groups of muscles that support glenohumeral joint stability. The rotator cuff muscles encompass the joint capsule and provide most of the centering force that retains the humeral head within the glenoid socket. The three regions of the deltoid lie superficially to the rotator cuff and mostly assist in movements when the arm is at higher elevations. The addition of these muscle forces to the computational model will provide the necessary stability and movement of the joint.

The anatomy of the insertions and origins of the tendons of these muscles were studied through the literature, electronic media, and anatomy textbooks [19,28,29,35,59-
A description of each of the origins and insertions can be found in the anatomy background in Chapter 2.2.

Each muscle relevant to the study was simulated with force elements. The number of force elements that represent each muscle depends upon the area of insertion on the humerus, as well as the overall breadth of the muscle closer to the origin. To determine how many force vectors are required to simulate each muscle an arbitrary value of 1 cm was chosen along with the size of the muscle at its origin. Measurements by Dugas of the areas of the tendon insertions of the rotator cuff muscles were used to approximate the number of vectors appropriate for each muscle [28]. Mochizuki more recently described the insertions of the supraspinatus and infraspinatus with greater detail [29]. For both the subscapularis and infraspinatus, three force vectors were utilized to simulate the muscles. For the supraspinatus, two vectors were employed. The teres minor and biceps used only one vector. Finally, one vector was used to represent each of the three regions of the deltoid: anterior, middle, and posterior.

For each force vector, an origin point on the scapula and an insertion point on the humerus were chosen to represent the muscle force for that region. In the human body, the rotator cuff muscles partially wrap around the glenoid, creating a new line of action. This anatomy was approximated by mounting blocks and pins to the scapula that redirect the forces in a more anatomically correct direction. These features were placed and oriented to direct the muscle force vector above the surface of the scapula, through the anatomic cross section of the muscle body.
3.7 Force Damping

When forces are applied in computational models, they can be either ramp forces or instantaneous forces. When the forces move solid bodies too quickly, the computation time can be significantly extended. Adding in dampers helps to slow the simulation down, without compromising accuracy. In other words, dampers react to changes in velocity, so when the solid bodies become stationary, the dampers apply no force. Initially, dampers were added into the tension-only spring code for ligament definition using a velocity dependent variable that controlled how much force was applied. This method was not as repeatable as adding separate dampers in parallel to each ligament, with a constant value damping factor. Each damper that was added had a value of 0.1 N/(mm/s), meaning for each mm/s of change in distance, a 0.1 N force was applied to
slow down this velocity. Additionally, torsional ligaments were attached to the pivot points of the rotating swing arms with a value of 1 N*mm/(deg/s), which helped to slow down their rotational velocity. This damper applies a torque to resist motion that is dependent on the rotational speed of the object. Adding dampers to the model helps to stabilize the solid bodies and smooth out the results, without altering the steady state value.

3.8 3D Contact

When solid bodies in a computational model come in contact with each other, the program determines the reaction forces by computing distances between the points that compose the surface geometries. Computational models do not take into account material stiffnesses of the bodies, as no deformation is allowed. Rigid solid bodies simply use the geometry of the surface to determine the correct reaction. When defining the 3D contact, there are a few parameters that can alter the results. Since the contact is calculated by the interaction between the surface geometries, these parameters allow for variation of contact activation. This penetration before contact can be altered, as well as the rebounding stiffness and damping of the contact. The changing of these values does not generally affect the overall results of the simulation since the bodies are not deformable, but they can change the way the two bodies interact. Multiple simulations were run, varying these 3D contact parameters to better understand the differences they cause (Figure 3-14).
Figure 3-14. Comparison of 3D contact settings for humerus to scapula contact. The four variables are stiffness (S), exponent (E), damping (D), and penetration (P). Changes in these values do not significantly affect the overall trends or magnitudes, though can provide a smoothing factor to the data points. This graph does not represent results of the experiment, but is just an attempt to determine 3D contact settings that will provide consistent data.

Changing the parameters of the 3D contact settings does not have a large effect on the trends or magnitudes of the data. Therefore, the settings that provided the smoothest, most consistent results were utilized for the computational model. Though the user can tailor each of these variables to the desired need, the right combination happened to be greasy aluminum, a preset in the program. For comparison to the other user defined settings, the stiffness for greasy aluminum was 33,300 N/mm, with an exponent of 1.5, a damping of 27.95 N/(mm/s), and a penetration depth of 0.1 mm.

At this point, the necessary structural features of the glenohumeral joint have been developed in the computational model. Solid bodies created from CT scans represent the geometry of the humerus, scapula, and clavicle. Ligament structures have been modeled
as tension-only springs with a redirection around the humeral head. The rotator cuff and three regions of the deltoid were added to the model as linear force elements. The next phase of the process reviews the cadaveric study to mimic the experimental methodology. The positioning of the bones, motion constraining devices, and external loading must now be added to the model so that a comparison of the results can be made.
4 ANTERIOR STABILITY STUDY

4.1 Overview

Validation of the computational model required direct comparison to a cadaveric experiment performed on the shoulder. A common form of injury to the shoulder involves an initial dislocation due to traumatic force followed by subsequent joint laxity. Typically, this type of injury causes damage that leads to an increased potential for redislocation in the future. Stability in the shoulder joint arises from a delicate balance of active (muscular) and passive (ligamentous) stabilizers that work together to center the humeral head within the glenoid. The large range of motion of the shoulder stems from the minimal bony constraint of the joint, but this requires the soft tissues to perform the majority of the joint’s stability. Many cadaver experiments have been performed to determine the relative contributions to stability that each of the structures in the joint provide, though many of these experiments are not as comprehensive as others [26,42,62,66,61]. An experiment that simultaneously studied the effects of muscles and ligaments and their respective roles in anterior stability was chosen to validate the developed computational model. “Anterior Glenohumeral Stabilization Factors: Progressive Effects in a Biomechanical Model” by Malicky et al. was selected as a comparison as this particular study provided a detailed description of the methodology of their cadaveric experiment [31].
The goal of Malicky’s study was to determine the individual contribution to glenohumeral stability of a set of defined muscles and ligament regions. The experiment essentially applied a constant displacement to the humeral head and measured the required force to achieve this displacement. Malicky hypothesized that as muscles and ligaments were removed from the shoulder, the force required to achieve the displacement would decrease. By comparing the decreases in force due to the absence of a particular structure, the most important stabilizers could be determined.

The experiment was conducted in two parts. The first part defined the values of the muscle forces used to stabilize the shoulder before performing the displacement part of the experiment. They began with eight shoulders from four cadavers. The shoulders were dissected free of the surrounding tissue, leaving intact the tendon stumps of the rotator cuff and deltoid muscles, as well as the entire ligamentous capsule. The scapula and humerus were injected with polyester resin to allow rigid mounting. Dacron cords were either clamped or sutured to the tendon stumps and directed through the estimated cross section of the muscle by using eyelets and pulleys (Figure 4-1).
Once the cords had been connected to the tendon stumps and insertions, forces were applied using pneumatic cylinders to elevate the arm to 90° of abduction in the scapular plane (30° anterior to the frontal plane). The prescription of muscle forces was based on “electromyographic data, lever arm information, cross-sectional area data, and clinical knowledge” [31]. Generally, forces were applied to the rotator cuff muscles in larger amounts during the lower angles of abduction, and then increased by smaller increments through the higher elevations. The deltoid muscles were used mostly at the higher ranges of abduction. The average force values required to elevate the humerus to 90° of abduction in the scapular plane were calculated for all eight shoulder specimens, and were used as the stabilizing forces for the next part of the experiment.

The second phase of the experiment required the humerus to be fixed at the distal end so as not to interfere with the motions of the humeral head as it reacted to the applied anterior displacement. The distal portion of the humerus was constrained to a ball joint
that allowed the humerus to slide along its axis, but was unable to move in a transverse
direction. Additionally, the rotation of the humerus was fixed with a tie-rod that was able
to move, but only within a single plane. These constrictions remove three degrees of
freedom, but still allow the humerus to pivot about the ball joint and slide along its long
axis (Figure 4-2).

Figure 4-2. Image depicting six translations of the humeral head (D) allowed by the constraints [42].
A tie rod (C) limits rotation of the humerus about its long axis, used for setting internal and external
rotation. A rod (A) was set in the intramedullary canal (B) which was used to position the distal
humerus within the scapular plane. The humeral head is only restrained in motion by simulated
muscle forces (F) and capsuloligamentous structures (E).

A servohydraulic testing machine was used to displace the humeral head 10 mm
in the anterior direction. During the displacement, the muscle forces defined earlier for
attaining the 90° abducted position continued to be applied. The “standard case” was
defined as 100% force load for each muscle. The displacement procedure was then
repeated after varying each muscle load (0%, 50%, or 150% of the standard case) or
performing a capsular cut of a ligament region. Both neutral and external rotations were
tested for each of the shoulder specimens. By varying each of the muscle forces and
repeating the displacement procedure, a difference in required force to subluxate

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(partially dislocate) the humeral head could be measured and compared across the varying combinations.

Initially, the first part of the experiment (the abduction phase) was simulated with the computational model. The average muscle forces calculated from the eight shoulders were applied to the humerus, causing it to rise from the initial CT scan position. Using the average values for each muscle, plus the standard deviation, only permitted the humerus to rise to 75° of abduction (Table 4-1). A few key differences may explain the lessened abduction. First, the forces applied were averaged over the eight shoulders. Each shoulder required a different combination of muscle forces to reach the required position, thus the forces could be altered so that the humerus would reach the required position. The table below shows the average set of muscle forces, compared with the set of muscle forces that the computational model required to approach the target position. Second, the geometry of the articular surfaces, the positions of the insertions and origins, and the ligament stress free lengths that differ from shoulder to shoulder may have played a role in the discrepancy. This comparison was more of an exercise to see how the computational model would react to the standard case of muscle forces, and if it would approach the experimental results.
Muscle Malicky et al (N) Computational Model (N)
Supraspinatus $67 \pm 7$ 74
Subscapularis $53 \pm 7$ 60
External Rotators $55 \pm 10$ 65
Biceps 31 31
Anterior Deltoid $37 \pm 25$ 62
Middle Deltoid $49 \pm 14$ 64
Posterior Deltoid $37 \pm 13$ 50

Table 4-1. Standard case (100%) muscle forces to abduct humerus to 90° in the scapular plane with the standard deviations from eight shoulders [31]. The computational model required the standard case forces, plus their standard deviations, to reach only 75° of abduction.

The discussion of the cadaveric experiment detailed the infinite number of combinations of muscle forces to abduct the humerus. The forces to raise the arm were applied in real-time by the investigators, varying the forces as necessary to achieve the desired position. Choosing values to cause the computational model to rise to 90° in the scapular plane would provide little useful information for the rest of the study. Since the results of the cadaveric study were presented as an average of the group, the average input muscle forces had to be used to generate results to which to compare to the cadaver results. Therefore, the computational model was configured to parallel only the second stage of the experiment, the anterior displacement phase, using the averaged standard set of muscle forces. The following sections describe the setup of the computational model with comparisons to the experimental setup.

4.2 Bone Positions

4.2.1 Setting the Scapula and Clavicle

The second stage of the cadaver experiment began with the humerus in a position of 90° abduction in the scapular plane. To mimic the correct anatomic positions of the bones, the scapula was rotated upwards to coincide with the upward rotation of the
humerus. Since the CT scan had been performed in the supine position with arms at the side, the positions of the bones had to be changed to match the second stage of the experiment.

To consistently position the bones in space, coordinate systems derived from anatomical landmarks on the bones were used as references. Published 3D kinematics data detailing the positions of the bones at 90° abduction in the scapular plane was then used to orient the 3D solid bodies within the computational environment. The scapula served as the reference for defining the anatomical coordinate system. The first step in this process involved defining the scapula coordinate system.

The International Society of Biomechanics (ISB) has recently defined local axes for many of the bones in the body to reduce the variability in data that occurs when each researcher defines their own coordinate systems [67]. This facilitates the comparison of data between researchers by standardizing one process of experimental methodology. Since older kinematics papers were used for orienting the scapula in this computational model, a slight departure from the ISB recommended coordinate system was used. Instead of placing the origin at the Angulus Acromialis (AA) as the ISB states, the origin was chosen as the Acromioclavicular (AC) joint to match to the kinematic data (Figure 4-3). The Z axis connects AC to the root of the spine (RS), pointing towards AC. The X axis is formed by a line drawn from AC pointing forward, normal to a plane defined by AC, RS, and the Angulus Inferior (AI). The Y axis points upward, and is perpendicular to both the Z and X axes [67,68]. The scapula coordinate system, along with published kinematic data was used to determine the orientation of the scapula when the arm is in
90° abduction in the scapular plane. Using this data to manipulate the solid bodies will present the scapula in the correct orientation with gravity in the computational environment, so the humeral head can be appropriately oriented to the surface of the glenoid.

![Coordinate system of scapula (posterior view). Origin lies at the acromioclavicular (AC) joint. The Z axis points away from root of spine (RS). The X axis is normal to a plane formed by the inferior angle (IA), RS, and AC. Rotations about the X axis are up/down rotation (UR/DR) of the scapula. Internal/external rotation (IR/ER) occurs about the Y axis. Rotations about Z are termed anterior/posterior tilt (AT/PT) [69].](image)

Physical therapists often utilize 3D electromagnetic tracking machines that can record the positions and orientations of an object in space. When tracking markers are attached to the skin, the orientations and positions of the bones can be recorded through a range of movement. As we have previously defined the scapular coordinate system, a series of rotations around the three axes will position the scapula in its true anatomic orientation. These are referred to as Euler rotations and must be performed in a specific sequence. For the scapula the axial rotations must first be internal/external rotation (around the Y axis), upward/downward rotation (around the X axis), then
anterior/posterior tilt (about the Z axis) [70]. Rotations were determined from a journal article that recorded the positions of the scapula during elevation in the frontal, scapular, and the sagittal plane [71].

Once the scapula was correctly oriented in reference to the anatomical coordinate system, it was “fixed,” meaning the scapula was made stationary. The rest of the solid bodies in the model were positioned in respect to the scapula and the defined anatomic coordinate system. The computational model was designed so that changes to the orientations could be easily made, allowing the scapula to be rotated to any position of interest.

The clavicle was treated as rigidly connected to the scapula, so any rotational changes that the scapula underwent were reflected in the rotations of the clavicle.

4.2.2 Positioning the Humerus

To position the humerus in the same orientation as used in the cadaver experiment, a similar coordinate system was defined again using anatomic landmarks. Following recommendations by the ISB, the origin of the humerus was placed at the center of rotation of the glenohumeral joint. Normally, this location is measured using a 3D motion tracking system with regression analysis, but another method was used in the computational environment. As a substitute, a sphere was fit to the head of the humerus, using the geometry of the humerus above the surgical neck. Mimics was used to calculate this sphere fit, and the coordinates of the center of the sphere were then used in SolidWorks to create a point to serve as the origin of the coordinate system of the humerus. At the distal end of the humerus, the medial and lateral epicondyles serve as the
anatomical landmarks. The Y axis links the origin to the midpoint of the line connecting the two epicondyles (Figure 4-4). The X axis is determined by the line normal to a plane formed by the origin and the two epicondyles. The Z axis is perpendicular to the two other axes [67]. Since the VHP image data does not include the data corresponding to the elbow, the geometry of the epicondyles was missing from the CT scans. To determine the correct orientation of the coordinate system, another 3D body of a different humerus was superimposed on the VHP humerus. The features of the humeral head, and distinctly the bicipital groove, were used to convey the approximate epicondylar axis to the VHP humerus.

Figure 4-4. Coordinate system of the humerus. Rotations around the Y (long) axis are internal and external rotation (IR/ER). Flexion and extension (Flex/Axt) occur around the Z axis, and abduction/adduction (AB/AD) around the X axis.[69]

With the scapula and clavicle in their correct orientations with respect to the anatomic coordinate system, and thus gravity, the humerus was positioned such that its long (Y) axis was perpendicular to the direction of gravity (therefore being 90°
abducted). The long axis was also made to lie parallel to the scapular plane, which was defined as 30° anterior to the frontal plane (Figure 4-5). The definition of the scapular plane varies, but is generally accepted as somewhere in the range of 30-45°. A more commonly agreed upon value is 30° [72] and was chosen for this computational model since the description of the method does not specify the exact scapular plane angle. Abduction in the scapular plane is used commonly as it approximates the mechanical axis of the humerus to be in line with the mechanical axis of the scapula [73]. This allows the abducting muscles (supraspinatus and deltoids) to operate in their most efficient position [74].

![Figure 4-5. Depiction of scapular plane, 30° anterior to frontal plane [75].](image)

The humerus was tested in both neutral and external rotations to study how the position of the humerus affected the anterior stability of the glenohumeral joint. Axial rotation of the humerus can be measured by the position of the forearm when the upper arm is positioned at 90° of abduction (Figure 4-6). If the humerus is in this position with the forearm pointing in the anterior direction, this is defined as 0° of rotation. Lifting the forearm is external rotation, while lowering it is internal rotation. In the cadaveric experiment, external rotation was defined as the position 10° internal from full external rotation. Neutral rotation was defined as midway between full internal and full external rotation.
rotation to allow maximum ligament laxity. Manual torquing was performed to determine the full rotation positions [40].

![Figure showing internal and external rotations of humeral shaft](image)

**Figure 4-6.** Figure showing internal and external rotations of humeral shaft. The humerus was tested in 101.45° and 32.175° of external rotation. The latter position (called neutral rotation) reflects the midway point between full internal and full external rotations to maximize ligament laxity. [76]

Since no data was provided in the cadaveric study for the average rotational positions of the arm, a separate paper was used to determine the angles relating to external and neutral rotation. Reagan et al. measured rotations in 54 college baseball players for both arms [77]. Before using the data, the dominant and non-dominant data were averaged. Neutral and external rotation were then computed according to the definitions given in the cadaver experiment. A line connecting the medial and lateral epicondyles was used as the reference to determine the angle of humeral shaft rotation [65]. A tie rod was then fit to the humerus in SolidWorks that represented the forearm position, perpendicular to the epicondylar axis. Thus, the angle of the tie rod with respect to the scapular plane measured the amount of internal/external rotation. External rotation was set at 101.45° from the transverse plane, while neutral rotation was defined as 32.18° from the transverse plane.
The humerus was then moved into contact with the glenoid surface by defining geometry on the scapula. A reference plane was created that approximated the cross sectional plane of the glenoid (Figure 4-7). A line was then drawn normal to this plane, passing through the deepest point of the glenoid. The center of the humeral head was made coincident with this line, and was moved along the line using the “Stop at collision” feature. This brought the humerus into contact with the scapula and defined their initial relationship. The humerus was fixed in this position until after the motion-restraining devices were positioned around it.

Figure 4-7. Initial positioning of humeral head to glenoid fossa (posterior view). A reference plane was fit to glenoid rim, with a line normal to this plane passing through the deepest point of the glenoid. The center of the humeral head was moved along this line until contact was made between the two bones.
4.3 Restraints to Motion

4.3.1 Ball Joint

From this position, the ball joint was placed at the distal end of the humerus. Instead of creating more 3D contacts to model the ball joint and the distal humerus relationship, mates were used to reduce the degrees of freedom given to the humerus. To mimic the cadaveric experiment, the translations of the distal humerus were removed. This was accomplished by mating the long axis of the humerus to be coincident with a single point (located at the center of the loop of the ball joint). Since this mate was a coincident mate, axial plunge movement was not restricted as the center line could move through the point. The placement of the center of the loop on the ball joint was determined by a diagram found in a later experiment by Blasier et al. that described a two degree rotation due to a 10 mm humeral head displacement (Figure 4-8) [61].

Trigonometric analysis determined a distance of 285.37 mm from the ball joint to the center of the humeral head.

Figure 4-8. A 10 mm displacement imparts a 2° rotation at the distal humerus [61].
The ball joint was further positioned by aligning the faces of the base to the anatomical planes, resulting in the bottom side of the ball joint normal to the direction of gravity (Figure 4-9). The loop through which the humerus passed is merely for visual purposes, as the long axis of the humerus was constrained to move only through the center point of the loop.

4.3.2 Axial Rotation Restraint

To prevent the humerus from deviating from the set position of neutral or external rotation, a restraint was designed to interact with the tie rod attached to the distal end of the humerus. The tie rod, described in Section 4.2.2, was positioned perpendicular to the long axis of the humerus and points in the direction of the forearm. Holding the tie rod in a particular rotation would thus prevent the humerus from rotating about its long axis. However, overly restricting the movement of the tie rod would prevent the humeral head from having the appropriate degrees of freedom. A bar with a slot was designed to fit over the tie rod to restrict rotational movement. The slot was made to allow 1 mm of space between the tie rod and the edge of the slot, allowing a slight change in axial rotation of the humerus. This change in rotation was considered negligible in comparison to the allowed change in translation at the humeral head [31,42,78,61]. The long edge of the bar was made to lie parallel to the long axis of the humerus (Figure 4-9). The bar was also positioned to be symmetric about the center of the tie rod, meaning the tie rod initially had the same amount of space between it and the sides of the slot. The restraining bar was positioned vertically on the tie rod, 90 mm from the center of the ball joint. This dimension was approximated from a picture in an earlier cadaver experiment,
relating the diameter of the humeral shaft to the placement of the tie rod restraining bar [42]. The length of the tie rod extending from the intramedullary canal of the humerus was altered at this time to extend to the center of the ball joint [42].

Figure 4-9. Relationship of ball joint and tie rod restraints to humerus in external rotation (lateral view). The long axis of the humerus is perpendicular to gravity and parallel with the scapular plane. The ball joint is positioned coincident with the long axis 286.37 mm from the center of the humeral head. The restraining bar is centered on the tie rod, which was set for either neutral or external rotation.

4.4 Anterior Displacement

Displacement of the humeral head during the cadaver experiment was achieved by a servohydraulic testing machine that applied “anterior displacements to the humeral
neck and measured the resultant forces required for subluxation” [31]. Within COSMOSMotion, a “linear motor” was set to apply a linear displacement of 10 mm to a separate piston-shaped body. The piston was positioned to contact the humeral head, with its line of movement passing through the center of the humeral head (Figure 4-10).

![Figure 4-10. Orientation of piston to humeral head (superior view). The anterior direction is straight up in the image. The movement of the piston passes through the humeral head and is normal to a plane parallel to the scapular plane (green line).](image)

The description of the exact direction and application of the displacement was quite vague in the cadaveric experiment. First, the paper states that the servohydraulic testing machine was used to “apply anterior displacements to the humeral neck” [31]. By many anatomical textbooks, there exist two “necks” of the humerus. The surgical neck lies just below the head of the humerus, where the bone begins to taper from the head to the shaft. Separately, the anatomical neck is described as the border of the articular surface. The only diagram with a clue comes from a different cadaver experiment by a few of the same authors, albeit a slightly different experimental methodology (see Figure 66).
Both approaches to displacement were attempted, with more consistent results coming from the anatomical neck application. This presented a larger surface on which to apply the piston, with less chance of it slipping inferior or superior on the thin aspect of the shaft of the humerus. Secondly, although the cadaveric experiment states “anterior” displacement, a true anterior displacement would tend to separate the humerus from the glenoid with the arm elevated in the scapular plane. This raises the question of whether the displacement was performed normal to the scapular plane or to the frontal plane. To produce a true dislocation (with all 10 mm of displacement causing the head of the humerus to slide across the glenoid) would require the displacement to occur normal to the scapular plane. Again, this theory is supported by the one diagram from a different cadaveric experiment (see Figure 4-8).

4.5 Summary of Procedure

Slight deviations from the initial positions of the 3D solid bodies will create abnormalities within the output data of the simulation. To acquire consistent results from the computational model, a regimen was developed to position the bones and fixture devices in their correct positions. The bones were positioned first, and then the fixture devices were positioned according to the bones. The scapula and clavicle as one unit were oriented to reflect the scapula’s position with respect to gravity and the anatomical planes of the body. Its position was determined from 3D kinematics data when the arm was in 90° abduction in the scapular plane. For initial positioning purposes, the long axis of the humerus was constrained to lie perpendicular to the direction of gravity and parallel to the scapular plane. The axial rotation of the humerus at this point was set for
either internal or external rotation at this point. Finally, a plane was created that approximated a cross section normal to the glenoid surface. A line was drawn normal to this plane, passing through the deepest point of the glenoid. The center of the humeral head was made coincident with this line, and the humerus was then moved until contact was made between the humeral head and the glenoid socket. Contact was defined by the “Stop at Collision” feature enabled in the “Move” menu. This procedure defined the initial position of the humerus with respect to the scapula.

Before any forces were applied to the humerus, the motion restraints were positioned and then “fixed” in space. The center of the loop of the ball joint was made to lie on the long axis of the humerus, 286.37 mm from the center of the humeral head. The three sides of the base were made to be parallel to the anatomical planes, with the restraining loop pointing superiorly. The tie rod restraint was positioned so its symmetrical center lied on the long axis of the tie rod. Each of these restraints was fixed in place at this point, allowing the humerus to move only constrained by the ball joint, tie rod, and scapula. The two relationships maintained by the humerus kept it centered within the ball joint and limited the axial rotation through the tie rod constraint.

In the cadaver experiment, the initial part of the experiment defined the muscle forces required to raise the arm to 90° of abduction. These same muscle forces were applied to center the humeral head within the glenoid after the distal end of the humerus was restrained by the ball joint. To mimic this centering in the computational model, the muscle forces were applied to the humeral head, allowing it to steady using the defined constraints above. The humeral head shifted slightly when the muscle forces were
applied, and this new position was marked for use during each anterior displacement simulation.

The piston was directed along a line normal to the scapular plane and through the center of the humeral head. The piston was moved along this line until contact was made utilizing the “Stop at Collision” feature of the Move command. A linear motor was then used to apply a 10 mm displacement, pushing the humeral head towards the anterior aspect of the glenoid socket.

One simulation run consisted of the humerus being set to its “muscle-centered” position within the glenoid, followed by displacement caused by the piston. Then the humerus was reset to the “muscle-centered” position, and another simulation was run after removing a selected muscle or ligament. Simulations were run that represented all of the cadaveric tests. Muscles and ligament groups were selectively removed to study their effect on the force required to displace the humeral head from the glenoid. A comparison of these forces reveals which muscles act to better stabilize the joint.
5 RESULTS

Validation of the accuracy of this computational model of the shoulder requires a results comparison to the cadaver experiment which it simulates. COSMOSMotion allows the user to define a wide range of parameters to record throughout the simulation such as the distance between two objects, the force generated through contact, velocities, etc. The program will record these values and present them at the conclusion of the simulation. One of the benefits of computational modeling is its ability to produce a wealth of simultaneous information concerning the simulation each time the program is run. The effect of altering one feature of the computational model can be quickly assessed by analyzing the output data. To directly compare the results of the computational simulation to the physical experiment only one parameter was tracked: force required to move the humeral head in an anterior direction. Additionally, results were recorded for parameters not found in the physical experiment. These included glenohumeral contact force, ligament displacements, and tension developed within the ligaments. Though these cannot be compared directly to the cadaver experiment, they provide further insight into the mechanics of the shoulder joint and can be compared to alternative cadaver studies for accuracy.
5.1 Subluxation-Force Relationship

The force required to subluxate (partially dislocate) the humeral head to a particular displacement was recorded during the simulation. The data below represents the standard case only, meaning all muscle forces were applied and the ligament capsule structures remained intact. In neutral rotation, the force displayed a near linear increase with displacement. The force required to move the humeral head at 10 mm was 127.0 N (Figure 5-1). The physical experiment reported 172 N at 10 mm of displacement, with a rapid increase in force from 0 to 4 mm and a slower increase above 4 mm.

![Figure 5-1. Subluxation force vs. displacement for neutral rotation with all muscles active and intact ligament capsule, computational model results. Force was computed to be 127.0 N at 10 mm of displacement. The cadaveric experiment described a rapid increase in force from 0 to 4 mm, and a slower increase above 4 mm reaching a peak of 172 N.]

In external rotation, the subluxation force peaked at 244.1N at 10 mm of displacement. The physical experiment stated 239 N as the maximum value for force. The computational model again showed a trend that was nearly linear, though a slight rise in force occurred near the 5 mm mark. The cadaver experiment described a slightly
different trend with a large and progressive rise in force above 4 mm of displacement, essentially an exponential rise from 0 mm.

![Graph showing subluxation force vs. displacement](image)

**Figure 5-2.** Subluxation force vs. displacement for external rotation with all muscles active and intact ligament capsule, computational model results. Force was computed to be 244.1 N at 10 mm of displacement. The cadaveric experiment described a large and progressive rise in force above 4 mm to a maximum value of 239 N.

The subluxation-force relationship was plotted for each removal of a muscle force. The first test was performed using all rotator cuff muscles at standard load. In each rotation, this results in five tests for stability (standard case and removal of one of four muscles). When these relationships are overlaid, the vertical distance between from the standard case to each line is the stabilizing force due that particular muscle. In neutral rotation, the trends for each muscle removal case tend to change as displacement increases (Figure 5-3). The chart is difficult to analyze, but at the mid-displacements, removal of either the supraspinatus or the biceps results in a much less stable joint. At 10 mm of anterior displacement, the important stabilizers are the supraspinatus and the subscapularis. The external rotators appear to have little effect on stability throughout the displacement range. The next section represents this same data in a different style graph.
Figure 5-3. Subluxation force vs. displacement for neutral rotation. Results are from the computational model. The difference from the standard case to each of the other muscle or ligament removal cases is the basis for the change-in-force deflection plots in the following section.

For external rotation, the graph shows that at 10 mm, removal of the subscapularis allows the joint to be subluxated with less force (Figure 5-4). Removal of the supraspinatus has the least effect at 10 mm, as the subluxation force required to reach 10 mm does not change significantly.

Figure 5-4. Subluxation force vs. displacement for external rotation. Results are from the computational model. The difference from the standard case to each of the other muscle or ligament removal cases is the basis for the change-in-force deflection plots in the following section.
5.2 Change-in-Force Deflection Curves

The following curves represent the change in force required to subluxate the joint due to the removal of a particular muscle force. In a graph of change-in-force vs. subluxation distance, this trend is represented by the X axis, a zero change from the standard load case. The data in these plots is essentially force-subluxation data from the previous section, but arranged to more clearly show the difference in contribution to joint stability by a particular muscle.

In the neutral rotation physical experiment, the biceps was found to be the most important joint stabilizer (Figure 5-5). The subscapularis and supraspinatus were nearly as important. All muscles showed similar trends, less of a stabilizing effect at low displacements and more so at high displacements. Initially, the presence of the external rotator muscle forces (infraspinatus and teres minor) caused the joint to destabilize. The plots depict removal of that particular muscle force, so decreases in the change of subluxation force actually translate into increases in joint stabilization due to that muscle force.

The computational model predicts the subscapularis to be the most consistent stabilizer over the range of displacement (Figure 5-5). The biceps and supraspinatus were found to be more important at the middle displacements, with their effectiveness at stabilizing the joint decreasing at the low and high displacements. Above 8 mm, it seems the supraspinatus and biceps, as well as the external rotators, begin a trend of stabilization. The external rotators (infraspinatus and teres minor) were consistently the least important stabilizers in external rotation.
Figure 5-5. Change-in-force deflection curves for the removal of a muscle force in neutral rotation. Cadaveric experiment results (left) compared to computational model results (right). The horizontal line at 0 N represents the standard case, and deviations from the X axis depict the difference in force required to displace the joint for the removal of a muscle.

For external rotation in the physical experiment, removal of the subscapularis muscle force had the most negative effect on joint stability at the middle displacements (Figure 5-6). Higher displacements lessened the importance of the subscapularis. The supraspinatus and external rotators also had large contributions to stability, though more at higher displacements. Removal of the biceps force had the least effect on joint stability, and removing it actually seemed to stabilize the joint at low displacements.

In external rotation, the computational model predicted the subscapularis, along with the external rotators, to be the most important stabilizing muscles (Figure 5-6). The biceps initially destabilized the joint, and has less of an effect on stability than the either the external rotators or the subscapularis. The supraspinatus, however, showed a different trend. It was predicted to be a destabilizer at low displacements, with a negligible effect at higher displacements.
Removing the ligamentous restraints in neutral rotation had very little effect on joint stability in both the physical experiment and the computational simulation (Figure 5-7). The physical experiment determined the anterior region (SGHL and MGHL) to be the most important at low and middle displacements. At higher displacements, the coracohumeral (CHL) and posterior regions (PC) showed only differences of about 10 N in subluxation forces. The contributions of the inferior zone (IGHL) were inconsequential in neutral rotation. Overall in the cadaver experiment, none of the ligament regions created a difference in subluxation force of more than 15 N. The computational model determined a similar result. There was little if any stability provided by the ligaments at the higher displacements. Initially, removing the inferior glenohumeral ligament resulted in a more stable joint, though by 3 mm this effect was minimal. There was no graph provided in the physical experiment for which to visually compare the effects of ligament stability in neutral rotation.
In the physical experiment, sectioning of the anterior and inferior zones resulted in the greatest loss of stability, with maximums of 44 and 47 N respectively (Figure 5-8). All regions of the capsule were determined to provide some degree of stabilization effect in this rotation. The coracohumeral ligament tended to stabilize as displacements increased, but after 6 mm its contribution was diminished. The posterior region did not play an important role in stability, though displayed some destabilization effect at low displacements in both rotations. The computational model predicted minimal stabilization effects for all ligaments except the IGHL region. After 3 mm, the IGHL displayed an enormously important role in stabilizing the joint throughout the displacement range.
5.3 Ligament Lengths

The lengths of the ligaments were recorded during anterior displacement using the computational model. The cadaver experiment did not perform displacement analysis for the ligaments, but these results can be used to compare with other experimental results in the literature. The values shown depict the displacement length between the origin and insertion of the ligament, which in some cases (MGHL and 6 IGHLs) includes the length of the rotating swing arms defined in Section 3.5.3. This value does not depict the actual length of the ligament since the ligament may be folded upon itself if the origin and insertion are brought within sufficient proximity. No ligaments used in the computational model had a SFL of less than 39.8 mm, so any values below the 39.8 mm threshold indicate some degree of ligament folding (Figure 5-9). For the vast majority of ligaments in neutral rotation, the displacement either decreased or remained relatively constant. The IGHL was divided into three sub-regions, with two ligament elements
representing each section. The three sections are the IGHL anterior band (IGHL Ant), IGHL axillary pouch (IGHL AP), and the IGHL posterior band (IGHL Pos).

![Figure 5-9. Ligament lengths during displacement in neutral rotation. Results are from the computational model. The IGHL is broken into 6 different regions, with “1” being anterior to “2” for each region (see Figure 3-11 for diagram of IGHL ligaments). Refer to Table 3-1 for the stress free lengths for each ligament used in the computational model.](image)

In external rotation, the ligament displacements showed a different trend (Figure 5-10). Most ligaments elongated with displacement, while only the CHL and the SGHL decreased in displacement.
Figure 5-10. Ligament lengths during displacement in external rotation. Results are from the computational model. The IGHL is broken into 6 different regions, with “1” being anterior to “2” for each region (see Figure 3-11 for diagram of IGHL ligaments). Refer to Table 3-1 for the stress free lengths for each ligament used in the computational model.

5.4 Ligament Tensions

Tension developed by each ligament structure was also analyzed throughout the displacement range in the computational model. The cadaveric study did not record this data, though it can be compared to other similar cadaveric or computational experiments. In neutral rotation, only three portions of the IGHL developed any tension, with only the posterior section of the IGHL posterior band providing any significant tension (Figure 5-11). By 3 mm of displacement, that portion of the posterior band had relaxed to its SFL.
The tension developed in the ligament regions during displacement in external rotation was much more significant. The five most anterior regions of the IGHL all developed some tension during the range of displacement (Figure 5-12). The two ligament regions with the most tension were both of the axillary pouch elements. The anterior part of the IGHL posterior band developed tension initially, while the elements within the IGHL anterior band did not produce tension until the latter regions of displacement.
Figure 5-12. Ligament tension during displacement in external rotation (see Figure 3-11 for diagram of ligaments). Results are from the computational model. Only ligament regions with non-zero tensions are shown.

5.5 Joint Contact Force

Another difficult parameter to measure experimentally, joint contact force, was easily calculated by the computational model. For neutral rotation at 90° of abduction in the scapular plane, the joint contact force began at 322 N and decreased to 290 N at maximum displacement (Figure 5-13).

Figure 5-13. Joint contact force vs. subluxation displacement in neutral rotation. Results are from the computational model. Standard muscle loads applied with an intact ligament capsule.
With the humerus in external rotation at 90° of abduction in the scapular plane, the joint contact force began at 319 N and progressed upwards to 486 N at 10 mm of displacement (Figure 5-14).

![Figure 5-14. Joint contact vs. subluxation displacement force in external rotation. Results are from the computational model. Standard muscle loads applied with an intact ligament capsule.](image)

5.6 Joint Contact Area

Though the contact areas shown below are not quantifiable, they provide a visual description to help explain some of the other results. The 3D contact specified in Section 3.8 is computed by using small amounts of penetration between the solid bodies. When the humerus and scapula come into contact with one another, this penetration can be visually displayed by highlighting the faces that are closest together. Since the model does not simulate the presence of cartilage, the contact areas shown below are generally smaller than in vivo contact areas. With compressible cartilage surfaces, the contact areas would be much higher. Still, the diagrams give an indication of where the humeral head is in relation to the glenoid cavity.
The standard load for the muscles was used to center the humeral head within the glenoid socket in both rotations. The humerus was positioned manually using a geometric procedure (4.2.2), but had this position been used as the starting location, the tendency of the muscles forces to move the humeral head slightly would have confounded the results. So, the procedure of allowing the humeral head first to center with the standard muscle forces, and then to apply the displacement was used to reduce initial force effects and to define a consistent starting position.

![Figure 5-15. Approximation of joint contact areas for neutral rotation (top row) and external rotation (bottom row) with humerus in 90° abduction in the scapular plane. Lateral view of glenoid, with anterior direction to the left (humerus would translate to the left in each image). The image on the far right is the initial contact area at 0 mm of displacement, the middle image depicts 5 mm of displacement, and the far left is 10 mm.](image)

In neutral rotation, the humeral head made initial contact on the posterior/superior region of the glenoid rim (Figure 5-15). With displacement, the head migrated to the anterior aspect of the glenoid cavity. In external rotation, the humeral head started in a
more centered location. At maximum displacement, the head rose out of the glenoid socket and rested on the anterior/superior lip.
6 DISCUSSION

The results show that there are many similarities between the predictions from the computational model and the results of the cadaveric experiment. The reaction of the glenohumeral joint to an anterior displacement was measured by plotting subluxation force vs. subluxation displacement. To determine the most important stabilizing muscles, the displacement procedure was repeated after removing a specific structure. By comparing the force-displacement relationship between the different muscle removal cases, the most effective joint stabilizers could be elucidated. The similarities between the results of the two methods are discussed below, as well as possible reasons for their differences.

Additionally, some parameters were included that were not assessed in the cadaveric study. This extra data was instead compared to similar cadaveric studies or computational model to assist in validation of the model. Variables such as ligament lengths and tensions, as well as contact forces, were recorded throughout the computational model for comparison to other values found in the literature.

6.1 Subluxation-Force Relationship

As the piston moved anteriorly, it encountered greater resistance against the structures of the shoulder joint. Initially, the muscles provide most of the resistance to movement as they hold the humeral head into the glenoid socket. A component of the
subluxation force likely arises from the interaction of the humeral head to the anterior aspect of the glenoid fossa. As the humeral head continues to move forward, the ligaments begin to stretch and provide additional restraint, causing an increase in force.

The force required to subluxate the joint was less in neutral rotation than external rotation due primarily to the differences in ligament tensions. Neutral rotation was set in the cadaver experiment to maximize ligament laxity by choosing the midpoint between full internal and full external rotation. The experimental results detailed a slight stabilization effect at higher displacements which was likely due to a tightening of the ligaments. The computational model exhibited the opposite in neutral rotation. If any stability existed, it occurred only initially as can be seen from Figure 5-7. This can best be explained by the location of the humeral head within the socket. The initial placement of the humeral head relied on a geometrical relationship before the muscle forces were “activated” to center the head. In neutral rotation, this centering by the application of the muscle forces instead caused a posterior/superior drift of the humeral head (Figure 5-15), thus increasing the distance from the anterior lip of the glenoid to the contact point of the humeral head. Since the starting position of the humeral head lies posterior and superior on the glenoid, the ligament regions on the anteroinferior portion of the capsule tightened. As the humeral head was displaced forward, the ligament could then relax. Had the humeral head been more appropriately centered by the application of the standard muscle loads, it is possible that the initial ligament tension in the inferior region would be reduced, and an increase in ligament tension would be observed at the higher displacements.
The magnitudes of subluxation force at 10 mm differ between the two studies in neutral rotation. The cadaveric model reported a value about 50 N higher than the computational model predicted [31]. This difference is best explained by the lack of ligament tension. The wide variation in attachments, lengths, and stiffness of ligaments in the literature likely had an effect in this situation. The definition of muscle forces also determined the initial centering of the humeral head. With its posterior/superior starting point, the humeral head probably was not displaced far enough to initiate tension in the ligament capsule. It is worth noting here that a separate cadaveric experiment by Blasier et al. reported higher forces required to subluxate the joint to 10 mm [42]. This experiment utilized less force for each of the rotator cuff muscles, and did not include the deltoid in their study. However, they reported 335 ± 9.8 N for neutral rotation, expanding the variation gap in biomechanical testing of the shoulder.

In external rotation in the computational model, the subluxation force rose almost linearly with changes in displacement. This is likely due to the increasing tension developed in the IGHL as the humeral head was displaced (Figure 5-8). Even at low displacements, portions of the IGHL were stretched in external rotation. As subluxation increased, the ligament lengths also increased, leading to more tension developed. This helps to explain the difference in magnitude between the neutral and external rotation cases. The cadaveric experiment described an exponential rise in force, increasing rapidly after 4 mm of displacement. This finding was also explained by the subsequent ligament tightening, but may also be partially due to impingement with the coracoacromial arch. This impingement was not noticed in the computational model due
to the complete absence of soft tissue bulk wrapped around the humeral head. For comparison, the similar cadaveric experiment performed by Blasier et al. reported 430 N ± 9.8 N to achieve a 10 mm subluxation [42].

6.2 Change-in-force Deflection Curves

Analyzing the effects of removing muscles on the stability of the joint also had many similarities. For neutral rotation, the computational model produced a rather inconsistent plot, though some conclusions could still be drawn. The subscapularis showed a steady increase in stabilization as displacement increased. This was reflected in the cadaveric experiment as well. The external rotators were calculated to be the least effective stabilizers in neutral rotation, matching the results of the cadaveric study. The biceps and supraspinatus provided stability up until the mid region of displacement, though their importance tapered off at higher displacements.

The ligaments in neutral rotation did not play a significant role in anterior stabilization, a result shared by the cadaver experiment. Neutral rotation was chosen to minimize the effects of the ligament capsule, as bisecting the internal/external range of motion of the humerus would lead to the greatest amount of ligament laxity. The cadaveric experiment detailed no ligament regions adding more than 15 N of stabilizing force. Only initially did the IGHL have an effect on stability, though the presence of it actually destabilized the joint. This may be explained by the initial position of the humeral head more posterior in the glenoid fossa Figure 5-15). This would tend to stretch the IGHL until forward displacement correctly centered the humeral head. Slight
changes in stability due to the ligaments would likely have been noticed if the humeral head was displaced further than 10 mm in neutral rotation.

For external rotation, the trends for muscle stability were more consistent. The subscapularis and external rotators were found to be important stabilizers, which correlates well with the cadaveric study. The biceps initially destabilized the joint in both methods of study, though it was deemed a more important stabilizer in the computational model. The trends were similar, with an increase in stabilization effect at higher displacements.

Also in external rotation, the only ligament region contributing to anterior stability was the IGHL. The cadaver experiment concluded that all regions assisted in stabilizing the joint, but the anterior (SGHL and MGHL) and inferior (IGHL) regions were the most important. A possible reason for why the IGHL’s importance was exaggerated may lie in the definition of the size of the IGHL. Following Bigliani’s diagram (Figure 3-8), the IGHL encapsulates nearly ½ of the entire circumference of the joint [44]. The positioning of the ligament elements was based largely on illustrations, but even literature with the same authors displayed a vast difference in relative size of the structures [44,78]. The definition of the IGHL in the computational model may be an overestimate of the average shoulder, leading to a wider IGHL that replaces the duties of the anterior region of the capsule. Decreasing the size of the IGHL while expanding or altering the location of the MGHL or SGHL in the computational model would likely have shifted some stabilizing responsibility towards those structures.
One of the findings of the cadaver experiment detailed the large contrast in the stabilization role of the biceps between the two rotations. The contribution to stability by the other muscles did not vary much by rotation, which can be attributed to less change in the direction of the muscle action line. The biceps, being located superiorly to the humeral head, undergoes a direction change between neutral and external rotation. In external rotation, the biceps force is directed anteriorly to the glenoid, while in neutral rotation it is directed posteriorly (Figure 6-1). In neutral rotation, this force tends to oppose the direction of displacement, causing an increase in stability. To a lesser degree, the computational model predicts a similar reaction with the supraspinatus, likely due to its superior positioning to the glenohumeral joint. These findings are seen in the change-in-force deflection curves for neutral rotation (Figure 5-5) as the biceps and supraspinatus display a large role in stability during a majority of the subluxation displacement. In external rotation, the biceps and supraspinatus are the two least important stabilizers (Figure 5-6). A few authors have described this importance of the biceps in neutral rotation [79-81]; however, the computational model predicts a similar importance for the supraspinatus in this rotation.
Figure 6-1. Role of biceps in stabilization. Superior view of glenohumeral joint, anterior direction is to the right of the image. In neutral rotation (left), the biceps tendon would oppose an anterior subluxation, while a component of this force would act to compress the joint. In external rotation (right), the biceps tendon provides no resistance to anterior subluxation [81].

Another important factor that may lead to discrepancies between the cadaveric experiment and the computational model relates to the application of the muscle forces. The shoulder is composed of complex bony anatomy around which many of the muscle tendons wrap. The rotator cuff muscles, for instance, connect from the medial portions of the scapula, around the glenoid rim and over the humeral head. Since COSMOSMotion is unable to simulate a wrapping force element (such as a rope) to redirect the force, alternative means were used to approximate the paths of the muscles (see section 3.6). Blocks were placed near the glenoid rim to which the muscle force vectors were attached to replicate their anatomical direction of force. These blocks were placed with the intention of more closely approximating the path of the muscle by directing them around the glenoid rim, though positioning them proved challenging when dealing with two separate rotations (neutral and external) of the humerus. The humeral head also posed a problem, as the muscle forces sometimes passed through the humeral head to their lateral attachment site. This was more of an issue for the ligaments, as their medial attachment
was closer to the glenoid rim, resulting in a less correct lever arm. Hence, the “swing arms” were created to remedy the problem (see section 3.5.3). The muscle forces suffered from this problem as well, but to a lesser degree. For modeling simplicity, similar “swing arms” were not used for the muscles, resulting in smaller muscle moment arms that likely influenced the results.

6.3 Ligament Lengths

The plots describing the ligament lengths over the displacement range contain two different measurements. The length of the ligaments that utilized the “swing arm” redirection around the humeral head (MGHL and six IGHLs) is a summed total of two distances. The first distance is the straight line distance from origin on the scapula to the end of the swing arm. The second distance is from the end of the swing arm to the insertion point on the humeral head, in an arc whose radius is equal to the average humeral head radius. The other ligaments (PC, SGHL, CHL) use simply a straight line that may at times slightly pass through the humeral head.

In neutral rotation, most of the ligaments decrease in length with an anterior displacement. One would assume that structures on the anterior and posterior aspects of the humeral head would increase in length, but that assumes that the humeral head initially starts in a glenoid-centered position. Since the humeral head migrates superiorly and posteriorly with application of standard muscle forces in neutral rotation, the humeral head initially moves towards the center of the glenoid, causing an overall decrease in length for the ligaments. This is shown in Figure 5-9, as the mid-displacement lengths are generally less than the low displacement lengths. Most structures have leveled off by
10 mm of displacement, though a few begin to increase again in length as the humeral head approaches the anterior aspect of the glenoid socket. In external rotation, the humeral head begins in a more centered location, thus most ligament lengths increase as the humeral head is pushed anterior and rides up the anterior edge of the glenoid lip. Only the structures on the superior aspect of the capsule (SGHL and CHL) show a decrease in length as displacement is applied, and is likely due to the proximity of the insertion and origin as the humeral head shifts in position.

Computer measurements from another study have some similarities to the ligament lengths. Utilizing SIMM to measure the ligament paths in neutral rotation at 90 degrees in the scapular plane with 0 degrees of displacement, Debski et al. found the IGHL-AB and MGHL to have a length of around 35 mm [18]. Measurements from this computational model show the MGHL to be 37 mm and the IGHL-AB to be about 34 mm. Debski et al. measured the SGHL to be about 5 mm in this position using SIMM, while the computational model showed 12 mm. Since these measurements are based on two different CT scanned specimens, the location of bony anatomy and the scaling of the bone sizes may not be identical. Additionally, the location of the determined insertion and origins for each of the ligaments can change if based on different information.

6.4 Ligament Tensions

In situ ligament tensions are difficult to measure experimentally and often rely on experimental tension-length relationships to determine the developed force. Due to the large range of motion of the shoulder joint and difficulty in elucidating these forces, there are few studies to which to compare ligament tensions. One study determined that the
IGHL-AB carried more than 30 N of force at 90° abduction in neutral rotation with an 89N anterior displacement force [66]. The computational model showed no force being carried by that ligament in this arm position, but the experiment did not apply muscular forces across the joint. Without stabilizing muscular forces, the ligament would tend to display more tension as the displacements would likely rise to a comparable subluxation force. With an 89 N anterior subluxation force with no rotator cuff or deltoids, the humeral head was predicted to translate about 20 mm [66]. In this same circumstance, the computational model predicts a translation of only 6 mm (due to an 89 N applied force), a displacement too small to generate any significant change in ligament length. Debski et al. found that under their loading conditions (no stabilizing muscles) the ligaments provided little resistance to anterior translations of less than 10 mm [66]. A similar study by Debski et al. concluded that the resistance to anterior displacement in neutral rotation at 90° abduction was due to mainly the IGHL-AB, MGHL, and the passive rotator cuff muscles [82]. This experiment also negated the active effect of the muscles, and thus a large displacement occurred, near 22 mm of anterior dislocation. These experiments found the ligament structures to aid in stability, but only at higher displacements. The computational model did not reach this magnitude of translation, and thus is inconclusive in its prediction of tension developed in the ligaments for neutral rotation.

6.5 Joint Contact Force

The compressive force in a joint is the resultant magnitude of all the forces crossing the joint boundaries. As the joint is repositioned, the contact force can also vary
due to the changes in muscle or ligament forces applied to the joint. Joint contact forces are challenging to measure experimentally, since the surfaces of interest lie within the joint. Access to the interior regions of the joint would likely disrupt the mechanics of the joint. The joint contact forces calculated with this computational model are a second way to verify what is happening in the stabilizing structures.

In neutral rotation, the joint contact force was higher at low displacement than at high displacement (322 N at 0 mm and 290 N at 10 mm). This at first seems counterintuitive, but can be explained by the initial tension developed in the ligaments in neutral rotation. By mid displacements, the tension in the ligaments had tapered off, resulting in a flattening of the contact force. In external rotation, the contact force began low and increased with displacement (319 N at 0 mm and 486 N at 10 mm). As displacement increased in this rotation, so did ligament tension, leading to an increase in the contact force. The magnitude of the contact force for both rotations is comparable to the summed total of the seven applied muscle forces (329 N). In comparison, Novotny et al. determined less than 100 N for the average contact force in external rotation at 90° of abduction due to the ligaments, without the presence of any muscles [4]. Terrier et al reported a contact force of about 600 N, though the magnitudes of the applied muscle forces were not specified [1]. It is important to note that both of these authors utilized computational models to provide these results, and were not directly from a cadaver study.
6.6 Joint Contact Area

The initial position of the humeral head in neutral rotation likely caused many of the differences between the cadaveric experiment and the computational model. A comparison of contact areas in the literature strengthens the argument of an improperly positioned humeral head in neutral rotation. Soslowsky et al. determined the humeral head to be well centered, if not slightly anterior to center when the humerus is abducted in the scapular plane in neutral rotation (Figure 6-2). The contact area for 90° was not reported, so an average of the two should be made to compare to the computational model results. A search of the literature could not find a comparison for external rotation contact area. It is likely that the humeral head is properly positioned in a central location to the glenoid cavity.

Figure 6-2. Comparison of contact areas on the glenoid from a cadaveric study (left, darker is more contact) and the computational model (right, contact highlighted in pink). For the left images, the arm is in neutral rotation at 60° or 120° of scapular abduction. An average of the two left images must be made to compare to 90° of abduction for the right image [65].
6.7 Overall Differences

These differences between the computational model and the cadaveric results overall are best explained by the variations in anatomy that exist in the shoulder. The bone structures used as the basis for positioning the insertions and origins of the ligaments and muscles were based on the VHP. Even though the VHP scan is meant to be a normal sized individual, variations in bony anatomy from an average specimen are possible and likely. The cadaver experiment was performed on four specimens, the results being averaged across eight shoulders total. These eight shoulders were from relatively old individuals compared to the VHP specimen.

Ligament lengths are another large source of error since they are widely varied in the population as well. Some ligaments are not even present in certain individuals, revealing the fact that the mechanics of one patient’s shoulder may not operate like another’s. Not only does the length of each ligament have a wide standard deviation, but the stiffness of each ligament also varies significantly. With some of the ligament structures being especially stout and stiff, a small change in length (variation in origin/insertion) would have a large effect on the tension produced.

Positioning of the bones in space is another aspect of the computational model that could be a cause of the differences in the data. An attempt was made to accurately recreate the experiment setup, but vagueness and measurement differences may have had an effect on the results. For instance, the procedure for mounting of the scapula’s orientation was not detailed in the cadaver experiment, so 3D kinematics measurements were used to position it in space. It was found in the computational model that slight
changes in scapular orientation (especially in anterior/posterior tilt – similar to axial rotation of the humerus at 90° abduction) had a large effect on the tightness of the IGHL in external rotation, which could lead to a variation in any of the results. Further, the rotation of the humerus was also approximated since the VHP scans did not include the elbow portion. This made it impossible to determine the true epicondylar axis used for measuring internal/external rotation. A solid body of a humerus from another cadaver was overlaid on the VHP humerus to approximate the correct epicondylar axis, though the degree of humeral retroversion (twist) may have varied between the two bones. Because of the sensitivity of humeral internal/external rotation, the results could have been significantly impacted.

Muscle moment arms are extremely important in the shoulder, as they have a large influence on a muscle’s effectiveness in producing a force [1]. In the shoulder, the larger the moment arm, the better a muscle can produce a torque about the joint center. The moment arm in most cases is approximately the radius of the humeral head, but varies with the position of the humerus with respect to the scapula. Since the computational model was not designed to wrap the muscles around the humeral head, some portion of the true moment arm is lost resulting in decreased muscle efficiency. This change would mainly influence the change-in-force deflection curves and subluxation-force relationship. For simplicity of the model, wrapping elements were used only for the ligaments as they seemed to bisect more of the humeral head (and thus had a greater impact on their moment arm). Future versions of the computational model would benefit from the ability to wrap muscle forces, which would likely increase joint
stability. Currently, COSMOSMotion does not have the ability to simulate wrapping elements, though most FEA based programs can handle this complexity. Moving the model over to a more comprehensive simulation package would require a rebuild from the ground up, but could potentially increase the overall accuracy of the model.
The computational model developed here has demonstrated its ability to predict important parameters in the kinematics of the shoulder joint, but has areas that can be improved upon. For computational efficiency, some structures of the shoulder and non-linear material properties were neglected. The labrum, a soft tissue cup that deepens the glenoid, was not included in the model. The thickness of the cartilage on the articulating surfaces was also not accounted for. Large variations in anatomy also had an effect on the computational accuracy of the model, and some ligament lengths that could not be found in the literature were approximated using the other ligaments. Also, a more detailed experimental methodology would have permitted a more accurate computational model setup.

Computational models of musculoskeletal systems have a wide range of use. They can assist in pre-operative planning. They have proved their usefulness in assisting orthopedic implant design. Large amounts of data can be extracted from these models in a short amount of time, and can be easily modified to perform different analyses. Even so, construction of these models is especially difficult as biologic tissues are less predictable than ordinary materials. The delicate nature of the balance between soft tissue, bone, and joint stability in the shoulder has been challenging for many health care professionals. Creating an accurate computational model of the glenohumeral joint has proven to be equally complex.
REFERENCES


## ABBREVIATIONS

<table>
<thead>
<tr>
<th>Acronym</th>
<th>Phrase or Name</th>
</tr>
</thead>
<tbody>
<tr>
<td>AA</td>
<td>Angulus Acromialis</td>
</tr>
<tr>
<td>AB</td>
<td>Abduction of Arm</td>
</tr>
<tr>
<td>AB-IGHL</td>
<td>Anterior Band of IGHL</td>
</tr>
<tr>
<td>AC</td>
<td>Acromioclavicular Joint</td>
</tr>
<tr>
<td>AD</td>
<td>Adduction of Arm</td>
</tr>
<tr>
<td>Al</td>
<td>Angulus Inferior</td>
</tr>
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<td>AP-IGHL</td>
<td>Axillary Pouch of IGHL</td>
</tr>
<tr>
<td>AT</td>
<td>Anterior Tilt of Scapula</td>
</tr>
<tr>
<td>CHL</td>
<td>Coracohumeral Ligament</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomography</td>
</tr>
<tr>
<td>DR</td>
<td>Downward Rotation of Scapula</td>
</tr>
<tr>
<td>ER</td>
<td>External Rotation of Scapula</td>
</tr>
<tr>
<td>FE</td>
<td>Finite Element</td>
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<tr>
<td>FEA</td>
<td>Finite Element Analysis</td>
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<td>GHJ</td>
<td>Glenohumeral Joint</td>
</tr>
<tr>
<td>IA</td>
<td>Inferior Angle of Scapula</td>
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<tr>
<td>IGL</td>
<td>Inferior Glenohumeral Ligament</td>
</tr>
<tr>
<td>IR</td>
<td>Internal Rotation of Scapula</td>
</tr>
<tr>
<td>ISB</td>
<td>International Society of Biomechanics</td>
</tr>
<tr>
<td>MGHL</td>
<td>Middle Glenohumeral Ligament</td>
</tr>
<tr>
<td>PB-IGHL</td>
<td>Posterior Band of IGHL</td>
</tr>
<tr>
<td>PC</td>
<td>Posterior Capsule</td>
</tr>
<tr>
<td>PT</td>
<td>Posterior Tilt of Scapula</td>
</tr>
<tr>
<td>RS</td>
<td>Root of the Scapular Spine</td>
</tr>
<tr>
<td>SB-IGHL</td>
<td>Superior Band of IGHL</td>
</tr>
<tr>
<td>SD</td>
<td>Standard Deviation</td>
</tr>
<tr>
<td>SFL</td>
<td>Stress Free Length</td>
</tr>
<tr>
<td>SGHL</td>
<td>Superior Glenohumeral Ligament</td>
</tr>
<tr>
<td>STL</td>
<td>Stereolithography</td>
</tr>
<tr>
<td>TS</td>
<td>Trigonum Spinae</td>
</tr>
<tr>
<td>UR</td>
<td>Upward rotation of Scapula</td>
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<tr>
<td>VHP</td>
<td>Visible Human Project</td>
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</tbody>
</table>

Table 7-1. List of acronyms used in this document.
APPENDIX A

The values for modulus of elasticity are presented in the table below, along with the corresponding geometric measurements of the ligaments. The stiffness can then be computed using the given parameters, and is displayed in the right column.

\[
\text{Stiffness} = \frac{\text{Modulus} \times \text{Width} \times \text{Thickness}}{\text{Length}}
\]

<table>
<thead>
<tr>
<th>Ligament</th>
<th>Author</th>
<th>Modulus (N/mm²)</th>
<th>Width (mm)</th>
<th>Thickness (mm)</th>
<th>Length (mm)</th>
<th>Stiffness (N/mm)</th>
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<tbody>
<tr>
<td>IGHL – Anterior</td>
<td>Bigliani [44]</td>
<td>38.74 ± 18.09</td>
<td>13.33 ± 2.66</td>
<td>2.79 ± 0.49</td>
<td>41.3 ± 4.5</td>
<td>34.89</td>
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<tr>
<td>IGHL – Ax. Pouch</td>
<td>Bigliani</td>
<td>30.33 ± 10.58</td>
<td>12.61 ± 3.05</td>
<td>2.34 ± 0.43</td>
<td>39.8 ± 5.6</td>
<td>22.49</td>
</tr>
<tr>
<td>IGHL - Posterior</td>
<td>Bigliani</td>
<td>41.91 ± 12.50</td>
<td>10.86 ± 2.94</td>
<td>1.70 ± 0.55</td>
<td>41.0 ± 4.7</td>
<td>18.87</td>
</tr>
<tr>
<td>PC - Inferior</td>
<td>Bey [48]</td>
<td>56.8 ± 39.8</td>
<td>8.47*</td>
<td>1.3</td>
<td>40.7**</td>
<td>15.36</td>
</tr>
<tr>
<td>PC - Middle</td>
<td>Bey</td>
<td>44.9 ± 22.8</td>
<td>8.47*</td>
<td>1.6</td>
<td>40.7**</td>
<td>14.95</td>
</tr>
<tr>
<td>PC - Superior</td>
<td>Bey</td>
<td>28.4 ± 16.5</td>
<td>8.47*</td>
<td>2.3</td>
<td>40.7**</td>
<td>13.59</td>
</tr>
</tbody>
</table>

Table 7-2. Table of parameters used to compute the stiffness for each ligament.

Data are mean ± SD, if listed
* Based on ratio of length of IGHL insertion on glenoid rim to length of PC insertion on the glenoid rim, assuming the IGHL spans 180 degrees of circumference, and the PC spans 120 degrees of circumference from the posterior margin of the posterior axillary pouch to the apex of the glenoid/biceps tendon insertion
** Approximated from mean of Bigliani’s IGHL lengths
APPENDIX B

Degree values for bounding regions of the ligament zones, along with the degree used for approximation of the origins and insertions.

<table>
<thead>
<tr>
<th>Ligament Region</th>
<th>Degree for first border (°)</th>
<th>Degree for second border (°)</th>
<th>Degree for insertion point(s) (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>MGHL</td>
<td>30</td>
<td>60</td>
<td>45</td>
</tr>
<tr>
<td>SGHL</td>
<td>60</td>
<td>90</td>
<td>75</td>
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<tr>
<td>PC – Superior</td>
<td>90</td>
<td>135</td>
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<tr>
<td>PC – Middle</td>
<td>135</td>
<td>165</td>
<td>150</td>
</tr>
<tr>
<td>PC – Inferior</td>
<td>165</td>
<td>210</td>
<td>180</td>
</tr>
<tr>
<td>IGHL – Posterior</td>
<td>210</td>
<td>270</td>
<td>225, 255</td>
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<tr>
<td>IGHL – Ax. Pouch</td>
<td>270</td>
<td>330</td>
<td>285, 315</td>
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<td>IGHL – Anterior</td>
<td>330</td>
<td>30</td>
<td>345, 15</td>
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</tbody>
</table>

*Refer to Figure 3-9 for pictorial representation of degrees on the scapula.

<table>
<thead>
<tr>
<th>Ligament Region</th>
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<th>Degree for second border (°)</th>
<th>Degree for insertion point(s) (°)</th>
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<tr>
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<td>40</td>
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<tr>
<td>SGHL</td>
<td>40</td>
<td>70</td>
<td>55</td>
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<tr>
<td>PC – Superior</td>
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<td>PC – Middle</td>
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<td>PC – Inferior</td>
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<td>IGHL – Ax. Pouch</td>
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<td>265, 295</td>
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<tr>
<td>IGHL – Anterior</td>
<td>310</td>
<td>10</td>
<td>355, 325</td>
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</tbody>
</table>

*Refer to Figure 3-9 for pictorial representation of degrees on the humerus.
VITA

Kevin Arthur Elmore was born September 25, 1983 in Richmond, Virginia. At an early age, he showed an interest in discovering how things worked. His grandfather, an engineer by trade, further developed his interests by designing and constructing models of cars and boats, woodworking projects, and remote control vehicles. As he matured, he further refined his skills, applying them to repairing vehicles or to helping fix things around the house. His desire to become an engineer like his grandfather was further realized while single-handedly maintaining his first car.

At St. Christopher’s High School, he excelled in the sciences, particularly mathematics and physics. His love of the outdoors was exemplified through canoeing, hiking, and rock climbing with the Watermen program and by attaining the rank of Eagle Scout. He also enjoyed participating on the Ampersand crew, designing and constructing stages for the school plays. Additionally, he competed on the indoor track team for a number of seasons until he graduated in 2002.

Kevin was accepted to the Mechanical Engineering program at Virginia Tech for the 2002 fall semester. An opportunity to participate in the senior year concentration of Biomedical Engineering peaked his interest by applying newly learned mechanics principles to the bodily systems. After graduating with a B.S. in Mechanical Engineering in 2006, he decided to pursue his new interests by enrolling in the M.S. Biomedical Engineering program at Virginia Commonwealth University.