Musculoskeletal Model of the Glenohumeral Joint

Modeling the contact pressure in the Glenohumeral Joint

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**Resum**

L’espatlla humana és sovint afectada per osteoartròsi, el que pot ser provocat per diferents condicions al metabolisme, inflamació o sobrecàrrega mecànica repetida afectant a un 8.5% de la població dels Estats Units en qualsevol articulació. Degut al alt cost que això significa pel sistema sanitari, es capdals aprofundir en la comprensió i prevenció d’aquesta malaltia.

En aquest projecte s’empra les contribucions de la mecànica a aquesta patologia emprant un model numèric per l’articulació glenohumeral. L’objectiu últim d’aquest estudi és quantificar la pressió de contacte i la seva distribució als cartilàlegs sense patologies per el moviment d’abducció utilitzant el metode dels elements finits (FEM).

El model 3-d és construït a partir de ressonàncies magnètiques (MRI) d’una espatlla sana, incloent la scapula (homoplat), húmer i la clavícula amb els cartilàlegs de l’articulació.

S’analitzen dues forces fisiològiques a la unió glenohumeral. La primera entre el centre del cap húmeral i el centre de la superfície de la glenoide amb un mòdul variable en funció de l’angle girat. Per aquesta, el màxim de pressió a la superfície és de 7.3 MPa (paral·lelsisme amb estudis experimentals) a 90° d’abducció. L’altra es porta a terme emprant una força vectorial, la qual no té una direCCIó constant. En aquest cas, s’ha simulat els primer 50° d’abducció i després apareix una pèrdua de contacte a la unió (inestabilitat) degut a la excentricitat d’aquesta força. També, s’aplica un estudi de estabilitat per pròtesis per aquest model resultant més estable verticalment que horitzontalment.

Per concloure, ha sigut possible simular el moviment d’abducció amb el ritme scapulo-humeral, la translació a la articulació i la pressió de contacte. Tot i això, seria necessari implementar els músculs al nostre model per obtenir un model més proper per aquest pacient. Així també, caldria incloure el teixit labral del cartilage scapular, a fi de obtenir major superfície a la glenoide com succeeix a la realitat.
Abstract

The shoulder joint is often affected by osteoarthritis, which may be induced by metabolism, inflammation or repeated mechanical overloading affecting 8.5% of the US population in any joint. Due to the high cost for the health-care in the worldwide, it is necessary to go in depth in the comprehension and prevention of that disease.

The study uses the mechanical contributions to that pathology with a numerical contact model of the glenohumeral joint. The goal is to quantify the contact pressure and its distribution on the healthy cartilage layers for the abduction motion using the finite element method (FEM).

The 3-D model is built up from Magnetic Resonance Imaging (MRI) of a healthy shoulder, including the scapula, humerus and clavicle as well as articular cartilages.

We simulated the abduction for two physiological forces in the glenohumeral joint: the first between centers of the humeral head and glenoid cartilage with the result of maximum contact pressure of 7.3 MPa in 90° of abduction (parallelism with experimental data). Thus, for the same modulus of that force but with the direction variable (closer to reality), we simulated the first 50° of abduction and after this it appears a loosening of the humeral head to the glenoid cavity.

To conclude, we could simulate the abduction movement, the translation and the contact pressure in the glenohumeral joint. However, it was assumed that the muscles should be implement to the model to achieve the actual conditions as well as find a proper way to include the labrum in the scapular cartilage.
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1. Introduction

The shoulder is a complex group of 3 articular joints. The Glenohumeral (GH) Joint within the shoulder establishes the contact of humerus and scapula. Similar to a ball and socket joint the GH joint has 3 rotational degrees of freedom.

There are two cartilage layers: one on the humeral head and the other in the glenoid cavity. The small joint socket allows a wide range of motion, but requires an active stabilization through the rotator cuff muscles.

The GH joint is often affected by osteoarthritis (OA), a pathologic condition which is characterized by an irreversible process of cartilage degeneration which affects in the knee for example more than a third of population of over 65 years [Dawson et al., 2004]. It does not exist a medical procedure to regenerate cartilage cells. Symptom oriented treatment are methods for treating eventual pain. In the worst case, an Arthroplasty (Joint replacement with prosthesis) is required to recover a proper functionality. Surgery operations have increased last years and all of these have a socio-economical impact in the health-care system. As a consequence, the investigation in the prevention and regeneration of cartilage disease will play a significant role in early future.

There are multi-causes for osteoarthritis, these may be provoked by metabolism, trauma or repeated mechanical overloading. Excessive stress superior than 14 MPa [Démarteau et al., 2006] on joint tissues in repeated motions is a risk factor for osteoarthritis. This may be caused by either overloading mechanically the extremity or unappropriated mechanical functionality of the joint.

Studies containing Shoulder Arthroplasty has been developed numerically [Terrier et al., 2012]. However, to our knowledge it only exist a source of numerical model to GH joint developed by Buchler et al. (2002) where contact pressure in internal-external rotation of the GH joint was studied.

The lack of many experimental studies may be assumed for the difficulty to carry out in vivo measurements with non-invasive techniques. A experimental approach was developed from cadaverous in terms of contact pressure in abduction by Conzen and Eckstein (2000) and another for strain Soslowsky et al. (1992). It also exists a model for locating the center of the contact in abduction of Boyer et al. (2008).

The aim of the present study is to quantify the contact pressure generated in healthy cartilage layers of the GH joint for the abduction using a finite elements model. This approach might help in better understanding of the physiological loading conditions in the GH joint, which are relevant for cartilage’s health.
2. State of the Art

2.1. Anatomy

The following information was taken from the upper limb section of anatomy atlas of Hansen (2010) and Reinhard et al. (2008), and from the monograph of the shoulder of Rockwood and Matsen (2009).

2.1.1. Shoulder Joint

The human shoulder links the upper extremities to the trunk and consists in three bones: clavicle, scapula and humerus (fig. 2.2). The links of these bones appear in three different joints: sternoclavicular, acromioclavicular and glenohumeral joint. A wide range of motions are achieved by this complex structure of three joints.

It also exists the scapulothoracic joint which cannot be considered as a synovial joint because the contact takes place in the convex surface of the posterior thoracic cage and the concave surface of the anterior scapula (sub-scapular zone).

The sternoclavicular joint lets the contact of the clavicle to the thorax (sternum). The clavicle movements can be considered as a connecting rod. The acromioclavicular joint consists in the union between clavicle and scapula (in the acromion) and guarantees the motion of the scapula.

The GH joint is formed by the link of the humerus and scapula and might be simplified as system ball (Humeral head) and socket (Glenoid Cavity). Nevertheless, the morphology between humeral head and glenoid does not concord completely with that mechanism. The scapulo-humeral rhythm does not only span 3 rotational degrees of freedom but also translation between both parts. An illustration of the shoulder anatomy is shown in the figure 2.1.

![Figure 2.1.: Shoulder Joints](image)

Indeed, the glenoid fossa has a small size versus the humeral head, therefore the rotator cuff (explained in section 2.1.3) pulls the humerus against the fossa to maintain the stable
contact and avoiding the dislocation. Thus, the translation appears as a consequence of shape and that reaction force to stabilize the humeral head in the convenient position for the contact. In the following image (fig. 2.2) we can observe the difference of shape as well as understand all the different parts of bones.

![Shoulder Bones](image)

**Figure 2.2.: Shoulder Bones from Hansen (2010)**

This potential instability compared to other joints makes the GH joint be considered as the most complex. To sum up, it sacrifices some of its stability for greater range of motion.
2.1.2. Cartilages

**Morphology**

The articular cartilages have the task of guaranteeing the proper superficial contact in joints thanks to their mechanical properties (chapter 2.3.2). The GH joint contains 2 important cartilages: glenoid and humeral head cartilage.

The glenoid cartilage is attached to the scapula divided in two parts: fossa and labrum cartilage. The glenoid fossa cartilage is an uniform soft tissue (hyaline cartilage) with concavity form predominated. Its shape is defined by the shallow form of the glenoid cavity. The labrum is fibro-cartilaginous rim around the fossa cartilage. Its function is protecting the edges of the bone and fossa cartilage in the glenoid. The stability is improved in the joint thanks to the increase of the cavity. It could be compared in some ways to menisci in the knee. It also links the biceps tendon to the glenoid cartilage in the upper part.

The humeral head cartilage has also the shape according to the external surface of the bone which is close to a sphere. It encloses the humeral head until the anatomical neck of humerus. The material micro-structure of this cartilage is the same as the fossa. All this parts may be understood in the next image 2.3:

![Cartilage GH joint from Reinhard et al. (2008)](image)

Cartilages in the GH joint are surrounded by a capsule called synovial bursa. It is attached along the outside ring of the glenoid cavity and the anatomical neck of the humerus(figure 2.2). The presence of synovial fluid within that capsule reduces the friction between the articular cartilages in the joint. Another task of the synovial fluid is transporting both nutrients and waste within the cartilage because cartilages are an avascular tissues (lack of blood supply). The articular capsule consists of two layers: the outer fibrous membrane that contains ligaments (capsular ligament) and the inner synovial membrane that secretes the lubrication.
Figura 2.4.: Sinovial bursa in the GH joint

**Thickness**

The thickness of cartilage in the GH joint has been studied by many authors. The methods used to quantify the thickness of healthy cartilages are taken from MRI, ultrasounds imaging and cadaveric samples. It is observed that the humeral head cartilage is thinner than in the glenoid fossa and the thickness depends on the position in the joint. The following table 2.2 describes the mean of the thickness distribution for all those approaches and may be observed a high variability between samples (high Standard deviation-SD):

<table>
<thead>
<tr>
<th>Cartilage Publication</th>
<th>MRI</th>
<th>Ultrasound</th>
<th>Cadaverous</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Humerus</td>
<td>Glenoid</td>
<td>Humerus</td>
</tr>
<tr>
<td>Fox et al. (2008) [18*]</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Graichen et al. (2003) [8*]</td>
<td>1.2(0.09)</td>
<td>1.7(0.3)</td>
<td>1.4(0.12)</td>
</tr>
<tr>
<td>Yeh et al. (1998) [17*]</td>
<td>1.07(0.47)</td>
<td>2.02(0.71)</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 2.1.: Cartilage thickness - Mean(SD) - *Number of samples

Note that for Fox et al. (2008) in the table 2.1 only appears the humerus because this article is focused exhaustively on the thickness distribution around all the humeral head. It separates that in 9 regions distributed by anterior-posterior and inferior-superior. It assumes that the central sector was significantly thicker than all other sectors of the humeral head with the exception of the central-inferior (P=0.2).

**2.1.3. Muscles**

The muscles in the shoulder are responsible of movements. They are spanned by the muscles attached to the scapula, humerus and clavicle. The link between bones and muscles is done by tendons. Indeed, the muscles in the shoulder play a vital role stabilizing the joint as well as applying forces to achieve the proper motion.

The shoulder complex is spanned by 22 muscles. These muscles either connect the scapula and clavicle to the trunk, or link the clavicle, scapula and body wall to the proximal (head) end of the humerus or from ulna and radius to the scapula.

Concerning to the GH joint, muscles may be divided in two groups, rotator cuff and deltoid muscles.
Rotator Cuff Muscles

The rotator cuff (fig. 2.5) is made up of four muscles, which are the supraspinatus, infraspinatus, teres minor, and subscapularis. The rotator cuff has the main task of stabilizing the GH joint thanks to pull the humerus against the glenoid cavity. It contributes less importantly in applying forces for elevating (abduction) and rotating (internal-external rotation) the arm.

Every muscle, apart from working with the other rotator cuff muscles in the stability of the GH joint, has the task of:

Supraspinatus (supra-épineux):
The supraspinatus muscle originates above the spine of the scapula and inserts on the head of the humerus. The supraspinatus elevates the shoulder joint.

Infraspinatus (infra-épineux):
The infraspinatus muscle originates below the spine of the scapula, in the infraspinatus fossa, and inserts on the posterior aspect of the greater tuberosity (head) of the humerus. The infraspinatus externally rotates the shoulder joint.

Teres Minor (petit rond):
The teres minor muscle originates on the lateral scapula border and inserts on the inferior aspect of the greater tuberosity of the humerus. The teres minor muscle externally rotates the shoulder joint.

Subscapularis (subscapulaire):
The subscapularis muscle originates on the anterior surface of the scapula, sitting directly over the ribs, and inserts on the lesser tuberosity of the humerus. The subscapularis muscle
works to depress the head of the humerus allowing it to move freely in the glenohumeral joint during elevation of the arm.

**Deltoid Muscles**

The deltoid group is formed by anterior, medial and posterior deltoid (fig. 2.6). The deltoid is the prime mover of arm abduction along the frontal plane.

The deltoid has three origins: the lateral end of the clavicle, the acromion of the scapula at the top of the shoulder, and the spine of the scapula. Each origin gives rise to its own band of muscle fibers with the anterior band forming at the clavicle, the lateral fibers forming at the acromion, and the posterior fibers forming at the spine of the scapula. The bands merge together as they approach the insertion point on the deltoid tuberosity of the humerus.

![Figure 2.6: Deltoid Muscles](image-url)
2.2. Medical Imaging

2.2.1. Magnetic resonance imaging (MRI)

Magnetic resonance imaging (MRI) is a medical technique used in to investigate the anatomy and function of the body in both health and disease. MRI scanners use strong magnetic fields and radio-waves to form images of the body. The technique is widely used in hospitals for medical diagnosis, staging of disease and for follow-up without exposure to ionizing radiation which is considered unhealthy in large exposure. This technique envelopes many different fields such as: Neuroimaging, Cardiovascular, Musculoskeletal and oncology, among others. Since MRI does not use any ionizing radiation, it is recommended in preference to computed tomography (CT) imaging when either modality may yield the same information.

MRI is based in a process of detecting a radio frequency signal emitted by excited hydrogen atoms in the body (present in any tissue containing water molecules) using energy from an oscillating magnetic field applied at the appropriate resonant frequency, one per each tissue. The orientation of the image is controlled by varying the main magnetic field using gradient coils. This information was taken from the review of Hollingworth et al. (2000).

2.2.2. MRI in Cartilages

Despite the low thickness of cartilage, it is possible to visualize the cartilages directly and even analyze the morphometry if MRI images with high contrast and resolution are available [Eckstein et al., 2001].

In the case of the GH joint, compared to knee, the articular surfaces are highly curved (particularly the humeral head) and partial volume effects are therefore more severe. The validation of this way of procedure in the shoulder was made by Graichen et al. (2003) and the results show a severe difference of thickness between MRI and ultrasound method (table 2.2) with an error of 15.6% and 20.7% for humeral head and glenoid cavity respectively.

2.3. Material Properties

2.3.1. Bones

The bone is a material with a non-homogeneous distribution which provides to the body its structural form. Flat bones as the scapula are comprised of thick layers of dense bone which are connected by a porous network of bony spicules called cancellous (also known as trabecular or spongy bone).

Long bones (humerus, femur or clavicle, among others) can be divided in terms of form and mechanical properties in two tissues, cancellous and cortical bone. Cortical bone is located in the tubular part (Diaphysis) and it has anisotropic behavior due to the fibrous distribution of their cells. However, in the extreme of the bone (Epiphysis), the structure of the cells in the cancellous zone lets the bone complex shapes as well as isotropic and matrix properties in this region (image 2.7).
Indeed, Kabel et al. (1999) validated the material property of bone described as isotropic linear elastic material. However, they found a quadratic correlation between the young’s modulus ($E$) and the density ($\rho$). In fact, linear elastic moduli (non-homogeneous) of the shoulder bones was used with the equation 2.1 in the research of Buchler et al. (2002):

$$E_b(\rho) = E_0 \left( \frac{\rho}{\rho_0} \right)^2$$  \hspace{0.5cm} (2.1)

$E_0 = 15000$ MPa,
$\rho_0 = 1.8$ g/cm$^3$,
$\rho = \text{[g/cm}^3\text{]}$ Bone density per region (Node),
$\nu_0 = \nu = 0.3$ \hspace{0.5cm} (Poisson’s Ratio)

As it has been explained, in the cancellous zone the distribution of density is constant [Shukla et al., 1987]. Thus, the elastic moduli might be assumed constant.

### 2.3.2. Cartilages

#### Material Properties

The hyaline cartilage is soft matrix anisotropic material. The mechanical properties of articular cartilages are dependent on the chemical composition and architectural arrangement. The material properties of the cartilages achieve an almost frictionless contact and a rubbery behavior. The friction coefficient ($\mu$) in the GH joint is 0.003 (negligible) according to Poitout (2004). This frictionless condition is reached in a fluid environment because composition of cartilage is primarily of water (70-80% of weight). The solid fraction of the tissue is spanned by collagens (50-75%) and proteoglycians (15-30%) depending on the deep in the tissue (figure 2.8). All these is balanced with minor protein molecules and chondrocytes [Athanasiou and Darling, 2010].
Figure 2.8.: Cartilage structure varies depending on the zone. Image from Athanasiou and Darling (2010)

**Biphasic Theory**

Mechanical properties of articular cartilage have been studied, and accurate models were developed for that porous permeable material. The biphasic theory considers the liquid flow under the influence of pressure gradients in the solid state related to the time (min). The interaction between these two phases, fluid and solid determines the overall deformation behavior of the tissue which is close to real conditions of cartilage. The most important approach of the theory was carried out by Mow et al. (1980). Besides, Wu et al. (1998) used the biphasic theory through the finite elements assuming its complexity.

**Cartilage. Neo-hookean hyperelastic tissue**

In the numerical model of Buchler et al. (2002), it is assumed the mechanical behavior of cartilage as Neo-hookean hyperelastic. This means that for the same stress, it is obtained a higher deformation compared to a linear elastic model and this effect increase proportionally with the stress.

The Neo-hookean hyperelastic law for human cartilages was studied by Kempson (1979) and it follows this equation 2.2:

\[
P_{zz} = 2 \cdot C_{10} \left( \lambda - \frac{1}{\lambda^2} \right) \tag{2.2}
\]

\( P_{zz} \) = stress [MPa]
\( \lambda \) = stretch [ ]
\( E_c \) = 10 MPa (Young’s Moduli)
\( \nu \) = 0.4 [ ] (Poisson’s Ratio)

The parameters \( C_{10} = 1.79 \) MPa and \( D_1 = 0.12 \) MPa\(^{-1} \) are calculated with the following equations (2.3 and 2.4):
Experimental data. Stress-strain in the GH joint

A experimental data of cartilages in the GH joint was done in the Laboratory of Biomechanical Orthopedics (LBO) in order to analyze the most proper law for stress-strain. In that source, Fuentes and Terrier (2005) studied linear and hyper-elastic methods (Neo-hookean, Mooney-Rivlin and exponential) versus an experimental data from two cadaveric GH cartilages. For this study, the hypothesis of incompressibility was assumed.

The output of this study shows both Mooney-Rivlin and exponential hyper-elastic as the most suitable to the experimental behavior.

In the figure 2.9, we can observe the precision of the curves of the exponential hyper-elastic against the experimental test:

![Figure 2.9.](image)

The equation 2.5 is the case of exponential hyper-elastic law and its adjusted parameters to the experimental curves:

\[
P_{zz} = 2 \cdot \left[ \left( \lambda - \frac{1}{\lambda^3} \right) \cdot \alpha_e \cdot \beta_e \cdot e^{\beta_e (\frac{2}{5} \lambda^2 - 3)} - \left( \lambda - \frac{1}{\lambda^3} \right) \cdot \frac{\alpha_e \cdot \beta_e^2}{2} \right]
\]

\( P_{zz} = \text{Stress}[\text{MPa}] \)
\( \lambda = \text{stretch}[\text{]}] \)
\( \alpha_e = 1.1485 \text{ MPa}, \beta_e = 1.7333 \text{ [ ]} \)
In the conclusions of this study, it is outlined that hyper-elastic exponential, homogeneous, isotropic and incompressible laws are sufficient for the quasi-static analyses of GH cartilage under relatively rapid compression rates. Despite the precision of the curves, it is necessary to carry out further experiments because it has made with only two samples. It is also known, that the fluid flow effects are relevant for time characteristics in the order of minutes but this effect is minimized with rapid compression rates (0.1 Hz).
2.4. Pathology

2.4.1. Subchondral Bone Growth

Subchondral growth is the process of calcification of the internal layer of cartilages attached to bone. That slow subchondral activity front throughout adulthood allows for modulations of cartilage thickness and bone architecture (external shape).

Articular cartilage in regions where do not suffer high contact pressure do not experience compressive stress. As a consequence, the subchondral growth is permitted. Cartilage destruction and ossification proceed slowly with increasing age in these unloaded areas, information extracted from Carter and Beaupré (2001).

2.4.2. Osteoarthritis (OA)

Arthritis means literally a state of joint inflammation. However it is generally used for a broad category of pathologies in the joint which involve damaged of cartilages, loss of function, change in soft tissues near the joint and pain. Osteoarthritis is the most common form of arthritis. The consensus definition for OA is according to Moskowitz et al. (2007): "OA disease are a result of both mechanical and biological events that destabilize the normal coupling or degradation and synthesis of articular cartilage chondrocytes and extracellular matrix, and subchondral bone."

The causes of OA has been categorized in primary and secondary osteoarthritis depending on the cause. In OA secondary is included the degenerative Joint Disease (DJD). In fact, osteoarthritis is often called arthrosis or osteoarthrosis in order to distinguish from the primarily inflammation (suffix -osis means degeneration) [Carter and Beaupré, 2001].

Multi-causes lead to OA (e.g. alkaptonuria, congenital disorders of joints, diabetes, hemochromatosis, etc.). Damage from excessive mechanical stress with insufficient self repair by joints is believed to be the primary cause of OA [Brandt et al., 2009]. Concretely, contact pressure over 14 MPa may lead to damage in cartilage according to the in vitro mechanical study of Démarteau et al. (2006).

Besides, the body’s innate process for repairing the damaged tissues cannot be effective in the face of the overwhelming mechanical abnormality. In fact, elite athletes are prone to incur in a early GH arthritis due to high demanding activity and overuse of the joint [John et al., 2008].

Diagnosis is made with reasonable certainty based on history and clinical examination by X-ray images. The main symptom is pain, causing loss of ability and often bone stiffness[Rockwood and Matsen, 2011]. The joint loses the low friction which is needed for the relative rotation between bones. In the figure 2.10, we can compare the cartilage state between healthy and pathological cartilage.

Medical procedure to regenerate cartilage cells is being investigate [Moghadam et al., 2014]. Within the conservatives methods, there are lifestyle modification, moderate exercise and medication. Medicines are methods for relieving the pain (Symptom oriented treatment).

If disability is significant and more conservative management is ineffective, an Arthroplasty (Joint replacement with prosthesis) is required to recover a proper functionality.
Figure 2.10.: Left: Healthy humeral cartilage. Right: Osteoarthritic cartilage. [Mat- sen III, 2011]

Arthroplasty is an effective in the shoulder for short-term, its survival rates free of revision by glenoid implant type at 5, 10, and 15 years were, respectively, 96%, 96%, and 95% [Fox et al., 2009].

2.4.3. Instability

Chronic instability usually manifests itself as recurrent amount of subluxation which follows an initial dislocation. This dislocation occurs when the humeral head has lost the position inside the glenoid cavity as a consequence of a reaction force between them too eccentric showed in the figure 2.11. Since reaction force in the joint is applied by the muscles, they play an important role in the active stabilization.

There are three types of dislocation: anterior, posterior and inferior. Over 95% of shoulder dislocation cases are anterior. Most anterior dislocations are sub-coracoid, sub-glenoid and subclavicular.

Although dislocation are not a frequent disease (0.5-1.7%), the probability of suffer OA is higher after dislocations according to Robert and Robert (2005). Another publication by Ogawa et al. (2006) presents 31.2% arthritis evidence of 88 dislocated shoulders.

Figure 2.11.: Anterior dislocation
2.5. Shoulder Kinematics

2.5.1. Definition of local Coordinate System

As any mechanism, shoulder needs a reference to define the movement (joint coordinate system, JCS). A standard coordinate system for the upper limb was defined by International Society of Biomechanics (ISB) which is explained in the publication of Wu et al. (2005). This symposium were made joining concepts of reference system of older publications [van der Helm (1996)].

Firstly, all the referenced points of each bone where defined as is shown in the figure 2.12. From these points it can be created the coordinate system and its origins.

![Figure 2.12.: Points of reference upper extremity](image)

Regarding to the three bones in the shoulder, these are the important points, axis and origins:

**Thorax**

Points of reference for the thorax (fig. 2.12)

- ProcessusSpinokus (C7): Spinal process of the 7th cervical vertebra.
- ProcessusSpinokus (T8): Spinal process of the 8th thoracic vertebra.
- ProcessusSpinokus (IJ) Deepest point of Incisura Jugularis (suprasternal notch).
- ProcessusXiphoideus (PX): Most caudal point on the sternum

Thorax coordinate system — $y_t,z_t,x_t$(fig.2.13)

- $O_t$: The origin coincident with IJ.
- $y_t$: The line connecting the midpoint between PX and T8 and the midpoint between IJ and C7, pointing upward.
- $z_t$: The line perpendicular to the plane formed by IJ, C7, and the midpoint between PX
and T8, pointing to the right.

\( x_t \): The common line perpendicular to the \( z_t \)- and \( y_t \)-axis, pointing forwards.

**Clavicle**

Points of reference for the clavicle (fig. 2.12)

*Sternoclavicular point* (SC): Most ventral point on the sternoclavicular joint.

*Acromioclavicular point* (AC): Most dorsal point on the acromioclavicular joint (shared with the scapula).

Clavicle coordinate system — \( x_cy_cz_c \) (fig. 2.13)

\( O_c \): The origin coincident with SC.

\( z_c \): The line connecting SC and AC, pointing to AC.

\( x_c \): The line perpendicular to \( z_c \) and \( y_t \), pointing forward. Note that the \( x_c \)-axis is defined with respect to the vertical axis of the thorax (\( y_t \)-axis) because only two bony landmarks can be discerned at the clavicle.

\( y_c \): The common line perpendicular to the \( x_c \) and \( z_c \)-axis, pointing upward.

![Figure 2.13.: Coordinate System for the clavicle and thorax](image)

**Scapula**

Points of reference for the scapula (fig. 2.12)

*Trigonum Spinae Scapulae* (TS): The midpoint of the triangular surface on the media border of the scapula in line with the scapular spine.

*Angulus Inferior* (AI): the lowest (most caudal) point of the scapula.

*Angulus Acromialis* (AA): most laterodorsal point of the acromial angle.

Scapula coordinate system — \( x_sy_sz_s \) (fig. 2.14)

\( O_s \): The origin coincident with AA.

\( z_s \): The line connecting TS and AA, pointing to AA.

\( x_s \): The line perpendicular to the plane formed by AI, AA, and TS, pointing forward. Note that because of the use of AA instead of AC, this plane is not the same as the visual plane of the scapula bone.

\( y_s \): The common line perpendicular to the \( x_s \) and \( z_s \)-axis, pointing upward.
2.5.2. Definition of Movements using ISB Coordinate System

Many rotations may be defined in the shoulder thanks to the ISB coordinate system described in the section 2.5.1 using Euler angles. It the publication of Wu et al. (1998) can be report two types of rotations system: joint and segment(bone) rotational reference.

In this text it will be described the motion per segment(bone) which is the rotation of the clavicle, scapula, or humerus relative to the thorax. The definitions of the rotational axis using the Euler rotation sequence are: \( \alpha \) is around the Z-axis, \( \beta \) around the X-axis, and \( \gamma \) around the Y-axis, irrespective of the order of rotation. The joint displacements
should be defined respect to the proximal segment (in the case of bone reference system is not helpful because all is referenced from the thorax, there is no relative moment between bones defined).

**JCS and motion for the clavicle relative to the thorax (fig.2.13):**

- **$e_1$:** The axis fixed to the thorax and coincident with the $y_t$-axis of the thorax coordinate system. Rotation ($\gamma_c$): retraction (negative) or protraction (positive).
- **$e_3$:** The axis fixed to the clavicle and coincident with the $z_c$-axis of the clavicle coordinate system. Rotation ($\alpha_c$): axial rotation of the clavicle; rotation of the top backwards is positive, forwards is negative.
- **$e_2$:** The common axis perpendicular to $e_1$ and $e_3$, the rotated $x_c$-axis. Rotation ($\beta_c$): Elevation (negative) or depression (positive).

Notice that the angles for the joint and segment system are the same since the proximal coordinate system of the clavicle is the thorax ($\alpha_s = \alpha_{sc}, \beta_s = \beta_{sc}, \gamma_s = \gamma_{sc}$).

**JCS and motion for the scapula relative to the thorax (Y–X–Z order - figure 2.14):**

- **$e_1$:** The axis fixed to the thorax and coincident with the $y_t$-axis of the thorax coordinate system. Rotation ($\gamma_s$): Retraction (negative) or protraction (positive).
- **$e_3$:** The axis fixed to the scapula and coincident with the $z_s$-axis of the scapular coordinate system. Rotation ($\alpha_s$): anterior (negative) or posterior (positive) tilt.
- **$e_2$:** The common axis perpendicular to $e_1$ and $e_3$. Rotation ($\beta_s$): lateral (negative) or medial (positive) rotation.

**JCS and motion for the humerus relative to the thorax (figure 2.16):**

- **$e_1$:** The axis fixed to the thorax and coincident with the $y_t$-axis of the thorax coordinate system. Rotation ($\alpha_h$): Depending on the plane of elevation, $0^\circ$ is abduction, $90^\circ$ is forward flexion (positive) and extension (negative).
- **$e_3$:** Axial rotation around the $y_h$-axis. Rotation ($\gamma_h$): axial rotation, (endo-)internal-rotation (positive) and (exo-)external-rotation (negative).
- **$e_2$:** The axis fixed to the humerus and coincident with the $x_h$-axis of the humerus coordinate system. Rotation ($\beta_h$): abduction (negative) and adduction (positive).

![Figure 2.16.: JCS and humerus rotation relative to thorax](image)

This definitions of this Euler angles has remained as close as possible to the clinical definitions of segment motion. The decompose the three independent variables (three angles) let determine the rotation followed by the coordinate system (equation 2.6) per bone as well as define medically each rotational direction [van der Helm, 1996]. Euler angles can be
interpreted as subsequent rotations around axes of the (local or global) coordinate system for rotations $\alpha$, $\beta$ and $\gamma$ about the $x$-, $y$- and $z$-axis respectively. Successive rotations $a$, $b$ and $g$ result in:

$$R_i = R_x(\alpha_i) \cdot R_y(\beta_i) \cdot R_z(\gamma_i)$$  \hspace{1cm} (2.6)

$$R_x(\alpha_i) = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\alpha_i) & -\sin(\alpha_i) \\ 0 & \sin(\alpha_i) & \cos(\alpha_i) \end{bmatrix}$$

$$R_y(\beta_i) = \begin{bmatrix} \cos(\beta_i) & 0 & \sin(\beta_i) \\ 0 & 1 & 0 \\ -\sin(\beta_i) & 0 & \cos(\beta_i) \end{bmatrix}$$

$$R_z(\gamma_i) = \begin{bmatrix} \cos(\gamma_i) & -\sin(\gamma_i) & 0 \\ \sin(\gamma_i) & \cos(\gamma_i) & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

$i \in c, s, h$ (clavicle, scapula and humerus)

Note that the result (equation 2.6) of this composition of movements is referenced in each bone.

### 2.5.3. Abduction Movement

The abduction movement is the lateral motion away from the torso (Sagittal plane) of the body; moving the upper arm up to the side away from the body. This motion actually can be divided into three motions: the humerus raising in the back plane, upward rotation of the scapula and, less extent, the clavicle [Rockwood and Matsen, 2009]. The combination of movements between the scapula and humerus is well-known as scapulohumeral rhythm, the abduction of the humerus is twice faster than the scapula as it is shown in the figure 2.17 In the abduction the humerus rise up near to the scapular plan (Plane $x_s = 0$).

![Figure 2.17](image)

Figure 2.17.: Scapulo humeral rhythm from McClure et al. (2001)

Many in-vivo studies have been developed after the publication of Poppen and Walker (1976) which evaluate the abduction angles through regression equations in radiographies(2-d). Later on, van der Helm and Pronk (1995) presented a measurement technique based
on palpating and recording positions of bony landmarks with a 3-electromagnetic digitizer.

Finally, McClure et al. (2001) carried out an extensive experimental data of the abduction, among other motions. The extreme Euler angles for that publication may be observed in the table 2.2:

<table>
<thead>
<tr>
<th></th>
<th>( \alpha_i )</th>
<th>( \beta_i )</th>
<th>( \gamma_i )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Clavicle (c)</td>
<td>30°</td>
<td>50°</td>
<td>-24°</td>
</tr>
<tr>
<td>Scapula (s)</td>
<td>0°</td>
<td>10°</td>
<td>-21°</td>
</tr>
<tr>
<td>Humerus (h)</td>
<td>0°</td>
<td>147°</td>
<td>0°</td>
</tr>
</tbody>
</table>

Table 2.2.: Euler extreme values for abduction from McClure et al. (2001)

2.6. Numerical Models

2.6.1. Kinematics of the Shoulder

A significant step in the biomechanical and mathematical study of the shoulder was made by van der Helm (1994a). Finite elements analysis was made theoretically to estimate the loads in the GH joint which nowadays is still used in many researches. Another different study of van der Helm (1994b), presented a FEA of a very complete with bones, muscles and joints which let simulate the kinematics of the shoulder.

As it has been explained that Wu et al. (2005) not only published the coordinate system for the upper limb but also defined the Euler angles all the different rotations in the shoulder (explained in the chapter 2.5.2).

The effect of translation in the GH joint was studied a numerical algorithm by Terrier et al. (2008) and also validating with an algebraic known.

2.6.2. Articular Contact in the GH joint

It also exists a work of Buchler et al. (2002) who compares the contact loads between a normal and osteoarthritic GH joint, the shoulder model were studied from a tomography of two cadaverous models. The objective was not only to assess the stability of the joint but also the external and internal rotation of the humerus for each case. The model of FEA had been working properly to study the proper thickness in terms of pressure in a glenoid prosthesis by Terrier et al. (2012).

2.6.3. Muscles Model (EMG base-method)

Many researches in the field of total Arthroplasty has been developed with the purpose of studying the reaction forces in shoulder girdle(muscles and joint). Engelhardt et al. (2014) carried out a comparison between EMG-based (Electromyography) and stress-based method for the abduction movement with a mass of the arm of 3.75 kg. These two methods are efficient to solve the equilibrium of moments:
\[ \sum M = \sum r_m \cdot f_m + \sum r_e \cdot f_e = 0 \] (2.7)

where \( r_m \) is the muscle moment arm, \( f_m \) the muscle force, \( f_e \) an external force and \( r_e \) the corresponding moment arm. There, the arm weight was the only external force. The summation was done over his model of 10 muscle units: the subscapularis, supraspinatus, infraspinatus combined with teres minor, middle deltoid, anterior deltoid and posterior deltoid. Since many muscles forces under-determinate the model, it is need to use either EMG-base or Stress-base methods for solving the indeterminacy of the equation 2.7.

Electromyography (EMG) is a technique for evaluating and recording the electrical activity produced by skeletal muscles. An electromyograph detects the electric potential generated by muscle cells, when these cells are electrically or neurologically activated.

The EMG values of muscle activation \( \bar{a}_m \) were taken for each 30° of abduction. To get muscle force \( F_m \), a Hill muscles model was implemented into the active muscle parts:

\[ F_m = k \bar{a}_m f_m(l_m)PCSA_m \] (2.8)

where \( f_m \) the isometric relationship of tension–length , \( k \) is the Fick constant, and PCSA\(_m\) the physiological cross-sectional area of the muscle. The activation \( \bar{a}_m \) can vary between 0 (null activation) and 1 (maximum). The experimentally measured muscle activities do not lead to muscles forces that fulfill the mechanical equilibrium (equation 2.7). A new variable of muscles activation \( a_m \) is created in order to reach the equilibrium of moments whereas remaining as close as possible to experimental measurements \( \bar{a}_m \). Thus, the following minimization algorithm (2.9) was implemented to get those aims.

\[ \min[G(a_m)] = \min \left[ \sum_{m=1}^{M} (a_m - \bar{a}_m)^2 \right] \] (2.9)

This minimization was constraint mechanical equilibrium in equation 2.7 and positive muscle activation, \( a_m > 0 \).
3. Methods

3.1. Anatomy Reconstruction

3.1.1. MRI protocol

Magnetic resonance imaging (MRI) is a medical technique as has been explained in the chapter 2.2.1. Amira 5.4.5® (Visage Imaging GmbH, 92130 San Diego, United States of America) is a multifaceted 3D software platform for visualizing, manipulating, and understanding biomedical data coming from all types of sources and modalities. Initially known and widely used as the 3D visualization tool of choice in microscopy and biomedical research, Amira has become a more and more sophisticated, delivering powerful analysis capabilities in all visualization and simulation fields in bioengineering.

3.1.2. Bones

The anatomy was reconstructed from MRI images taken from 3-T scan of a volunteer (27 years old, height 180 cm and weight 75 kg) and using the software Amira. Amira lets reconstruct volumes using many different slices normal the three Cartesian Axis. For this model, there are 288 parallel images to the top view (transverse plane, z=0) and lateral view (Sagittal plane, x=0), and 224 parallel images to the front view (coronal plane, y=0).

The procedure followed is to fill all the slices where may be recognized the contour of the bones (figure 3.1). The humerus and scapula are the main bones which take part in the model. Thus, the clavicle is represented in order to make more intuitive the model and its rhythm. The anatomic distinction of bones was helped with an atlas of sectional anatomy made by Moeller and Reif (2009).

![Amira Reconstruction. Red: Humerus, Green: Scapula and Blue Clavicle](image-url)
The volume is built from voxels whose form is cubic. Voxels are basically pixels extruded from the filled contours in the images. This cubic form does not generate smoothed surfaces as may be seen in the figure 3.1(Right Bottom).

### 3.1.3. Post-processing

The reconstruction through voxels creates tiered and rough surfaces. Thus, it exists the need to smooth the surfaces in order to get the surface closer to reality. Furthermore, the modeling and finite element programs should be capable to compute those volumes and surfaces. In order to achieve the exact surface, the solids smoothing tools in Amira and Geomagic 13® (Geomagic GmbH, 27560 Morrisville, United States of America) software are used getting the surfaces of the bones (fig.3.2). This tools smooth the external surfaces thanks to interpolation of the tiered shape. In the end, these closed external surfaces generated are converted to volumes.

![Image](image.png)

Figure 3.2.: Post-processing. Left: Amira Model, Center: Amira Smoothing, Right: Geomagic Computed Volume.

### 3.1.4. Cartilages

**Anatomical Reconstruction**

The cartilages may be reconstructed following the analogous procedure for bones. A MRI images with a high contrast between bones and cartilages were chosen, different images of bones. In this case, we paid special attention to the contact between cartilage layers as may be observed in the figure 3.3.

![Image](image.png)

Figure 3.3.: MRI slice of Cartilage. Orange: Humeral head c. and green: glenoid c.
As may be seen (fig. 3.3), the glenoid cartilage was reconstructed with the labrum as well as the humeral head cartilage is thinner than in the glenoid fossa (consistent with publication in the chapter 2.1.2). In the following images, we exhibit the reconstruction of cartilages and it can be noticed the rough shape of both cartilages, glenoid and humeral head cartilage (fig. 3.4).

![Figure 3.4.: Reconstruction of cartilage. Left and right: Humeral head and Glenoid cartilage](image)

To sum up, the reconstruction of cartilages is not as accurate as in the case of bones due to the relation between the resolution of images and thickness of GH joint cartilages (1-2 mm described in the chapter 2.1.2). That precision makes the mean error in general of 0.25 mm, as the size of cartilage is quite smaller the relative error will be bigger.

**Smoothed Model**

After the situation described in the last chapter, we try to find a way to smooth the surface. In the early beginning, we dismiss the geomagic procedure because it may be too aggressive in terms of smoothing and we could lose a critic volume in the outer parts because of the interpolation in that areas.

As it has been explained in the chapter 2.1, the shape of the internal cartilage depends on the subchondral part of the bone (external surface) where it is attached. Thus, we decide to built up cartilages from those surfaces. Solidworks® is used to adding a thickness to that external surface. We look at publications of Graichen et al. (2003) and start creating those cartilages with constant thickness of: 1.2 mm for the humeral head and 1.7 mm for glenoid cartilage (without labrum). In the next images(3.5) are showed the shape of this cartilages:

![Figure 3.5.: Smoothed Cartilage. Left and right: Humeral head and Glenoid Cartilage](image)
3.1.5. Optimization of Cartilages Thickness

A comparison of both methods (Reconstructed and smoothed cartilages) is made in order to quantify and minimize difference of thickness between models. We carry out in this chapter an analysis of the thickness from making a boolean operation which consists in removing the volumed shared between each cartilage.

After optimizing the thickness for both cartilage, the humeral head cartilage remains at the same value (1.2 mm). Nevertheless, for the case of the glenoid we set the value of thickness from 1.7 mm to 2 mm in order to minimize the error of thickness. In fact that last amounts of thickness are closer to the publication of Yeh et al. (1998) for cadaverous.

In the next image (fig. 3.6), it is showed that analysis for the **glenoid cartilage**. The error or difference of thickness is lower than 0.35 mm without taking into account the labrum.

![Figure 3.6.: Difference of thickness between MRI and built up Glenoid Cartilage thickness](image)

For the case of the **humeral head cartilage**, the same process is done. Nonetheless, the thickness and shape remaining after the removing the shared volume is too complex to be computed by Solidworks (see image 3.7 left). In this case, the analysis of thickness should be made comparing the MRI model to the constant value of 1.2 mm as is showed in figure 3.7. The central area is mostly near to that thickness. We may also observe the high irregularity of the MRI reconstruction explained before (chapter 3.1.4).

![Figure 3.7.: MRI Humeral head Cartilage thickness compared to 1.2mm](image)
3.1.6. Model of the shoulder

In order to complete the skeletal appearance of the shoulder, the thorax (Sternum, ribs and column) was reconstructed following the same proceeding. All the bones and cartilages were assembled to ensure that everything is in the right position, so we show in the following image (fig. 3.8) all the anatomical model.

Figure 3.8.: Skeletal model of the Shoulder. Cartilage: Humeral head and glenoid. Bones: Thorax, humerus and clavicle

3.2. Coordinate System

To achieve the last objective of generating the movement of abduction, it is necessary to define a clear reference axis (csys). We follow the ISB recommendation for coordinate system explained by Wu et al. (2005) (chapter 2.5.1). It is important to notice that the points were located in the MRI images which are close to reality.

The csys of the clavicle depends on the thorax, so it is another reason to have reconstructed the thorax. For the case of the points in the sternoclavicular (SC) and acromioclavicular (AC) joint are placed in zones closer to clavicle with the origin in AA. For the scapula: the AI, TS and AA are selected as the protocol establishes with creating 3 points for the TS and calculating the center of the triangle of those point. The origin of the csys is located in AA.

Finally, the center of the humerus is found creating an adjusted sphere from several landmarks in the humeral head surface and calculating its center with the least squared method (regression of Meskers et al. (1997)). The MRI images do not contain the elbow, so we used MRI images for the same patience with elbow. We obtained the EL and EM points from it and $x_h$-axis is defined according to the publication. For the $y_h$-axis, we decided to choose the direction between humeral head center and the Diaphysis (tubular) humerus center. The coordinates system for our model is shown in the following figure (fig. 3.9):
3.3. Finite Elements Analyze (FEA)

3.3.1. Mathematic-Mechanical Concept

In mathematics, the finite element method (FEM) is a numerical technique for finding approximate solutions to boundary value problems of differential equations. Analogous to the idea that connecting many tiny straight lines can approximate a larger circle, FEM encompasses all the methods for connecting many element equations over many small sub-domains (nodes), named finite elements, to approximate a more complex equations over a larger domain.

FEM is best understood from its practical application, known as finite element analysis (FEA). FEA as applied in engineering is a computational tool for performing engineering analysis, in the field of thermodynamics, fluids, electromagnetism and mechanics.

For structural analysis, FEM uses virtual work principle approach which is applicable to both linear and non-linear material behaviors.

The principle of virtual displacements for the structural system expresses the mathematical identity of virtual work $W^*$ produced by external and internal force:

$$ W^* = \int_V \delta \epsilon^T \sigma dV \quad (3.1) $$

The virtual internal work $F$ in of the above equation (3.2) may be found by summing the virtual work in the all individual elements. In the case of linear-elastic behavior we can obtain any variable solving the matrix linear equation 3.2).

$$ E = Kq + F \quad (3.2) $$
\( E = \) Sector of nodal forces, representing external forces.
\( K = \) System stiffness matrix
\( q = \) Vector of system’s nodal displacements
\( F = \) Vector of equivalent nodal forces

This equations establish a relation between stress and strain. The elements are interconnected only at the exterior nodes, and altogether they should cover the entire domain as accurately as possible. Nodes will have nodal vectors which include displacement and rotations. In our case, this material properties are hyper-elastic which means a much more complex equation of equilibrium.

### 3.3.2. Computing FEA

Abaqus FEA 6.13-3® (SIMULIA, Dassault Systèmes, 78140 Vélizy-Villacoublay, France) is a software suite for finite element analysis and computer-aided engineering, originally released in 1978.

The Abaqus FEA offers powerful and complete solutions for engineering problems covering a vast spectrum of industrial applications. For example, in the automotive industry engineering work groups are able to consider full vehicle loads, dynamic vibration, multi-body systems, impact/crash, nonlinear static, thermal coupling, and acoustic-structural coupling using a common model data structure and integrated solver technology.

In our point of view, this software is the most convenient for such a complex 3-d shape of anatomy as well as capable to solve with hyper-elastic problems. It is also able to establish the contact between two solids and compatible with the other softwares used before.

The simulation made during this thesis was considered a quasi-static non-linear (chapter 3.4) solid mechanics simulation. Implicit or explicit methods to solve the system are taken into account.

The implicit solver of Abaqus® is the choice for this simulation due to the straightforward performing of quasi-static analysis within the implicit solver rather than the explicit [ABAQUS, 2005]. Moreover, the studied system has also only one high deformation component.

The implicit solver (standard solver in ABAQUS) uses a time implicit integration scheme. It means that it calculates the state of a system at the time \( t + \Delta t \) not only based on values at \( t \) but also on values at \( t + \Delta t \). Thus, it requires solving a system of equations to find the state of the system at \( t + \Delta t \). Here is a simplified equation of the way of iterating to the converged solution.

\[
\frac{\partial x}{\partial t} = f(x, t) \quad (3.3)
\]
\[
x_{n+1} = x_n + h_n \cdot f(x_{n+1}, t_{n+1}) \quad (3.4)
\]

This is an example of a differential equation 3.3. The calculation of \( x_{n+1} \) implies an equation 3.4 that must be solved in order to obtain the values of the system in the iteration \( n + 1 \).
The solution is obtained as series of increments; several iterations to obtain convergence may be needed in each time increment. Abaqus® implicit solver has an automatic accuracy check which leads the system to be unconditionally stable. However, since in each iteration a system of equations has to be solved, these increments are computationally expensive. Furthermore, for highly non-linear systems the number of increments can be extremely large, which can lead to long solving times.

3.4. Material Properties

Cartilages

Since experimental data explained in the chapter 2.3.2 is reliable source for our study, our model of cartilages is defined as exponential hyper-elastic using the parameters of that approach of Fuentes and Terrier (2005), they are $\alpha_e = 1.1485$ MPa, $\beta_e = 1.7333$. Despite its porosity and anisotropy, the cartilage in the glenoid cavity and humeral head can be assumed as isotropic and homogeneous.

Regarding to the mesh of these solids, we use quadratic tetrahedral elements which are more capable to define complex surfaces and shapes. It is well-agreed that the quadratic is more convenient for this case due to bigger number of nodes per element.

In the beginning, in order to get used to working with Abaqus® we test our model for the resting position at $60^\circ$ and normal force of 400 N. we try to assess the mesh which leads a low error with an efficient time. We observe in the figure that there is not a big error using the coarsest against the finest mesh 0.013 MPa (0.68%).

![MaxContact Pressure - Elements](image.png)

Figure 3.10.: Contact pressure - Total number of elements

For our model, last optimization is taken into account with a total of elements 9434 (close to validated before in fig. 3.10). We use a quite fine mesh for the glenoid and humeral head cartilage of 3436 and 5998 elements respectively for reaching a feasible simulation on terms of mesh. For the case of humeral cartilage, the size of seed was finer in the central-vertical part as is expected where the contact pressure will be more critic. The humeral head is also finer due to it is the slave surface in the contact.
Musculoskeletal Model of the Glenohumeral Joint

Figure 3.11.: Meshed of cartilages

Bone
The mechanical properties of the bone has been explained in the chapter 2.3.1. As the young’s moduli of the bone is greater than for the cartilages \( E_c \ll E_b \) according to section 2.3.1. Taking into account that I define the bones as rigid bodies. We are conscious that for huge stresses or deformations the contact between bone and cartilage may be significant.

3.5. Simulations
The simulations to achieve the goal follow a progression which is stated in this chapter. The following measures are common for all the simulations.

The 3-d model created from the reconstruction placed the bone out of the beginning of the abduction, concretely the arm is in the front thigh. So, we rotate the humerus 20° around its \( y_h \)-axis to put the \( x_h \) perpendicular to the scapular plane which is the right position to start the abduction. The interaction between cartilage layers is assumed frictionless sliding \((\mu = 0.003\) explained in the section 2.3.2\)) and normal hard contact. The master surface was defined in the glenoid because we prioritize its shape against the humeral head cartilage. Thus humeral cartilage is the slave surface and will be adapted to the glenoid. In order to achieve the simulation we meshed finer the humeral head cartilage (section 3.4).

As it exists interference between cartilage layers in the model. The software Abaqus is not able to establish the contact for that matter. In the beginning of every simulation, It appears two steps to compute the contact between layers: first, separating humerus and its cartilage of scapula and second to get the contact thanks to a low force. All this process minimizing the dissipation of energy.

We try to simulate a quasi-static (implicit) movement of abduction in the scapular plane from a rest position in 0° to 150° of elevation for the following cases:

3.5.1. Abduction Motion
The Euler rotation angles of clavicle, scapula and humerus were implemented following the experimental study of McClure et al. (2001). The sternoclavicular point \((SC)\) is fixed and the clavicle may rotate as a connecting rod. The scapula share two joints; glenohumeral
(GH) and acromioclavicular (AC) joint. The link between scapula and humerus is achieved thanks to the reaction force between centers and the shape in both parts (fig:3.12). That reaction force was set between the humeral head center and the geometric center of the glenoid, so these points follow the motion associated to these bones. The translation between the humeral head and the glenoid fossa was allowed.

Figure 3.12.: Mechanism of the model. AA: Angle of Acromion, GH: Glenohumeral Center, SC: Sternoclavicular Joint, AC: Acromioclavicular Joint and $F - GH$: Reaction Force in the GH joint.

3.5.2. Articular Contact in Normal-Transversal Force

The reaction force in the glenohumeral joint is constrained from the GH center to the geometry center of the glenoid cartilage. For the cases of prosthesis, it exists a test for analyzing the loosing of humeral head to glenoid cavity (stability). We decide to test our model with the specifications taken from Anglin et al. (2001). It consists in apply a force between centers as well as a transversal force. In that research prosthesis should be able to be stable for normal force (750 N) and transversal displacement (50 mm) of the glenoid, getting an eccentric reaction force with a mean of 865 N.

This eccentric force can be a composition of two forces. We apply the normal force of 750 N (mean) between centers and the transversal force is 430 N. Force is constant during the whole abduction so we define the motion in one step. As the transversal force in the publication is only set perpendicular to the normal, we simulate one oriented vertical (downward) and the other horizontal (backward). Those directions of the force may be observed in the following image (fig. 3.13):
3.5.3. Articular Contact in the Abduction with Physiological Force (EMG) between Centers

The reaction force in the glenohumeral joint has been widely studied by many authors like Westerhoff et al. (2009). In our case, we have a resource of a musculoskeletal model from the LBO obtained solving the model with EMG-based method of Engelhardt et al. (2014) explained in the chapter 2.6.3. This force \( F_{EMR_{rr}} \) is obtained each 30° though the summation of all nodal contact pressures multiplied its area. We set this force in many different steps of the software which interpolates the value in every increment between different step (abduction defined in one step of abduction per each 30°). This force follows the curve of the figure 3.14:

![Figure 3.14.: Force \( F_{EMR_{rr}} [N] \) depending on rotation[°] in abduction(\(x_h\)-axis) taken from EMG](image)

We use this amount of force to establish a physiological force between the humeral center and the center of the glenoid surface as it’s described in the figure 3.13, in this case without
transversal force (only $F - N$ force).

### 3.5.4. Articular Contact in the Abduction with Physiological Force (EMG).

#### Variable direction

In this part, we try to adjust the direction of the force from the EMG-base model of the last chapter 3.5.3 which means to use a vectorial force. It is known that the glenohumeral reaction force varies its direction and modulus during the abduction movement [Westerhoff et al., 2009].

Thus, we need to apply this vectorial model close to reality of physiological force. In the first view, we try to adjust the direction of this force according to compare both scapula coordinate systems, but we reject this issue because the morphology is too distinct.

After reflexion, it was agreed to fit the reaction force for $0^\circ$ to a direction between humeral head and one point in the surface taking into account the publication of Boyer et al. (2008), about the center of contact pressure for $0^\circ$ of abduction.

To align both vector, we make combine two lineal application of rotation around two canonical axis, the two combined rotations that fit better the rotation are these two angles, $\theta = 10^\circ$ and $\phi = 38.5^\circ$:

$$F_{EMR}^j = R_z(\phi) \cdot R_y(\theta) \cdot F_{EMG}^j \quad (3.5)$$

$$R_y(\theta) = \begin{bmatrix} \cos(\theta) & 0 & \sin(\theta) \\ 0 & 1 & 0 \\ -\sin(\theta) & 0 & \cos(\theta) \end{bmatrix}$$

$$R_z(\phi) = \begin{bmatrix} \cos(\phi) & -\sin(\phi) & 0 \\ \sin(\phi) & \cos(\phi) & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

$\theta = 10^\circ \quad \phi = 38.5^\circ \quad j = 0^\circ, 30^\circ, 60^\circ, 90^\circ, 120^\circ \text{ and } 150^\circ$

Finally, we should put the force vector ($F_{EMG}^j$) in reference to the coordinate system of the scapula, because this force should rotate synchronized to the scapula. We get the force ($F^j$) to apply in the 6 different steps ($i$) of abduction. $R_s = [x_s y_s z_s]$

$$F^j = R_s^{-1} F_{EMG}^j \quad (3.6)$$
4. Results

4.1. Abduction Motion

The results for the movement of abduction described before (section 3.5.1) is in the following figure 4.1. We can observe the scapulo-humeral rhythm where the humerus rotates twice faster than the scapula. In abduction, the humerus describes the trajectory in the scapula plane ($x_s = 0$). The clavicle is rotated less extent than the other bones.

Figure 4.1.: Abduction. Scapulo-humeral rhythm for $0^\circ$, $30^\circ$, $60^\circ$, $90^\circ$, $120^\circ$ and $150^\circ$. 
4.2. Articular Contact in Normal-Transversal Force

4.2.1. Vertical Transversal Force

We achieve the simulation for the vertical force with the values proposed in the chapter 3.5.2. We observe that there is no contact loosening and this model would pass the test. The highest contact pressure is located in the lower part(fig. 4.2), it is quite big value (15.14 MPa) due to the great reaction force. The contact pressure starts in the center of the glenoid and goes down due to the force direction and as a consequence the translation in the GH joint.

![Contact pressure(CPRESS) last point of abduction(150°), vertical transversal reaction Force. *Values in the table in MPa.](image)

Figure 4.2.: Contact pressure(CPRESS) last point of abduction(150°), vertical transversal reaction Force. *Values in the table in MPa.
4.2.2. Horizontal Transversal Force

In this case, the model looses the humerus of the scapula for that amount of reaction force. So, we decided to decrease the value of transversal force until it would be stable. That happens for a transversal force of 50N, generating a contact pressure of 6 MPa (fig. 4.3). The pressure begins in the geometrical center of the glenoid and goes backwards (body reference) due to the force and as a consequence the translation (same issue of figure 4.2). We can observe the humeral head resurfacing the glenoid cartilage changing its shape importantly.

![Contact Pressure](image)

Figure 4.3.: Contact Pressure (CPRESS) last point of abduction (150°), horizontal transversal reaction Force. *Values in the table in MPa.
4.3. Articular Contact in the Abduction with Physiological Force (EMG) between centers

4.3.1. Contact Pressure in Humeral Head Cartilage

The contact pressure in the humeral head moves from the lower part to the highest due to a humeral rotation. Its peak value of 7.3 MPa occurs at 90° of humeral abduction.

Figure 4.4.: Humeral head cartilage contact pressure (CPRESS) in abduction for 0°, 30°, 60°, 90°, 120° and 150°. *Values in the table in MPa
4.3.2. Contact Pressure in Glenoid Cartilage

Redundant to the humeral head, the peak contact pressure at 90° with a value of 7.3 MPa. The center of pressure remains almost at the same position due to the constant direction of the force, it appears a little effect of translation. The model is stable for this case, contact is guaranteed for all the abduction.

Figure 4.5.: Glenoid cartilage contact pressure (CPRESS) in abduction for 0°, 30°, 60°, 90°, 120° and 150°. *Values in the table in MPa
4.4. Articular Contact in the Abduction from EMG Variable Direction

This model do not work as expected, Abaqus is only able to simulate the first $50^\circ$ of abduction due to the loss of contact on that position. The pressure rises from the right lower to the center top (fig. 4.6). The peak of pressure takes place at the last position, if we compare this value to the model before (section 4.3.2) between $30^\circ$ and $60^\circ$, logically they are close (same modulus of force for both cases) to 3.5 MPa.

Figure 4.6.: Glenoid cartilage contact pressure (CPRESS) in abduction for variable direction force for $0^\circ$, $15^\circ$, $30^\circ$, $45^\circ$ and $50^\circ$. *Values in the table in MPa.
5. Social Impact

Nowadays, osteoarthritis has become a significant problem for the health-care system and is growing up with the aging population affecting in the knee for example more than a third of the people aged over 65 years old [Dawson et al., 2004]. According to Lawrence et al. (2008), the 8.5% (27 million of people) of the US population suffered OA in any joint, being the GH joint the third with more disease after the hip and the knee. The cost of osteoarthritis in the United States (US) was over $65 billion per year published by Jackson et al. (2001).

Indeed, the shoulder arthroplasties are increasing faster, the study of Kim et al. (2011) tells that around 27000 shoulder arthroplasties were carried out in 2008 only in the US. Moreover, the shoulder arthroplasties are predicted to increase between 192% and 322% by 2015 and also the price rate of the total shoulder Arthroplasty from $900 (700€) to $1700 (1260€) [Day et al., 2010].

Many studies biomechanical research should be released in terms of prevention and better understudying of this pathology. This approach should be useful to increase the knowledge about what conditions of pressure are generated in cartilage layers. This research is directly related with the understanding of osteoarthritis as it has been explained in the chapter 2.4.2.

This study is truly a short step compared to great challenge of develop a population databases of the contact pressure and muscles forces. Afterwards, this source would be able to predict through measuring the forces in muscles an overload in the GH joint to detect potential osteoarthritis. Thus, detecting early this disease might decrease the costs of shoulder osteoarthritis, making the health care system more sustainable.
6. Analysis of costs (Budget)

In any work, the project management is essential. In the following gantt (table 6.1), we schedule the project to get a proper organization. It is also necessary to evaluate the hours spent, 640 hours and their cost of the work. In the case of an academic project, the time expected was 540 hours however an extra time was taken to achieve the goal.

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<td>40</td>
<td>40</td>
<td>40</td>
</tr>
</tbody>
</table>

Table 6.1.: Gantt Scheduling

Once we have quantified the time spent in the project, we present the budget. Since, it is a project of research we are not able to estimate the direct economical benefits of our research. Thus, in the budget it appears only the costs in the table 6.2. The final cost estimated is 25807.33€.

<table>
<thead>
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<td>Contingency (10% Unexpected)</td>
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<td>Sub-total after Unexpected</td>
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<td></td>
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<tr>
<td>Total Projected Cost</td>
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<td>25807.33€</td>
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</table>

Table 6.2.: Total budget of the project at June 3, 2014

Those unit costs in the table 6.2 are mainly related to the salary and informatics’ equip-
ment. The cost of the software is related to the yearly license of the program (e.g., retail of Abaqus® 17000€, Amira® 3000€ or SolidWorks® 8000€). The hardware costs include the depreciation.
7. Discussion

In this approach, the simulation of the abduction movement was achieved from $0^\circ$ to $150^\circ$. From this point onwards, this model should be able to display many different motion in the shoulder apart from the abduction following the same procedure.

Furthermore, we could quantify the contact pressure in between cartilage layers in varied conditions. In fact, it exists a similarity of our research (section 4.3) and the experimental data of strain in the GH joint from Soslowsky et al. (1992). The humeral head cartilage follows the same evolution of our simulation as well as there is a parallelism to the center of the pressure in the glenoid. The peak of contact pressure of 7.3 MPa is close to the mean of 5.1 MPa from Conzen and Eckstein (2000) and it occurs at the same position $90^\circ$. Despite some consistence of our approach, we state some discordance between all those experimental data. Therefore, more in-vivo measurement of the GH joint should be assessed including the effect of muscles in GH joint.

Indeed, the phenomena of translation in the GH joint may also have been observed and it is stated the difference between a ball and socket mechanism and the GH joint.

Nevertheless, in the future analysis a finer resolution is required in order to get a smoothed surfaces as well as the reconstruction of the labrum. In our case, that resolution (0.5 mm) was too rough compared to the validation (0.125 mm) of Yeh et al. (1998).

The impact of the cartilage to bone has to be evaluated for high contact pressures due to the hyper-elastic properties. Even though, the exponential hyper-elastic is described in reliable experimental data, it is agreed that the biphasic theory is the most convenient to estimate the behavior of cartilage. We are not able to ensure if it will change the results. For this case, it would be recommended to use explicit method (dynamic analysis) in the software Abaqus®.

It has been observed that the glenoid labrum may play an important role in stability [Fehringer et al., 2003]. It is not only for the last simulation (section 4.4) where the contact pressure is too eccentric (loosening the contact) but also the influence to the peak of pressure. Greis et al. (2002) assessed a decrease of 28% for the peak pressure including the labrum. However, the great challenge would be to combine fibrous conditions with isotropic hyper-elastic exponential properties.

Otherwise, last simulation might mean that the vector force taken from a different model does not apply rightly for our work. Thus, muscles must be included for getting more precise work according to the functionality both mechanical and physiological of the GH joint. In fact, this process is on-going in the LBO using the EMG-based method to carry out this model including muscles.
8. Conclusion

The objective of the project was achieved because we could quantify the contact pressure between cartilage layers in the GH joint for the arm abduction. This step should be useful to increase the knowledge about what conditions of pressure are generated in cartilage layers. This issue is directly related with the pathogenesis’ comprehension of osteoarthritis.

Since this model is able to simulate abduction, further motions may be carried out with this procedure to study the mechanical behavior of cartilages during the wide broad of movements in the Glenohumeral joint.

Nevertheless, the model has some limitations regarding the reconstruction of cartilage and especially the physiological force applied. The labrum may play an important role, so it should be implemented despite its complexity. For future investigation of this model, it is strongly recommended to implement to the model the action of muscles.

To conclude, the coupling of biomechanics and numerical analysis is going to be a key topic within bioengineering in the next decades, which has already been proved by the non-stopping increasing amount of published papers during these last years.
9. Acknowledgments

This project was supported by the Laboratory of Biomechanical Orthopedics (LBO), inside the Swiss Federal Institute of Technology in Lausanne (Ecole Polytechnique Fédérale de Lausanne, EPFL).

I thank Prof. Alexandre Terrier and Prof. Dominique Pioletti for giving me the chance to work in such a magnificent environment. I express my deep gratitude to the PhD Christoph A. Engelhardt for the training and his helpful advice in the whole project.

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To conclude, I am really thankful to all those who have encouraged during my studies, especially to my family and girlfriend, María Salort.
Bibliography


Appendices
A. Computing Code

A.1. Exponential Hyper-elastic law (Python®)

```python
SUBROUTINE UHYPER(BI1,BI2,AJ,U,UI1,UI2,UI3,TEMP,NOEL,
1 CMNAME,INCMPFLAG,NUMSTATEV,STATEV,NUMFIELDV,FIELDV,
2 FIELDVINC,NUMPROPS,PROPS)
C
INCLUDE 'ABA_PARAM.INC'
C
CHARACTER*80 CMNAME
DIMENSION UI1(3),UI2(6),UI3(6),STATEV(*),FIELDV(*),
2 FIELDVINC(*),PROPS(*)
A1 = 1.14853
A2 = 1.17333
U = A1*DEXP(A2*(BI1-3.0D0))-(A1*A2/2)*(BI2-3)
UI1(1) = A1*A2*DEXP(A2*(BI1-3.0D0))
UI1(2) = -0.5*A1*A2
UI1(3) = 0.0D0
UI2(1) = A1*A2*A2*DEXP(A2*(BI1-3.0D0))
UI2(2) = 0.0D0
UI2(3) = 0.0D0
UI2(4) = 0.0D0
UI2(5) = 0.0D0
UI2(6) = 0.0D0
UI3(1) = 0.0D0
UI3(2) = 0.0D0
UI3(3) = 0.0D0
UI3(4) = 0.0D0
UI3(5) = 0.0D0
UI3(6) = 0.0D0
RETURN
END
```

A.2. Least quadratic Method for Humeral Origin (Matlab®)

```matlab
function [center, radius, residuals, R2] = fitSphere2(x,y,z)
% SPHEREFIT find least squares sphere
% Fit a sphere to a set of xyz data points
% [center, radius, residuals] = shperefit(X)
% [center, radius, residuals] = spherefit(x,y,z);
% Input
% x,y,z Cartesian data, nx3matrix or three vectors(n x 1 or 1
% x n)
```
% Output
% center: least squares sphere center coordinates, == [xc yc zc]
% radius: radius of curvature
% residuals: residuals in the radial direction
% Fit the equation of a sphere in Cartesian coordinates to
% a set of xyz data points by solving the overdetermined
% system, ie, x^2 + y^2 + z^2 + ax + by + cz + d = 0
% The least squares sphere has radius R = sqrt((a^2+b^2+c^2)
% /4-d)
% and center coordinates (x,y,z) = (-a/2,-b/2,-c/2)
error(nargchk(1,3,nargin)); % check input arguments
if nargin == 1 % n x 3 matrix
  if size(x,2) ~= 3
    error('input data must have three columns')
  else
    z = x(:,3); % save columns as x,y,z vectors
    y = x(:,2);
    x = x(:,1);
  end
elseif nargin == 3 % three x,y,z vectors
  x = x(:); % force into columns
  y = y(:);
  z = z(:);
  if ~isequal(length(x),length(y),length(z)) % same length ?
    error('input vectors must be same length');
  end
else % must have one or three inputs
  error('invalid input, n x 3 matrix or 3 n x 1 vectors expected');
end
% need four or more data points
if length(x) < 4
  error('must have at least four points to fit a unique sphere');
end
% solve linear system of normal equations
A = [x y z ones(size(x))];
b = -(x.^2 + y.^2 + z.^2);
a = A \ b;
% return center coordinates and sphere radius
center = -a(1:3)/2;
radius = sqrt(sum(center.^2)-a(4));

% calculate residuals
if nargout > 2
    residuals = radius - sqrt(sum(bsxfun(@minus,[x y z],center.').^2,2));
end

% whichstats = {'adjrsquare' 'rsquare'};
% stats = regstats(b,A(:,1:3),'linear',whichstats);
% R2 = stats.rsquare;
% R2adj = stats.adjrsquare;

sse = sum(residuals.^2);
meanX = [mean(x) mean(y) mean(z)];
for i=1:size(x)
    X = [x(i) y(i) z(i)];
    tot(i) = norm(meanX-X);
end

sst = sum(tot.^2);
R2 = 1-sse/sst;

A.3. Coordinate System (Matlab®)

C7 = [70.5729 -35.7244 150.764]
T8 = [78.9481 -29.5985 -40.6273]
IJ = [74.1129 -74.8066 68.9446]
PX = [82.5171 -133.113 -76.9724]
yt = ((C7+IJ)-(PX+T8))/2 %TORAX needed for Clavicle yc

ACc = [-79.3788 -17.6388 128.459] %Art. Acromioclaviculare
SC = [53.2266 -68.0611 61.4247] %Art. Sternoclaviculare
Oc=SC
zc=ACc-SC
xc = \texttt{cross(yt, zc)}
yc = \texttt{cross(zc, xc)}

Xc = xc / \texttt{norm(xc)}
Yc = yc / \texttt{norm(yc)}
Zc = zc / \texttt{norm(zc)}

% ABAQUS REFERENCE POINT OF AXIS

Xca = Oc + Xc
Yca = Oc + Yc
Zca = Oc + Zc

% ABAQUS CLAVICULE

%---

Osg = [−69.6059, −7.9796, 81.5891]

TS1 = [−4.5294, 61.6920, 78.4544] \% Trigonum Scapulae 1
TS2 = [0.3576, 58.2558, 90.1397] \% Trigonum Scapulae 2
TS3 = [−11.2639, 57.7059, 87.3115] \% Trigonum Scapulae 3
ACs = [−86.1877, −16.8160, 127.6564] \% Art. Acromioclaviculare (Scapula)
AI = [−2.9947, 5.4159, 50.5310, 58.6] \% Angulus Inferior
AA = [−1.0332, 5.8052, 59.2773, 43.8] \% Angulus Acromialis

Os = AA \% Centre of scapular rotation
TS = (TS1 + TS2 + TS3) / 3 \% Centre of the 3 referenced points of Trigonum Scapulae

zs = AA − TS \% External Axis
xs = \texttt{cross}(zs, AI − TS) \% Forward axis
ys = \texttt{cross}(zs, xs) \% Vertical Axis

Xs = xs / \texttt{norm(xs)} \% Unit Vector
Ys = ys / \texttt{norm(ys)}
Zs = zs / \texttt{norm(zs)}

% ABAQUS REFERENCE POINT OF AXIS

Xsa = Os + Xs
Ysa = Os + Ys
Zsa = Os + Zs

% ABAQUS SCAPULA
Musculoskeletal Model of the Glenohumeral Joint

%---------------------------------------------

%%HUMERUS%%

%--POINTS TAKEN FROM RESTING POSITION (r),
%--NOT FROM BEGINNING OF ABDUCTION

Oh = [−86.3699, −25.1763, 85.8332] %Glenohumeral Rotation Centre from MRI
EL = [−127.285, −54.8861, −208.496] %From Christoph %Lateral Epicondyle from MRI
EM = [−80.2709, −11.761, −203.232] %Medial Epicondyle from MRI
EMP = [−101.298, −17.604, −119.178] %Epicondyle midpoint

% RESTING POSITION %

Xh = xh / norm(xh) %Unit Vector -- HUMERUS REFERENCE
Yh = yh / norm(yh)
Zh = zh / norm(zh)

%% AXIS IN THE SCAPULAR PLAIN %%

zhsc = Osg−TS %External Axis
xhsc = cross(zs, AI−TS) %Forward axis
yhsc = cross(zhsc, xhsc)

Zhsc = zhsc / norm(zhsc) %Unit Vector -- HUMERUS REFERENCE
Yhsc = yhsc / norm(yhsc)
Xhsc = xhsc / norm(xhsc)

%%MATRIX OF VECTORS%%

XH = transp([[Xh; Yh; Zh]]) %Matrix of change of basis

%%ABAQUS REFERENCE POINT OF AXIS%%

Xha = Oh + Xh
Yha = Oh + Yh
Zha = Oh + Zh

%%ABAQUS%% HUMERUS
% Initial Displacement

\[ \text{Zhadi} = -\text{Zh} \times 0.4 + \text{Oh} \]
\[ \text{Zhaf} = \text{Zh} \times 2 + \text{Oh} \]

% Euler Angles

% Final Position Angles
% Clavicle:

\[ \text{UR1c} = -0.11 \]
\[ \text{UR2c} = -0.2251 \]
\[ \text{UR3c} = 0.4119 \]

% Scapula:

\[ \text{UR1s} = -0.5794 \]
\[ \text{UR2s} = 0.0367 \]
\[ \text{UR3s} = 0.274 \]

% Humerus:

\[ \text{UR1h} = -2.5045 \]
\[ \text{UR2h} = 0 \]
\[ \text{UR3h} = 0 \]

% Abduction Not Constant
% Humerus (Others equal)

\[ \text{UR1ha} = -30 \times 2 \times \pi /360 \]

\[ \text{FV} = [131.9068 \ 2.552402 \ -29.14157] \]
\[ \text{F} = \text{FV} / \text{norm(FV)} \]
\[ \text{Pos} = -\text{F} \times 2.5 + \text{Oh} \]
\[ \text{Posf} = \text{F} \times 0.3 \]
\[ \text{POSF} = ((\text{XH}) \times \text{Posf'})' \]

% Scapula

\[ \text{UR1hb} = \text{UR1h} - \text{UR1ha} \]
\[ \text{UR1cf} = \text{UR1c} / 5 \]
\[ \text{UR2cf} = \text{UR2c} / 5 \]
\[ \text{UR3cf} = \text{UR3c} / 5 \]

% Scapula:

\[ \text{UR1sf} = \text{UR1s} / 5 \]
A.4. Force Direction (Matlab®)

\[
O_{sg} = [\begin{bmatrix} -69.6059 & -7.9796 & 81.5891 \end{bmatrix} \\
O_{h} = [\begin{bmatrix} -86.3699 & -25.1763 & 85.8332 \end{bmatrix} \\
Z_s = [\begin{bmatrix} -0.8644 & -0.4703 & 0.1780 \end{bmatrix} \\
\beta Z = 38.5 \times \pi /180 \quad \%\text{Angle of rotation} \\
R_z = [\begin{bmatrix} \cos (\beta Z) & -\sin (\beta Z) & 0 \\
\sin (\beta Z) & \cos (\beta Z) & 0 \\
0 & 0 & 1 \end{bmatrix} \\
\beta Y = 10 \times \pi /180 \\
R_y = [\begin{bmatrix} \cos (\beta Y) & 0 & \sin (\beta Y) \\
0 & 1 & 0 \\
-\sin (\beta Y) & 0 & \cos (\beta Y) \end{bmatrix} \\
AA = [\begin{bmatrix} -20.17027 & 39.64398 & 15.54523 \end{bmatrix} \\
TS = [\begin{bmatrix} -123.27052 & 1.10895 & 7.43461 \end{bmatrix} \\
AI = [\begin{bmatrix} -119.92752 & 1.10895 & -95.43585 \end{bmatrix} \\
\%\text{Abaqus origin seen from ISB reference} \\
\%AA \\
O_{sj} = [\begin{bmatrix} 44.01314 \\
& -16.37370 \\
& -3.85809 \end{bmatrix} \\
\%\text{Along X} \\
X_A = [\begin{bmatrix} 47.50615 \\
& -17.10188 \\
& -13.19985 \end{bmatrix} \\
xa = (X_A - O_{sj}) / \text{norm}(X_A - O_{sj}) \quad \%\text{Axis vector abaqus EMG}
\]

%Along Y
YA = [ 34.64372  
    -16.52436  
    -7.34969 ]

ya = (YA−Osj)/norm(YA−Osj) %Axis vector abaqus EMG

%Along Z
ZA = [ 44.12665  
    -6.40139  
    -4.59298 ]

za = (ZA−Osj)/norm(ZA−Osj) %Axis vector abaqus EMG

A = [ Xs’ Ys’ Zs’ ] %ISB Csys Scapula EMG

Fj0 = [ 131.9068 2.5524 −29.1416] %Reaction Force from EMG
Fj30 = [552.5045 −2.5667 64.816713]
Fj60 = [839.2533 −13.4401773 88.2986]
Fj90 = [869.1611 −12.0567 −82.8264]
Fj120= [635.85486 −11.8358 −308.0513]
Fj150= [325.3548 −9.3876 −431.0138]

Fjry0 = (Ry∗Fj0)’ %Reaction Force from EMG, rotation y.
Fjry30 = (Ry∗Fj30)’
Fjry60 = (Ry∗Fj60)’
Fjry90 = (Ry∗Fj90)’
Fjry120= (Ry∗Fj120 )’
Fjry150= (Ry∗Fj150 )’

Fjr0 = Rz∗Fjry0 %Reaction Force from EMG, rotation x.
Fjr30 = Rz∗Fjry30
Fjr60 = Rz∗Fjry60
Fjr90 = Rz∗Fjry90
Fjr120= Rz∗Fjry120
Fjr150= Rz∗Fjry150

%Change of reference system to system scapula

F0 = (inv(A) * Fjr0)’
F30 = (inv(A) * Fjr30)’
F60 = (inv(A) * Fjr60)’
F90 = (inv(A) * Fjr90)’
F120= (inv(A) * Fjr120 )’
F150= (inv(A) * Fjr150 )’

Fc=(Osg−Oh)/norm(Osg−Oh) %UnitVector
FJR0=Fjr0/norm(Fjr0)

%Validation

angle0 = atan2(norm(cross(Fc, Fjr0)), dot(Fc, Fjr0))*180/pi
angle30 = atan2(norm(cross(Fc, Fjr30)), dot(Fc, Fjr30))*180/pi
angle60 = atan2(norm(cross(Fc, Fjr60)), dot(Fc, Fjr60))*180/pi
angle90 = atan2(norm(cross(Fc, Fjr90)), dot(Fc, Fjr90))*180/pi
angle120 = atan2(norm(cross(Fc, Fjr120)), dot(Fc, Fjr120))*180/pi
angle150 = atan2(norm(cross(Fc, Fjr150)), dot(Fc, Fjr150))*180/pi

Aip=[Xh’ Yh’ Zh’]

%Initial Displacement

ip= inv(Aip)*Fjr0/norm(Fjr0)*.2
ip2=inv(Aip)*Fc’*0.25 %Starting from the center

anglesc = atan2(norm(cross(za’, Zs)), dot(za’, Zs))*180/pi
anglesc1= acos(dot(za’, Zs))*180/pi
anglesc2= asin(dot(za’, Zs))*180/pi
anglesc3= atan(dot(za’, Zs))*180/pi

angle0ref=atan2(norm(cross(Fc,FJR0)), dot(Fc,FJR0))*180/pi