A Computational Investigation Into Acromial Fractures After Reverse Total Shoulder Arthroplasty

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Abstract

Acromial fractures are a debilitating complication following reverse total shoulder arthroplasty (RTSA). The purpose of this work is to (1) improve the current state of FE bone stress estimates after RTSA and to investigate the effects of (2) plane of elevation, (3) hand loads, and (4) baseplate screw position on scapular spine stress. The FE method used was validated against an in-vitro strain gauge based experiment and found to be accurate in the prediction of stress increases and decreases after RTSA. The coronal plane of elevation was found to increase scapular spine stress compared to more central planes of elevation. Hand load increased stress levels above the fatigue threshold of healthy cortical bone. A divergent screw angle decreased bone stress and if the screw penetrated the scapular spine then leaving the screw in place decreased bone stress compared to withdrawal. This work can help in determining the optimal surgical protocol.

Keywords

Reverse total shoulder arthroplasty, acromial fractures, biomechanics, finite element analysis, implant placement, bone stress/strain.
Summary for Lay Audience

Reverse total shoulder arthroplasty (RTSA) is a type of shoulder replacement surgery. As common with surgeries there is a chance of complication, one such complication associated with RTSA is acromial fractures which occur in up to 7% of cases. The acromion is the upper structure of the scapula, and a fracture is a break or crack in the bone. The purpose of this work was to investigate causes of these fractures by determining how altering the surgical procedure changes bone stress.

In Chapter 2 different implant placements were investigated over a range of arm positions to determine their effect on bone stress. Implant position was varied by shifting the scapular component away from the scapula (0,5,10mm), and downwards(0,2.5,5mm), along with shifting the humeral component towards and away from the body(-5,0,5mm). This was done using computer models of 10 scapulae with a simulated implant. A custom algorithm estimated muscle forces in the shoulder using anatomic dimensions and weights. The different implant configurations were compared to determine bone stress patterns. It was found that shifting the scapular component away from the body increased bone stress and shifting it downward decreased bone stress.

In Chapter 3 the computer modelling method was used to investigate the position of screws that are used in securing the shoulder implant. Screw position was altered by changing the angle (0°,15°), using 3 screws as opposed to 4, and altering the screw length to see the effect of proximity of screws to the acromion. The results found that screws did increase bone stress and a 15° angle decreased bone stress compared to 0°.

In Chapter 4 the custom computer modelling method was validated against the gold standard method of measuring strain in bone. An experiment was performed on 6 cadaveric scapulae and recreated with computer modelling. The results of both methods were compared to prove the computer modelling results are meaningful. No significant difference was found between the two methods.

This work can be used to design future experiments with the goal of determining the optimal surgical procedure to minimize the likelihood of unwanted surgical complications.
Co-Authorship Statement

Chapter 1:  JS Lockhart – sole author

Chapter 2:  JS Lockhart – Study design, model creation, data collection, statistical analysis, wrote manuscript

MT Wong – model creation, reviewed manuscript

GS Athwal – Study design, reviewed manuscript

GDG Langohr – Study design, reviewed manuscript

JA Johnson – Study design, reviewed manuscript

Chapter 3:  JS Lockhart – Study design, model creation, data collection, statistical analysis, wrote manuscript

GS Athwal – Study design, reviewed manuscript

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Chapter 4:  JS Lockhart – Study design, model creation, data collection, statistical analysis, wrote manuscript

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Chapter 5:  JS Lockhart – sole author
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List of Nomenclature

° Degree
mm millimeter
MPa Megapascal
2D Two-dimensional
3D Three-dimensional
COR Center of rotation
RTSA Reverse Total Shoulder Arthroplasty
Chapter 1

1 Introduction

This chapter provides an overview of the anatomy and function of the shoulder, followed by the structure and material properties of bone. This lays a foundation to review the various types of orthopaedic shoulder replacements used to treat common end stage diseases that are described within. A more thorough review of reverse total shoulder arthroplasty and its common complications is given, as the prime focus of this work. The empirical methods utilized to perform the computational and in-vitro experiments are described. Finally, the motivation, objectives, and hypotheses for the following investigations are outlined.
1.1 The Shoulder

The function of the shoulder is to provide a wide range of positioning for the hand\textsuperscript{1}. The shoulder is made up of 3 bones, 4 joints, and a myriad of soft tissue that both passively supports and actively participates in joint motion. The bony structures in the shoulder are the humerus, scapula, and clavicle, which together along with the sternum and rib cage form the glenohumeral joint, acromioclavicular joint, sternoclavicular joint, and the scapulothoracic joint (Figure 1-1). The glenohumeral joint followed by the scapulothoracic joint are the most significant joints active in shoulder motion\textsuperscript{2,3}. The 2 clavicular joints allow the clavicle to provide support to the upper limb, similar to a strut\textsuperscript{4}. The joint of focus in this biomechanical work is the glenohumeral joint and corresponding musculature.
Figure 1-1: The Shoulder Anatomy

The shoulder with bony anatomy and joints labelled. (adapted from Reeves et al\textsuperscript{5})
1.1.1 Bony Anatomy

1.1.1.1 Humerus

The humerus is a long bone that plays a major role in the function of the shoulder and elbow along with helping to form the upper limb. The proximal end of the humerus articulates as part of the glenohumeral joint. Significant anatomic landmarks on the proximal humerus that contribute to the glenohumeral joint consist of; a spherical head that contacts the scapula in the glenohumeral joint. The head is approximately one third of a sphere, allowing itself to rotate and translate on the concave surface of the scapula\(^4\). Distally adjacent to the humeral head are the greater and lesser tubercles, which are the insertion points for tendons connecting major rotator cuff muscles. The other major abduction muscle, the deltoid, inserts onto the deltoid tuberosity or proximal lateral humerus\(^2\) (Figure 1-2).
Figure 1-2: Humerus

Anterior (left) and posterior (right) views of the humerus.
1.1.1.2 Scapula

The scapula is the interface of the upper limb to the axial skeleton. The scapula is a flat bone with a complex geometry that contributes three major landmarks to the glenohumeral joint (Figure 1-3, Figure 1-4). The glenoid fossa, a shallow concave socket that forms the articular surface for the humeral head, the acromion and scapular spine provide surface area for tendon insertion of the deltoid, and the coracoid process protrudes medially relative to the glenohumeral joint for soft tissue insertion\(^4,6\). Additionally, the scapula plays a critical role in articulation of the glenohumeral joint by rotating along with the humerus during arm elevation\(^3\).

The acromion originates from the scapular spine and extends laterally above the glenohumeral joint. It provides a large lever arm and surface area for the deltoid to insert above the joint, increasing its mechanical advantage during elevation of the arm. The acromion has a unique shape similar to that of a diving board\(^7\).

1.1.1.3 Clavicle

The clavicle is a long bone that interacts with the glenohumeral joint through the acromioclavicular joint. The clavicle is the insertion site for a portion of the anterior deltoid, and the coracoclavicular ligament, which joins the clavicle and coracoid process\(^2,4,6\). Medially the clavicle acts as a strut and provides stability for the shoulder during arm elevation. It is uncertain whether the clavicle’s role is altered by shoulder replacement and how that might effect outcomes\(^8-10\).
Figure 1-3: Anterior Scapula

Anterior view of the scapula and clavicle, including annotation of significant anatomic landmarks (adapted from Reeves et al\textsuperscript{5}).
Figure 1–4: Posterior Scapula

Posterior view of the scapula and clavicle, including annotation of significant anatomic landmarks (adapted from Reeves et al⁵).
1.1.2 Muscles

Two layers of muscle surround the glenohumeral joint to provide joint stability and aid with arm elevation. The inner layer is made up of 4 muscles that form the rotator cuff, and the outer layer is composed of the deltoid (Figure1-5).

1.1.2.1 Deltoid

The deltoid is the largest and outermost muscle structure of the gleno-humeral joint. Proximally it connects to the scapular spine, acromion, and lateral portion of the clavicle. The distal deltoid inserts onto the mid humerus at the deltoid tuberosity. Commonly, the deltoid is described as 3 segments; the posterior deltoid that inserts onto the scapular spine, the middle deltoid that inserts onto the lateral acromion, and the anterior deltoid that inserts onto the lateral third of the clavicle and anterior acromion. Research models have been developed to describe the deltoid in more detail as 7 individual segments, each with a unique line of action.

The deltoid plays a significant role in elevation of the arm as a result of its large cross-sectional area. Each segment’s participation varies in different planes of elevation, however, the middle deltoid is typically the most active.

1.1.2.2 The Rotator Cuff

The rotator cuff is a group of 4 muscles that form the deepest layer surrounding and stabilizing the glenohumeral joint. Subscapularis, supraspinatus, infraspinatus, and teres minor are the four muscles, each originating on the scapula and inserting onto the proximal humerus. Rotator cuff tears have been shown to change kinetics of arm elevation, compromise joint stability, and restrict shoulder range of motion.

Rotator cuff tears are a common source of shoulder pain, and are most often associated with the supraspinatus. Tears can range in magnitude from partial to full thickness tears, traditionally treatment has been both operative and non-operative. A focus has been put on early intervention in hopes of yielding the progression of tears. In severe cases,
typically in the presence of full thickness tears and after arthroscopic repairs have failed, reverse total shoulder arthroplasty is performed\textsuperscript{1,15}.
Figure 1-5: Musculature of the shoulder

Musculature of the shoulder including the rotator cuff and deltoid, which are the major musculature that surrounds the glenohumeral joint (adapted for Kerrigan et al17).
1.1.3 Arm Elevation

During elevation of the arm both the humerus and scapula are mobile. The humerus rotates about the glenohumeral joint relative to the scapula, and the scapula rotates superiorly relative to the thorax at the scapulothoracic joint\(^3\) (Figure 1-1). The humerus and scapula have been found to rotate at a consistent ratio of about 1:1 respectively above and beyond 30° of elevation\(^{12}\). For the first 30° of elevation the scapula remains relatively stationary with respect to superior rotation\(^{18,19}\). The shoulder enables the arm to be manipulated in 3 axes; the flexion/extension axis, the abduction/adduction axis, and the internal/external rotation axis (Figure 1-6).

The glenohumeral joint loosely follows the behaviour of a ball and socket joint. However, the humerus is able to translate on the glenoid fossa as a result of the glenoid fossa’s shallow nature\(^{20}\). A true ball and socket joint is a purely rotary joint, the glenohumeral joint will be treated and referred to as a shallow ball and socket joint accordingly. The shallow nature of the glenoid fossa increases the importance of the soft tissue socket of the glenohumeral joint, as the musculature is relied on to provide stability by constraining the humerus.
Figure 1-6: Shoulder Motions

Basic motions of the shoulder and their axes (I) Abduction/adduction axis (II) Internal/External Rotation Axis (III) Flexion/Extension Axis.
1.2 Bone Properties, Structure, and Health

1.2.1 Bone Structure

Bone has a composite and dynamic structure, which provides a structural framework for the body. The appendicular skeleton or limbs mainly consist of long bones that are composed of cortical and trabecular bone. Cortical bone is also referred to as compact bone as it is the dense layer surrounding long bones that acts as a protective barrier. On a macroscopic scale cortical bone is uniform in composition, has a low surface area to volume ratio, and makes up 80% of bone in the body by mass. In contrast trabecular bone, also called cancellous bone, is non-uniform and sponge-like, with a high surface area to volume ratio that allows for the distribution of loads, and accounts for 20% of bone by mass. Trabecular bone is commonly found on the interior of long bones concentrated near the articulating surfaces in order to contribute compressive strength and flexibility\textsuperscript{21–23}.

Cortical bone has a layered cylindrical structure made up of lamellae (Figure 1-7). Lamellae are centered by canals that allow for nerve and vasculature access to provide bone cells with the necessities of life. Bone cells and extracellular matrix are organized in cylindrical patterns surrounding these canals and forming the lamellae. Trabecular bone has a complex structure of lamellae forming a network of tubular branches that intersect with each other and have an average thickness of 200 micrometers. The large degree of connectivity allows for the increased dispersion of force. Bone is consistently deconstructed and remodeled to effectively resist loading over time. Osteoclasts and osteoblasts are the major cells responsible for the remodeling cycle allowing bones to adapt to loading changes within our body. The natural cycle of bone remodeling can be impeded by bone related diseases\textsuperscript{21–23}.
Figure 1-7: Bone Structure

*Visualization of bone structure including cortical and trabecular bone.*
1.2.2 Structural Properties of Bone

Bone is a composite material consisting of cortical and trabecular bone, which have unique mechanical properties\textsuperscript{22}. The material properties of bone as a whole depend on the bone’s composition, which includes the ratio of trabecular to cortical bone. Many studies have been performed to link the apparent density of bone to its mechanical properties, where apparent density is the mass of the bone over the entire volume it occupies. Apparent density is also affected by bone porosity as it alters the volume of the sample. A previous study showed that cortical bone has an apparent density $>1.3 \ [g/cm^3]$, and trabecular bone consistently has an apparent density $< 1.3 \ [g/cm^3]$, with an overall range of 0.1-2.1 $[g/cm^3]$\textsuperscript{22}. Trabecular bone was found to have a much larger range of apparent density compared to cortical bone, which was relatively constant\textsuperscript{22}.

Human bone is a complex material with variable material properties as described above. Both cortical and trabecular bone have been shown to have anisotropic strength, meaning that bone strength is dependent on the location and direction of loading. Bone is a viscoelastic material, consequently its behaviour is dependent on the rate of applied load. Additionally, the mechanical properties of bone are influenced by many factors such as age, weight, activity levels, and sex\textsuperscript{22,23}.

1.2.3 Osteoarthritis (OA)

Osteoarthritis is responsible for $27.5 \ billion$ of direct and indirect health care costs in Canada, it affects 12\% of the Canadian workforce, and is a major issue worldwide, especially in the aging population\textsuperscript{24,25}. Osteoarthritis is the degeneration of the articular joint, with an emphasis on articular cartilage. Typically, osteoarthritis is seen in weight bearing synovial joints, such as the shoulder, hip, and knee. Currently there is no cure for osteoarthritis, consequently the existing interventions are focused on slowing or minimizing damage onset by the disease. The current standard of care for patients with end stage osteoarthritis greater than 65 years of age is joint arthroplasty\textsuperscript{24}. 
1.3 Arthroplasty

The three most common forms of shoulder arthroplasty are hemiarthroplasty (HA), total shoulder arthroplasty (TSA), and reverse total shoulder arthroplasty (RTSA).

1.3.1 Hemiarthroplasty (HA)

Hemiarthroplasty is the reconstruction of only the humeral side of the glenohumeral joint. Indications traditionally included younger patients with arthritis, avascular necrosis, and rotator cuff tears. Hemiarthroplasty’s aim was initially to maintain the native joint as much as possible, however, its effectiveness has been questioned. It was shown that hemiarthroplasty was particularly ineffective for patients with severe rotator cuff tears. Accordingly, the need for an arthroplasty procedure that improved shoulder function for rotator cuff deficient patients arose.

1.3.2 Total Shoulder Arthroplasty (TSA)

Total shoulder arthroplasty is the anatomic reconstruction of the articulating surfaces of the glenohumeral joint. It is most commonly used to treat end stage osteoarthritis in patients with intact rotator cuffs and sufficient bone quality. The reconstruction of the articulating surfaces relieves pain and prevents impingement caused by osteoarthritis. Most commonly the humeral head is replaced with a metal implant (typically titanium) that is cemented or press fit into the proximal humerus via a stem or anchor, and the glenoid fossa is replaced with a shallow polyethylene cup.

1.3.3 Reverse Total Shoulder Arthroplasty (RTSA)

Reverse total shoulder arthroplasty (RTSA) is the reconstruction and reversal of the articular surfaces of the glenohumeral joint. Reverse total shoulder arthroplasty was first designed to replace the glenohumeral joint for end stage rotator cuff tears. Severe rotator cuff tears can lead to destabilization of the glenohumeral joint and inferior shoulder range of motion. Traditional total shoulder arthroplasty and hemiarthroplasty were found to be inconsistent at restoring range of motion in these cases, as soft tissue function is not intentionally altered by these procedures.
Figure 1-8: Common shoulder arthroplasties

A comparison of common shoulder replacements and the native shoulder joint. (I) Native shoulder (II) Hemiarthroplasty (III) Total shoulder arthroplasty (IV) Reverse total shoulder arthroplasty.
Reverse total shoulder arthroplasty was designed to alter the biomechanics of the soft tissue and consequently restore shoulder function. In reverse total shoulder arthroplasty, the humeral head is replaced with a constrained polyethylene cup, and the glenoid fossa is replaced with a metal glenosphere (typically cobalt chromium or titanium). In a ball and socket joint the center of rotation is located at the center of the ball, in the case of the native shoulder this is the center of the humeral head. Thus, the reversal of anatomy shifts the joint’s center of rotation medially to the glenoid fossa from the center of the humeral head, which increases the length of the deltoid’s moment arm and stabilizes the joint (Figure 1-9). The increased moment arm decreases the force required by the deltoid to elevate the arm, enabling the deltoid to lift the arm in the absence of the rotator cuff\textsuperscript{1,14,29}. Increased deltoid participation and efficiency helps to restore the glenohumeral joint’s range of motion. Furthermore, deltoid tension is restored by the humerus’ position being shifted inferiorly during reconstruction, creating a stable ball and socket joint even in the absence of the rotator cuff\textsuperscript{14}.

The US Food and Drug Administration first approved reverse total shoulder arthroplasty in 2004\textsuperscript{29}. Since then reverse total shoulder arthroplasty is becoming an increasingly popular option to treat end stage osteoarthritis in rotator cuff deficient patients, revisions of total shoulder arthroplasty, and severe fractures of the proximal humerus. A review of shoulder replacements in high volume hospitals indicates that reverse total shoulder arthroplasty now represents approximately 78% of all shoulder arthroplasties\textsuperscript{1,26,29–31}.

Common complications of reverse total shoulder arthroplasty include implant loosening, glenoid notching, and acromial fracture\textsuperscript{31–35}. Implant loosening is defined as migration of the implant as a result of insufficient mechanical fixation that can lead to implant failure and revision surgery. Glenoid notching is the deterioration of the inferior edge of the glenoid fossa as a result of repeated contact between the glenoid fossa and humeral component of the reverse total shoulder arthroplasty implant\textsuperscript{32}. Acromial fractures occur as a result of the altered deltoid loading and has been found to cause pain and poor clinical outcomes\textsuperscript{35}. As reverse total shoulder arthroplasty is in its infancy limited biomechanical research has been performed to completely understand and mitigate
complications, the current literature has a large focus on implant loosening, which has been significantly improved since the introduction of reverse total shoulder arthroplasty implants\textsuperscript{1}.
Figure 1-9: Biomechanics of RTSA

Biomechanical rationale behind reverse total shoulder arthroplasty for rotator cuff deficient patients
1.3.3.1 Implant Placement

Implant placement during reverse total shoulder arthroplasty can be altered at the discretion of the surgeon, both through the glenoid component and the humeral component (Figure 1-10). The glenosphere can be shifted laterally, inferiorly or tilted in the sagittal plane. The placement of the implant on the glenoid has been shown to have a relationship with implant loosening, glenoid notching, and acromial fracture\textsuperscript{36,37}. Humeral cup position can be altered via a shift in the medial/lateral axis, neck shaft angle, and version angle. Research regarding positioning of the humeral component and its effect on clinical outcomes is ongoing. Neck shaft angle has been shown to influence glenoid notching through alteration of impingement free range of motion, but may not effect joint stability\textsuperscript{38,39}. Humeral version studies have focused on optimization of impingement free range of motion, suggesting a neutral version angle\textsuperscript{40,41}. 


Figure 1-10: RTSA Implant Placement
Reverse total shoulder arthroplasty implant placement variables (I) glenosphere lateralization (II) glenosphere Inferiorization (III) glenosphere Inferior tilt (IV) humeral cup lateral/medialization
The placement and trajectory of baseplate screws are also up to the discretion of the surgeon. Variation in the number of screws used, and angle of screw trajectory is seen in reverse total shoulder arthroplasty procedures (Figure 1-11). Baseplate screw placement is manipulated with the goal of increasing screw purchase to maximize implant stability. Regions of high bone density are targeted for screw placement\textsuperscript{42,43}. Research has largely focused on the effect screws have on implant stability and avoiding major nerve structures to prevent further injury\textsuperscript{44,45}. 
Figure 1-11: RTSA screw placement

*A depiction of RTSA baseplate and fixation screw placement in the scapula. (A) depicts a divergent screw angle of 15° (B) depicts a screw angle of 0°, perpendicular to the baseplate surface.*
1.3.3.2 Implant Loosening

Implant loosening is a highly studied and devastating complication after shoulder arthroplasty\textsuperscript{2,13,44,46,47}. Glenoid component loosening has been reported to cause up to 63% of revision surgeries for total shoulder arthroplasties\textsuperscript{46}. Humeral component loosening is less common, but has been recently investigated with the introduction of new stemless reverse total shoulder arthroplasty humeral components\textsuperscript{48–50}.

An implant’s mechanical fixation is relied on for stability during the first three months post operation, as implant motion above a certain threshold prevents the bonding between implant and bone. Osseointegration results in the connection that is formed between bone and the implant that provides long-term stability. After osseointegration occurs the implant’s mechanical fixation is less crucial. Previous studies have shown that implant micromotion greater than 150 microns impedes osseointegration and leads to implant loosening\textsuperscript{44,51}.

Implant loosening is thought to be a result of many factors such as implant design, position, and bone quality, which leads to excessive movement of the implant relative to the bone\textsuperscript{52}. The effect of bone quality on reverse total shoulder arthroplasty implant stability has been examined by altering both bone density and bone loss\textsuperscript{44,47}. Superior glenoid bone loss was found to compromise implant stability at 50% footprint erosion\textsuperscript{43}, and bone density was found to have a significant correlation with implant instability when tested in saw bone specimen\textsuperscript{44}. Studying the effect of bone quality on implant stability can help influence inclusion criteria and surgical procedure in order to mitigate implant failure via instability. Implant position has also been linked to implant stability, studies have shown that inferior inclination of the glensphere along with lateralization and glensphere size may effect implant motion\textsuperscript{37,39}. Implant position has been shown to alter bone remodeling via stress shielding that also impacts long-term implant stability\textsuperscript{50}. Implant design parameters such as baseplate peg length, screw length, and number of screws have been investigated in terms of implant stability\textsuperscript{44,53,54}. Further research in regard to implant stability is integral in increasing implant longevity, and consequently decreasing the financial burden of revision surgeries on the healthcare system. Both
future implant design and surgical procedure can be influenced by implant stability research.

### 1.3.3.3 Acromial Fracture

Acromial fracture is a common complication following reverse total shoulder arthroplasty, occurring in 0.8-7.2% of populations \(^{32,33,55-57}\). Acromial fracture has been found to cause pain and lead to poor clinical outcomes as it impedes deltoid function by decreasing deltoid tension.

Reverse total shoulder arthroplasty implants rely on tension in the deltoid in order to function. As a major insertion site for the deltoid, damage to the acromion can compromise deltoid tension leading to deleterious function of the joint. Studies have shown limited range of motion in patients that experience acromial fractures compared to patients that do not\(^{58-60}\). One study in particular found the fracture group to have inferior range of motion by a total ASES score of 58 postoperatively compared to a score of 74 in patients without acromial fractures (p=0.001). ASES scoring is a clinical tool for surgeons to score post-operative performance via range of motion that has been validated for patients with rotator cuff disease and arthritis in the glenohumeral joint\(^{58}\). Acromial fractures after reverse total shoulder arthroplasty are a debilitating complication because of their high risk of non-union regardless of intervention technique\(^{59-61}\). Non-union is a common occurrence because of the geometry of the acromion, the population that undergoes reverse total shoulder arthroplasty, and the loading of the acromion during arm elevation after reverse total shoulder arthroplasty. The acromion is a relatively small bone with small surface area in its cross section in the sagittal plane, this small surface area provides limited bony contact for bone remodeling and regrowth to properly heal the fracture, thus non-union is common. Furthermore, patients of greater age have a limited capacity for bone healing on average. The large forces the deltoid applies to the acromion after reverse total shoulder arthroplasty also increased chance of non-union due to the distracting forces at the fracture site impeding bony growth.

Current literature on acromial fractures has been focused on documenting and classifying their occurrence. Levy et al has classified acromial fractures into 3 types based on clinical
observation of their location on the acromion related to deltoid insertion\textsuperscript{55} (Figure 1-12). Type I fractures include fractures caused by the anterior and middle deltoid segments, type II fractures involve the entire middle deltoid segment, and type III fractures involve both the posterior and middle deltoid segments. Type II fractures have been found to occur most commonly. Crosby et al proposed another classification system of acromial fractures after reverse total shoulder arthroplasty based on the location of the fracture site on the acromion. Type I fractures are defined as fractures of the anterior acromion near the base of the coracoacromial ligament, type II fractures occur on the acromion posterior to the acromioclavicular joint, and type III fractures occur on the scapular spine\textsuperscript{62}. Type II fractures were found to be the most commonly observed fractures. However, a study showed that classification of acromial fractures after reverse total shoulder arthroplasty with the Crosby et al system has low intra rater reliability\textsuperscript{63}. The Levy et al system was then created and validation showed improved intra rater reliability, it has since been used to classify acromial fractures after reverse total shoulder arthroplasty\textsuperscript{36,60}. 
Figure 1-12: Levy classification of the acromion

Levy et al regions for classification of clinically observed acromial fractures after reverse total shoulder arthroplasty.
The mechanism of postoperative acromial fracture is thought to be fatigue failure due to increased stress levels above the operational microdamage (fatigue) threshold of bone (approximately 60 MPa), rather than a traumatic fracture caused by a single stress above the ultimate strength of bone (approximately 120 MPa)\textsuperscript{64,65}. Fatigue failure is caused by repetitive high stress events that leads to a buildup of micro damage occurring at a rate greater than the bone’s ability to remodel. The cause of acromial fractures after reverse total shoulder arthroplasty is likely multifactorial and linked to the alteration of muscle moment arms that increase deltoid participation and tension during elevation of the arm.

The acromion has a unique shape similar to that of a diving board acting as a lever arm for the deltoid during arm elevation. The acromion is commonly categorized into three groups by its shape in the sagittal plane. Type I acromia are flat, type II acromia are curved, and type III acromia are hooked (Figure 1-13). Acromial shape has been shown to be related to joint impingement and may effect occurrence of acromial fractures\textsuperscript{7}. The bending moment that the deltoid creates on the acromion and scapular spine is thought to lead to the clinically observed stress fractures after reverse total shoulder arthroplasty. Although deltoid efficiency is increased after reverse total shoulder arthroplasty; deltoid tensioning, and deltoid participation contribute to altering the direction and force that the deltoid applies to the acromion\textsuperscript{40,66}. Resting arm length is increased by RTSA implants to create joint stability through tensioning of the deltoid, possibly resulting in greater resting forces at the insertion sites of the deltoid on the acromion. Additionally, RTSA is largely used in patients with compromised rotator cuffs that can no longer aid in arm elevation leaving the deltoid to be the sole major elevator of the arm.
Figure 1-13: Common shapes of the acromia

Acromia categorized by shape in the sagittal plane. (A) Type III, hooked (B) Type II, curved (C) Type I, flat.

Baseplate screw placement during reverse total shoulder arthroplasty has been linked to increased occurrence of acromial fractures when the screw intersects the base of the scapular spine\textsuperscript{45,56}. Kennon et al. specifically showed that the superior baseplate screw increased acromial fracture occurrence from 0\% to 4.4\%\textsuperscript{45}. The screw tip located in the scapular spine is possibly acting as a stress riser that is the fracture propagation site\textsuperscript{56}. However, the factors causing screw related acromial fractures are not fully understood. Determining optimal baseplate screw placement may help mitigate acromial fractures following reverse total shoulder arthroplasty.

1.4 Methods to Assess Shoulder Biomechanics with Special Interest in the Assessment of Loading and Bone Stresses

1.4.1 Finite Element Analysis

Finite element (FE) analysis is the discretization of objects to quantify the mechanical behaviour given a set of defined conditions. The FE outcome in this work is primarily principal stress and strain experienced by bone. Finite element analysis enables the estimation of several physical phenomena such as stress and strain of bone, contact area and pressure of a joint, and bone remodeling rates. Objects are discretized into a set of elements and nodes\textsuperscript{67,68} (Figure 1-15). Tetrahedral and hexahedral elements are two
types of elements commonly used in FE models. Tetrahedral elements are triangular prisms, hexahedral elements are rectangular prisms, and nodes are the points that connect them (Figure 1-14).

![Figure 1-14: Basic FE elements](image)

**Figure 1-14: Basic FE elements**

*Basic element shapes used in finite element analysis cornered by nodes (I) Tetrahedral element (II) Hexahedral element.*

Objects are constrained by user-defined boundary conditions and acted upon by user-defined forces. Each element in the object is represented by a set of partial differential equations. Boundary conditions provide enough restrictions to create a determinant problem allowing the finite element method to calculate one unique solution. User defined conditions are set to mimic the physiologic case. Element properties including stiffness are a required input for the finite element method. Many studies have been performed to determine the appropriate element properties to accurately model bone. These studies have related the mechanical stiffness to the apparent density of bone. Apparent bone density is determined from CT attenuation data, which has been related to apparent density through the use of phantoms. Cortical and trabecular bone are treated separately as it has been found that the cortical bone can be modelled with a constant young’s modulus of 20 GPA, and a variable young’s modulus is required for trabecular
bone\textsuperscript{5}. Traditionally, trabecular Young’s modulus has been applied as a constant to each element in FEA, however recent studies have shown an increase in model accuracy when Young’s modulus is applied to each element as a gradient\textsuperscript{5}. 
Figure 1-15: FE model of the scapula

A discretized scapula using tetrahedral elements based on CT geometry and material data.
At the introduction of RTSA the FE method was utilized to determine the effect of varying rotator cuff tears on joint reaction forces\textsuperscript{69,70}. RTSA was designed for rotator cuff deficient patients, accordingly, there is a need to investigate the effectiveness of the implant placement for varying degrees of rotator cuff tears. The FE method was then extended to investigating the effect of implant design and surgical procedure on implant loosening as it was initially the most prevalent complication associated with the implants\textsuperscript{37,71}. A previous study looked at the effect of inferior tilt on implant stability\textsuperscript{71}. Studies were also performed to investigate the validity of the finite element method and assumptions made in applications of reverse total shoulder arthroplasty\textsuperscript{72}. Differences in outputs as a result of some model assumptions emphasized the importance of validation of finite element models. As a better understanding of implant stability has developed further research has incorporated other common complications associated with reverse total shoulder arthroplasty, such as acromial fractures\textsuperscript{36}. Finite element analysis is an important tool in biomechanics research of the shoulder as it is low cost and allows for a flexibility to test a high number of cases comparatively to in-vitro cadaveric testing.

1.4.2 Computed Tomography (CT) Imaging

Computed tomography imaging is a medical imaging modality that produces a 3D map of attenuation coefficients, creating cross sectional images of the body. CT scanners function via a source that emits x-rays, the x-ray beams travel through the body, and are absorbed by a detector, similar to an x-ray image. CT scanners differ from x-ray machines as the whole system is mounted on a gantry and rotated around as the body is translated through the gantry in order to create an array of 2D attenuation coefficient maps, resulting in a 3D map of attenuation coefficients. Attenuation coefficients represent the percentage of beam energy that was dissipated as the x-ray transmits through the body, the attenuation coefficient is linked to the density of the object it transmits through as higher density objects absorb more energy. Phantoms of known apparent densities have been used to correlate CT attenuation coefficients to the apparent density of bone to use for finite element modelling\textsuperscript{73}. Further, the 3D maps created by CT imaging are often used to provide the geometry and material properties required to create FE models for biomechanical applications.
1.4.3 Muscle Force Estimation

Mathematically determining forces acting on the body during activities of daily living has been a constant challenge for the field of biomechanics. The complexity of joint function creates an indeterminate problem, consequently there is no unique solution. Forces are important to estimate such that they can enable biomechanical research, for instance, as an input to finite element and in-vitro studies. Finite element analysis relies on user-defined forces to determine the resulting stress in objects, and in-vitro studies often involve loading cadaveric specimen with pneumatic actuators where the applied forces are user defined. Both tools have been instrumental in building a knowledge base and driving innovation in the field of biomechanics. Therefore, it is critical for the biomechanics research to have an accurate understanding of the forces that act on anatomic bodies.

Static optimization models calculate muscle forces using moment and force balances. In order to calculate muscle forces assumptions must be made to make the problem determinant. Existing models have made assumptions of how the muscles share the load at the joint. An example of these assumptions is that muscle force is proportional to cross-sectional area of the muscle, which maximizes muscle efficiency. Once an assumption is made anthropometric data for segment length and weight is used to predict force application for each muscle for a subject at the 50th percentile of the population. Other factors are also accounted for such as passive lengthening of the arm as a result of reverse total shoulder arthroplasty.

When implants are the focus of biomechanics research the gold standard of force estimation is telemetry instrumented implants. These implants are instrumented with strain gauges and load cells to measure the loads experienced by the implant in-vivo. Telemetry circuits have made it possible to utilize instrumented implants surgically and communicate with them in a non-invasive nature. Westerhoff et al have created and implemented telemetry implants for total shoulder arthroplasty and other joint replacements. Unfortunately, there are currently no published telemetry instrumented implant data for reverse total shoulder arthroplasty. Some studies have used force values from instrumented total shoulder arthroplasty implants, while others have created inverse
dynamic models to estimate the force that muscles are applying to the joint. When the acromion is the focus of research, using data from total shoulder arthroplasty instrumented implants is inappropriate because of the altered deltoid moment arm in reverse total shoulder arthroplasty that affects the deltoid force. The change in force consequently alters the stress levels that the acromion experiences in reverse total shoulder arthroplasty compared to total shoulder arthroplasty.

1.4.4 Direct Strain Measurement

Strain gauges are a common experimental tool for measuring surface strain of an object. Strain gauges are composed of an electrical circuit embedded in an elastic backing. As strain is applied to the circuit the output voltage is altered proportionally to the change in length experienced by the circuit. The manufacturer provides a gauge factor for each gauge to define the relationship between strain and change in signal. Strain gauges measure the local strain of a material for the area contained underneath the circuit, and the strain is measured in the direction of the electrical circuit. Multiple circuits can be placed together to provide multiple directions of strain, combinations of 1-3 circuits are commonly available for purchase (Figure 1-16). A strain gauge with 1 circuit is referred to as a uniaxial strain gauge, and a gauge with multiple circuits is often referred to as a rosette.

For optimal experimental measurement the surface must be smoothed and cleaned for proper gauge adherence, in order for the gauge to be properly coupled with the bone. Proper procedure for applying strain gauges to bone has been experimentally outlined; the specimen is denuded (often including the use of sandpaper) and degreased using an acid in order to provide a clean surface for strain gauge application. The gauge is then adhered to the specimen using an industrial grade adhesive, such as cyanoacrylate adhesive, and is optionally sealed to provide water proofing.
Motivation

The body of biomechanical research regarding RTSA has been growing since its inception in 2004, however, there still exists a need to better understand implant design in order to mitigate complications. A previous study has investigated the effect of implant placement on acromial stress to gain insight on acromial fractures after RTSA, however, the results were limited to a single plane of elevation\textsuperscript{36}. Thus, there is a need to better understand how implant placement acts over a range of arm positions that represents a wider set of activities of daily living. Additionally, investigating a variety of hand loads will provide insight on the effect of lifting during activities of daily living.

Another design aspect that has been linked to implant instability is variability in baseplate screw placement. Studies have focused on investigating the position of screws and the quality of bone they are placed in. Screw angle is one placement variable that has not been heavily studied. A previous study has investigated screw angle in saw bone specimen in terms of implant stability and no significant relationship was determined,
possibly due to the lack of structural complexity in saw bone blocks\textsuperscript{44}. Further investigation into screw angle and its effect on acromial fractures is required.

Understanding the effect of angle on both acromial fracture and implant stability will more likely provide significant insight into the optimal screw placement in reverse total shoulder arthroplasty. Moreover, incorporating both complications will aid in providing a more complete basis for the design of clinical studies and possibly lead to the influence of surgical procedure and implant design.

In order to investigate the effect of experimental factors on acromial fracture measurements of strain in the acromion will need to be acquired. Strain gauges have been proven to be an effective experimental tool for measuring strain. However, the measurements are limited to the area underneath the strain gauge making it difficult to determine strain patterns throughout the entire acromion. The development and use of finite element models will enable a more detailed observation of strain patterns in the acromion. A validation of these finite element models against in-vitro strain gauge measurements will ensure that the assumptions made in the finite element models mimic the physiologic case accurately, provide confidence in study results, and enable further use of the finite element method in studying acromial fractures after RTSA.

1.6 Objectives and Hypotheses

Objectives:

1. To advance the current state of force estimation for an inverse model of the shoulder following reverse total shoulder arthroplasty to more accurately predict deltoid forces exerted on the acromion, to cover;
   a) Multiple planes of elevation between the coronal and sagittal planes (a 90\textdegree range).
   b) Account for activities that involve lifting weights in the palm.
   c) To validate the inverse and finite element models against strain measurements in an in-vitro setting with identical loading and boundary conditions.
2. To investigate how implant placement interacts with variable planes of elevation and hand loads. Specifically looking at glenosphere lateralization (0,5,10mm), glenosphere inferiorization (0,2.5,5mm), and humeral offset (-5,0,5 mm).

3. To investigate the effect of the number of baseplate screws and screw angle on stress levels in the acromion and consequently the likelihood of acromial fracture, with the aims of:
   a) Comparing perpendicular (0°) and divergent (15°) angles for bone screws placed in the glenoid during reverse total shoulder arthroplasty, and the resulting peak stress in the acromion.
   b) Determine if altering the absence of the superior or posterior glenoid baseplate screws effects acromial stress.
   c) In the case of screw penetration of the scapular spine, compare acromial stress when the screw is left in place and when it is withdrawn from the scapular spine.

Hypotheses:
1. The inverse and finite element models will produce forces in agreement with in-vitro strain gauge measurements in a comparative capacity rather than absolute.

2. Glenosphere lateralization will significantly increase acromial stress, while glenosphere inferiorization, and humeral offset will not significantly alter acromial stress. It is expected that plane of elevation will alter the effect of implant placement on acromial stress, as a result of the acromion’s unique morphology.

3. Hand loads associated with activities of daily living will increase stress levels above the fatigue threshold (60 MPa) of cortical bone.

4. Acromial stress will be decreased as a result of (a) a divergent screw angle (b) absence of the posterior screw (c) the presence of a screw if the scapular spine is penetrated as opposed to screw removal.
1.7 Thesis Overview

Chapter 2 investigates acromial stress as a function of RTSA implant placement over a range of planes of elevation (0°, 30°, 60°) and hand loads (0, 2.5, 5 kg). With the objective of better understanding the impact of implant placement in a range of scenarios that represent the activities of daily living for patients after RTSA. Implant placement variables included; glenosphere lateralization (0, 5, 10 mm), glenosphere inferiorization (0, 2.5, 5 mm), and humeral offset (-5, 0, 5 mm).

Chapter 3 describes the relationship between acromial stress and bone screw placement used for implant fixation in RTSA. Screw placement was investigated through altering; the angle (0°, 15°), the presence and absence of the superior and posterior screws, and if the screw penetrated the scapular spine the effect of leaving the screw within the scapular spine or withdrawing the screw from the spine and leaving the screw within the glenoid vault.

Chapter 4 describes a validation of the FE method and muscle wrapping algorithm used in Chapters 2 and 3 by using the FE method to reconstruct the conditions of an in-vitro strain gauge study on acromial stress as a function of implant placement variables similar to that of Chapter 2, and directly comparing the results.

Chapter 5 cumulates the results from the three studies outlined in Chapters 2-4 and summarizes how the findings may impact further biomechanical or clinical research, reverse total shoulder arthroplasty implant design, and reverse total shoulder arthroplasty surgical protocol.
1.8 References


62. Crosby LA, Hamilton A, Twiss T. Scapula fractures after reverse total shoulder


Chapter 2

The Effect of Load and Plane of Elevation on Acromial Stresses After Reverse Total Shoulder Arthroplasty

In this chapter a finite element method and custom muscle wrapping algorithm was utilized to further the current state of knowledge on the effect of reverse total shoulder arthroplasty implant placement on acromial stress. Currently, implant placement has been studied in the unloaded scenario in the scaption plane of elevation. There is a need to further understand implant placement in a set of scenarios that more closely represents activities of daily living after RTSA. A range of hand loads (0, 2.5, 5kg) typically associated with tasks of daily living and multiple planes of elevation (the coronal plane 0°, the scaption plane 30°, and a forward flexion plane 60°) were investigated to cover a range of motion that more closely mimics the shoulder’s range of motion.

A version of this chapter has been accepted for publication in Shoulder & Elbow¹.

2.1 Introduction

As stated in Chapter 1, Reverse total shoulder arthroplasty (RTSA) is a surgical procedure that reverses the native shoulder joint to restore stability and range of motion to the rotator cuff deficient patient. RTSA shifts the center of rotation (COR) compared to the native shoulder. The artificially created COR’s medial position increases deltoid participation and consequently decreases the force required for elevation of the arm². RTSA is becoming a common procedure in shoulder replacement, accounting for up to 78% of replacements³.

As previously described in sections 2 and 2.3, the shoulder is a versatile joint that manipulates the arm to place the hand with 6 degrees of freedom. The shoulder joint acts
in three axes; the flexion/extension axis, the abduction/adduction axis, and the internal/external rotation axis. The native shoulder’s range of motion covers roughly 225° of flexion and extension, 160° of internal and external rotation, and 230° of abduction and adduction. The extensive range of motion of the shoulder highlights the need to test experiments across a wide range of arm motions, to better understand how implant factors will affect behaviour commonly associated with activities of daily living.

Postoperative acromial fractures are a common complication after RTSA that are reported in up to 7% of patients. Acromial fractures have been found to lead to inferior clinical outcomes, resulting in decreased range of motion, weakness, and pain. The cause of postoperative fracture is thought to be fatigue failure due to increased stress levels above the operational microdamage (fatigue) threshold of bone (approximately 60 MPa), rather than a traumatic fracture caused by a single stress above the ultimate strength of bone (approximately 120 MPa). Note, the bone thresholds provided are approximated for healthy bone and can vary by case. Significant factors that affect acromial fatigue fracture include the alteration of moment arms created by the deltoid due to RTSA biomechanics, and osteopenia, which is common in the elderly RTSA patient population. The effect of implant placement on acromial stress during elevation in the 30° scaption plane has been previously studied. It was found that an inferior medial glenosphere, and a medialized humeral implant decrease acromial stress during scaption (where scaption is arm elevation in the scapular plane). However, knowledge related to the effect of implant placement on acromial stress is limited to the unloaded scenario in the scaption plane of elevation. The purpose of this finite element analysis study was to determine the effect of implant position on acromial stress when factors such as the plane of elevation (0°, 30°, 60°) and loading (0, 2.5, 5 kg) were varied. We hypothesized that (1) due to the unique morphology of the scapula and acromion, varying the plane of elevation would affect stresses, and that (2) acromial stresses would increase to above the fatigue threshold of cortical bone with increased hand load.
### 2.2 Materials & Methods

Three dimensional models of the glenohumeral joint were created from CT data of 10 cadaveric specimens (Figure 2-1, average age: 68±19 yrs, N=10). Wickham et al. identified seven independent deltoid segments and their insertions, which were used to model deltoid loading as seven force vectors applied to their anatomic areas of insertion on the scapula and humerus\textsuperscript{11}. The scapula and humerus were both oriented based on the International Society of Biomechanics standards\textsuperscript{12}. To match in-vivo position the scapula was rotated 10° about the scapula coordinate system’s Z axis to account for the anterior tilt\textsuperscript{13}. For arm elevation in the scaption plane (30°) the humerus was constrained to the glenoid using CT data for accurate positioning, and the humeral XY plane was set parallel to the XY plane of the scapula to initialize the humerus in the adduction position. The humerus was rotated about the X axis of the scapula's coordinate system to perform arm elevation in other planes of elevation, the humeral coordinate system was rotated about the Y axis by the corresponding angle (-30° or 30°). See Figure 2-1 for the positioning of the scapular and humeral coordinate systems.
Figure 2-1: Geometric Modelling Parameters

Anatomical mapping used for the mathematic algorithm (described later). I) The three planes of elevation studied. II) The seven lines of action for the seven Wickham et al segments of the deltoid in the 0° adduction position\(^{10}\). III) The scapular (S) and humeral (H) coordinate systems with origins.
A traditional reverse total shoulder implant (38mm glenosphere, 155° neck-shaft angle, 20mm humeral offset) was reconstructed using SolidWorks (Dassault Systèmes, France) to represent a shoulder post-RTSA. In the baseline implant configuration, the glenoid baseplate was placed at the inferior edge of the glenoid fossa with zero lateralization. RTSA implant position was altered by increasing glenosphere lateralization (0, 5, 10 mm) and inferiorization (0, 2.5, 5 mm), as well as lateralizing the humeral component relative to the humerus (15, 20, 25mm, Figure 2-2).
Figure 2-2: Implant Placement Parameters

The three implant placement parameters that were varied in the model: I) Glenoid lateralization (0, 5, 10 mm) II) Glenoid inferiorization (0, 2.5, 5 mm) III) Humeral lateralization (15, 20, 25 mm).
The force that each of the seven deltoid segments applies to the acromion was calculated by a mathematical algorithm (Matlab, MathWorks, Massachusetts) at static positions during arm elevation between 0°-120° at 15° increments for RTSA models in all implant configurations and hand loads. The arm elevation angles and planes of elevation were chosen to cover the most commonly utilized arm positions in activities of daily living⁴,¹⁴. The algorithm utilizes a moment balance based on the vectors applied by each deltoid segment, hand load, and arm weight as well as their respective distances from the joint’s COR, as previously reported¹⁰.

The force of each deltoid segment was composed of two components, the active and passive forces. The active force is created by muscle contraction of each deltoid segment, and the passive force is from the tension in each deltoid segment created by the RTSA implant lengthening the arm. Both components were derived from an optimal deltoid force as found in literature¹⁵ and proportioned to optimal deltoid segment force via a ratio of each segment’s cross-sectional area. The optimal segment force was converted to the passive force component by multiplying by the ratio of arm length before and after reconstruction of the joint with the RTSA implant. The active force component was modelled as the optimal segment force multiplied by an activation constant \( (\alpha_n) \). The seven segments were assumed to minimize overall activation during arm elevation, therefore, the static moment balance was solved for by optimizing the sum of the squared activation constants, as follows in equation (2-1)¹⁶,¹⁷:

\[
\sum_{n=1}^{7} \alpha_n^2
\]

(equation 2-1)

The opposing forces in the moment balance included the force of gravity acting at the arm’s center of gravity, and the simulated hand loads. Hand Loads were applied to model arm lifting during everyday tasks. Loads of 0, 2.5, and 5 kg were applied to the hand. The force of gravity was assumed to be that of a 75 kg patient, which corresponds to the mass of a 50th percentile male¹⁸. The resultant moment was calculated as a cross product of the
weight of the applied load and the distance from the COR of the shoulder joint to the center of mass of the hand. The distance between the COR of the shoulder joint and the center of mass of the hand was determined using anthropometry\textsuperscript{19}.

The direction of each deltoid segment was calculated using a custom wrapping algorithm, based on the obstacle set method\textsuperscript{20}. A line of action was calculated for each muscle segment, where the distal end of the line of action is the geometric center of the insertion footprint of each deltoid segment onto the deltoid tuberosity on the humerus and the proximal end was the geometric center of the muscle insertion sites on the acromion and clavicle. The midpoints of the line of actions were determined by minimizing the length of the line of action while required the line to pass within 0-15 mm of the humeral head which was modelled as a sphere capping a cylinder. This constraint prevented non-physiological lines of action.

The deltoid forces output from the moment balance both in magnitude and direction were input into Abaqus (Dassault Systèmes, France), a finite element (FE) software to compute the stress on the acromion. The FE model consisted of the scapula segmented into 160,000±20,000 quadratic tetrahedral elements, sized according to a previously performed mesh convergence study\textsuperscript{21}. The inferior portion of the scapula was rigidly constrained and the deltoid forces were tied to each segment’s physiologic insertion region on the acromion\textsuperscript{11}, visualization of loading conditions is provided in appendix B. The outcome variable was the maximum principal stress in cortical bone. The maximum principal stress is the variable most closely related to the ultimate failure behavior of cortical bone. Outputting peak stress levels in the acromion provided trends in acromial stress by investigating relative results between configurations, however, maximums represent a small area of the bone, meaning the stress levels may exceed the threshold of bone strength without the acromion fracturing.

This process was completed for each of the 0° (abduction), 30° (scaption), and 60° (forward elevation) planes of elevation. For each plane of elevation, the paths between the deltoid origins and center of rotation vary. In the investigation of plane of elevation and loading 2 187 individual cases (9 angles, 9 implant positions, 3 planes of elevation, 3
hand loads) were tested on 10 different cadaveric based models for a total of 21,870 simulations.

Acromial stress was also evaluated based on the location on the acromion. The acromion was split into three regions as defined by Levy et al.\textsuperscript{22} (Figure 2-3) based on the types of fractures that are clinically observed. Region I involve fractures caused by the anterior and middle deltoid segments, region II involves the entire middle deltoid segments, and region III involves both posterior and middle deltoid segments.

![Figure 2-3: Acromial Region Classification](image)

*Visualization of the Levy regions on the acromion. Each region is defined by the deltoid segments that cause fractures localized to said region (Region I: anterior and middle, region II: middle, region III: posterior, middle)*\textsuperscript{22}.

Three-way repeated-measures ANOVAs (abduction angle, plane of elevation, hand load) were performed for each implant position across each specimen. The ANOVAs enabled comparison between configurations and to the failure thresholds of bone (60 MPa fatigue threshold, and 120 MPa ultimate threshold.) to determine each factors effect on acromial fracture.
2.3 Results

All percentages herein are reported as percent difference to compare between experimental measures and all values are reported as a mean ± 1 standard deviation.

2.3.1 Plane of elevation

The peak acromial stress observed for neutral implant positioning during unloaded humeral elevation from 0° to 120° in the scaption plane was 24±4 MPa occurring at 45° of elevation (Figure 2-4). In abduction, the peak acromial stress was 61 ± 6 MPa occurring at 75° of elevation, while in forward elevation, the peak stress was 24 ± 3 MPa occurring at 90° of elevation.

Humeral elevation in abduction resulted in an increase in peak acromial stress of 58 ± 3.0% (+20.0 ± 14 MPa, p=0.002) over all elevation angles compared to the scaption plane, whereas elevation in the forward plane decreased peak acromial stress insignificantly by 10 ± 30% (-3 ± 8 MPa, p=0.28), compared to the scaption plane.

Figure 2-4: Acromial Stress by Angle of Elevation

*Maximum acromial stress during elevation (0-120°) for the baseline implant configuration in all three planes of elevation (I) abduction 0°, II) scaption 30°, III) forward elevation 60°) with loading (0, 2.5, 5 kg).*
2.3.2 Loading

Increasing hand load from 0 to 5 kg in the scaption plane increased peak acromial stress by 109±4% (39 ± 9 MPa) over all elevation angles, compared to the unloaded state (p<0.001, Figure 2-4). During elevation in the abduction plane, increasing hand load from 0 to 5 kg increased the peak acromial stress by 102±6% (59±9 MPa) over all elevation angles, compared to the unloaded state (p<0.001, Figure 2-4). Increasing hand loading in the forward elevation plane increased the peak acromial stress by 98±14% (38±10 MPa) over all elevation angles, compared to the unloaded state (p<0.001, Figure 2-4).

Increasing hand weight generally affected all planes of elevation similarly and raised stress levels above the fatigue threshold (60 MPa) for cortical bone.

![Figure 2-5: Acromial Stress by Implant Placement](image)

*Change in maximum acromial stress as a function of implant placement and applied load in each plane of elevation (I) 0°, II) 30°, III) 60°). Implant placement includes glenosphere lateralization (GLAT), glenosphere inferiorization (GINF), humeral component medialization (HMED) and humeral component lateralization (HLAT)*
2.3.3 Glenosphere Lateralization

For all planes of elevation (0°, 30°, 60°) and hand loads (0, 2.5, 5kg) investigated, glenoid lateralization consistently increased acromial stress (Figure 2-5).

The increase in maximum stress on the acromion as a result of glenosphere lateralization in abduction was 7±1% (4±1 MPa, p<0.001) for 5mm, and 16±2% (9±1 MPa, p<0.001) for 10mm; compared to the scaption plane where the increases were 9±1% (4±0 MPa, p<0.001) for 5mm, and 19±2% (8±1 MPa, p<0.001) for 10mm; and in the forward plane 11±3% (4±1 MPa, p=0.009) for 5mm, and 21±3% (8±1 MPa, p<0.001) for 10mm.

2.3.4 Glenosphere Inferiorization

For all planes of elevation (0°, 30°, 60°) and hand loads (0, 2.5, 5kg) investigated, glenoid inferiorization consistently decreased acromial stress.

The decrease in peak stress in the acromion as a result of glenosphere inferiorization in abduction over all elevation angles was 3±1% (1.8 ± 0.3 MPa, p=0.001) for 2.5mm, and 5±1% (3±1 MPa, p=0.002) for 5mm; compared to the scaption plane where the decrease was 3.6±0.4% (1.4±0.2 MPa, p<0.001) for 2.5mm, and 6±1% (2.5±0.3 MPa, p<0.001) for 5mm. The decrease in acromial stress in the forward elevation plane was 2±1% (0.8 ± 0.4 MPa, p=0.3) for 2.5mm, and 5±2% (2±1 MPa, p=0.041) for 5mm.

2.3.5 Humeral Component Medialization and Lateralization

Humeral medial and lateral offset did not significantly affect acromial stresses (Figure 2-6).
Figure 2-6: Acromial Stress by Implant Placement as a Function of Elevation Angle

Maximum acromial stress as a function of implant placement I) glenoid lateralization (GLAT) II) glenoid inferiorization (GINF) III) humeral component medialization (HMED) and lateralization (HLAT) and abduction angle compared to standard positioning averaged over all loads, regions, and planes of elevation.
2.3.6 Stress by Location on the Acromion

The average stress acting on each Levy region\textsuperscript{22} of the acromion during arm elevation over all implant configurations, loads, and planes of elevation was found to consistently be highest in region II (p<0.001), followed by region III (p=0.05) with the lowest stress occurring in region I of the acromion (II>III>I, Figure 2-7).

Figure 2-7: Acromial Stress by Acromial Region

*Maximum acromial stress as a function of acromial regions and abduction angle averaged over all loads, implant placements, and planes of elevation*\textsuperscript{22}. 
2.4 Discussion

Variations in the plane of elevation (abduction [0°], scaption [30°], and forward elevation [60°]) and the amount of weight the hand is lifting have significant effects on the stresses the acromion is experiencing. Acromial stresses were significantly higher for humeral elevation in the abduction plane as compared to more forward planes of elevation. This was likely a result of the unique shape of the scapular spine and acromion, as it is relatively unsupported laterally. As the middle aspect of the deltoid is largely responsible for abduction, it would follow that greater stresses would be experienced by the acromion in abduction as it overhangs the shoulder analogous to a diving board. Additionally, indirect muscles paths associated with the abduction plane may be introducing a more aggressive loading state. Understanding high stress arm positions may be beneficial in the early post-operative period to minimize exposure of the acromion to stresses that are in range of the failure thresholds of cortical bone (60 MPa fatigue threshold, and 120 MPa ultimate threshold).

Inferior glenosphere positioning significantly reduced acromial stress in all planes and for all loads. Glenosphere inferiorization decreases acromial stress likely because the shoulder joint's center of rotation is shifted causing the moment arm of the deltoid to lengthen. In a moment balance, an increased moment arm reduces the corresponding deltoid force required to combat gravity and hand loads. Decreased deltoid forces lead to decreased acromial stress because deltoid forces are applied directly to the surface of the acromion. The effect of glenosphere inferiorization on stress increases in magnitude in a linear trend with respect to applied load (Figure 2-5) and exhibits similar behavior in all 3 planes of motion suggesting that this behavior is consistent over the entire range of motion.

Glenosphere lateralization resulted in a significant increase in acromial stress, likely a result of the alteration in stress in the opposite manner of inferiorization, by shifting the joint’s center of rotation laterally the deltoid’s moment arm is shortened. Similar to glenosphere inferiorization, the effect of lateralization scales in magnitude with respect to
applied load. Recall that the increased stress levels associated with applied load approach and exceed the fatigue threshold of cortical bone (60 MPa). Therefore, lateralization plays a bigger role when the stress levels are at a level that leads to failure. Humeral medial and lateral offset produced an insignificant effect on stress over varying planes of elevation. The effects of humeral offset on stress does not scale with applied load making humeral offset less of a factor when hand loads are greater, which is when stress levels are at their highest. Therefore, humeral offset may have a diminishing effect on acromial stress as stress levels approach the fatigue threshold (60 MPa). The fatigue (operational microdamage) threshold of 60 MPa for cortical bone is the level at which the rate of damage is greater than the rate of repair. If bone is subject to stress above this threshold for an extended period of time or over multiple events a stress fracture may occur. Elevation of the arm under the application of hand loading shows that the acromion of an RTSA patient experiences stress above the operational microdamage threshold as a result of hand loads consistent with everyday tasks. Furthermore, the ultimate strength of cortical bone (120 MPa) was exceeded in the coronal plane (0°) and approached in the scaption (30°) and forward flexion planes (60°) at larger hand loads (5kg). Based on the results from this computational study hand loads introduce stress to the acromion that is consistent with multiple mechanisms of cortical bone failure of cortical bone. Additionally, when incorporating required hand lifting with plane of elevation, patients may be counselled to lift in forward elevation, if required, rather than in abduction.

The limitations of this study are associated with implementing a mathematical model and the necessary assumptions. The shoulder joint is modeled using the assumption that it is a static one-dimensional moment balance. The only muscle considered for elevation in this model is the deltoid, and it is assumed that contraction of the deltoid follows the activation-squared-minimization principle. These assumptions have been found to mimic the physiologic case after reverse total shoulder arthroplasty with reasonable accuracy. The mass of the patient is assumed to be 75 kg with average anatomical segmentation in order to represent the 50th percentile male, and hence results may vary as patient mass varies. However, due to the parametric nature of this study, we are confident that relative results would be similar.
The strength of using a mathematical model is the capability to test an extensive number of implant scenarios with precise positioning. In this study, comparisons are made between configurations, which improves accuracy as relative results eliminate some systematic error associated with the assumptions at the foundation of the model. The model’s accuracy is also strengthened by the fact that average stress in acromial regions follows the expected magnitude order of regions II>III>I determined by Levy et al, suggesting appropriate proportioning of force relative to the anterior, middle, and posterior deltoid\textsuperscript{22}.

2.5 Conclusion

Overall, the results of this study show that humeral elevation in the forward planes produce lower acromial stresses than elevation in the abduction plane. The addition of weight to the hand increased acromial stresses exceeding the fatigue threshold and approaching the ultimate strength of cortical bone for all planes of elevation investigated. The results of this study extend current knowledge regarding RTSA implant designs, in that inferior positioning of the glenoid has a positive effect on acromial stress, while lateralization has a negative effect. These effects scale as load is increased making it important in loaded scenarios that may lead to stress fractures. Additionally, humeral component offset, both lateral and medial, were observed to have minimal effect on acromial stresses for the planes of elevation and hand loads examined. These results confirm that implant parameters, plane of elevation, and the weight an arm lifts have significant effects on acromial stresses.
2.6 References


Chapter 3

3 The Effect of Glenoid Baseplate Screw Placement on Acromial Stresses After Reverse Total Shoulder Arthroplasty

In this chapter the finite element method was applied to investigate if a relationship exists between the placement of bone screws used for mechanical fixation of reverse total shoulder arthroplasty (RTSA) implants and acromial stress. The relevance of this work is driven by the rise in usage of RTSA, the prevalence of acromial fractures, and the debilitating consequences these fractures have on clinical outcomes. Acromial fractures are thought to be caused by repetitive high stress levels, thus, determining if screw placement alters acromial stress will provide insight into the optimal surgical protocol for RTSA. The FE method was used to model eight cadaveric specimens with RTSA reconstructions. Arm positions of nine quasi-static elevation angles (0°-120°) in the scapular plane of elevation were assessed. The bone screw related variables investigated were the screw angle (0°, 15°), the presence or absence of the posterior and superior screws, and if a screw penetrated the scapular spine the screw was left within the spine versus withdrawn from it.

3.1 Introduction

As stated previously in Chapters 1 and 2, Reverse total shoulder arthroplasty (RTSA) is a surgical procedure that replaces the articular surfaces of the glenohumeral joint while reversing the native anatomy for the treatment of severe rotator cuff tears, severe proximal humeral fractures, and revision surgeries. RTSA aims to restore the shoulder’s range of motion while relieving pain and is becoming increasingly popular, accounting for up to 78% of shoulder replacement surgeries. Up to 7.2% of RTSA patients experience an acromial fracture post operation that can decreases shoulder range of motion and is a source of pain. Acromial fractures are a debilitating complication and often lead to revision surgery. These fractures occur most frequently within the first year after surgery, and the mechanism of fracture is thought to be fatigue of the acromion due to increased stress as a result of the RTSA reconstruction altering the forces.
experienced by the acromion. Fatigue failure of bone occurs when the rate of microdamage is greater than the rate of micro repair. 60 MPa is the load at which healthy cortical bone has been shown to fatigue, known as the fatigue threshold\textsuperscript{11}. The cause of increased stress levels in the acromion following RTSA is not fully understood. Chapter 2 has shown that arm position and implant configuration, specifically the glenoid component, has a direct impact on acromial stress.

As documented earlier in Section 1.4.3, RTSA implants are composed of two main components, the glensphere and the humeral cup. The glensphere is fixed to the glenoid fossa via a metal baseplate that is secured with 4-6 bone screws (depending on the commercially available implant, Figure 3-1). The bone screw positioning within the glenoid fossa is at the surgeon’s discretion with regards to point of entry and angle. Research with regards to fixation screws has been largely focused on the effect of screw position on implant stability\textsuperscript{12–15}. However, the posterior and superior bone screws are in close proximity to the acromion and it has been shown that bone screws create high stress concentrations that drop off rapidly with radial distance from the screw-bone interface\textsuperscript{16}. Incidences of clinically observed acromial fractures that are aligned with the screw axis have been reported, suggesting bone screws potentially act as a stress riser\textsuperscript{13,17}. Therefore, the position of the screw with respect to the acromion may cause a difference in the stress levels experienced by the acromion and consequently bone screws may be affecting the probability of acromial fractures in patients after RTSA.

Previous studies investigating the effect of bone screws on complications after RTSA have focused on implant stability. One study examined various bone screw factors such as angle, number of screws inserted, length of screw, and bone density on glenoid baseplate micromotion. It was found that increased screw length and bone density increased stability significantly, while angle did not have a significant effect on stability\textsuperscript{12}. Other studies have been performed to identify areas of high bone density within the scapula for optimal screw placement\textsuperscript{14,15}. It was found that divergent angles were often required to place the bone screws in higher density bone\textsuperscript{15}. Finite element (FE) analysis studies have been performed investigating how best to model the screw-bone interface\textsuperscript{16,18,19}. 

There is a lack of understanding with regards to how bone screw placement after RTSA affects acromial stress. Accordingly, the purpose of this study was to determine if there is a significant relationship between acromial stress and the placement of glenoid baseplate bone screws by investigating the screw angle, the absence of posterior and superior screws, and the depth of screws that penetrate the scapular spine.
3.2 Materials & Methods

Three-dimensional FE models of eight unilateral cadaveric scapulae (N=8, average age 74±19yrs) were created using CT data (Mimics, Materialise, Belgium). The population consisted of five left, and three right, all male shoulders. The geometry of the glenohumeral joint was recreated for each model in Solidworks (Dassault Systèmes, Vélizy-Villacoublay, France) as described by the International Society of Biomechanics. To recreate the anterior tilt of the scapula, a 10° tilt was applied.

The glenohumeral joint was reconstructed with an RTSA implant, similar to the RTSA reconstruction described in Chapter 2, consisting of a 38mm glensphere affixed to the corresponding glenoid baseplate, which was placed at the inferior edge of the glenoid with 0 mm of lateralization. The glenoid baseplate fixation consisted of a 15 mm central post and four 2.5mm inner diameter 48 mm cortical bone screws, where length was maximized to increase screw purchase. The bone screws were modeled with the FE method as simplistic cylindrical rods with a full body tie to the surrounding bone.

Screw placement was varied within the models to determine the effect on acromial stress. The angle of the bone screws was varied between 0° (parallel to the baseplate post) and 15° divergent (Figure 3-1). The absence of the posterior and superior screws was investigated to determine if their proximity to the spinoglenoid notch and the scapular spine was related to acromial stress. After the screws were placed in the scapula each case was inspected to determine if the scapular spine had been penetrated. If so, the model was replicated and the screw in question was withdrawn from the scapular spine while remaining in the glenoid vault (Figure 3-2). This was done to investigate the effect of the screw penetrating the scapular spine. Additionally, acromial stress as a function of radial distance from the screw shaft, both along the length of the screw and medial of the screw, was investigated (Figure 3-3).
Figure 3-1: Screw Placement

A shoulder model reconstructed with a RTSA glenoid baseplate and fixation screws. A) depicts the divergent angle of 15° relative to the central peg, shown in the coronal plane. B) Depicts the 0° angle that is parallel with the central peg, shown in the coronal plane. C) depicts the glenoid baseplate mounted on the scapula with the bone screws labelled relative to their anatomic position.
Figure 3-2: Scapular Spine Penetration by a drill hole and screw

In models where the scapular spine was penetrated by a drill and a screw the effect of withdrawing the screw from the scapular spine while leaving the drill hole in the spine unfilled was tested. A) Model of a clinical scenario where the posterior hole of the baseplate is drilled into the scapular spine. The drill hole is filled with a longer screw that completely fills the hole in the scapular spine. B) A model of the clinical scenario where a shorter screw is inserted, filling only the glenoid vault while leaving the drill hole in the scapular spine empty.
Figure 3-3: Regions of Interest With Respect to the Screw Shaft

Views of the scapula in the coronal plane and a cross sectional view in the transverse plane are provided to visualize the regions of interest relative to the bone screws that are used to fix the glenoid baseplate to the scapula. The outcome of acromial stress by radial distance to the screw shaft along the length of the screw (Figure 3-8) is indicated as the region “radial to screw shaft” in the cross sectional view. The outcome of acromial stress that is radial and medial of the screw shaft is within the region labelled as “radial and medial of the screw shaft” (Figure 3-9).
The primary outcome variable was principal stress for each element within the acromion and scapular spine. The acromion and scapular spine were segmented into three regions as defined by Levy et al⁸ as previously described in section 1.4.3.3 (Figure 3-4) to determine if stress levels were consistently higher in the region where clinically observed fractures most frequently occur.
The Levy classification system divides the scapular spine into three regions based on fracture location relative to the deltoid segment insertion sites. Region I involves the anterior and middle deltoid, region II involves the entire middle deltoid with a portion of the posterior deltoid, and region III involves the middle deltoid and entire posterior deltoid.
To determine the forces acting on the scapula during abduction, nine static angles (0-120°) were considered in the 30° plane of elevation (scaption) internally rotated from the coronal plane. The deltoid was assumed to be the sole muscle participating in abduction with other muscles simply stabilizing the joint, as this has been shown to mimic the physiologic case\textsuperscript{20}. The deltoid was modeled as seven individual segments as described by Wickham et al\textsuperscript{21}. A custom algorithm written with Matlab (MathWorks, Massachusetts, USA) was used to perform a moment balance at the center of rotation of the glenohumeral joint, which output the seven deltoid forces acting on the acromion\textsuperscript{22}. As performed and described in Chapters 2 and 4.

The MATLAB algorithm, as previously described in chapter 2, was used to estimate both the force the deltoid applies to the arm to enable shoulder motion and the direction in which the deltoid muscle acts. As stated, the force calculation is performed via a one dimensional moment balance about the reconstructed joint’s center of rotation, which is located at the center of the base of the glenosphere, Figure 3-6. The line of action for each deltoid segment was defined using the obstacle set method\textsuperscript{23} as a single line between the geometric center of the segment’s insertion sites on the scapula and humerus. The mid points of the line segment were determined through simulated wrapping around the glenohumeral joint by the implementation of constraints that represent the bony anatomy of the glenohumeral joint and the connective tissue that contains the upper edge of the muscle. The bony anatomy was simplified to the that of a sphere capped cylinder and the lines of action were required to pass with 0-15 mm of the sphere while minimizing the muscle’s length. The seven lines of action were then projected into the plane of elevation, and a static moment balance was performed for the seven muscle segments against the force of gravity acting due to the weight of the arm. The seven muscle segment’s lines of action are shown in Figure 3-5.
Figure 3-5: Deltoid Lines of Action

The lines of action for each of the seven individual deltoid segments are mapped onto the scapula and humerus to display how the deltoid forces act on the acromion within the moment balance.
Figure 3-6: RTSA Biomechanical Rationale

*The joint’s center of rotation after RTSA as shifted from the humerus to the scapula, and the resulting deltoid moment arm that is increased by the center of rotation being shifted medially.*
The force of each muscle segment was composed of two components, in the same methodology of Chapter 2, the passive force as a result of the stretched muscle after RTSA, and the active force as a result of the muscle contraction. A peak force for the deltoid in optimal conditions was taken from literature\textsuperscript{24}, the peak force was converted into a peak optimal force for each of the seven segments by proportioning according to cross sectional area as reported in literature\textsuperscript{21}. The passive force component was determined from the optimal peak force adjusted by the ratio of the post RTSA muscle length divided by the muscle length in the native state at a 0° elevation angle. In order to make the moment balance determinant the muscle behavior was assumed to optimize the activation of the muscle segments. Accordingly, the active force component was calculated from the peak optimal force for each segment and an activation constant \( \alpha_n \).

The activation constant for each segment was restricted between 0 and 1, where 0 represents no muscle contraction and 1 represents a full contraction of the muscle segment \( (0 \leq \alpha_n \leq 1) \). The sum of the squared activation constants were minimized to mimic muscle behavior as follows in equation 3\textsuperscript{-1}\textsuperscript{25,26}:

\[
\sum_{n=1}^{7} \alpha_n^2 \\
\text{(equation 3-1)}
\]

The calculated deltoid forces along with the arm weight were then used to determine the joint reaction force that acts on the glenoid baseplate.

All forces were utilized in Abaqus (Dassault Systèmes, Vélizy-Villacoublay, France) to perform the finite element analysis. Each deltoid segment force was applied to the geometric center of the proximal deltoid insertion site as defined by Wickham et al\textsuperscript{21}, and the geometric center was tied to the entire insertion site of the respective deltoid segment. The joint reaction force was applied to the glenoid baseplate at the reconstructed joint’s center of rotation. To define the model’s boundary conditions the inferior portion of the scapula below the glenoid fossa was rigidly constrained. The scapula was recreated with
the FE method by importing the geometry from CT data (Mimics, Materialise, Belgium) and meshing the body with tetrahedral quadratic elements as these elements have been shown to sufficiently mimic the behavior of bone\textsuperscript{27}. A maximum element side length of 2 mm was utilized, as a previous mesh convergence study has shown that an element side length of 2 mm is the greatest length that does not alter the output stress results significantly\textsuperscript{27}. The stiffness of each element was assigned using the CT attenuation data to calculate apparent density, which was converted to Young’s Modulus as described in equation 3-2\textsuperscript{28}. Equation 3-2 covers a range of densities that represents both cortical and trabecular bone.

\[
E = 8920 \times \rho_{\text{apparent}}^{1.83}
\]  
(3-2)

The peak 1\% of principal stress values in each Levy region of the acromion were averaged and output as the primary outcome variable, which in turn was analyzed in a three-way repeated measures ANOVA. The peak 1\% of values were utilized to investigate peak stresses while minimizing the potential effect of outliers, when investigating radial distance from the screw shaft the peak 10\% of values were used to account for the small number of elements in proximity to the screw shaft. Further, the peak 1\% of principal stress values concentric to the screw axis and medial of the screw tip were utilized to investigate if the screw is acting as a stress riser that may lead to acromial fractures in line with the screw axis.
3.3 Results

Altering screw angle between 0° and 15° significantly affected scapular spine stress (p=0.009, Figure 3-7). The use of a perpendicular angle resulted in an average scapular spine stress of 35 ± 5 MPa compared to a divergent angle (15°), which produced 33 ± 5 MPa. In direct comparison, the divergent angle decreased acromial stress by 6.5% (2.2 ± 0.6 MPa, p=0.036). Exclusion of the superior screw did not significantly affect acromial stress for the divergent angle (33 ± 5 MPa, p=0.89) nor for the perpendicular angle (35 ± 5 MPa, p=0.78). Similarly, exclusion of the posterior screw had no significant effect on scapular spine stress. The scapular spine was not penetrated in any of the perpendicular angle models. The scapular spine was penetrated by the posterior screw in all specimen when a divergent angle was used. In 5/8 specimen the screw intersected both regions II and III of the acromion as defined by Levy8, region III was solely punctured in 2/8 specimen, and in 1/8 specimen region II was solely punctured. The average acromial stress after screw removal was 38 ± 6 MPa, which is higher than the 33 ± 5 MPa when the screw remained in the spine. In direct comparison removing the screw from the screw spine increased acromial stress by 14% (5 ± 2 MPa, p=0.04).
Figure 3-7: Acromial Stress by Screw Placement

The average of the peak 1% of acromial stress values is depicted for each screw configuration. The left figure shows acromial stress as a function of screw angle; divergent (15°) and perpendicular to the bone surface (0°). The right graph shows acromial stress for the divergent angle when the screw was withdrawn and when it remained within the scapular spine.
The relationship between acromial stress and radial distance from the screw shaft along the length of the screw is shown in Figure 3-8.
Figure 3-8: Acromial Stress by Radial Distance from The Screw Shaft

(A) The average of the peak 10% of acromial stress within 2.5 mm radial intervals between 0-2cm from the screw-bone interface. This investigates the influence of the bone screws at their boundary. (B) The regions of bone that were considered to be radial from the screw-bone interface relative to anatomic landmarks of the scapula.
The relationship between acromial stress in bone that is concentric to the screw shaft and radial distance from the screw shaft is given in Figure 3-9. There was no significant relationship between the distance intervals (p=0.197).
Figure 3-9: Peak acromial stress by radial distance from the screw shaft, in regions medial of the screw.

(A) The average of the peak 1% of stress values medial of the screw shaft as a function of radial distance from the screw axis. (B) The regions of bone that were considered to be radial from and medial of the screw relative to anatomic landmarks of the scapula.
Investigating stress location within the acromion showed that the greatest stress levels occurred in Levy region II, averaging a peak 1% stress of $62 \pm 9$ MPa, which was $52 \pm 9$ MPa greater than region I ($p=0.002$) and $28 \pm 6$ MPa greater than region III ($p=0.007$). (II>III>I, Figure 3-10).

![Figure 3-10: Acromial Stress by Region](image)

The average of the peak 1% acromial stress was determined for each acromial region as defined by Levy et al\textsuperscript{8} over all screw placements.
3.4 Discussion

The primary finding of this study is that the positioning of glenoid baseplate screws utilized in RTSA can alter acromial stresses significantly, and thus acromial stress should be considered when employing bone screws in the glenoid during RTSA. The most crucial factor to consider for this is proximity of the bone screw to the scapular spine and acromion. This is evident from the change in acromial stress as radial distance from the screw shaft increases, Figure 3-8. Stress rapidly reduces with distance from the bone screw, which is consistent with the literature\(^\text{16}\). The importance of proximity is furthered by the absence of the superior and posterior screws. The only configuration where the absence of a screw affected acromial strain was the divergent posterior screw, which is the closest screw in proximity to the scapular spine and acromion.

The importance of baseplate screw position is further highlighted by Figure 3-9, which shows a non-significant increase in peak acromial stress concentric to the screw axis. The highest stress was seen in the range 0.625 - 1.25 mm, which is within the 2.5 mm diameter of the screw. An increase in acromial stress concentric to the screw axis is consistent with clinically observed fractures along the screw axis\(^\text{13,17}\). A divergent angle that results in posterior or superior screws penetrating the scapular spine may lead to an increased probability of acromial fracture after RTSA, through the increase of stress concentric to the screw’s axis.

When utilizing a divergent angle for the posterior screw acromial stress must be considered. If the scapular spine is penetrated, the results suggest that leaving the screw within the scapular spine is more advantageous than withdrawing the screw in terms of acromial stress. The increase in acromial stress caused by a withdrawn screw is likely a result of the compression of the hole when the scapular spine and acromion are under load, causing the hole to act as a stress riser, which is localized largely to acromial region II and III where the screw typically penetrates. Further, a divergent angle of 15° decreased acromial stress compared to the perpendicular angle, likely due to the bone screw acting as a buttress against the bending of the acromion during loading of the
deltoid. However, the beneficial decrease in maximum stress may be offset by the linear stress riser concentric to the screw shaft as previously mentioned. Further, load could be transferred to the baseplate as a result and this may affect implant micromotion. Further research into angle and its effect on implant stability in a setting that accounts for deltoid pull on the acromion may be required to determine if a divergent angle is beneficial to all major complications following RTSA.

The lack of change in acromial stress between configurations when utilizing a perpendicular angle (0°) suggests that acromial stress is not a concern at a perpendicular angle and the exclusion of screws can be performed as deemed necessary to improved results or to mitigate other complications.

When considering acromial regions I, II, and III8 (Figure 3-4) in the intact state, as done in chapter 2, it is expected that region II will have the highest acromial stress because it is the region that experiences the most bending while under deltoid load. However, when introducing bone screws that puncture the scapular spine and in many cases are nearer region III than region II it is no longer clear that region II will contain the highest stress levels, as the screws may introduce new peak stresses. The fact that peak stress remained highest in region II over all screw placement configurations, which is consistent with the most commonly observed clinical fractures, suggests that it is plausible that screw placement is playing a role in acromial fractures after RTSA.

The strength of this study is the ability to directly compare different screw placement configurations within the same specimen. In-vitro studies commonly compare different configurations between different specimen because of the inability to reconstruct the same specimen with multiple implants, meaning the results could be skewed by differences between each specimen. Further, the computational nature of this study allows for precise placement of the baseplate and bone screws to achieve the desired angle, implying a higher likelihood of repeatability.

This study is limited by the assumptions used to create the custom algorithm that estimates deltoid force and direction, such as the deltoid being considered as the sole muscle participating in a one-dimensional moment balance, and muscle contraction
minimizing activation. Literature suggests that these assumptions mimic the physiologic case\textsuperscript{20,25,26}. The deltoid has been shown to account for the largest muscle force after RTSA, while other muscles are important for stability\textsuperscript{20}. Previously performed studies have shown accurate simulation of muscle activity by assuming that muscles minimize activation\textsuperscript{25,26}. Further, this study is a time zero study that does not consider the effects over time. The sample population is all male, however, the relative comparison between results will more likely apply to the whole population.

The study is also limited by the method of screw modelling. Screws have been previously modeled in 3 levels of simplifications\textsuperscript{16}: modeled with full detail including re-creation of the thread, modeling the screw as a cylindrical rod while mimicking the thread through tying the portion of the rod that the thread would occupy, and modeling the screw as a cylindrical rod with a full body tie to the bone. The more detailed modeling is more realistic, however, requires more computational power. A study was performed by another lab with the aim to find what applications the more simplistic models could be used to save computational time while maintaining accuracy in outcome variables\textsuperscript{16}. The most simplistic model of a cylindrical rod with a full body tie was found to accurately model peak strains consistent with other methods (<5\% difference)\textsuperscript{18}. However, the simplistic model created a fictional tensile force when sheared because the whole body tie prevents lift off of the bone that creates an incorrect stress distribution at the screw-bone interface of the screw head. Thus, the limitations of this modeling method should not significantly affect the results of this study of peak stress in the acromion.
3.5 Conclusion

The placement of bone screws is variable during RTSA, and the effect of screw placement on acromial stress is not well understood. The results of this study show that the placement of bone screws has a significant relationship with acromial stress and should be considered when performing RTSA. The impact of bone screws on acromial stress is largely dependent on the screw’s proximity to the scapular spine and acromion, as acromial stress was found to be noticeably higher within 0.5 mm of the screw shaft. Further, a divergent angle may decrease acromial stress compared to a perpendicular angle but may increase stress in the region concentric to the screw axis. Further knowledge of a divergent angle with respect to implant stability and stress concentric to the screw axis is required to determine if a divergent angle is beneficial in all aspects. Finally, if a bone screw penetrates the scapular spine, leaving the screw in place may decrease acromial stress compared to withdrawing the screw from the scapular spine.
3.6 References


Chapter 4

4 Validation of the Finite Element Method to Predict Acromial Surface Strains: An In-Vitro Analysis

In this chapter the finite element method and muscle wrapping algorithm, from Chapters 2 and 3, are validated against experimental surface strain measures. An experiment was conducted simulating cadaveric shoulder motion with a validated shoulder motion simulator to measure the resulting strain in the acromion with uni-axial strain gauges. The boundary and loading conditions were recreated with the finite element method. The finite element method’s resulting estimated strain values were compared directly to the experimental strain to determine the validity of the finite element method.

4.1 Introduction

As documented in Chapters 2 and 3 reverse total shoulder arthroplasty (RTSA) is an increasingly popular procedure for treating rotator cuff deficiency, severe proximal humeral fractures, and revision arthroplasty\(^1-\text{4}\). Acromial fractures are a debilitating complication associated with RTSA that has been found to occur in up to 7.2\% of the patient population\(^5-9\). Acromial fractures have been shown to lead to inferior clinical outcomes and pain\(^10-12\). Therefore, there is a need to better understand the multifactorial cause of acromial fractures.

As described in section 1.4.1, finite element (FE) analysis has been commonly used as a research tool in biomechanics, as it enables the investigation of many factors in a cost effective and time efficient manner. The FE method has been used in shoulder arthroplasty to investigate joint reaction forces of the shoulder pre and post operation, stress shielding, implant design factors on stability, and a range of rotator cuff tear severity\(^13-17\). The FE method utilized in Chapters 2 and 3 has been used to investigate the effect of implant position on acromial strain in a variety of planes of elevation and hand loads along with the effect of screw placement on acromial strain\(^18\). FE outcomes are impacted by the assumptions made at the model’s boundary conditions\(^10-21\), which could compromise the accuracy of the results. While individual assumptions in this procedure
have been found to mimic the physiological case by other studies\textsuperscript{14,22–24}, validation is required for the entire method to provide confidence that the results are both meaningful and useful.

As described in section 1.4.4, strain gauges are the current gold standard for measuring strain experimentally\textsuperscript{25–27}. Strain gauges are composed of an electric circuit that is adhered to the surface of an object; they measure strain by outputting a voltage signal that is proportional to the change in length of the surface that the gauge is adhered to. Strain gauges are limited to measuring surface strain of the object that is directly underneath the area of the strain gauge, and in the direction the circuit is laid out\textsuperscript{27}. Therefore, the direction and placement of the strain gauge must be considered in order to compare the two measurements directly.

The FE method is commonly used in orthopaedics for comparative studies, to show the effect of a design or procedure change. As a result, the difference between simulation predictions or trends are more critical than the absolute value of the model’s prediction, this was considered in the design of the methodology for this validation study. Further, as a model that is used to predict trends it is important to be aware of possible errors in the resulting conclusions. Incorrect conclusions can be classified as either type I errors (false positives) or type II errors (false negatives)\textsuperscript{30}. When comparing design or procedure changes type I errors are considered more detrimental as they suggest false confidence in a design or procedure change, which could lead to an increased risk of adverse complications in clinical studies or at the implementation into clinical practice. Whereas type II errors do not suggest the adoption of one method over the other, inferring further knowledge is required before clinical practice is changed.

Previous experiments have employed strain gauge based measurements to validate the FE method in other areas of the body such as the femur, however, there is still a need for validation of shoulder models dealing with RTSA\textsuperscript{13,28,29}. Strain gauges measure surface strain on an object, while the FE method commonly measures strain over the volume of three-dimensional elements. One study directly compared the FE method and strain gauge outcomes by outputting displacement with the FE method and using the chang in
surface length to compare to the experimentally measured strain\textsuperscript{13}. A different study that investigated strain in the metatarsal compared strain gauge measurements to FE by including FE strain measurements from elements within a 4 mm radius of the center of the strain gauge and transforming the FE strain to the local coordinate system of the strain gauge. This method resulted in good agreement between FE and strain gauge measurements achieving R-squared values ranging from 0.94-0.96 with a sample size of thirty-three (N=33).

As previously stated there is a lack of literature validating FE models for the shoulder after RTSA. Hence, the aim of this study is to validate the previously described FE models with a custom muscle wrapping software by comparison to an experimental in-vitro strain gauge study. We hypothesized that the differences in strain values between implant configurations predicted by the FE method will be the same as the differences between configurations measured experimentally. Our model also investigated different implant positions to ensure that a range of clinical scenarios were included.
4.2 Materials & Methods

The methodology of this study is described in 3 sections; the in-vitro experiment, finite element method, and data analysis and statistical methods.

An in-vitro experiment was performed by Kerrigan et al. to investigate the effect of implant position, at a variety of arm positions, on acromial strain\textsuperscript{31}. The in-vitro protocol has been previously described\textsuperscript{31} and is summarized below.

Six (6) cadaveric specimens (avg age 73\textpm9 yrs) were reconstructed with a custom adjustable RTSA implant. Full thickness rotator cuff tears were simulated in the supraspinatus and upper infraspinatus muscles, while the subscapularis, teres minor, and lower infraspinatus were held tight with a constant 10 N force applied by a pneumatic actuator to stabilize the implant. Each cadaveric shoulder was mounted to a validated shoulder motion simulator\textsuperscript{32} that utilizes pneumatic actuators to simulate muscle contraction as seen in Figure 4-2. The scapula was rigidly mounted to the active motion simulator by affixing the portion of the scapula inferior of the glenoïd fossa directly to the simulator. The deltoid was severed from the deltid tuberosity and the 3 major deltid segments (anterior, middle, posterior), as described by Sakoma et al\textsuperscript{33}, were individually attached to pneumatic actuators by cables. Arm elevation was achieved via the actuators to maintain a constant torque of 1.5 Nm, the resulting deltid forces were recorded with a load cell located in the stem of the humerus, which was also fixed to the simulator. Deltoid forces were proportioned between the deltid segments according to literature at constant ratios; 15% anterior, 70% middle, 15% posterior in the scaption plane and 40% anterior, 50% middle, 10% posterior in the forward flexion plane\textsuperscript{14}. Arm motion was simulated at static elevation angles of 0°, 90° in the scaption and forward flexion planes of elevation (30°, 90°), Figure 4-1. An elevation are guide restricted the humerus to the desired plane of elevation and allowed for elevation to the desired angle. The experimental setup is displayed in Figure 4-2.
Figure 4-1: Arm Positions

The arm was positioned in 2 planes of elevations and 2 elevation angles in each plane. I) Arm elevation angles of 0° and 90° (the scapulohumeral rhythm is displayed) II) The scaption and forward flexion planes of elevation.
Figure 4-2: Experimental Setup

The experimental set up to gather in-vitro strain measurements of the acromion. A cadaveric shoulder mounted in the shoulder motion simulator.
The acromion was outfitted with 4 uniaxial strain gauges measuring strain in the direction of the scapular spine. Strain gauge measurements were collected in real time through ethernet connection, Figure 4-2. The strain gauges were placed in each of the major acromial regions as described by Levy et al, with 2 gauges placed in region II where acromial fractures most commonly occur. Anatomic landmarks were used to place the strain gauges consistently across all specimen, Figure 4-4.

Nine (9) implant configurations where tested, by varying humeral offset (-5, 5, 15mm) and glenoid lateralization (0, 5, 10mm), Figure 4-3. Each configuration was placed in the 4 arm positions previously stated, Figure 4-1. Deltoid force was applied for 10 seconds in 3 separate trials, this was repeated for each configuration while strain measurements were collected, averaged, and reported as micro strain.
Figure 4-3: RTSA Implant Placement

Reverse total shoulder arthroplasty implant placement variables (A) glenosphere lateralization (B) glenosphere inferiorization (C) humeral cup lateral/medialization
Figure 4-4: Strain Gauge Placement

Strain gauge 1 was placed in the middle of the acromial clavicular joint and the lateral/medial edges of the acromion, strain gauge 2 was placed midway between the lateral edge of the acromion and strain gauge 3, strain gauge 3 was placed above the spinoglenoid notch, and strain gauge 4 is placed 2 cm medial of strain gauge 3.
FE models were constructed to mimic the in-vitro biomechanical study that measured surface strain on the acromion after RTSA for a range of implant configurations and arm positions. FE models of 6 fresh frozen cadaveric shoulders (ave age 73+/-9 yrs.) used in the in-vitro study were constructed using CT data (Mimics, Materialise, Belgium). The scapula was segmented, similarly to Chapters 2 and 3, into quadratic tetrahedral elements as they have been found to adequately model the behavior of bone\textsuperscript{13}, with a maximum side length of 2 mm applied to the acromion and superior scapula. The size of interior elements was increased in the region of the scapula that was immobilized to decrease computation time, while preserving accuracy in the areas of interest. Material properties were assigned based on Young’s Modulus calculated using apparent density from CT data, according to equation 3 - 2.

The boundary conditions of the FE models were duplicated from Kerrigan et al; the inferior scapula was immobilized by restricting translation and rotation in elements below the glenoid fossa, and forces were simulated for the anterior, middle, and posterior deltoid segments\textsuperscript{33}. The motion simulator was programmed to control forces by maintaining a constant torque of 1.5 Nm to maintain a static arm position at the desired angle of elevation, the resulting forces were recorded and input to the FE models for each arm and implant configuration. The direction of the force for each deltoid segment was determined using a custom muscle wrapping algorithm, as described in Chapters 2 and 3.

The purpose of the custom muscle wrapping algorithm was to determine the path that the deltoid occupies between insertion sites on the scapula and the humerus. The deltoid’s distal insertion site was identified as a single point on the deltoid tuberosity and was kept constant across the deltoid segments. Each deltoid segment’s proximal insertion site was defined as the entire region where the muscle attaches to the surface of the acromion and clavicle, as described by Sakoma et al\textsuperscript{7,33}. The geometric center of each proximal insertion site of the deltoid was used in the muscle wrapping algorithm to determine each deltoid segment’s line of action. The deltoid force was applied to the bone over the whole proximal insertion site by tying the geometric center to every element within the insertion
site on the surface of the bone. The wrapping algorithm determined muscle path by minimizing the length of the deltoid between the two insertion sites, while requiring each muscle segment to pass within the range of 0-15 mm of the humeral head to mimic physiological wrapping directions. Constraining the deltoid to within 0-15 mm of the humeral head simulates the deltoid sitting on top of the humerus to prevent pathways that intersect with the bone. The wrapping algorithm was based off of a previously validated obstacle-set method that simulates the humeral head as a sphere\textsuperscript{18,34,35}.

All experimental configurations were reproduced via FE modelling according to the experimental study as follows: implant position was varied in the FE models by offsetting the glenoid baseplate laterally (0, 5, 10 [mm]), and altering the humeral offset (-5, 5, 15 [mm]) to match the experimental configurations, Figure 4-4. The arm was elevated in two planes of elevation; in line with the scapula or the scaption plane of elevation (30°), and in the forward flexion plane of elevation (90°), and the arm was abducted to 2 angles (0°, 90°) in both planes of elevation, Figure 4-1.

The principal strain components for each strain gauge location was output from the FE models, which were compared locally to the experimental strain measures in order to validate the FE method and muscle wrapping algorithm previously used in Chapters 2 and 3.

To compare the strain measurements directly, CT scans of the cadaveric specimen were taken after the in-vitro experiment, the strain gauge locations were then overlaid onto the FE models. The four corners of each strain gauge were identified in the FE model’s global coordinate system, a local strain gauge coordinate system was then defined using the 4 corners of each gauge. Element sets were defined for the mesh elements at the surface of the acromion that were within the boundaries of each strain gauge. The output principal strain components for each element set were transformed into the local strain gauge coordinate system. The resulting principal strain in the direction of the uni-axial gauge was averaged over all elements in the corresponding strain gauge element set, strain gauge direction is highlighted in Figure 4-5. This procedure was performed for
each of the 4 strain gauges. The averaged strain measurements were then compared to the in-vitro measurements directly.
Figure 4-5: Strain Gauge Coordinate Systems

A depiction of the global coordinate system and the local coordinate systems defined for each strain gauge with the x axis aligned with the strain gauge’s uniaxial circuit.
In order to assess the FE method’s ability to predict trends in strain results repeated measure ANOVAs were performed for the output of both the FE and in-vitro methods. The ANOVA tested differences in strain between strain gauge location, humeral offset, glenoid offset, plane of elevation, and angle of elevation. A success rate for the FE method’s predictions of trends was calculated to validate the FE method, where a predicted trend was considered successful if the sign matched that of the in-vitro measured trend. Additionally, the magnitudes of the predicted trends were compared within an ANOVA to determine if a significant difference in the trend magnitudes existed between the FE method and in-vitro study. As hypothesized, it is expected that the FE method will predict the same trends as the in-vitro experiment, and no significant difference will exist.
4.3 Results

The measurement of strain for both methods as a function of strain gauge location on the acromion is summarized in Figure 4-6.

![Figure 4-6: Strain as a Function of Strain Gauge Placement](image)

*The graph on the left displays strain values for both the FE and in-vitro methods at each strain gauge location averaged over all other factors, with uncertainty of ±1 standard deviation. The graph on the right shows the differences (trends) for the strain gauge locations as quantified by both methods.*
The measurement of strain for both methods as a function of humeral component offset is summarized in Figure 4-7.

**Figure 4-7: Strain as a Function of Humeral Offset**

The graph on the left displays strain values for both the FE and experimental methods for each humeral offset, with uncertainty of ±1 standard deviation. The graph on the right shows the differences (trends) between the different humeral offsets as predicted by both methods.
The measurement of strain for both methods as a function of glenoid component offset is summarized in Figure 4-8.

Figure 4-8: Strain as a Function of Glenoid Offset

The graph on the left displays predicted strain values for both the FE and experimental methods for each glenoid offset, with uncertainty of ±1 standard deviation. The graph on the right shows the differences (trends) between the different glenoid offsets as predicted by both methods.
The measurement of strain for both methods as a function of plane of elevation is summarized in Figure 4-9.

**Figure 4-9: Strain as a Function of Plane of Elevation**

The graph on the left displays predicted strain values for both the FE and experimental methods in each plane of elevation, with uncertainty of ±1 standard deviation. The graph on the right shows the differences (trends) between the two planes of elevation as predicted by both methods.
The measurement of strain for both methods as a function of angle of elevation is summarized in Figure 4-10.

**Figure 4-10: Strain as a Function of Angle of Elevation**

The graph on the left displays predicted strain values for both the FE and experimental methods at each elevation angle, with uncertainty of ±1 standard deviation. The graph on the right shows the differences (trends) between the two elevation angles as predicted by both methods.
The strain values and trends between experimental factors as predicted by both methods are displayed in Figures 4-6 to 4-10. The FE method was able to predict significant trends with a success rate of 100% (5/5), while predicting both significant and non-significant trends at a success rate of 86% (12/14). There was no significant difference in the predicted trend magnitudes between the FE and experimental methods (p=0.198).
Table 4-1: Summary of Predicted Strain Values

Green values indicate trend agreement between in-vitro and FE measures, while red values indicate disagreement.

<table>
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<tr>
<th>Finite Element Method - Strain Gauge Location</th>
<th>In-vitro Method - Strain Gauge Location</th>
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</thead>
<tbody>
<tr>
<td>-------</td>
<td>------</td>
</tr>
<tr>
<td>SG 1</td>
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<tr>
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<td>3</td>
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<td></td>
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<tr>
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<td>2</td>
</tr>
<tr>
<td></td>
<td>4</td>
</tr>
<tr>
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</tr>
<tr>
<td></td>
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</tr>
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<td>3</td>
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<thead>
<tr>
<th>Finite Element Method - Humeral Offset</th>
<th>In-vitro Method - Humeral Offset</th>
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</thead>
<tbody>
<tr>
<td>HO (mm)</td>
<td>Mean</td>
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<tr>
<td>-------</td>
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</tr>
<tr>
<td>-5</td>
<td>166.327</td>
</tr>
<tr>
<td>5</td>
<td>156.004</td>
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<td>159.653</td>
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<th>In-vitro Method - Glenoid Offset</th>
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</thead>
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<td>GO (mm)</td>
<td>Mean</td>
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<td>------</td>
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<td>0</td>
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<th>In-vitro Method - Plane of Elevation</th>
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<tr>
<td>-------</td>
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<td>abd</td>
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<th>In-vitro Method - Angle of Elevation</th>
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<td>------</td>
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<tr>
<td>0</td>
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4.4 Discussion

The current work shows that the FE method predicts the trends of strain between experimental factors very consistently (Figure 4-6 to Figure 4-10). The trends predicted by the FE method with significance (p<0.05) were found to agree with experimental strain measurements at a success rate of 100%. This shows that the combination of the FE method and wrapping algorithm is a sufficiently accurate research tool for determining which experimental factors such as implant design factors and arm positions will lead to higher acromial stresses/strains. Further, with an 86% success rate at predicting trends including non-significant trends shows that the FE method is accurate at predicting both significant and non-significant trends, resulting in a higher level of confidence in the given results.

This validation study is limited by the ability to directly compare the results. The strain gauges measures surface strain while the FE method measures strain over the entire volume of elements. This was accommodated for by only considering one layer of elements, which has a maximum thickness of 2 mm. One layer of quadratic tetrahedral elements was used rather than using shell elements that more closely mimic surface strain because quadratic tetrahedral elements are the elements used in the models for other studies. Therefore, the utilization of quadratic tetrahedral elements better validates the FE protocol used in Chapters 2 and 3. The study is also limited by the fact that the sample population was all male right shoulders and may not represent the entire population of RTSA patients, however, relative results are more likely applicable than absolute results.

A major strength of this work is the ability to make comparisons between the same specimen over multiple configurations. Comparing 144 cases per specimen (4 locations, 3 humeral offsets, 3 glenoid offsets, 2 planes of elevation, and 2 angles of elevation) provides a large database and allows for the comparison between many factors and trends that this model has been used to predict in previous studies, giving confidence in the results of the validation. The validation is pragmatic in design as it incorporates both the muscle wrapping algorithm and the FE method to give confidence in the full protocol rather than just the FE method\textsuperscript{13,28,29}.
4.5 Conclusions

Overall, the FE method and muscle wrapping algorithm is an accurate method for predicting increases and decreases in acromial strain for experimental factors and can be used as a biomechanical research tool to investigate changes in surgical procedure or implant design in a cost effective, and repeatable manor.
4.6 References


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Chapter 5

5  Thesis Closure

In this chapter the thesis is brought to a close by summarizing the objectives and hypotheses. A general discussion of the results follows. The strengths and limitations of the body of work as a whole is recapped. Future directions are outlined, and the significance of the results is stated.

5.1  Summary

Reverse total shoulder arthroplasty (RTSA) is a common end stage treatment with the aim of restoring shoulder joint range of motion and relieving pain. Acromial fractures are a common and debilitating complication associated with RTSA. This dissertation utilizes mechanical and biomechanical engineering methods to explore the likely multifactorial cause of acromial fractures, with the aim to mitigate the probability of acromial fractures in future patients.

The first objective (Chapter 2 and Chapter 4) was to improve on the estimation of muscle forces acting on the acromion after RTSA, by extending the model to multiple planes of elevation (Objective 1a), to account for multiple hand loads that correspond to typical weights lifted in activities of daily living (Objective 1b), and to validate the FE model against in-vitro measures (Objective 1c). It was hypothesized that the differences in stress between implant configurations would be similar in both the FE and the in-vitro method (Hypothesis 1). Further it was hypothesized that the addition of hand loads consistent with activities of daily living would increase stress above the fatigue threshold (60 MPa) of cortical bone and that plane of elevation would alter bone stress levels (Hypothesis 2).

The validation of the FE method by comparison to a strain gauge based in-vitro study showed acceptable accuracy for comparing trends between configurations as predicted in Hypothesis 1. With an overall trend prediction success rate of 86% and no significant difference detected between the magnitude of trend predictions between methods (p=0.198), the FE methodology was shown to be an acceptably accurate research tool for
predicting increases and decreases in acromial strain. Significant increases in acromial stress were noted as applied load was increased in all planes of elevation acromial stress was elevated within the range of the fatigue threshold for cortical bone (60 MPa) as hypothesized. Plane of elevation was seen to effect acromial stress significantly as more external planes (0°, 90°) increased stress compared to more central planes of elevation (30°, 60°; Hypothesis 2).

The second objective (Chapter 2) was to determine how the variable placement of RTSA implants (glenosphere lateralization, glenosphere inferiorization, and humeral offset) affects acromial stress over a set of arm positions and hand loads representative of the shoulder range of motion during a variety of activities of daily living (Objective 2). It was hypothesized that the only implant placement parameter that would affect acromial stress significantly would be glenosphere lateralization (Hypothesis 3).

Exploring the effects of implant position over the range of planes of elevation, and hand loads showed that glenosphere lateralization and glenosphere inferiorization had a significant impact. Glenosphere lateralization consistently increased acromial stress likely a result of shortening of the deltoid’s moment arm. Glenoid inferiorization consistently decreased acromial stress, likely by lengthening the deltoid’s moment arm. The results support Hypothesis 3 in the sense that glenosphere lateralization had a significant impact on acromial stress, however, the impact of glenosphere inferiorization was not hypothesized. Glenosphere inferiorization’s effect in previous studies was not shown to be significant, however, the magnitude of its impact was magnified with applied load leading to significance. Further humeral offset in the configurations investigated did not impact acromial stress significantly.

The third objective (Chapter 3) was to investigate the placement of bone screws used to fixate RTSA implants onto the scapula and how screw placement alters stress levels in the acromion. Screw placement variables investigated include: the drilling angle used to place the screws (Objective 3a), the presence or absence of the posterior and superior screws (Objective 3b), and the presence or withdrawal of a screw within the scapular spine (Objective 3b). It was hypothesized that a divergent drilling angle (hypothesis 4a),
the absence of the posterior screw (Hypothesis 4b), and leaving the screw within the scapular spine if it was penetrated (Hypothesis 4c) would all decrease acromial stress.

A divergent drilling angle decreased acromial stress compared to a perpendicular drilling angle. The absence of the posterior screw significantly increased acromial stress. If the scapular spine was penetrated by a bone screw it was found to be beneficial to leave the screw within the scapular spine rather than withdrawing the screw from the spine in terms of acromial stress. The results of this study show that proximity of the bone screws to the scapular spine plays a major role on its impact on acromial stress. If a perpendicular drilling angle is used, it is unlikely that acromial stress will be altered.

5.2 Strengths and Limitations

A major strength associated with performing FE based experiments is the ability to directly compare implant placement or design factors within the same specimen. The invasive nature of reconstructing a joint with an implant in both the in-vivo and in-vitro environments prevents multiple reconstructions of the same joint, however, multiple FE models of the same joint with different reconstructions can be made virtually. The benefit of comparison between the same specimen is the ability to isolate the change in the implant from the anatomic differences that exist between different specimens. Furthermore, the FE method allows for placement of objects in space in a quantitative and automated way, which may increase the repeatability of implant placement between models compared to manually placed implants.

The validation study described in Chapter 4 strengthens the results of Chapters 2 and 3 as it provides confidence in the methodology used. Commonly in validations of the FE method both the boundary conditions and loading conditions are replicated identically, which does not validate the loading conditions used in the respective FE studies. The pragmatic nature of validating both the FE method and the wrapping algorithm is in the author’s opinion a strength of this work.

A limitation of using the FE method is that the biomechanical models are often complex and indeterminate, meaning that the models require assumptions of how the bony and soft
tissue structures behave. In order to have confidence in the assumptions that are made validation studies are required. To satisfy this previously performed validations of individual assumptions have been referenced and a pragmatic validation of the process as a whole was performed in Chapter 4. Another limitation of this work is that the population used for experimentation were all male and some were younger than the average age of the RTSA population, to account for this all studies were designed to be parametric such that relative differences between configurations would more likely represent the behavior of the whole RTSA population. The estimation of muscle forces after RTSA is limited by the lack of a golden standard for comparison, the typical gold standard for joint reaction forces is telemetry-based measurements from in-vivo implants which is lacking with respect to RTSA. Therefore, the validity of estimated forces can only be compared against other models and experiments rather than actual values. To account for this, relative values between models were utilized rather than absolute predictions.

5.3 Future Directions

The objectives laid out in Chapter 1 have been sufficiently met to better understand how implant placement variables effect acromial stress after RTSA. However, acromial fractures after RTSA remain a complex problem that is not completely understood.

To further the current state of knowledge on complications after RTSA the FE method used can be improved, specifically with the intention to achieve accurate absolute values rather than solely accurate trends. Telemetry based joint reaction forces for RTSA could be gathered to improve the loading conditions in FE models for RTSA. Additionally, digital volume correlation is a recent method that has been utilized to improve FE boundary conditions for increased accuracy in the absolute value of stress and strain predictions.

To better understand acromial fractures after RTSA, the effect of acromion shape in the sagittal plane could be investigated to determine if a subset of the RTSA population is more susceptible to acromial fractures as a result of their anatomy. Acromial shape has been shown to affect joint impingement and may affect acromial stress possibly as a
result of altering the deltoid moment arm. Further, the results of this biomechanical FE and in-vitro work can be used to design in-vivo retrospective studies as the population size of RTSA grows, to ascertain if the effects detected at this level are translatable to clinical results.

5.4 Significance

As the prevalence of RTSA continues to grow, it is important to understand how all major complications can be mitigated. This work shows the importance of implant and screw placement on acromial stress after RTSA. While implant placement is being used to address other complication such as implant stability, these results highlight that their effect on acromial fractures should also be considered. This work can help guide the optimal surgical procedure with the goal of minimizing the occurrence of acromial fractures.
Appendices

Appendix A: Custom MATLAB code for force magnitude and direction estimation

The custom MATLAB algorithm used to estimate Deltoid forces is presented below:

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```
function ForceCalculationsF(Load_Force)

profile on

% Specimen IDs
Specimen_List{1} = '11-03057L';
Specimen_List{2} = '12-02045L';
Specimen_List{3} = '11-03052R';
Specimen_List{4} = '10-01021R';
Specimen_List{5} = '11-03045R';
Specimen_List{6} = '11-03053L';
Specimen_List{7} = '09-12052L';
Specimen_List{8} = '10-08004L';
Specimen_List{9} = '11-09004L';
Specimen_List{10} = '08-02050L';

n_Specimen = size(Specimen_List,2);

% run for the desired specimen
for i_Specimen = [5]

    Load = num2str(Load_Force)
    Specimen = Specimen_List{i_Specimen}
```

Points to import
% naming conventions
LandmarksFile_NAT = [Specimen,'-NAT.xlsx'];
LandmarksFile_RSA = [Specimen,'-RSA.xlsx'];
AbaqusFileName_NAT = [Specimen,'-NAT.inp'];
AbaqusFileName_RSA = [Specimen,'-RSA.inp'];
File_Humerus_NAT[1] = [Specimen,'-NAT-HUM.stl'];
n_hum_NAT = size(File_Humerus_NAT,2);
File_Humerus_RSA[1] = [Specimen,'-RSA-HUM.stl'];
n_hum_RSA = size(File_Humerus_RSA,2);
File_Scapula_NAT = [Specimen,'-NAT-SCAP.stl'];
File_Scapula_RSA = [Specimen,'-RSA-SCAP.stl'];
SaveName_NAT = [Specimen,'-Loaded-',Load,'kg-NAT-Flexion.mat'];
SaveName_RSA = [Specimen,'-Loaded-',Load,'kg-RSA-Flexion.mat'];
SaveName_NAT = fullfile('Increments',SaveName_NAT);
SaveName_RSA = fullfile('Increments',SaveName_RSA);

% start timer
tic

% Functional variables :=run 0=skip
Abaqus = 0;
Visualize = 0;
AbaqusTransform = 0;
SaveResults = 0;
CT_COR = 0;
n_Type = 2; %1 = native joint 2 = RTSA joint
for i_Type = 1:n_Type

  % set up for calculations
  if i_Type == 1
    LandmarksFile = LandmarksFile_NAT;
    File_Humerus = File_Humerus_NAT;
    n_hum = n_hum_NAT;
    File_Scapula = File_Scapula_NAT;
    SaveName = SaveName_NAT;
    AbaqusFileName = AbaqusFileName_NAT;
    NoOffset = 1;
  else
    LandmarksFile = LandmarksFile_RSA;
    File_Humerus = File_Humerus_RSA;
    n_hum = n_hum_RSA;
    File_Scapula = File_Scapula_RSA;
    SaveName = SaveName_RSA;
    AbaqusFileName = AbaqusFileName_RSA;
    NoOffset = 0;
  end

  % load in landmarks for the specimen of interest
Landmarks = xlsread(LandmarksFile, 'B2:T4');
Landmarks_CT = Landmarks(1:3,:);
Landmarks = Landmarks(4:end,:);
Radius = xlsread(LandmarksFile, 'U2:U2');

% Set all possible configurations
Abd_Angle = [0 15 30 45 60 75 90 105 120];
% Rhythm_Angle = [0 0 15 30 45 60 75 90]*4/5; %Poppen 1976
%Ackland 2011 based on Poppen 1976
Rhythm_Angle = [0 0 15 30 45 60 75 90]*4/5;
Elev_Angle = [0 0 15 30 45 60 75 90]*1/2;
Lat_Offset = [0 5 10];
Inf_Offset = [0 2.5 5];
Hum_Offset = [0 -5 5];

% For testing
%     Abd_Angle = [0,120];
%     Elev_Angle = Abd_Angle;
%     Lat_Offset = 0;
%     Inf_Offset = 0;
%     Hum_Offset = 5;
if NoOffset
    Lat_Offset = [0];
    Inf_Offset = [0];
    Hum_Offset = [0];
end

% Offset = [0 0 0; 0 0 5; 0 0 10; 0 5 0; 0 10 0];
n_Abd_Angle = length(Abd_Angle);
n_Elev_Angle = length(Elev_Angle);
n_Deltoid = 7;
n_Lat_Offset = length(Lat_Offset);
n_Inf_Offset = length(Inf_Offset);
n_Hum_Offset = length(Hum_Offset);
n_Movements = 2; % 1=abduction 2=elevation

% Pad list of angles if not the same. For ease of storing in arrays
movement_diff = n_Abd_Angle - n_Elev_Angle;
if movement_diff > 0
    n_Elev_Angle = n_Abd_Angle;
    n_Elev_Angle = [n_Elev_Angle, zeros(1, movement_diff)];
elseif movement_diff < 0
    n_Abd_Angle = n_Elev_Angle;
    n_Abd_Angle = [n_Abd_Angle, zeros(1, -movement_diff)];
end

cases = [n_Abd_Angle, n_Deltoid, n_Lat_Offset, n_Inf_Offset, n_Hum_Offset, n_Movements];
%Set parameters for force calculations
PCSA = [4.2 3.9 6.0 5.0 4.2 2.0 2.4]; %Wickham 2002 in cm^2
F_max = 2045*PCSA/sum(PCSA); %Garner 2003
%  F_max = PCSA * 33; % 330kPa max stress Garner 2003 ref 39 dont use
Gravity_MomentArm = 320; %values from Terrier, 2008
Gravity_Force = 37.5; %values from Terrier, 2008 (g=10 m/s^2 was used)
Load_MomentArm=703; % value from Wiley 1990
%  Load_Force=10; % N for a 1 kg load (g=10 m/s^2 is used)
options_old = optimset('fmincon');
options = optimset(options_old,'Algorithm', 'interior-point', 'Display', 'off');
%delete later
% PassiveCurve = csvread('PassiveForce.csv');
% ActiveCurve = csvread('ActiveForce.csv');
PassiveCurve = xlsread('PassiveForcePoints.xlsx');
fit_passive = fit(PassiveCurve(2:end,1),PassiveCurve(2:end,2),'exp1');
ActiveCurve = xlsread('ActiveForcePoints.xlsx');

%Properties to calculate
WrapVecs = cell(cases);
ForceComponents = cell(cases);
Lengths = zeros(cases);
AllWrapPoints = cell(cases);
AbductionTmats = cell(cases([1,3:end]));
MomentArms = zeros(cases);
MomentScale = zeros(cases);
Forces = zeros(cases);
Forces_Passive = zeros(cases);
Length_Ratios = zeros(cases);
Passive_Ratios = zeros(cases);
Active_Ratios = zeros(cases);
n_ScapWrap = zeros(cases);

Define local coordinate systems
%
[ Tmat_Scap ] = CS_Scap(Landmarks(2,:), Landmarks(3,:), Landmarks(1,:), Landmarks(5+n_Deltoid,:));

%define the transform from Abaqus/CT coordinates to SW
[ Abaqus_translation, Abaqus_roll, Abaqus_pitch, Abaqus_yaw, Tmat_CT_SW ] = rel_Tmat(Landmarks_CT, Landmarks(1:3,:));
%Tmat\b takes local points and transforms them to global
%Tmat\b takes global points to local
if AbaqusTransform
    disp('Perform x,y,z rotations then translation as follows: 
    Abaqus_roll
    Abaqus_pitch
    Abaqus_yaw
    Abaqus_translation
end

%to move COR to position from CT
if i_Type == 1 && CT_COR == 1
    COR_HumCT = xlsread(LandmarksFile,'V2:V4');
    COR_HumSw = Tmat_CT_SW*[COR_HumCT,1]';
    COR_HumSw = COR_HumSw(1:3,:);  
    for i_dim = 1:3
        Landmarks(end-4:end,i_dim) = Landmarks(end-4:end,i_dim)+COR_CT2SW(i_dim);
    end
end

Landmarks_Transform = [Landmarks, ones(length(Landmarks),1)]';
Landmarks_ScapCS_Transform = Tmat_Scap\Landmarks_Transform;
Landmarks_ScapCS = Landmarks_ScapCS_Transform(1:3,:);

% Tilt scapula forward to achieve physiologic state
Scap_Tilt = 10;
x_dir = Tmat_Scap(1,1:3);

%for L shoulder reverse rotation angle. Looking at x coord of anterior
%and posterior deltoid to see if x is positive in anterior direction
%(means its the left shoulder)
if (Landmarks_ScapCS(5,1)-Landmarks_ScapCS(4+n_Deltoid,1)) > 0
    Left = 1;
    Scap_Tilt = -Scap_Tilt;
else
    Left = 0;
end

Tmat_ScapTilt = RotationMatrix(0,0,Scap_Tilt);
Tmat_ScapGlobal = Tmat_ScapTilt/Tmat_Scap;

Landmarks_TrueScapCS_Transform = Tmat_ScapTilt^Landmarks_ScapCS_Transform;
Landmarks_TrueScapCS = Landmarks_TrueScapCS_Transform(1:3,:);'

Load Humerus vertices to tie wrap points to.

Hum_v = [];
Hum_v_part = cell(n_hum,1);
Hum_f_part = cell(n_hum,1);
for m = 1:n_hum
    [Hum_v_part{m},Hum_f_part{m}] = stlread(File_Humerus{m});
    [Hum_v_part{m},Hum_f_part{m}] = patchslim(Hum_v_part{m},Hum_f_part{m});

    % to move COR to CT position
    if i_Type == 1 && CT_COR == 1
        for i_dim = 1:3
            Hum_v_part{m}(:,i_dim) = Hum_v_part{m}(:,i_dim)+COR_CT2SW(i_dim);
        end
    end
Hum_v = [Hum_v; Hum_v_part{m}];

end

Tmat_HumOffset = eye(4);

Run Through Abduction Angles in all configurations

for i_Abd_Angle = 1:n_Abd_Angle
    for i_Hum_Offset = 1:n_Hum_Offset
        Landmarks_Post_HumOffset = Landmarks_TrueScapCS_Transform;
        Landmarks_Post_HumOffset(3,6+n_Deltoid:end) =
        Landmarks_Post_HumOffset(3,6+n_Deltoid:end)-Hum_Offset(i_Hum_Offset); %Translates HumPoints laterally by Hum_Offset
        Tmat_HumOffset(3,4) = -Hum_Offset(i_Hum_Offset);
        for i_Movements = n_Movements
            %rotate humerus points to chosen abduction angle
            if i_Movements == 1
                [ Tmat_Abduction ] = RotationMatrix(Abd_Angle(i_Abd_Angle),0,0);
            elseif i_Movements == 2
                %rotate about y so that x is normal to elevation
                %plane, then rotate about x by elevation angle
                [ Tmat_Abduction ] =
                RotationMatrix(0,30,0)\(RotationMatrix(Elev_Angle(i_Abd_Angle),0,0)*RotationMatrix(0,30,0)

                if ~Left
                    [ Tmat_Abduction ] = RotationMatrix(0,-30,0)\(RotationMatrix(Elev_Angle(i_Abd_Angle),0,0)*RotationMatrix(0,-30,0));
                end
                [ Tmat_Abduction ] = RotationMatrix(0,-60,0)*RotationMatrix(Elev_Angle(i_Abd_Angle),0,0);
            end

            for i_Lat_Offset = 1:n_Lat_Offset
                Tmat_Abduction(3,4) = -Lat_Offset(i_Lat_Offset);
            end

            for i_Inf_Offset = 1:n_Inf_Offset
                Tmat_Abduction(2,4) = -Inf_Offset(i_Inf_Offset); %apply to COR
            end
        end
    end
end

AbductionTmats{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} = Tmat_Abduction;

Landmarks_Abd_Transform =
Tmat_Abduction*Landmarks_Post_HumOffset;
Landmarks_Abd = [Landmarks_Post_HumOffset(1:3,1:4+n_Deltoid),
Landmarks_Abd_Transform(1:3,5+n_Deltoid:end)]';
%only apply abduction to humerus points & COR

% define IA, SE, DI, PD, TS
TS = Landmarks_Abd(2,:);
IA = Landmarks_Abd(3,:);
SE = Landmarks_Abd(4,:);
DI = Landmarks_Abd(end-2,:);
DI_resting = Landmarks_Post_HumOffset(1:3,end-2)';
HH = Landmarks_Abd(end-3,:);
HH_resting = Landmarks_Post_HumOffset(1:3,end-3)';
HumBottom = HH;
HumBottom(2) = HH(2) - Radius;
HumLat = HH;
HumLat(3) = HH(3) - Radius;
Scap_Edge = zeros(100,3);
Joint_Space = zeros(100,3);
Arc = zeros(100,3);
HumLine = zeros(100,3);
hum_transition = 1;
for i_dim = 1:3
    Scap_Edge(:,i_dim) = linspace(SE(i_dim),IA(i_dim),100);
    Joint_Space(:,i_dim) = linspace(HumBottom(i_dim),SE(i_dim),100);
    Arc(:,i_dim) = linspace(HumLat(i_dim),HumBottom(i_dim),100);
    HumLine(:,i_dim) = linspace(HumLat(i_dim),HH(i_dim),100);
end

for i_arc = 2:length(Arc)
    Arc(i_arc,2) = HH(2) - sqrt(Radius^2 - (Arc(i_arc,1) - HH(1))^2 - (Arc(i_arc,3) - HH(3))^2);
end

inside_line = [Arc(1:end-1,:);Joint_Space(1:end-1,:);Scap_Edge];
inside_line_global = Tmat_Scap'(Tmat_ScapTilt\Tmat_HumOffset\(((inside_line,ones(size(inside_line,1),1)])'))
inside_line_global = inside_line_global(1:3,:)';
IDX = knnsearch(Hum_v,inside_line_global);
inside_line2_global = Hum_v(IDX,:);
inside_line2 = Tmat_Abduction'(Tmat_ScapTilt\Tmat_Scap\[inside_line2_global,ones(size(inside_line,1),1)])
inside_line2 = inside_line2(1:3,:)';
inside_line2(hum_transition:end,:) = inside_line(hum_transition:end,:);

% Wrap point on scapula if needed
for i_Deltoid = n_Deltoid:-1:1
    inside = 0;
    Scap_wrap = Landmarks_Abd(4+i_Deltoid,:);
    L_Scap = 0;
end
%ObstacleSet( Humerus_Head, Deltoid, 
%Humerus_Tub, Med_Epi, Lat_Epi, Radius );
[ AllPoints, Length_ObstacleSet]...
   = obstacleSet( Landmarks_Abd(6+n_Deltoid,:), 
ScapWrap(1,:), ... 
   = obstacleSet( Landmarks_Abd(7+n_Deltoid,:), 
Landmarks_Abd(8+n_Deltoid,:), Landmarks_Abd(9+n_Deltoid,:), Radius );

if i_Deltoid > 4 && i_Movements == 2 && i_Abd_Angle > 4

PD = Landmarks_Abd(4+i_Deltoid,:);
    u = SE-IA;
    u = u/norm(u);
    t = SE-TS;
    v = DI-PD;
    v = v/norm(v);

plane_n = cross(u,t);
plane_n = plane_n/norm(plane_n);
plane_d = dot(plane_n,SE);
SN = -dot(plane_n,PD-SE);
SD = dot(plane_n,v);
SI = SN/SD;
muscle_projection = PD + SI.*v;

if muscle_projection(3) > inside_line(1,3)

    line_index = knnsearch(inside_line(:,3),muscle_projection(3));
    if muscle_projection(2) > inside_line(line_index,2)
        proj_index = knnsearch(inside_line,muscle_projection);
        inside = 1;
        L_tot = 10^9;
        for i_line = 1:size(inside_line,1)
            if Left
                if i_Deltoid == 5
                    FixedPoint = [5.164; -0.779; 29.801]';
                    FixedPoint = Tmat_Scap^T(Tmat_ScapTilt\(Tmat_HumOffset\(((\text{FixedPoint,ones(size(FixedPoint,1))})')))));
                    FixedPoint = FixedPoint(1:3,1)';
                elseif i_Deltoid == 6
                    FixedPoint = [5.228; -10.274; 23.991]';
                    FixedPoint = Tmat_Scap^T(Tmat_ScapTilt\(Tmat_HumOffset\(((\text{FixedPoint,ones(size(FixedPoint,1))})'))));
                    FixedPoint = FixedPoint(1:3,1)';
                else

else


FixedPoint = [-0.265; -15.726; 21.958]';
FixedPoint = Tmat_Scap'*(Tmat_ScapTilt'(Tmat_HumOffset(((FixedPoint,ones(size(FixedPoint,1))))')));
FixedPoint = FixedPoint(1:3,1)';
end
else
if i_Deltoid == 5
FixedPoint = [1.52802; 0.58151; 35.31221]';
FixedPoint = Tmat_Scap'*(Tmat_ScapTilt'(Tmat_HumOffset(((FixedPoint,ones(size(FixedPoint,1))))')));
FixedPoint = FixedPoint(1:3,1)';
elseif i_Deltoid == 5 && i_Abd_Angle == 9 && i_Type == 1
FixedPoint = [-5.41; 4.82; 25.74]';
FixedPoint = Tmat_Scap'*(Tmat_ScapTilt'(Tmat_HumOffset(((FixedPoint,ones(size(FixedPoint,1))))')));
FixedPoint = FixedPoint(1:3,1)';
elseif i_Deltoid == 6
FixedPoint = [-1.703; -9.569; 28.072]';
FixedPoint = Tmat_Scap'*(Tmat_ScapTilt'(Tmat_HumOffset(((FixedPoint,ones(size(FixedPoint,1))))')));
FixedPoint = FixedPoint(1:3,1)'
else
FixedPoint = [1.046; -15.191; 23.132]';
FixedPoint = Tmat_Scap'*(Tmat_ScapTilt'(Tmat_HumOffset(((FixedPoint,ones(size(FixedPoint,1))))')));
FixedPoint = FixedPoint(1:3,1)';
end
end
ScapWrap = [FixedPoint(1,:);Landmarks_Abd(4+i_Deltoid,:)];
%inside_line2(i_line,:)
[AllPoints_Scap, Length_Scap]... = ObstacleSet( Landmarks_Abd(6+n_Deltoid,:), ScapWrap(2,:), ... FixedPoint, Landmarks_Abd(8+n_Deltoid,:), Landmarks_Abd(9+n_Deltoid,:), Radius );

[ AllPoints, Length_ObstacleSet]... = ObstacleSet( Landmarks_Abd(6+n_Deltoid,:), FixedPoint, ... Landmarks_Abd(7+n_Deltoid,:), Landmarks_Abd(8+n_Deltoid,:), Landmarks_Abd(9+n_Deltoid,:), Radius );
AllPoints_resting = Tmat_Abduction*[AllPoints,ones(size(AllPoints,1),1)]';
AllPoints_resting = AllPoints_resting(1:3,:)
;if (AllPoints(end-1,2) > Landmarks_Abd(5+n_Deltoid,2)) %change wrap direction
  if (size(AllPoints,1) > 2)
% change wrap direction
WrapDir = cross(AllPoints(end,:),AllPoints(end-2,:)-AllPoints(end-1,:));
CylDir = WrapDir(1);
SphereDir = WrapDir(3);
if ~Left
    SphereDir = -SphereDir;
end
%
Radius_FixWrap = Radius;
Fix = 0;
%
if CylDir < 0 && ((AllPoints(end-1,2) > HH(2)) && (AllPoints(end-2,2) > HH(2) && AllPoints(end-2,3) > HH(3)))
    Radius_FixWrap = Radius_FixWrap +1i;
    Fix = 1;
end
if Fix
    OldPoint = AllPoints(end-1,:);
    [AllPoints, Length_ObstacleSet]...
        = ObstacleSet(Landmarks_Abd(6+n_Deltoid,:), ScapWrap(1,:), ..., Landmarks_Abd(7+n_Deltoid,:),
        Landmarks_Abd(8+n_Deltoid,:), Landmarks_Abd(9+n_Deltoid,:), Radius_Fixwrap);
    AllPoints_resting =
        Tmat_Abduction\[AllPoints,ones(size(AllPoints,1),1)';
    AllPoints_resting =
        AllPoints_resting(1:3,:)';
    if (norm(AllPoints_resting(end,[1,3])-DI_resting([1,3])) < norm(AllPoints_resting(end-1,[1,3])=DI_resting([1,3]))) &&
        HH_resting(3) < AllPoints_resting(end-1,3) && HH_resting(2) < AllPoints_resting(end-1,2)
        Radius_FixWrap = -Radius_Fixwrap;
        [AllPoints, Length_ObstacleSet]...
            = ObstacleSet(Landmarks_Abd(6+n_Deltoid,:), ScapWrap(1,:), ..., Landmarks_Abd(7+n_Deltoid,:),
                Landmarks_Abd(8+n_Deltoid,:), Landmarks_Abd(9+n_Deltoid,:), Radius_FixWrap);
end
end
end
if (Length_Scap + Length_ObstacleSet) < L_tot
    L_tot = Length_Scap + Length_ObstacleSet;
    AllPoints_Scap_min = AllPoints_Scap;
    AllPoints_min = AllPoints;
    Length_ObstacleSet_min = Length_ObstacleSet;
L_Scap_min = Length_Scap;
ScapWrap_min = ScapWrap;
line_min = i_line;
end
end

AllPoints = AllPoints_min;
AllPoints_Scap = AllPoints_Scap_min;
Length_ObstacleSet = Length_ObstacleSet_min + L_Scap_min;

end
if line_min < hum_transition || size(AllPoints_Scap,1) > 2 % ScapPoint is on humerus
    ScapWrap = Landmarks_Abd(4+i_Deltoid,:);
end
end

% end

n_ScapPoints = size(ScapWrap,1);

n_ScapWrap(i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = n_ScapPoints;

Lengths(i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = Length_ObstacleSet;

AllPoints_Global = Tmat_Scap*(Tmat_ScapTilt\(Tmat_HumOffset\(Tmat_Abduction\([AllPoints,ones(size(AllPoints,1),1)]\)\)\)\)\)\)\);

ScapWrap_Global = Tmat_Scap*(Tmat_ScapTilt\(Tmat_HumOffset\(([ScapWrap,ones(size(ScapWrap,1),1)]\)\)\)\)\)\);
ScapWrap_Global = ScapWrap_Global(1:3,:);

AllWrapPoints{i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} = [AllPoints_Global(1:3,1:end-n_ScapPoints)'; ScapWrap_Global];
undo the humeral transformation to delt origin which was applied

% Tie Wrap Points to points on Humerus STL
IDX = knnsearch(Hum_v, AllWrapPoints{i_Abd_Angle, i_Deltoid, i_Lat_Offset, i_Inf_Offset, i_Hum_Offset, i_Movements}(1:end-n_ScapPoints,:));

AllWrapPoints{i_Abd_Angle, i_Deltoid, i_Lat_Offset, i_Inf_Offset, i_Hum_Offset, i_Movements}(1:end-n_ScapPoints,:) = Hum_v(IDX,:);

wrapPoint_Global = AllWrapPoints{i_Abd_Angle, i_Deltoid, i_Lat_Offset, i_Inf_Offset, i_Hum_Offset, i_Movements}(end-1,:);

% Last point of contact before deltoid origin
MomentPoint_Global = AllWrapPoints{i_Abd_Angle, i_Deltoid, i_Lat_Offset, i_Inf_Offset, i_Hum_Offset, i_Movements}(end-n_ScapPoints,:);

% Last contact on humerus
if n_ScapPoints == 2
    wrapPoint = (Tmat_ScapTilt*(Tmat_Scap\[wrapPoint_Global,1]'));
else
    wrapPoint = Tmat_Abduction*(Tmat_ScapTilt*(Tmat_Scap\[wrapPoint_Global,1]'));
end
wrapPoint = wrapPoint(1:3)';
MomentPoint = Tmat_Abduction*(Tmat_ScapTilt*(Tmat_Scap\[MomentPoint_Global,1]'));
    MomentPoint = MomentPoint(1:3)';

% Calculate moment arm
if i_Movements == 1
    Landmarks_Moment = Landmarks_Abd;
elseif i_Movements == 2
    Tmat_Elev = RotationMatrix(0,30,0); % rotate the coordinate system such that elevation is in the z plane
    Landmarks_Moment = Tmat_Elev*([Landmarks_Abd, ones(length(Landmarks_Abd),1)]');
    Landmarks_Moment = Landmarks_Moment(1:3,:);
    MomentPoint = Tmat_Elev*[MomentPoint, 1]';
    MomentPoint = MomentPoint(1:3)';
    ScapWrap = Tmat_Elev*[ScapWrap, ones(size(ScapWrap,1),1)]';
Scapwrap = Scapwrap(1:3)';

end

perp_dir = Landmarks_Moment(5+n_Deltoid,:) - MomentPoint;
%direction from COR to wrapping point perpendicular to moment arm
perp_dir(1) = 0; %since moment only in scapular plane
WrapLine = Scapwrap(1,:) - MomentPoint;

MomentScale(i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) =
norm(WrapLine);

WrapLine(1) = 0;

MomentScale(i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = ...

MomentScale(i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements)/nor
m(WrapLine);

WrapLine = WrapLine/norm(WrapLine);
moment = cross(WrapLine,perp_dir);

MomentArms(i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) =

end

%                         if i_Movements == 2 %Based on Brown
2007 Post Delt not active during abduction. Wrapping Points not calculated correctly

% MomentArms(i_Abd_Angle,5:7,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = 0;

r =

MomentArms(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements)./ ...

MomentScale(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements); %since
taking perpendicular component of total force

%prepare tensioning effects
Lengths_Current =

Lengths(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements);
Lengths_Current = reshape(Lengths_Current,size(F_max));
if i_Type == 1 && i_Abd_Angle == 1 && i_Movements == 2

Lengths_Resting = Lengths_Current;
end

Lengths_Ratio = Lengths_Current./Lengths_Resting;

ForceRatio_Active =

interp1(ActiveCurve(:,1),ActiveCurve(:,2),Lengths_Ratio,'pchip'); %Zajak 1989
ForceRatio_Passive =

interp1(PassiveCurve(:,1),PassiveCurve(:,2),Lengths_Ratio,'pchip');

Active_Extrap_Over = Lengths_Ratio > ActiveCurve(end,1);
Passive_Extrap_Over = Lengths_Ratio > PassiveCurve(end,1);
Active_Extrap_Under = Lengths_Ratio < ActiveCurve(1,1);
Passive_Extrap_Under = Lengths_Ratio < PassiveCurve(1,1);
ForceRatio_Active(Active_Extrap_Over) = 10^-5;
ForceRatio_Active(Active_Extrap_Over) = 10^-5;
ForceRatio_Passive(Passive_Extrap_Over) = fit_passive(Lengths_Ratio(Passive_Extrap_Over));
ForceRatio_Passive(Passive_Extrap_Under) = 0;

for i_Lengths = 1:n_Deltoid
  if Lengths_Ratio(i_Lengths) == 1
    ForceRatio_Active(i_Lengths) = 1;
    ForceRatio_Passive(i_Lengths) = 0;
  elseif Lengths_Ratio(i_Lengths) < 1
    ForceRatio_Passive(i_Lengths) = 0;
  end
end

ForcePassive = ForceRatio_Passive.*F_max; %Passive component of force due to stretch
ForceMax_Active = ForceRatio_Active.*F_max; %Maximum active component with full activation

Fn_guess = (ForceMax_Active./ 10) + ForcePassive;
UB = ForceMax_Active + ForcePassive;
LB = ForcePassive;
minimization = @(Fn)Activation_Optimization_Tension(Fn, ForceMax_Active, ForcePassive);

%Find the opposing moment due to gravity and load
arm_dir = (Landmarks_Moment(end,:)+ Landmarks_Moment(end-1,:))/2 - Landmarks_Moment(5+n_Deltoid,:);
arme_dir(1) = 0; %since Moment about X direction
arm_dir = arm_dir/norm(arm_dir);
down_dir = [0 -1 0];
arm_angle = acosd(dot(arm_dir,down_dir)); %angle with respect to down in CS

gravity_angle = arm_angle + (Rhythm_Angle(i_Abd_Angle));

Opp_Moment = ((Gravity_MomentArm + dot((Landmarks_Moment(6+n_Deltoid,:)- Landmarks_Moment(5+n_Deltoid,:)),arm_dir))* Gravity_Force ...
+ (Load_MomentArm + dot((Landmarks_Moment(6+n_Deltoid,:)- Landmarks_Moment(5+n_Deltoid,:)),arm_dir))*Load_Force) * sind(gravity_angle);
%compensating for arm lengthing
%Equalities satisfy moment balance equation
%Bounds make sure forces are above or equal to
% passive, below maximum for that length
if r*ForcePassive' > Opp_Moment

Forces(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = ForcePassive;
else

Forces(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = fmincon(minimization, Fn_guess,[],[],r,Opp_Moment, LB, UB, [], options);
end

Forces_Passive(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = ForcePassive;

Length_Ratios(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = Lengths_Ratio;

Passive_Ratios(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = ForceRatio_Passive;

Active_Ratios(i_Abd_Angle,:,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements) = ForceRatio_Active;

    for i_Deltoid = 1:n_Deltoid

ForceComponents{i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} = ...

wrapvecs{i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} * ... 

Forces(i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements);
end
end
end
end

% lap timer
Calculation_Time = toc;

Visualization

% Import STLS and wrapping info

% File_Scapula = 'SHOULDER_JOINT2 - SCAP-1 11-03057_SCAP-1.stl';
% File_Humerus = 'SHOULDER_JOINT_NAT_CORRECTv2 - 11-03057_HUM-1.stl';

if Visualize

    i_Lat_Offset = 1;
    i_Inf_Offset = 1;

i_Hum_Offset = 1;
i_Movements = 2;

for i_Abd_Angle = 1:9

figure

Tmat_Abduction = AbductionTmats{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements};
Tmat_HumOffset = eye(4);
Tmat_HumOffset(3,4) = -Hum_Offset(i_Hum_Offset); %apply to only Hum points
Tmat_HumGlobal = Tmat_Abduction*Tmat_HumOffset*Tmat_ScapGlobal;

[Scap_v,Scap_f]=stlread(File_Scapula);
[Scap_v,Scap_f]=patcheslim(Scap_v,Scap_f);

Scap_V = Tmat_ScapGlobal*([Scap_v,ones(length(Scap_v),1)]');
Scap_v = Scap_V(1:3,:');
ScapSTLPatch = patch('faces',Scap_f,'vertices',Scap_v,'FaceColor',[0 1 1],'FaceAlpha',... 
, .8,'EdgeAlpha', 0,'FaceLighting','gouraud', 'AmbientStrength', 0.15);
hold all

RedFactor = 1000;
reducepatch(ScapSTLPatch,RedFactor);
Hum_V = cell(n_hum,1);
HumSTLPatch = cell(n_hum,1);
for i = 1:n_hum
    Hum_V{i} = Tmat_HumGlobal*([Hum_v_part{i},ones(length(Hum_v_part{i}),1)]');
    Hum_V{i} = Hum_V{i}(1:3,:');
    HumSTLPatch{i} = patch('faces',Hum_f_part{i},'vertices',Hum_V{i},'FaceColor','g','FaceAlpha',... 
, .6,'EdgeAlpha', 0,'FaceLighting','gouraud', 'AmbientStrength', 0.15);
    reducepatch(HumSTLPatch{i},RedFactor);
end

Path = cell(n_Deltoid,1);
for i_Deltoid = 1:n_Deltoid
    Path_Global = [AllWrapPoints{i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements},...
    ones(size(AllWrapPoints{i_Abd_Angle,i_Deltoid,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements}),1,1)]';
    Path_Scap = Tmat_ScapGlobal*Path_Global;
    Path_Hum = Tmat_HumGlobal*Path_Global;
    if
n_ScapWrap(i_Abd_Angle, i_Deltoid, i_Lat_Offset, i_Inf_Offset, i_Hum_Offset, i_Movements) == 1
Path{i_Deltoid} = [Path_Hum(1:3,1:end-1),Path_Scap(1:3,end)]';
else
    Path{i_Deltoid} = [Path_Hum(1:3,1:end-2),Path_Scap(1:3,end-1:end)'];
end

plot3(Path{i_Deltoid}(:,1),Path{i_Deltoid}(:,2),Path{i_Deltoid}(:,3),'Marker','.', 'MarkerSize',15, 'Linewidth',1, 'Color','k');
end

axis('image')
view([190,-70])
camlight('headlight');
material('dull');
hold off
end

end

% lap timer
Visualization_Time = toc - Calculation_Time;

Add Forces to Abaqus file

if Abaqus && n_Movements==2
    AbaqusFile = fopen(AbaqusFileName, 'r');
    DirName = [AbaqusFileName(1:end-4), '_Incr_Increments_Flex'];
    StorePath = 'G:\Jason';
    mkdir(StorePath,DirName)
    %read current abaqus file to AbaqusData
    l = 1;
    tline = fgetl(AbaqusFile);
    AbaqusData{l} = tline;
    ChangeIndex = [];
    Delt_Numbers = [];
    Dir_Numbers = [];
    while ischar(tline)
        l = l+1;
        tline = fgetl(AbaqusFile);
        AbaqusData{l} = tline;
        if strcmp(tline,'*Cload')
            ChangeIndex = [ChangeIndex; l+1;l+2;l+3];
            Delt_Numbers = [Delt_Numbers; str2double(AbaqusData{l-1}(11));str2double(AbaqusData{l-1}(11));str2double(AbaqusData{l-1}(11));str2double(AbaqusData{l-1}(11))];
            Dir_Numbers = [Dir_Numbers ; 1; 2; 3];
        end
    end
    %check to see if correct line and write to a variable 'Line'
end
fclose(AbaqusFile);
AbaqusData{l-1} = '**';
AbaqusData{l} = '*EL Print, ELSET=A1, POSITION=INTEGRATION POINTS, FREQUENCY=1';
AbaqusData{l+1} = 'SP3';
AbaqusData{l+2} = '**';
AbaqusData{l+3} = '*EL Print, ELSET=A2, POSITION=INTEGRATION POINTS, FREQUENCY=1';
AbaqusData{l+4} = 'SP3';
AbaqusData{l+5} = '**';
AbaqusData{l+6} = '*EL Print, ELSET=A3, POSITION=INTEGRATION POINTS, FREQUENCY=1';
AbaqusData{l+7} = 'SP3';
AbaqusData{l+8} = '**';
AbaqusData{l+9} = '*End Step';
AbaqusData{l+10} = -1;

BatchName = sprintf('%s_%s_Batch.bat', AbaqusFileName(1:end-4),Load);
Batch = cell(n_Abd_Angle,n_Lat_Offset,n_Inf_Offset,n_Hum_Offset,n_Movements);
AbaqusFileName_New = cell(n_Abd_Angle,n_Lat_Offset,n_Inf_Offset,n_Hum_Offset,n_Movements);
AbaqusFile_New = cell(n_Abd_Angle,n_Lat_Offset,n_Inf_Offset,n_Hum_Offset,n_Movements);
MovementNames = { 'Abduction'; 'Elevation' };
for i_Abd_Angle = 1:n_Abd_Angle
    for i_Lat_Offset = 1:n_Lat_Offset
        for i_Inf_Offset = 1:n_Inf_Offset
            for i_Hum_Offset = 1:n_Hum_Offset
                for i_Movements = 1:n_Movements
                    AbaqusFileName_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} = ...
                    sprintf('%s_%s_%gdegrees%s_0x_%gy_%gzmmCupOffset_%gmmHumOffset.%s',
                        AbaqusFileName(1:end-4),Load,Abd_Angle(i_Abd_Angle),MovementNames{i_Movements},Inf_Offset(i_Inf_Offset),
                        Lat_Offset(i_Lat_Offset),Hum_Offset(i_Hum_Offset),AbaqusFileName(end-2:end));
                    Batch{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} = sprintf('abaqus job=%s interactive',
                        AbaqusFileName_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements}(1:end-4));
                    AbaqusFileName_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} = ...
                    fullfile(DirName,AbaqusFileName_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements});
                    AbaqusFile_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements} = ...
                    fopen(AbaqusFileName_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements}, 'w');
for l = 1:numel(AbaqusData)
    [Lia,Locb] = ismember(l,ChangeIndex);
    if AbaqusData{l+1} == -1
        for i_Abd_Angle = 1:n_Abd_Angle
            for i_Lat_Offset = 1:n_Lat_Offset
                for i_Inf_Offset = 1:n_Inf_Offset
                    for i_Hum_Offset = 1:n_Hum_Offset
                        for i_Movements = n_Movements
                            fprintf(AbaqusFile_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements},
                                    '%s',AbaqusData{l});
                        end
                    end
                end
            end
        end
    end
    elseif Lia
        str_end = strfind(AbaqusData{l}, ',',');
        for i_Abd_Angle = 1:n_Abd_Angle
            for i_Lat_Offset = 1:n_Lat_Offset
                for i_Inf_Offset = 1:n_Inf_Offset
                    for i_Hum_Offset = 1:n_Hum_Offset
                        for i_Movements = n_Movements
                            NewData = [AbaqusData{l}(1:str_end+4),num2str(ForceComponents{i_Abd_Angle,Delt_Numbers(Locb),...}
                                          i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements}(Dir_Numbers(Locb)))];
                            fprintf(AbaqusFile_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements},
                                    '%s\n',NewData);
                        end
                    end
                end
            end
        end
    else
        for i_Abd_Angle = 1:n_Abd_Angle
            for i_Lat_Offset = 1:n_Lat_Offset
                for i_Inf_Offset = 1:n_Inf_Offset
                    for i_Hum_Offset = 1:n_Hum_Offset
                        for i_Movements = n_Movements
                            fprintf(AbaqusFile_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements},
                                    '%s\n',AbaqusData{l});
for i_Abd_Angle = 1:n_Abd_Angle  
    for i_Lat_Offset  = 1:n_Lat_Offset  
        for i_Inf_Offset = 1:n_Inf_Offset  
            for i_Hum_Offset = 1:n_Hum_Offset  
                for i_Movements = n_Movements  
                    fclose(AbaqusFile_New{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements});  
                end  
            end  
        end  
    end  
end

BatchName = fullfile(DirName,BatchName);  
BatchFile = fopen(BatchName,'w');  
fprintf(BatchFile,'\n');  
for i_Abd_Angle = 1:n_Abd_Angle  
    for i_Lat_Offset  = 1:n_Lat_Offset  
        for i_Inf_Offset = 1:n_Inf_Offset  
            for i_Hum_Offset = 1:n_Hum_Offset  
                for i_Movements = n_Movements  
                    fprintf(BatchFile,'%s\n',Batch{i_Abd_Angle,i_Lat_Offset,i_Inf_Offset,i_Hum_Offset,i_Movements});  
                end  
            end  
        end  
    end  
end
fprintf(BatchFile,'\n');  
fclose(BatchFile);

if SaveResults  
    save(SaveName)  
end

% lap timer  
AbaqusOutput_Time = toc - Visualization_Time - Calculation_Time;  
Total_Time = toc;
end
end
profile viewer

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Appendix B: FE Boundary and Loading Condition Details

Figure B1 – FE boundary conditions

On the left are the deltoid forces applied to the geometric center of the proximal insertion sites, and tied to the entire insertion site, as well as the joint reaction force applied to the glenoid component of the implant. On the right are the boundary conditions applied to nodes below the glenoid fossa holding them rigidly in place. Note, the arrows used, and their directions are for representation purposes only and are not to scale.
Figure B1 highlights the loading conditions and boundary conditions of the finite element models that are described herein. Concentrated forces were applied by the deltoid to the acromion and scapular spine at the geometric centroid of each deltoid segment’s proximal insertion site. The geometric center of each insertion site was coupled to the entire insertion site to allow the force to act on the entire area. The glenoid component of the implant was tied to the surrounding bone with a surface tie, and the joint reaction force was applied to the center of the baseplate and the baseplate center was coupled to the face of the baseplate. The portion of the scapula inferior to the glenoid fossa was rigidly constrained, all nodes in this region were rigidly fixed so that they could not rotate or translate.
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