Doctoral Thesis

A model to study active shoulder motion and stability

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A MODEL TO STUDY ACTIVE SHOULDER MOTION AND STABILITY

A dissertation submitted to

ETH ZURICH

for the degree of

DOCTOR OF SCIENCES

presented by

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2011
THIS WORK IS DEDICATED TO

CAROLE, AXEL, AURÉLIEN

AND MY PARENTS
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SUMMARY

Shoulder stability is primarily ensured by coordination of muscular action. While the influence of passive anatomical structures on shoulder stability has been studied to some extent, the mechanisms involved in active muscular stabilization of the shoulder remain poorly understood. The aim of this thesis was to develop a realistic and anatomically precise three-dimensional numerical model to simulate active joint balance and stabilization.

In a first step, a novel method is presented to automatically compute the muscle path, relying on contact detection capabilities of commercial FE packages. Realistic muscle length and moment arm were predicted when compared to experimental measurements over a wide range of shoulder motion. This method eliminates many of the weaknesses associated with available methods, providing a valuable tool for both general and subject-specific musculoskeletal modeling.

Second, this anatomical information was used to estimate the muscle forces needed to balance and stabilize the glenohumeral joint. An innovative algorithm distributes infinitesimal forces to selected groups of shoulder muscles (those possessing most favorable moment arms at the given static position) to balance and stabilize the joint. Realistic muscle forces can be predicted for any given position of the humerus in equilibrating an external force acting in any arbitrary direction.

Finally, the resultant force was applied in a 3D finite element model of the glenohumeral joint for simulation of joint contacts and humeral head translations. The framework was exploited to simulate elevation as a composite of instantaneous positions and thoroughly validated.

Muscle path, muscle forces and joint contact characteristics can now be computed automatically, at any joint position. This is the first model able to cope with the wide glenohumeral joint range of motion in all degrees of freedom and the considerable variability in external loading conditions as encountered in daily life activities. This framework may be used to address clinical hypotheses related to shoulder joint stability that cannot be pursued using simplified modeling approaches.
ZUSAMMENFASSUNG


Dieses Werkzeug kann nun zur Testung von klinischen Hypothesen über die Stabilität des Schultergelenkes angewendet werden, die mit vereinfachten Modellen nicht untersucht werden können.
“It deserves to be known how a shoulder which is subject to frequent dislocations should be treated. For many persons owing to this accident have been obliged to abandon gymnastic exercises, though otherwise well qualified for them; and from the same misfortune have become inept in warlike practices, and have thus perished. And this subject deserves to be noticed, because I have never known any physician treat the case properly; some abandon the attempt altogether, and others hold opinions and practice the very what is proper”

On the Articulations

By Hippocrates 400 B.C
CHAPTER 1
1.1. INTRODUCTION

FOUNTAIN OF NEPTUNE, PIAZZA NAVONA, ROME
The shoulder complex consists of the humerus, the clavicle, the scapula and the sternum/thorax. Movements at the shoulder are the result of an intricate coordination of motions taking place at the sternoclavicular, acromioclavicular and glenohumeral (GH) joints and through gliding of the scapula on the thorax (Figure 1).

![Diagram of shoulder complex with labels: Acromioclavicular joint, Sternoclavicular joint, Acromion, Scapula, Clavicle, Sternum, Glenohumeral joint, Scapulothoracic articulation, Humerus.]

**Figure 1: Bones and joints of the shoulder complex.** Adapted from Rockwood et al, The Shoulder, 4th Edition, Saunders Elsevier, Amsterdam, the Netherlands, 2009.

Together these structures form the most mobile part in the human body. Humeral rotations of about 180° in three different planes are possible, allowing the range of motion of the shoulder to cover 65% of a sphere (Engin and Chen, 1986). These large motions take place for the greatest part at the GH joint, enabled by the very little coverage of the humeral head by the small glenoid (the narrow socket on the scapula, see Figure 2).
Introduction

Unfortunately, this high mobility comes at a cost: the GH joint is the most frequently dislocated major joint of the body, affecting at least 1.7% of people in the course of their lifetime (Hovelius, 1982). As much as 8% remain recurrent 25 years after the first dislocation (Hovelius et al., 2009). It is established that recurrence rates depend strongly on the age at the first dislocation (Table 1) but the underlying reasons leading to healing or recurrence remain unclear (Hovelius et al., 2008).

<table>
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<th>Age at 1st dislocation</th>
<th>Recurrence (2 or more)</th>
<th>Follow up</th>
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<tr>
<td>≤ 22 years</td>
<td>55 %</td>
<td>5 years $^1$</td>
</tr>
<tr>
<td>23-29 years</td>
<td>37 %</td>
<td>5 years $^1$</td>
</tr>
<tr>
<td>30-40 years</td>
<td>12 %</td>
<td>5 years $^1$</td>
</tr>
<tr>
<td>&gt; 60 years</td>
<td>22 %</td>
<td>7 years $^2$</td>
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Dislocations have disastrous consequences for the patients’ quality of life, potentially evolving into arthropathy (Hovelius and Saeboe, 2009). Joint instability also represents one of the major complications after conventional GH arthroplasty (Franta et al., 2007; Wirth and Rockwood, 1996) and reverse shoulder replacement (Sirveaux et al., 2004; Wall et al., 2007). A deeper understanding of the mechanisms involved in shoulder stability is therefore clearly required.
The role of the passive stabilizing structures has been studied extensively (Debski et al., 1999a; Debski et al., 1999b; Motzkin et al., 1998; O’Connell et al., 1990; Terry et al., 1991). However, in contrast to the hip, the congruency of the humeral head and the glenoid cavity is too low to withstand large dislocating forces (Soslowsky et al., 1992). In contrast to the knee, the ligaments are generally lax throughout the greater part of GH rotation and become functionally relevant only at range of motion extremes. The ligaments are therefore inactive for stabilization in most joint positions (Bigliani et al., 1996) and their relative importance in overall stability remains uncertain (Veeger and van der Helm, 2007).

GH stability is primarily ensured by coordinated muscular action, which provides concavity compression (Lippitt and Matsen, 1993). Study of active muscular stabilization remains relatively neglected (Apreleva et al., 1998; McMahon and Lee, 2002; Schiffern et al., 2002) probably because of the number of involved muscles and their infinite potential interactions. Nevertheless, it is widely accepted that the rotator cuff muscles (Figure 3) move the humerus and simultaneously prevent dislocation (McMahon and Lee, 2002), and that shoulder muscles can either stabilize or destabilize the joint depending on GH position (Labriola et al., 2005; McMahon and Lee, 2002; Werner et al., 2007).

Figure 3: Muscles of the rotator cuff.
These muscles press the humeral head within the glenoid to stabilize the joint.


1.2. SPECIFIC AIMS

The aim of this thesis was to develop a realistic and anatomically precise three-dimensional (3D) numerical model of the GH joint stabilized by muscular action. A three steps approach was taken. First the muscle path was computed for the joint position of interest. Second, this anatomical information was used to estimate the muscle forces required to balance and stabilize the joint. Finally, the forces were applied on a realistic representation of the joint contacts to simulate stability. The model will lay an important technical foundation to later investigate open and essential questions in shoulder biomechanics and surgery. The influence of anatomy, articular surface defects, muscle dysfunctions, implantation of prosthesis, etc., on the behavior of the joint can then be examined.

Specific aim 1

Development of a new method for muscle path prediction based on contact detection capabilities of finite element softwares.

Specific aim 2

Refinement and implementation of a novel algorithm to predict forces for any given position of the humerus in equilibrating an external force acting in an arbitrary direction, while keeping the joint stable.

Specific aim 3

Development of a realistic FE model of the GH joint, with touching contacts allowing rotations and translations of the humerus within the glenoid (without a priori constraints on GH kinematics). Integration of the geometric input data of Aim 1 and the static force calculation of Aim 2 in this model to simulate motion as a succession of static steps, with joint stability tested in each iteration.
THESIS OUTLINE

Chapter 1: Introduction
A general introduction of the problem, the specific aims and the outline of the thesis are presented.

Chapter 2: Background
The state of the art in numerical models of the shoulder is depicted. The capabilities of the different model types are reviewed, and the need for a novel model of GH stability is demonstrated.

Chapter 3: Automated muscle wrapping using finite element contact detection
A novel method to represent realistic muscle paths is established relying on the contact detection capabilities of finite element softwares. This approach obviates the weaknesses associated with previously available methods (time consuming pre-processing, muscle penetration through the bone).

Chapter 4: A new paradigm for shoulder muscle force estimation
In order to estimate the muscle forces required to simultaneously stabilize and equilibrate the GH joint, a novel algorithm centered on relative muscle mechanical advantage is developed. It activates muscles that are best positioned to balance the external moment, and attributes supplementary rotator cuff muscle force if required for joint stability.

Chapter 5: An integrated model of active glenohumeral stability
There exists no physiologically and anatomically true model of the GH joint. To enable investigations of the influence of anatomical parameters that affect GH joint stability (such as glenoid orientation or defect, labrum or cartilage integrity, etc.), an anatomically precise 3D model of GH joint contacts is built. It is then combined with the muscle path and force prediction methods previously described to allow automatic simulation of joint contact characteristics in any joint position.

Chapter 6: Synthesis
A synthesis summarizes and concludes the presented work. Potential future developments and applications of the model are suggested.
REFERENCES


CHAPTER 2
2. BACKGROUND

Shoulder ‘organ’ (Mollier 1899)
Cords were attached at the muscle insertions and muscular action simulated by pressing the piano keys.

State of the art finite element simulation of the strains in the scapula
2.1. NUMERICAL MODELS OF THE SHOULDER

Since the 19th century, in vitro modeling by means of cadaver experiments has been widely used to examine various aspects of shoulder biomechanics (Fick and Weber, 1877; Mollier, 1899; Motzkin et al., 1998; Nyffeler et al., 2006a; O'Connell et al., 1990; Werner et al., 2007) with the usual limitations regarding tissue decay and simplified analysis restricted to a subset of muscles. In addition, cadaver experiment can only deliver values at the sensor locations. To obtain a dense distribution of values, one would have to use a very large number of sensors, which is extremely technically demanding and time consuming.

In vivo measurements of several biomechanical parameters in humans remain to date technically and ethically challenging. For example, the measurement of muscle force is limited to specific muscles and mostly restricted to laboratory settings (Erdemir et al., 2007).

Numerical models tackle many of these issues. They are non-invasive, can provide a continuum of values instead of discrete measurements, can calculate several outputs in one simulation, allow parameter studies to be readily performed and can be used in a predictive manner before invasive implementations are performed on human patients. Models can be used to simulate phenomena that would otherwise be impossible to carry out experimentally, and are generally more cost-effective than clinical, animal or laboratory testing methods (Huiskes and Hollister, 1993).

**Increasing interest in shoulder modeling**

Most numerical simulations in orthopaedics have historically focused on the hip and the knee, with fewer on the shoulder. From a clinical perspective, the hip and knee have largely occupied the interests of clinical and industrial researchers because the vast majority of joint replacements are performed at these joints. However, the number of shoulder replacements has tripled from 1994 to 2004 in the USA, reaching 29 000 partial and total shoulder replacements in 2004 (Kozak et al., 2006). From a modeling standpoint, the complexity of the shoulder joint itself may be a disincentive to modelers who have to cope with intricate active and passive stabilizing mechanisms and an extremely large range of joint motion (Hogfors et al., 1995). Upper-extremity motions are by necessity more variable than the locomotive movements of the lower limb (a well defined cyclic motion), and while
a 2D analysis of gait can reasonably characterize leg kinematics, such a simplified treatment of the shoulder is inadequate (Rau et al., 2000).

Nevertheless, lessons learned from other joints can now be applied to the shoulder, and improvements in software and computational power have facilitated the building and solving of ever more complex models. All these factors have driven the exponential increase in the number of publications involving numerical models of the shoulder (Figure 4). Numerical models of the shoulder have already played a major role in enhancing shoulder knowledge in surgical planning of transfers, prosthesis design, prevention of overload and injuries, clinical diagnosis, treatment and rehabilitation.

![Figure 4: International publications based on numerical models of the shoulder.](image)

The scale on the horizontal axis is not linear. Pubmed database was searched for the keywords “computer model shoulder”, “rigid body model shoulder”, “muscle force model shoulder” and “finite element shoulder”. All articles were controlled for inclusion, and only original articles were considered. Old publications on analytical models (earlier than 1980) were found in the references of the more recent articles.

**Models of GH stability**

Most available numerical models of the GH joint do not allow translations of the humeral head within the glenoid fossa. The GH joint is generally simplified as a ball-and-socket joint with a fixed center of rotation, prohibiting humeral head translations within the glenoid. Translations can be large in the unstable shoulder, limiting the use of such models in the study of stability. Joint stability can be estimated by considering the intersection of the
resultant force with the glenoid boundaries (Favre et al., 2005; van der Helm, 1994a; Yanagawa et al., 2008). However, the constrained humeral head translations preclude analysis of tissue deformations, contact area or pressure (Buchler et al., 2002; Hopkins et al., 2006) and may lead to underestimated muscular forces, joint reaction forces and stresses (Terrier et al., 2008; Veeger and van der Helm, 2007).

A first step away from the ball-and-socket assumption has been made by allowing limited humeral translations during axial rotation only (Buchler et al., 2002). Abduction and flexion were not permitted and vertical translations were restricted by a non-physiological spring. The joint was centered actively in another model (Terrier et al., 2007; Terrier et al., 2008), but was limited to 2D, to the deltoid and rotator cuff muscles and to abduction. Shoulder muscles can generate considerable joint moment components in the other two rotational degrees of freedom (DOF). These must be compensated by other muscles that may exert large and possibly antagonistic moments that cannot be neglected (Favre et al., 2009b; Veeger and van der Helm, 2007).

Although kinematically unconstrained models of the GH joint are clearly required to investigate stabilization and motion at the GH joint (Favre et al., 2009b; Hill et al., 2008; Veeger and van der Helm, 2007), validated six DOF models are lacking from the literature.

The purpose of this study is to develop the first (to our knowledge) 3D finite element (FE) model of the GH joint to implement active muscle driven humeral positioning and stabilization at any joint position, without a priori defined kinematic constraints. The model is intended to function in an automated fashion, with the advantage that simulations can be easily computed for any position of the joint at any time, as opposed to those measured experimentally. In this approach, the FE method is used as a framework for the integrated model. Modern commercial FE software packages allow simulation of large motions and can implement the simplifying aspects of rigid body assumptions in a hybrid fashion, permit simultaneous inclusion of muscle force estimation and recruitment strategies, and determine component deformations in critical locations.

This integrated model consists of three distinct components (Figure 5). First, the path of the muscles that cross the GH joint is determined for the given joint position of interest (Figure 5A). Second, knowing the spatial direction of the muscles and the length of their moment
arms, muscle forces which would equilibrate and stabilize the joint can be estimated (Figure 5B). Third, the computed muscles forces can be implemented in an anatomically precise 3D model of the GH joint to avoid artificial restrictions on humeral translations and simulate joint contact characteristics.

Figure 5: Schematic of the three components constituting the integrated model. Anatomical data is first generated for the position of interest (A) and is then used to estimate the muscle forces required to balance and stabilize the joint (B). Finally, the resultant force is applied in a deformable model of the joint contacts (C).

In this chapter, the currently available methods for modeling these three components (simulation of muscle path, muscle forces and joint contacts) are described. Their advantages and applications are reviewed. The weaknesses that prevent their inclusion to a realistic model of GH stability are illustrated, demonstrating the need for the novel methods that will be described in the following chapters. Finally, in order to ascertain that the model delivers realistic values, a validation must be performed. In the closing part of this chapter, possible validation techniques are briefly described.

### 2.2. SIMULATION OF MUSCLE PATH

Realistic muscle path representation is central to musculoskeletal system modeling, because muscle moment arms and lengths directly affect muscle force and moment production.
capacity, and finally joint contact forces (Dul et al., 1984; Herzog, 1992; Raikova and Prilutsky, 2001). However, our ability to measure or predict muscle path remains limited, particularly in joints like the GH joint with a complex anatomy and large range of motion. A useful biomechanical model of the shoulder complex must realistically represent the lines of action of muscles for all 3D positions of the arm (Johnson et al., 1996; Otis et al., 1994; Wickham and Brown, 1998).

In an experimental setting, moment arms can be assessed with the tendon travel methods, based on the fact that tendon excursion (corresponding to change in muscle length) is proportional to the length of its moment arm for a given change in joint angle (Equ. 1) (Hughes et al., 1998).

\[
\text{Moment arm} = \frac{\partial (\text{tendon excursion})}{\partial (\text{angle})} \quad \text{Equ. 1}
\]

The tendon travel method offers the advantage of determining moment arms without knowing the position of the center of rotation.

Moment arms have been measured experimentally on fresh shoulder cadaver specimens (Hughes et al., 1998; Kuechle et al., 1997; Kuechle et al., 2000; Langenderfer et al., 2006; Liu et al., 1997; Nyffeler et al., 2004; Otis et al., 1994; Poppen and Walker, 1978), but tissue deterioration generally confined such measurements to a single moment arm component for a few muscles in a restricted number of positions. This hampers the use of this data in muscle force prediction models, where all three components of the muscle moment arms are necessary for a broad range of joint positions. Artificial models replacing bone with full-scale epoxy replicates and musculo-tendon units with braided cords (Figure 6) overcome the problem of tissue deterioration and have yielded wider ranges of data (Favre et al., 2005; Favre et al., 2008; Favre et al., 2009a). Data collection was as time and equipment intensive as for cadaveric measurements. Experimental assessment of moment arms and lines of action of 27 muscle segments took about 4 hours for one single position, preventing the use of this technique in an automatic fashion to generate data for any arbitrary position.
Figure 6: Artificial model of the shoulder.
The bones were epoxy reproductions of real human bones, and the muscles were simulated by strings. This model was used to determine muscle lever arms and directions of muscle action (Favre et al., 2005; Favre et al., 2008; Favre et al., 2009a).

Moment arms have also been measured in vivo on active muscles, using MRI systems (Graichen et al., 2001; Juul-Kristensen et al., 2000), but difficulties with measuring dynamic movements for wide ranges of motion and large joints present critical limitations (Ruckstuhl et al., 2009). Moreover, the technique remains expensive, data post-processing time-consuming, and with the muscle path computed by a straight line, the muscle could penetrate the bone when wrapping occurred (Graichen et al., 2001). This disadvantage was tackled by combining MRI with computer aided design models, but the necessary extensive user intervention confined this method to discrete joint positions (Ruckstuhl et al., 2009).

Computer models can be automated to provide reliable prediction of moment arms over wide ranges of motion (Gatti et al., 2007). However, modeling becomes complicated when a muscle path deviates from a straight line as it wraps over surrounding anatomical features. Computerized approaches to muscle wrapping can be roughly classified into two categories, depending on the representation of the muscle and bone anatomies. In the first category, muscles are represented by deformable line segments. Here muscle segments are usually constrained to pass through via points (Delp and Loan, 1995) or, in the obstacle-set method, to wrap over simplified geometries approximating the centroidal muscle path (Charlton and Johnson, 2001; Garner and Pandy, 2000) (see Figure 7). Obstacle-set and via points have occasionally been combined (Holzbaur et al., 2005). These methods solve quickly and can deliver realistic moment arms. However, to provide valid predictions, the number and position of via points, or the appropriate obstacle type (sphere, cylinder etc.), size, and
orientation must be determined for each muscle segment at various joint positions. This becomes even more complex in joints with many muscles and multiple degrees of freedom.

Figure 7: Muscle string representation. The muscle (red line) passes through via points (left panel) or wraps on an underlying simplified obstacle (right panel). Illustration made with our own model.

In a second approach, wrapping was achieved using precise numerical interfitting of bone and muscle. For instance, an earlier attempt was made to approximate the bone as a series of cross-sectional boundaries (Gao et al., 2002). This approach solved very quickly but necessitated substantial pre-processing to define the bony cross-sections and motion path delimiters. Later approaches have used elaborate 3D volumetric finite element (FE) models to compute moment arms for certain lower limb muscles (Blemker and Delp, 2005). Many artificial boundary conditions (such as via points) were avoided by constraining the muscle within defined contact regions. This approach generally involved large computational expense and labor intensive data processing that could preclude its integration to larger musculoskeletal models (Blemker and Delp, 2005; Grosse et al., 2007; Marsden et al., 2008; Vasavada et al., 2008). These limitations limit modeling of the entire GH joint, where over 20 muscles segments being regularly considered (Charlton and Johnson, 2006; Favre et al., 2005; Garner and Pandy, 2001; Holzbaur et al., 2005; Karlsson and Peterson, 1992; van der Helm, 1994). Despite apparent higher “biofidelity”, FE approaches have predicted moment arms generally different from experimental findings (Blemker and Delp, 2005).

An intermediate technique relying on contact detection capabilities of commercial FE packages is presented (Chapter 3). The muscles will be modeled as deformable strings that wrap on 3D anatomically precise representations of the bones. Information of muscle moment arms and line of action can then be used as input for the estimation of the muscle forces required to equilibrate and stabilize the joint. In this manner, advantageous features of both approaches can be combined (Table 2).
### Background

<table>
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<tr>
<th>Model characteristics</th>
<th>Muscle representation</th>
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<th>Contact detection</th>
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<td>For joint positioning and visualization</td>
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<td>Possible by activation inactivation of via-points</td>
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<td></td>
<td>Deformable string</td>
<td>Simplified underlying geometry</td>
<td>Embedded in FE software</td>
<td>Always present and difficult to control</td>
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<td>3D volume</td>
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<td>Embedded in FE software</td>
<td>Permitted, constrained by contact between the volumes</td>
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<td></td>
<td>Deformable string</td>
<td>3D volume</td>
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<th>Muscle drifting on bone in simulated joint motion</th>
<th>Penetration of muscles through the bones</th>
<th>Multi-object wrapping</th>
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<td>Possible in multiobject wrapping</td>
<td>Requires additional algorithms</td>
<td>Simple geometric shapes</td>
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<td></td>
<td>Always present and difficult to control</td>
<td>None</td>
<td>Automatic</td>
<td>Any shape</td>
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<td>Permitted, constrained by contact between the volumes</td>
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<td>Permitted, constrained by glued node constraint.</td>
<td>None</td>
<td>Automatic</td>
<td>Any shape</td>
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| Required pre-processing for new joint positions | Very high | Very high | High | Very low |
| Computation efficiency | Very high | High | **Very low** | High |

**Table 2: Features of the obstacle-set and via-point methods, volumetric and proposed muscle wrapping approaches. The features are classified as advantageous (normal), neutral (italic) and disadvantageous (bold).**
2.3. ESTIMATION OF MUSCLE FORCE

It is established that stability of the shoulder joint must primarily be effected through muscular action because joint capsule and ligaments are generally lax throughout the greater part of GH rotation, while the joint surfaces themselves offer little resistance to dislocation (Gerber, 1992; Jössel, 1880; Kronberg et al., 1990; Lippitt et al., 1993; McMahon et al., 1995; Perthes, 1906). In addition to particular muscles called upon to equilibrate an external force, shoulder joint stability might require additional activity by the rotator cuff muscles to provide what is now commonly termed ‘concavity compression’ (Lippitt et al., 1993). Therefore, it becomes important to understand the strategy adopted by muscles in balancing a given external force, while preventing dislocation of the GH joint.

The complexity of the shoulder joint, with its extensive range of motion and large number of muscles, necessitates a close examination of the possible muscular stabilization strategies. This can help understanding several shoulder disorders and treat these more effectively. Estimation of muscle loading is critical in predicting when tendon tears will occur. When tears have already been diagnosed, this information is required to develop strategies for safely reconstructing them, or to optimize rehabilitation by focusing on the remaining viable muscles. It is also necessary to understand how the central nervous system controls the musculoskeletal system during specific tasks in order to best treat neurological or muscular deficiencies. Finally, accurately estimating the forces at the joint contact interface is fundamental to prosthesis design in preventing accelerated implant wear and premature failure.

To date, direct experimental measurement of muscle forces in vivo remains restricted to superficial tendons (Achilles tendon), or during an operation when the tendon is exposed (Erdemir et al., 2007). These measurements are in nature not applicable in a clinical setting.

In order to simulate how muscles are coordinated to move the arm or to hold it in a certain position, two parameters must be defined (Figure 8): first, an appropriate set of active muscles (muscle recruitment) and second, the distribution of force among these muscles to reach equilibrium.
Figure 8: Muscle force determination.
The external force must be balanced by a defined set of active muscles that apply a given force. Drawing by Leonardo Da Vinci, Anatomical studies of the shoulder (1510-11).

Several investigators, from the second half of the 19th century up to recently (Bassett et al., 1990; De Luca and Forrest, 1973; Fick, 1850; Fick, 1910; Murray et al., 2000; Pauwels, 1965), have considered the muscle force to be directly proportional to the muscle physiological cross-sectional area, under the assumption that a muscle with a larger cross-sectional area can exert a greater force.

Early models of shoulder muscle force balance were two-dimensional systems that considered only a few muscles, and could be solved by simple analytical methods. The deltoid and a resultant rotator cuff force required for arm elevation were estimated based on a vector analysis (Inman et al., 1944). Contact forces were approximated using anatomical data measured on cadavers and EMG signals from living subjects (Poppen and Walker, 1978). Also, the shoulder was simplified with the assumption that the middle deltoid was the only actuator during planar elevation (Ringelberg, 1985). Deltoid, supraspinatus and contact forces were estimated for working postures requiring elevated upper arm positions (Dul, 1988). In these models, the system was governed by the same number of equations and unknowns, so that a unique solution could be found. While these models were limited to a simple GH movement (elevation) and included a restricted set of muscles, they nonetheless delivered important insight into some basic working principles of shoulder function. More importantly, they provided a technical foundation on which later
models could be built. Because of asymmetry and degrees of freedom of the shoulder joint, 3D models were soon needed. Also, the necessity to simulate more complex aspects of shoulder biomechanics led to correspondingly more complicated models. Nowadays, the use of sophisticated mathematics and the large amount of data to be treated have made computers mandatory.

When comprehensively modeling the muscular forces at the shoulder, the main hurdle is of a mathematical nature. In the simplified case that the GH joint is represented as a ball-and-socket (thus removing/constraining the three translational degrees of freedom), the system is governed by three equations describing the equilibrium in each rotational degree of freedom (see Eq. 9, Chapter 4). The muscle forces are treated as unknowns in these equations. This system of three equations would have one unique solution only if no more than three unknowns were present. In actuality, the many muscles that cross the GH joint present a much less defined system. Muscles with large origins are often subdivided to allow for differentiated action because all fibers in a same muscle do not necessarily contract simultaneously (Fick and Weber, 1877; Wickham and Brown, 1998). Additionally, moment arm length and direction of tendon action can vary considerably within an individual muscle (Favre et al., 2005; Favre et al., 2009a; van der Helm and Veenbaas, 1991). Therefore, current shoulder models regularly include more than 20 muscle segments, yielding a mathematically indeterminate system with more unknowns than equations, and are characterized by an infinite number of possible solutions that yield equilibrium.

The main issue is to find a method that systematically isolates a suitable solution among this infinite number. Most computational methods tackle the indeterminacy issue using mathematical optimization. The forces are distributed among muscles in a way that a chosen “cost function” is minimized, while ensuring that these forces satisfy given physical constraints. The cost function can be based upon muscle stress, force, energy, fatigue, etc. and the constraints usually dictate that the muscle forces can apply tension only, must satisfy the equilibrium equations and fall within reasonable upper (tetanic muscle force) and lower (negligible force) muscle force limits (Karlsson and Peterson, 1992). The first optimization models were developed to simulate muscular load sharing at the wrist (Penrod et al., 1974) or during gait at the hip or knee (Crowninshield and Brand, 1981; Hardt, 1978; Seireg and Arvika, 1975). Since then, these techniques have been applied to most joints,
but again, certainly because of the joint complexity, the shoulder represents one of the most recent applications. Only in the mid-1990’s, optimization methods were applied to estimate shoulder muscle forces. Karlsson and Peterson first extensively described the method to build a 3D model of the shoulder (Karlsson and Peterson, 1992), followed by van der Helm who developed the so far most cited and used shoulder model (van der Helm, 1994b). In this last model, the results obtained when minimizing the sum of the squared forces, the sum of the squared stresses, the sum of the squared forces normalized to the maximal muscle force or the maximal stress in the entire mechanism were compared. The sum of the squared forces criterion led to unsatisfying results while the three others did not show great differences. Minimization of the sum of squared muscle stresses was the preferred criterion, based on computational efficiency. In later studies, muscular energy consumption (metabolic cost) during fast goal directed movements was minimized (Happee and Van der Helm, 1995). Other models followed, using minimization of the sum of squared muscle force (Happee, 1994; Nieminen et al., 1995a) or squared muscle stresses (Hogfors et al., 1995; Hughes and An, 1996). Mixed optimization criteria have then been used for analyzing endurance-type activities. At low load levels, the sum of the squared muscle forces was minimized. At higher load levels, time elapsed from the start of the activity decreased the allowable muscle stress on the basis of stress-endurance time curves, and a minimum-fatigue criterion was used (Niemi et al., 1996; Nieminen et al., 1995b). Optimal posture (Schouten et al., 2001) and trajectory (Ohta et al., 2004) of the arm were studied on two-dimensional models including 6 muscles, also using combined minimization criteria.

The previous paragraph illustrates that one of the major problems and difficulties in optimization techniques is the choice of an appropriate cost function. Comparisons with EMG activity have shown that some cost functions predict on-off muscle recruitment patterns better than others, but it is not yet clear what principles would dictate how muscles are recruited in vivo (Erdemir et al., 2007; Karlsson and Peterson, 1992; van der Helm, 1994a). Furthermore, muscle load sharing strategy is likely to differ between subjects, depends to some extent on the type of activity, and varies in response to fatigue, mental demands, visual feedback or in the presence of musculoskeletal disorders, such as rotator cuff tear (Jensen et al., 2000; Kronberg et al., 1991; Steenbrink et al., 2006). Finally, co-contraction of antagonistic muscles can theoretically be predicted (Jinhaa et al., 2006), but it
has been shown to confound conventional optimization-based models (Cholewicki et al., 1995; Dickerson et al., 2008).

In an attempt to overcome the drawbacks of analytical optimization techniques, experimentally based, EMG-driven models that were first developed for the lower back (McGill, 1992) have been extended to the shoulder (Koike and Kawato, 1995; Langenderfer et al., 2005; Laursen et al., 1998). In such models the set of active muscles is identified in recordings of EMG activity for isolated shoulder positions. The muscle force is then estimated by assuming a linear relationship between EMG and force under isometric conditions.

\[ \text{Force} = f(\text{EMG signal}) \]  
\text{Equ. 2}

One advantage is that the recruitment of a particular muscle is directly indicated by the EMG measurements, so that realistic muscle recruitment patterns can be immediately implemented into the model. In this way, co-contraction of antagonistic muscles may be directly accounted for and simulated. The main weakness of such models is that reliable EMG recordings of all relevant muscles must be available for the simulated position. This is technically demanding (if even possible) and brings along the well known difficulties inherent in EMG signal measurement and processing (De Luca, 1997; Inman et al., 1952; McMahon et al., 1995).

Despite limitations of these methods, and the difficulties in validating such models (to be discussed later), muscle force estimation models have been applied successfully to investigate a variety of clinical questions. They have been used to evaluate the influence of a prosthesis on the muscular forces (de Leest et al., 1996), to assess the influence of scapular neck malunion on shoulder function (Chadwick et al., 2004), to test tendon transfers (Magermans et al., 2004a; Magermans et al., 2004b) and for prevention of overload injuries in wheelchair design and propulsion (Lin et al., 2004; van der Helm and Veeger, 1996; van Drongelen et al., 2005; van Drongelen et al., 2006; Veeger et al., 2002). Because muscle force estimation models must necessarily entail a muscle recruitment strategy, they give explicit insight into how the central nervous system drives the musculoskeletal system. Changes in the applied cost function or recruitment criteria directly affect the prediction of
recruited muscles, and can be varied to explore neuromuscular control hypotheses. The effect of a neurological or muscular deficiency can be assessed by scaling the allowed maximum relative force of selected muscles (van Drongelen et al., 2006). In a forward dynamics approach, muscle forces can be used to calculate joint torques and trajectory (Koike and Kawato, 1995), and combined with rigid body models with the advantage of motion visualization.

Most of the above mentioned computational methods have unfortunately been tailored to meet special cases of loading (usually to balance an external moment acting in one principal direction). They therefore do not readily lend themselves for application in a clinical environment where one wishes to apply an arbitrary combination of external loading conditions. In an effort to avoid the previously described limitations that accompany traditional muscle force estimation methods (choice of an appropriate cost function, co-contraction prediction, requirements for EMG measurements) we have developed in this thesis an algorithm to predict the muscle forces required for shoulder joint equilibrium and stabilization. This algorithm implements a novel recruitment strategy that focuses on selecting muscles with a relative mechanical advantage, and a corresponding set of muscles that counterbalance secondary joint moments. The muscle forces computed with this method are then used as input in a deformable model of GH joint contacts.

2.4. MODELING OF JOINT CONTACTS

Tissue deformations at the joint are relevant to very important questions in shoulder biomechanics. Location, extent and magnitude of joint contact pressure can for instance predict onset of potential arthritis (Buchler et al., 2002). Quantifying mechanical bone stresses is essential in arthroplasty, where excessive stress concentration may indicate failure of the implant, while stress shielding could point towards bone resorption (Buchler and Farron, 2004; Couteau et al., 2001; Farron et al., 2006; Gupta et al., 2004b; Gupta et al., 2004a; Hopkins et al., 2004; Lacroix and Prendergast, 1997; Lacroix et al., 2000; Murphy and Prendergast, 2005; Terrier et al., 2005; Terrier et al., 2006). Realistic modeling of cartilage and labrum deformations is fundamental for simulating GH joint stability, as these
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structures adapt to the joint reaction force and motion of the humeral head within the glenoid (Soslowsky et al., 1992).

The FE method can simulate the deformations of complex systems that are otherwise difficult to assess and has been used to address a broad range of problems in the field of biomechanics and orthopaedics (Huiskes and Hollister, 1993). Most FE models of the shoulder have been developed to specifically investigate implant glenoid fixation and loosening, generally relating excessive bone or cement stresses to failure (Buchler and Farron, 2004; Couteau et al., 2001; Farron et al., 2006; Gupta et al., 2004b; Gupta et al., 2004a; Hopkins et al., 2004; Lacroix and Prendergast, 1997; Lacroix et al., 2000; Murphy and Prendergast, 2005; Terrier et al., 2005; Terrier et al., 2006).

The main limitation of existing deformable models lies in the definition of boundary conditions that are employed. There exist to date no 3D deformable model of the GH joint implementing realistic constraints that allow 3 rotational and 3 translational degrees of freedom of the humerus. These simplifications prevent translations at the GH joint, and thus limit computation of accurate contact characteristics in both the stable and unstable joint. A first step away from the ball-and-socket assumption has been made by allowing limited humeral translations during axial rotation (Buchler et al., 2002). The influence of the shape of the humeral head on the stress distribution in the scapula was quantified with the intent to compare a normal to an osteoarthritic shoulder. Abduction and flexion were not permitted and vertical translations were restricted by a non-physiological spring. The joint was actively centered in another model (Terrier et al., 2007; Terrier et al., 2008), but was limited to 2D analysis of the deltoid and rotator cuff muscles in abduction. Shoulder muscles can generate considerable joint moment components in the other two rotational DOF, and these cannot be neglected (Favre et al., 2009b; Veeger and van der Helm, 2007). Moreover, the large variability in daily life movements and loading conditions at the GH joint call for models that have not been tailored to meet special cases of loading and motions.

Although realistic, kinematically unconstrained models of the GH joint are clearly required to investigate GH joint contacts (Favre et al., 2009b; Hill et al., 2008; Veeger and van der Helm, 2007), validated six DOF models are lacking from the literature. The purpose of this study was to develop (to our knowledge) the first 3D finite element (FE) model of the GH
joint without \textit{a priori} defined kinematic constraints. The anatomically precise FE model is combined with the muscle path prediction and muscle force estimation approaches developed in this thesis to implement active muscle driven humeral positioning and stabilization at any joint position to realistically simulate joint contacts. This framework is finally exploited to simulate elevation as a composite of instantaneous positions.

2.5. MODEL VALIDATION

To efficiently address scientific and clinical questions through a modeling approach, the goal is not to build a perfectly realistic model, but a sufficiently accurate representation of the aspect to be investigated. The complexity of a biological system like the shoulder is such that some simplifications must be made in the modeling process, although each assumption introduces a potential source of error. To ensure that the imposed simplifications have not diminished the veracity of the simulation, validation is essential. The validation process confronts simulation results with reality, or when this is not directly possible, with controlled experiments that approximate reality. Simulation results are compared against measureable parameters with the aim to identify inappropriate assumptions or simplifications that can then be corrected to improve the fidelity of the model.

Validating muscle path

A first validation can be done qualitatively, ensuring visually that all muscles follow a physiologic path and that the muscles do not pass through the bones. Quantitatively, models used for the generation of moment arms can be directly compared to \textit{in vitro} experiments (Ackland et al., 2008; Favre et al., 2009a; Hughes et al., 1998; Kuechle et al., 1997; Langenderfer et al., 2006; Liu et al., 1997; Nyffeler et al., 2006a; Otis et al., 1994; Poppen and Walker, 1978) or \textit{in vivo} measurements using MRI (Graichen et al., 2001; Juul-Kristensen et al., 2000; Ruckstuhl et al., 2009).

Validating muscle force estimation models

In an indeterminate system of equations, it falls on the modeler to restrict the number of considered solutions to those that can realistically occur in vivo. As described above, muscle
force estimation models deliver two main pieces of information: the set of recruited muscles and the force that these exert. EMG signals can be used to compare the timing of the muscle activity (Happee and Van der Helm, 1995) and verify that the model activates a reasonable set of muscles (Happee, 1994; Happee and Van der Helm, 1995; Karlsson and Peterson, 1992; Niemi et al., 1996; Nieminen et al., 1995a; van der Helm, 1994b) or could also allow comparing paralysis experiments (either pathological or neurotoxin induced) against model behavior when specific muscles are removed from consideration. On the other hand, we are currently limited in our ability to quantify individual muscle force magnitude since no in vivo muscular force measurement devices are currently available (Erdemir et al., 2007) and EMG amplitude is a poor measure for validating the forces obtained by musculoskeletal models (Inman et al., 1952; van der Helm, 1994b). In vivo measurements of maximum isometric muscle torques exerted at the joint and/or the resultant kinematics can be used for comparison, although these only validate the global simulation quality and do not yield specific information on individual muscle forces (Garner and Pandy, 2001). Given the lack of reliable muscle force measurement techniques, results have been generally compared to those obtained with previous models. However, this approach cannot be considered an adequate muscle force validation, but merely serves as an indication that two models represent reality in a similar way. Since this approach cannot root out problems with widely held assumptions, there is obviously a pressing need for better techniques to quantify in vivo muscle forces. Validation of the forces predicted by numerical musculoskeletal models of the shoulder has now been enhanced with direct measurements of GH contact forces using an instrumented implant (Bergmann et al., 2007).

Validating models of joint stability

In the past, validation often compared model results to experiments performed on cadavers. In order to validate the geometry of the joint contacts, stability ratios can be compared with experimental values (Halder et al., 2001; Lazarus et al., 1996; Lippitt and Matsen, 1993). Emerging technologies in imaging techniques and instrumented implants now allow direct comparison to selected in vivo measurements, bringing the models to a higher level of accuracy. Their use for validation of shoulder models is nascent (Dubowsky et al., 2008; Terrier et al., 2008a), but is destined to increase. Rapid progress in the field of medical imaging opens new possibilities to measure rotation and translation of the GH joint
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during in vivo movements on radiographs (Chopp et al., 2010), bi-plane x-ray images (Bey et al., 2008), open MRI (Graichen et al., 2000; Graichen et al., 2005; Sahara et al., 2007) or fluoroscopy (Teyhen et al., 2010). These quantities provide useful global criteria for a partial assessment of model performance of contact models.

2.6. SUMMARY

The currently available numerical methods for simulating muscle path, muscle forces and joint contacts are impaired by critical weaknesses that prevent their inclusion to a realistic model of GH stability.

Muscle path prediction techniques are currently associated with inaccuracies, computational expense or labor intensive pre-processing. Existing muscle force estimation techniques are limited by the necessary intricate choice of an appropriate cost function, their lack of ability to predict co-contraction, or the major requirements inherent to EMG measurements. State-of-the-art joint contact models simplify the GH joint by constraining humeral head kinematics, limiting their use in the study of joint stability. In addition, most of the above mentioned computational methods have unfortunately been tailored to meet special cases of loading in given joint positions or consider only a subset of muscles crossing the GH joint.

In the next chapters, novel methods are developed for muscle path prediction (chapter 3), muscle force estimation (chapter 4) and joint contacts simulation (chapter 5), alleviating most weaknesses of the current approaches. These components are combined to provide a realistic, integrated model of GH joint stability (chapter 5) that should be able to cope with the wide GH joint range of motion in all degrees of freedom and the considerable variability in external loading conditions as encountered in daily life activities. This is especially important for clinical applications, where a realistic representation in more than one single special case of loading or for more than one unique motion is necessary. Each model component and the integrated model itself are thoroughly validated.
REFERENCES


Background


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Background


CHAPTER 3
3. AUTOMATED MUSCLE WRAPPING USING FINITE ELEMENT CONTACT DETECTION

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Automated muscle wrapping using finite element contact detection.
ABSTRACT

Realistic muscle path representation is essential to musculoskeletal modeling of joint function. Algorithms predicting these muscle paths typically rely on a labor intensive predefined definition of via points or underlying geometries to guide wrapping for given joint positions. While muscle wrapping using anatomically precise 3D FE models of bone and muscle has been achieved, computational expense and pre-processing associated with this approach exclude its use in applications such as subject-specific modeling. With the intention of combining advantageous features of both approaches, an intermediate technique relying on contact detection capabilities of commercial FE packages is presented. We applied the approach to the GH joint, and validated the method by comparison against existing experimental data.

Individual muscles were modeled as a straight series of deformable beam elements and bones as anatomically precise 3D rigid bodies. Only the attachment locations and a default orientation of the undeformed muscle segment were pre-defined. The joint was then oriented in a static position of interest. The muscle segment free end was then moved along the shortest Euclidean path to its origin on the scapula, wrapping the muscle along bone surfaces by relying on software contact detection. After wrapping for a given position, the resulting moment arm was computed as the perpendicular distance from the line of action vector to the humeral head center of rotation.

This approach reasonably predicted muscle length and moment arm for 27 muscle segments when compared to experimental measurements over a wide range of shoulder motion. Artificial via points or underlying contact geometries were avoided, contact detection and multiobject wrapping on the bone surfaces were automatic, and low computational cost permitted wrapping of individual muscles within seconds on a standard desktop PC. These advantages may be valuable for both general and subject-specific musculoskeletal modeling.

Keywords: muscle wrapping, muscle path, moment arm, musculoskeletal modeling, shoulder, muscle length.
3.1. INTRODUCTION

As described in the Background chapter, existing numerical approaches for muscle path modeling involve a high degree of complexity and an extensive need for user intervention. This prevents their use in a more automatic fashion, severely limiting their utility for parametric joint studies or subject-specific modeling. In the via point (Delp and Loan, 1995) and obstacle-set approaches (Charlton and Johnson, 2001; Garner and Pandy, 2000), the complexity lies in representing the underlying guiding geometries, as well as implementing rules for wrapping and the associated algorithms for contact definition. Furthermore, multiobject wrapping is often complicated by muscle penetration through bone (Audenaert and Audenaert, 2008; Charlton and Johnson, 2001; Marsden and Swailes, 2008a; Marsden et al., 2008b). In a volumetric 3D FE approach (Blemker and Delp, 2005), the complexity rather lies in a computationally intensive modeling of the muscle. We hypothesized that multi-object wrapping on any shaped bone can be managed while avoiding a priori definition of via points or wrapping geometries by relying on automatic contact detection capabilities in commercial FE software. Further, by using a simplified muscle representation, we assumed that low computing costs would allow the model to solve quickly on a standard desktop PC. These advantages would be valuable in both general and subject-specific modeling, where efficient, flexible, and automatic multi-object wrapping is often required.

3.2. METHODS

In the presented approach (Figure 9), 3D bone surfaces are first oriented in a static position of interest. The muscles are modeled as a straight series of beam elements with very low bending stiffness. They are anchored at their insertion to the humerus but not yet attached at their origin on the scapula. A wrapping step then displaces the muscle origin node toward its predefined origin on the scapula. Along the way, the muscle elements wrap around relevant bone surfaces, relying on contact detection of the FE software. This sequence can later be repeated for any joint position and can be implemented incrementally to simulate motion as a series of static positions.
Figure 9: Schematic of the basic working principles of the wrapping method. The muscle path is computed in one static position. For a full motion decomposed to a series of static joint positions, the joint angle was changed incrementally. The shaded steps were automated by means of Matlab routines.
Geometries of the humerus and scapula were imported from the Bel repository (Van Sint Jan et al., 2004) into Geomagic Studio 8 (Geomagic, Inc., Research Triangle Park, NC, USA). From the initial triangulated surface mesh, a smoothed NURBS surface was created for each bone using the autosurface function with default parameters. These surfaces were imported as rigid bodies into Marc Mentat 2007r1 (MSC Software, Santa Ana, CA).

The articular surface of the humeral head is part of a sphere (Boileau and Walch, 1997). A sphere was therefore fitted to the humeral head in a least square sense following the method described in (Meskers et al., 1998). A set of 40 points was selected on the humeral joint surface. The radius and the center of a sphere leading to the minimal error between the radius and the distance of each of these 40 points to the sphere center were then sought by minimizing the function:

\[
\sum_{i=1}^{40} \sqrt{((x(i) - C(x))^2 + ((y(i) - C(y))^2 + ((z(i) - C(z))^2) - r)^2}
\]

where \(r\) is the radius of the sphere; \(x(i), y(i), z(i)\) are the coordinates of the point \(i\) chosen on the humeral joint surface, and \(C(x), C(y), C(z)\) are the coordinates of the sphere center.

A 24 mm radius was estimated with this method, which falls within reported values for average adult shoulders (Boileau and Walch, 1997; Iannotti et al., 1992). The humeral head rotation center found by sphere fitting was then positioned with respect to the scapula by linear regression (Meskers et al., 1998).

The coordinate system for both bones and their relative position in the joint was defined according to the recommendations of the International Society of Biomechanics (Wu et al., 2005), (see Figure 10). Based on bony landmarks, a coordinate system was defined for each bone. These were then aligned to orient the humerus relative to the scapula.
Figure 10: Definition of the bone coordinate systems.
(Xs, Ys, Zs) stand for scapula, (Xh, Yh, Zh) for the humerus. The coordinate systems were aligned according to the recommendations of the ISB (Wu et al., 2005).

The humerus was rotated relative to the scapula around the humerus center in the joint configuration of interest, assuming a fixed rotation center (Meskers et al., 1998; Veeger, 2000). Rotation sequences of the shoulder were described using Euler angles (plane of elevation, elevation, axial rotation) as shown in Figure 11, i.e. using a spherical coordinate system (Wu et al., 2005).

Figure 11: Definition of GH motion components.
This was done in Euler angles, according to the International Society of Biomechanics (Wu et al., 2005). The plane of elevation motion takes place about the fixed Zs axis (see Figure 10), elevation occurs around Yh, and axial rotation takes place around the Zh axis.
Because the FE software uses a cartesian coordinate system, Euler rotations were decomposed in a series of rotations in a Cartesian system to rotate the humerus. A series of three rotations must be performed, one for each rotational degree of freedom, in the sequence plane of elevation (angle $\alpha$), elevation (angle $\beta$) and axial rotation (angle $\gamma$). First, the humerus is rotated by an angle $\alpha$ for the plane of elevation about $Z_s$.

In matrix notation, this gives:

$$
\begin{bmatrix}
\cos \alpha & -\sin \alpha & 0 \\
\sin \alpha & \cos \alpha & 0 \\
0 & 0 & 1
\end{bmatrix}
$$

Equ. 4

After this first rotation, the axis for the second rotation (elevation about $Y_h$) was rotated together with the humerus by an angle $\alpha$. To then achieve elevation about this rotated axis, the simplest manner is to rotate the humerus again to its initial configuration, perform the elevation about the known axis $Y_h$, and rotate back in the plane of elevation. Rotations being orthonormal, we have

$$R^{-1} = R^T$$

Equ. 5

To achieve elevation, the matrix form is then

$$
\begin{bmatrix}
\cos \alpha & -\sin \alpha & 0 \\
\sin \alpha & \cos \alpha & 0 \\
0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
1 & 0 & 0 \\
0 & \cos \beta & -\sin \beta \\
0 & \sin \beta & \cos \beta
\end{bmatrix}
\begin{bmatrix}
\cos \alpha & \sin \alpha & 0 \\
-\sin \alpha & \cos \alpha & 0 \\
0 & 0 & 1
\end{bmatrix}
$$

Equ. 6

In a similar manner, to achieve axial rotation of the humerus around its axis, the humerus is set back in its initial configuration to rotate about $Z_h$ and then rotated again by $\alpha$ and $\beta$ using Equ. 7:

$$
\begin{bmatrix}
\cos \alpha & -\sin \alpha & 0 \\
\sin \alpha & \cos \alpha & 0 \\
0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
\cos \gamma & -\sin \gamma & 0 \\
\sin \gamma & \cos \gamma & 0 \\
0 & 0 & 1
\end{bmatrix}
\begin{bmatrix}
\cos \alpha & \sin \alpha & 0 \\
-\sin \alpha & \cos \alpha & 0 \\
0 & 0 & 1
\end{bmatrix}
$$

Equ. 7

All muscles crossing the GH joint were considered and were segmented to allow for differentiated actions. Some were divided into two, others into three parts. This was especially necessary in muscles with broad insertions because moment arm length and direction of tendon action vary considerably within a given muscle (Favre et al., 2009;
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Wickham and Brown, 1998). Each muscle segment was modeled as 150 deformable, two-node, straight, elastic beam elements in series (Element number 98, (MSC.Software)) as shown in Figure 12.

Figure 12: Deformable string representation. Each muscle is modeled by a series of 2D elements (close-up)

The teres major, coracobrachialis, triceps, biceps short and long head were each modeled with one segment. The rotator cuff muscles were represented using two supraspinatus, three infraspinatus, three subscapularis, and two teres minor segments. The latissimus dorsi and the pectoralis major were each composed of three segments and the deltoid was divided into six segments (one anterior, three middle, and two posterior deltoid segments). Origin and insertion sites for each muscle segment were identified on the bone surface as described in previous studies (Favre et al., 2005; Favre et al., 2009)(see Figure 13).

Figure 13: Muscles origin and insertion. Initial configuration of all muscle segments. The origin points are depicted by the stars. A: Supraspinatus, B: Subscapularis, C: Infraspinatus, D: Teres minor, E: Deltoid, F: Triceps, G: Coracobrachialis H: Teres major, I: Latissimus dorsi, J: Pectoralis major, K: Biceps.
Each attachment footprint was divided into equal parts according to the number of muscle segments. Typically, if the muscle was segmented into medial and lateral segments, the medio-lateral width of the attachment area was halved, and each segment was first positioned to attach at the respective center of the subarea. The exact origin and insertion sites, not being known in advance, were varied within the designated muscle attachment footprint until a suitable agreement was reached with experimentally reported moment arms. All origin and insertion coordinates are listed in Table 3.

<table>
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<th>Normalized vector</th>
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<td>Biceps short head</td>
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<td>0.0559</td>
<td>-0.7372</td>
</tr>
</tbody>
</table>

Table 3: Origin and insertion coordinates, and normalized initial orientation vector of all muscle segments. The origin and insertion coordinates are given in a reference frame according to the ISB guidelines (Wu et al., 2005), with the joint in the resting position. All values are normalized to the length of the humerus (323.2mm).
After each segment was attached to the humerus at its insertion site, the initial segment orientation was set as perpendicular to the humeral surface. In a few cases, the default segment orientation was adjusted to avoid non-physiological bone surface contact during the wrapping process. Here, the humerus was rotated to the end of the physiological range of motion in axial rotation (Rundquist and Ludewig, 2004) and elevation (limited by inferior and superior impingement). If a pre-wrapped muscle segment was seen to penetrate the scapula, or would later obviously wrap on an inappropriate bone surface, the segment was rotated backward in the plane of motion until the problem disappeared. For the infraspinatus and teres minor, which insert close to the glenoid border, adjustment of initial orientation was necessary to prevent scapula penetration in extreme external rotation. The posterior deltoid was similarly adjusted to avoid contact in external rotation. All other segments were left in their perpendicular orientation. All initial orientation vectors are given in Table 3. Once set, the same pre-wrapping orientation relative to the coordinate system was utilized for all subsequent wrapping simulations, regardless of joint position. To assess sensitivity to initial orientation, both supraspinatus segments were systematically re-oriented 10° from the default orientation towards the medial, lateral, anterior and posterior directions. The change in moment arm and muscle length was monitored during scapular elevation from 0 to 80 degrees, and then averaged.

Insertion nodes were constrained to prohibit translations, but allowed three degrees of freedom in rotation. The free end node of each segment was then moved along the shortest Euclidean distance towards its designated point of origin on the scapula. Contact between the bone surface and the muscle nodes were defined. Non-physiologic muscle wrapping on the acromion and coracoid process was avoided by removing them from the scapula contact body (Figure 14), allowing the muscles to penetrate these structures when wrapping.
During simulations the muscle nodes were automatically controlled for contact with the bone surface. The muscle segments deformed and wrapped automatically on contacting bone surfaces. A glued node constraint between contacting muscle-bone nodes restricted relative tangential motion after contact. This prevented muscle drift on the bone during the wrapping process for a given static joint position of interest.

Muscle moment arms were computed at 95% completion of the wrapping process, slightly shifting the muscle origin from the bone surface (Bremer et al., 2006; Favre et al., 2005; Favre et al., 2008; Favre et al., 2009). The vector from the last muscle node in contact with the bone (automatically detected using the contact status option (MSC.Software)) to the tenth node towards the origin defined the muscle line of action (Figure 15).

**Figure 14:** Removal of acromion and coracoid process from scapula contact body.

**Figure 15:** Wrapping of one muscle segment on the humeral bone surface. The muscle is shown in its starting (A), intermediate (B) and final (C) positions. The muscle line of action (D) was computed by taking the vector linking the last point of contact on the bone to the tenth node towards the origin.
The moment arm was computed as the perpendicular distance from this vector to the humeral head center. A Matlab routine (v.7.0.1, The MathWorks, Natick, MA, USA) automated the entire procedure and calculated muscle wrapping for incrementally applied changes in joint position.

The method was validated against published experimental data of shoulder muscle lengths (Holzbaur et al., 2007; Klein Breteler, 1996; Langenderfer et al., 2004; Veeger et al., 1997) and moment arms (Favre et al., 2005; Hughes et al., 1998; Kuechle et al., 1997; Kuechle et al., 2000; Langenderfer et al., 2006; Liu et al., 1997; Nyffeler et al., 2004; Otis et al., 1994; Poppen and Walker, 1978). The moment arms were computed over a series of discrete joint positions in 10° increments to describe the corresponding experimental motion. Muscle segment length was calculated as the summation of individual element lengths in the joint rest position (0° elevation, neutral axial rotation).

Finally, moment arms of the rotator cuff muscles in two joint positions were compared with previous modeling results (Charlton and Johnson, 2006; Dickerson et al., 2007; Garner and Pandy, 2000; Holzbaur et al., 2005; van der Helm, 1994; Webb et al., 2007).

3.3. RESULTS

Wrapping for a given joint position was determined within 10 seconds for single muscle segments and in less than 4 minutes for 27 muscle segments on a standard desktop PC (Intel Pentium 4, 3GHz, 1 GB RAM).

Qualitative comparison of predicted muscle paths against an established experimental shoulder model (Favre et al., 2005; Favre et al., 2009) indicated that the approach consistently generated anatomically reasonable results, without muscle penetration of bone surfaces (Figure 16).
Figure 16: Visual validation of muscle path with epoxy model. Wrapping for the humerus in 10° (top) and 50° (bottom) elevation in the scapular plane and in neutral axial rotation. Percentages indicate the portion of the wrapping process (0% represents the initial configuration, 95% the configuration at which muscle moment arms were computed). The right panels show the artificial full-scale replicate (Favre et al., 2005) in the same joint position.

Quantitative comparison indicated realistic abduction moment arm predictions during humerus elevation in the scapular plane (Figure 17), as well as axial rotation moment arms during humerus axial rotation at 0° elevation (Figure 18) when compared to published in vitro data.
Figure 17: Quantitative validation of moment arm for elevation.
Elevation moment arms of all muscle segments (black lines) compared to published data measured in vitro (individual studies shown with grey lines, area of experimental values is shaded) during elevation of the humerus in the scapular plane (Favre et al., 2005; Hughes et al., 1998; Kuechle et al., 1997; Liu et al., 1997; Nyffeler et al., 2004; Otis et al., 1994; Poppen and Walker, 1978). The joint angles are given for a fixed scapula. Positive moment arm values stand for abduction, negative for adduction. The description of the horizontal axis is shown for the first panel only. Bottom right: model at the corresponding 10° steps. Note the permitted muscle sliding between distinct positions.
Figure 18: Quantitative validation of moment arm for axial rotation. Axial rotation moment arm of all muscle segments compared to published data measured in vitro (individual studies shown with grey lines, area of experimental values is shaded) during axial rotation of the humerus with the arm at 0° elevation (Kuechle et al., 2000; Langenderfer et al., 2006). The joint angles are given with a fixed scapula, negative values of humerus position (x-axis) and moment arm (y-axis) stand for internal axial rotation. The description of the horizontal axis is shown for the first panel only. Bottom right: model at the corresponding 10° steps. Note the allowed muscle sliding between distinct positions.

Moment arms of the rotator cuff muscles in two joint positions (Figure 19) were also in agreement with previous modeling approaches.
Figure 19: Comparison of moment arms with other numerical models. The moment arms of the rotator cuff muscles obtained with other published models, as reported previously (Gatti et al., 2007) for 30 and 60° GH elevation yield similar values. The models by Garner, DSEM, Newcastle, and Dickerson implemented wrapping on underlying geometries. The model by Holzbaur combines the obstacle-set and via-point methods. The model by Webb represents a volumetric model of the supraspinatus. The shaded area represent the range obtained from experimental studies (Gatti et al., 2007).

All predicted muscle lengths (Figure 20) fell within one standard deviation of corresponding experimental values. Supraspinatus moment arm and length were relatively insensitive to initial segment orientation. The moment arm varied an average of 1.2 mm and the length 1.1 mm in response to 10° perturbations from the default initial orientation.
Figure 20: Muscle length validation.
Comparison of predicted muscle lengths with reported data obtained for the resting arm position using magnetic resonance imaging (Holzbaur et al., 2007) or from cadaver dissections (Klein Breteler, 1996; Langenderfer et al., 2004; Veeger et al., 1997). The shaded gray area spans the plus minus standard deviation range of experimental values. The abbreviation sup stands for superior, inf for inferior, mid for middle, ant for anterior, post for posterior.
3.4. DISCUSSION

The presented approach for predicting muscle path yielded realistic muscle lengths and moment arms in an automatic fashion for 27 muscle segments over a wide range of shoulder motion. This method combines a computationally efficient muscle representation with commercial FE contact capabilities, eliminates the need for predefined via points and wrapping surfaces, and thus enables an automatic, penetration-free, multi-object wrapping of muscle on complex bone geometries.

Four steps have been suggested to test musculoskeletal model accuracy (Delp and Loan, 1995), with two steps related to muscle path modeling. First, the muscle path should be physiological, with no interpenetration of bone and muscle. Bone-muscle penetration was successfully (and automatically) avoided by the FE software contact detection algorithms. Although inter-muscle contact constraints were not imposed, no aberrant muscle paths were visually identified over the tested range of joint positions.

In the second test of accuracy, muscle moment arms should be compared to experimental data. Appropriate moment arms were predicted when compared against a large experimental dataset as well as against published modeling results. In a few isolated cases (the middle and anterior deltoid in Figure 17, the deltoid and infraspinatus in Figure 18) the average moment arm curve fell outside of the envelope defined by published experiments, but the error remained within ranges that have previously been considered tolerable (Arnold et al., 2000; Blemker and Delp, 2005; Holzbaur et al., 2005). Anatomical variations including size, age or bone geometry may account for such discrepancies. With its origin on the acromion, the path of the middle deltoid depends on the degree of lateral extension (Nyffeler et al., 2006) and acromial shape (Bigliani et al., 1986; Gagey et al., 1993). The anterior deltoid, the biceps short head and coracobrachialis are influenced by the anatomy of the coracoid process (Bhatia et al., 2007; Gerber et al., 1987). Methodologically, discrepancies may also stem from variations in muscle segmentation or designation of the joint center and relative bone positioning. This will diminish as the ISB recommendations (Wu et al., 2005) gain more widespread usage. While we considered all Medline indexed studies that included the muscles of interest with experimentally measured moment arms over a rotation range of at least 60°, there was a scarcity of experimental data for the
Muscle wrapping

triceps, coracobrachialis, both heads of the biceps (Figure 17), the subscapularis, teres major, latissimus dorsi, and for the pectoralis major (Figure 18), for which only one study could be found describing scapular elevation (Favre et al., 2005), and one study describing axial rotation (Kuechle et al., 2000).

As with volumetric models, the main benefit of our method over the obstacle-set method is the ability to wrap around complex geometries without the need to define artificial obstacles or via points. This is helpful since objective, anatomical guidelines for these methods do not exist and unique wrapping algorithms must be implemented depending on the underlying shape (Charlton and Johnson, 2001; van der Helm, 1994; Vasavada et al., 2008), or depending on joint position (Garner and Pandy, 2000; Vasavada et al., 2008). In our method, wrapping is done automatically for all degrees of joint motion by exclusively relying on contact detection algorithms available in commercial FE software. The wrapping process is performed for each static joint configuration, and the muscle-bone contact points are automatically updated at each simulated position. Our method also avoided muscle penetration through bone, which remains a major challenge associated with the obstacle-set method (Delp and Loan, 1995; Vasavada et al., 2008). This issue has been analyzed using custom algorithms (Charlton and Johnson, 2001), the geodesics theory (Marsden and Swailes, 2008a) or optimization algorithms (Audenaert and Audenaert, 2008; Marsden et al., 2008b). These methods still rely on simple underlying geometries and encounter issues in numerical accuracy, computational efficiency, and occasional interpenetrations that require user intervention.

As with the obstacle-set and via point methods, the major benefit compared to volumetric models is drastically lower computational expenses. A quantitative comparison of computational costs associated with available methods is difficult, since previous reports include a wide range of muscles numbers, and different computer processors. However, a fairly recent volumetric hip model was reported to solve in 5-10 hours on a supercomputer (Blemker and Delp, 2005), while the obstacle-set (Audenaert and Audenaert, 2008) and present methods solve within seconds for a single muscle on a desktop computer. This allows the complete shoulder model to run on a standard desktop PC and permits simultaneous use of a wrapping algorithm in conjunction with muscle force estimation models. Of course, this gain in computational efficiency is made at the cost of a simplified muscle representation. However, it has been shown that reliable modeling results can be
achieved by dividing muscles with large attachment areas into no more than 6 muscle line-segments (Van der Helm and Veenbaas, 1991). In any case, the intended application of the model should dictate the number of segments, striking a balance between reduced complexity and functional accuracy.

The model shares several limitations with previous models. The GH joint was represented as an ideal ball-and-socket joint (Charlton and Johnson, 2006; Garner and Pandy, 2001; Holzbaur et al., 2005; Karlsson and Peterson, 1992; van der Helm, 1994). Although humeral head translations within the glenoid remain small in vivo (Schiffers et al., 2002; Werner et al., 2006) and are generally neglected in modeling (Veeger, 2000), they may influence the moment arms in an unstable joint. Muscle thickness was neglected (Gao et al., 2002; Ruckstuhl et al., 2009), representing only the muscle fibers closest to the joint (Favre et al., 2005; Favre et al., 2009). For muscles that cross the joint with deep bellies, in particular the deltoid, the obtained moment arms are therefore minimal values. However, because of the extreme variability in the morphology of these muscle bellies, and want of relevant details with respect to the path taken by the centroids of cross-section in various humeral positions, these minimal values (worst conditions) have been considered sufficient. Further work is required to produce more realistic data for deltoid action.

The muscle path obtained with this method are not mathematically exact solutions for the shortest path problem, which can be obtained with optimization methods (Audenaert and Audenaert, 2008; Marsden and Swailes, 2008). However, the close agreement with experimental evidence indicates that the simplified muscle representation did not adversely affect accuracy. Intra and intermuscular interactions were not accounted for (Gao et al., 2002; Marsden et al., 2008), a limitation typically associated with representation of muscles by “strings”. However, while muscle interactions have been explicitly considered by volumetric muscle models, inclusion of volumetric muscle interactions comes at considerable computational expense, and without obvious improvement in accuracy (Blemker and Delp, 2005).

While the current model was developed for use as a general model, and was based on the anatomy of an average sized adult, further work could apply this method within subject-specific modeling studies. Such models can be used to account for anatomical variability in surgical planning or to assess pathological deformities. For subject-specific modeling, determining a muscle path without first defining wrapping surfaces and via points would be
valuable, as these have been directly associated with muscle penetration and moment arm inaccuracies in subject-specific lower limb models (Arnold et al., 2000; Scheys et al., 2008a). Anatomies based on patient specific image data should be employed rather than scaling of a general model, as has been shown for the lower limb (O’Brien et al., 2009; Scheys et al., 2008a; Scheys et al., 2008b). For these models, muscle attachment sites could be determined using either anatomical descriptions or more elaborate techniques (Kaptein and van der Helm, 2004; Matias et al., 2009). While modeling of anatomical deformities may necessitate adjusting initial muscle orientations, the sensitivity analysis presented here indicates that this would only slightly affect the results.

The present model will be combined in the next chapter with a muscle force estimation algorithm. While the approach we describe was developed for this purpose, the current study represents a general proof of principle for use in joints with a wide range of motion and numerous degrees of freedom. It is reasonable to assume that the proposed wrapping method could be applied to other joints, even those for which the joint center position cannot be assumed whereby the virtual velocity method (Delp and Loan, 1995) could be used to compute the moment arms. Thus the proposed method may prove useful for a wide range of musculoskeletal modeling applications.

REFERENCES


Muscle wrapping


MSC.Software, Marc Mentat User Documentation.


Muscle wrapping


4. A NEW PARADIGM FOR SHOULDER MUSCLE FORCE ESTIMATION

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ABSTRACT

The shoulder joint represents an indeterminate mechanical system, making it difficult to predict individual muscle forces required to equilibrate a given arbitrary external force. Although considerable work has been published on this matter, no model exhibits the adaptability required for the analysis involving different positions of the humerus and for any external load. An algorithm involving decision-making loops is developed to predict forces exerted by muscles that cross the shoulder joint in equilibrating a given external force acting in an arbitrary direction.

Lever arms and directions of action of shoulder muscles are used together with the external force as input. The algorithm selects an appropriate group of muscles and step by step attributes as little force increments as possible to withstand the external moment. Each muscle force increment is stored after every loop and eventually summed up. Stability of the glenohumeral joint is the final determining factor.

Six worked-out examples demonstrate that the model correctly activates muscles when compared to EMG measurements, and show interesting features of probable muscular activity. In opposition to optimization techniques, this algorithm is able to simulate co-contraction. This approach can cope with any physiologic combination of joint position and external load. Muscle segmentation is of paramount importance for spatial control. Although stability can be achieved by increasing the overall rotator cuff activity (co-contraction), this is rarely necessary.

The strategy of force sharing among the muscles opens up the possibility to examine the outcome of muscle deficiencies and to investigate causes of joint instability as encountered in clinical practice.

Keywords: shoulder model, shoulder muscle forces, muscle recruitment, glenohumeral stability, moment arm lengths, direction of muscular force.
4.1. INTRODUCTION

The GH joint capsule and ligaments are lax throughout the greater part of GH rotation, and hence inactive for stability, while the joint surfaces themselves offer little resistance to shear forces (Gerber, 1992; Jössel, 1880; Kronberg et al., 1990; Lippitt, 1993; McMahon et al., 1995; Perthes, 1906). Stability of the shoulder joint is primarily ensured by active muscular contraction. A realistic estimation of muscle forces involved in balancing a given external force, while preventing dislocation of the GH joint is therefore mandatory when one wishes to model GH joint stability.

As described in the Background section, available methods to derive muscle forces include optimization techniques or EMG-driven models. However, these models are confronted with limitations that may hinder their use for the purpose of this thesis (choice of cost function, inability to simulate co-contraction, inherent difficulties with EMG measurements or simulation limited to a few given joint configurations).

In order to tackle these disadvantages, a novel paradigm to predict muscle forces and test joint stability was developed. Originally, this algorithm was based on input data (muscle moment arms and lines of action) that had been measured in advance off an established full-size epoxy model of the thorax, scapula and arm (see Figure 6). This allowed investigations of potentially any joint position (Favre et al., 2005; Favre et al., 2008; Favre et al., 2009), but experimental measurement of moment arms and lines of action of 27 muscle segments took about 4 hours for one single position.

The results of the present chapter are based on anatomically data gathered from the plastic artificial model. In the integrated model (described in the next chapter), the algorithm uses the input data delivered by the wrapping method described in the previous chapter. This allows a fully automated computation of muscle moment arm, line of action and force for any combination of joint position and external load.
4.2. METHOD

Definition of motion components, position, and directional signs

Joint motions and positions are described according to the ISB recommendations and decomposed into plane of elevation, elevation and axial rotation components relative to the scapula. A directional sign is added to this definition to specify the sense of rotation (Figure 21). “Positive elevation” denotes an upward rotation, “negative elevation” a downward rotation; “positive plane of elevation” is rotation to the front or anterior, “negative plane of elevation” rotation to the back or posterior; “positive axial rotation” implies medial or internal axial rotation, and “negative axial rotation” lateral or external axial rotation.

![Figure 21: Motion definition. The motion is decomposed in plane of elevation, elevation and axial rotation components. Each component can be positive or negative, depending on the sign of rotation. This Figure shows a right shoulder.](image)

Definition of “Initial” external moment and “virtual” external moment

The initial external moment is the moment induced by the initial external force or external pure torque. As for motion components, the external moment is decomposed in three orthogonal components (Figure 21). The moment is named according to the direction of movement it would initiate; e.g. a positive elevation moment causes a positive elevation, etc..
The virtual external moment is defined as the difference (error) between the initial external moment and the partially exerted muscle moment, as determined at any intermediate step of the iteration procedure. It is therefore the remaining external moment that the muscles still have to balance. At the end of the algorithm, when equilibrium is reached, the sum of all iterate muscle moments opposes in direction and equals in magnitude the initial external moment, within the range of the chosen acceptable error (Equ. 8 and Figure 22).

\[
\text{External moment} + \sum \text{muscular moment} = 0
\]

\[\Leftrightarrow \text{External moment} = -\sum \text{muscular moment} \quad \text{Equ. 8}\]

Figure 22: Graphical representation of equilibrium. Equilibrium is reached when each component of the sum of the muscular moments reaches the same magnitude (within the chosen tolerance) in the opposite direction as the corresponding external moment component.

Definition of protagonists and antagonists
Protagonists are muscles exerting a moment that counteracts a given component of the external moment; e.g. the supraspinatus muscle would be a protagonist in equilibrating a negative elevation moment. Antagonists are muscles exerting a moment in the same sense as that of the external moment component, thereby increasing the imbalance for this component. Any muscle having three components of action can well be a protagonist for one external moment component, but an antagonist for another. For example (Figure 23), a muscle opposes the external moment arm components for the first component, and is therefore a protagonist for this component. On the other hand, the same muscle is an
antagonist for the other two components as it exerts moments in the same sense (mathematically speaking, with the same sign) as the external moment components.

![Figure 23: Definition of protagonist and antagonist. Examples of a protagonist (first component) and an antagonist (second and third component) muscle component.](image)

**Muscle segments**

The muscles of the shoulder have been segmented to allow for differentiated actions. As described in chapter 3, this was especially necessary in muscles with broad insertions because moment arm length and direction of tendon action vary considerably within a given muscle (Favre et al., 2009). Moreover, neuromotor control strategies differ across the breadth of muscle and all fibres in a given muscle must not necessarily contract simultaneously (Fick, 1877; Johnson et al., 1996; van der Helm and Veenbaas, 1991; Wickham, 1998).

**Synopsis of the algorithm**

Because each muscle generally gives rise to turning moments in all three orthogonal planes simultaneously, choice of muscles and the magnitude of force demanded from each to equilibrate a given external moment becomes an elaborate issue. Solving the problem therefore requires a system of arithmetical operations that have to finally fulfil several boundary conditions, some of which are anatomically and physiologically set.

The algorithm for shoulder forces estimation (“ASFE” for Algorithm for Shoulder Force Estimation) involves decision-making iteration loops (Figure 24). An external moment at the joint center of rotation is decomposed into three orthogonal components. In the main loop,
the algorithm first recruits the muscles that have the largest potential mechanical advantage in offsetting the greatest of the three external moment components, without any regard for the two other components. Here potential mechanical advantage is defined by the muscle moment arms for the current position, multiplied by muscle cross-section. In order to offset the two remaining components of the external moment, muscles that simultaneously oppose all three external components are recruited. Muscles that do not fit within these two groups are not active in the current loop, but may become active in later loops if the greatest component of the remaining external moment is another than the initial component (thereby simulating co-contraction). The equilibrium force equations attribute forces to the set of recruited muscles in proportion to each muscle’s cross-sectional area, making the system mathematically determinate. This approximation is necessary to only start the process of iteration, i.e. to express a first set of equations (one for each of the three internal orthogonal moment components). After further modification of these muscle forces through sub-routines to suit specific boundary conditions (such as range in tension from zero to the tetanic muscle force) a better approximation of the required muscle forces is obtained. The resultant muscle forces from this loop are then arbitrarily scaled down (typically by 95%) to ensure that the model converges and the external moment is not overshot. The whole process is iteratively repeated using the remaining external moment as input to the next loop, until all three components of the external moment are balanced. At the end, the determined muscle force stored after every loop is finally summed up to give the final estimate. The resultant force is determined, and if the joint is unstable (joint reaction force points outside of the glenoid or stability ratio is too high), the simulation is restarted, but first giving all rotator cuff muscles a supplementary force.

With this approach, a solution for the indeterminate problem can be found. In opposition to optimization techniques, the algorithm activates muscles that are not optimal in a mathematical sense, but are physiologically important (such as antagonistic action/co-contraction).
Figure 24: Flowchart of the ASFE.
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**Input (Figure 24A)**
The input consists of the initial external force and the humerus position. According to the humerus position, the corresponding predetermined muscle moment arm and line of action data is loaded.

**External moment calculation (Figure 24B)**
From the input data, components of the initial external moment for elevation, plane of elevation and axial rotation are calculated in magnitude and direction (positive or negative).

In the following main iteration loop, suitable muscle segments are selected, equilibrium equations solved, and muscle forces attributed.

**Muscle recruitment (Figure 24 C)**
The strategy used to solve the problem of muscle recruitment is to first focus on the greatest external moment component which calls for a choice of muscles that strongly oppose this in the first place and to then select other muscles that will counteract the two remaining moment components, too. The selection of essential muscle segments therefore requires grouping muscles into two categories (Figure 25):

- **Group A**: Muscle segments that can exert a significant moment to oppose the greatest of the three external moment components, while totally disregarding the two lesser ones.
- **Group B**: Muscle segments that can exert moments which oppose all three external components simultaneously, totally disregarding their magnitude. The main criterion for this group is the direction of muscle action.
Figure 25: Muscle recruitment based on the external moment components. In this example, the largest external moment component is the second component. In group A, muscles with a large potential to oppose this largest component are selected, without considerations for the other two components. In group B, muscles with potential to balance all three components of the external moment are selected.

From group A, all muscles with a high potential (large moment magnitude) to exert a protagonistic function in opposing the greatest external moment component are primarily chosen. To estimate the maximal potential turning moment of the muscle, moment arms are multiplied by the maximal potential force (defined as the cross-sectional area multiplied by a constant factor, as explained below, see Table 4). Such muscle segments are selected when their individual maximum muscle moment in the direction of interest is greater than the average of all maximum muscle moments in the same direction multiplied by a threshold factor. If the threshold value is high, few muscles with high potential are recruited to counteract the external moment. If the threshold value is low, several muscles of lower potential share the task, resulting in finer tuning of the muscular moment. With this threshold factor, a selection between larger or finer muscles can be accomplished. Since it would be logical to assume that neuromotor control strategy is such that muscles are loaded to the least extent in performing a given task, it follows that the issuing magnitude of the joint resultant would also be as low as possible. Tests have shown that a minimum joint
resultant force issues with a threshold factor between 0.2 and 1.4. The algorithm is run iteratively varying the threshold between these limits and the threshold leading to the lowest value of resultant force is kept final.

To group B belong those muscles that do not exhibit a sufficiently large moment potential in the major direction, but which oppose simultaneously all three components. This favors convergence in all three directions simultaneously, especially since muscles from group A may increase the imbalance for the remaining two components. This category has been found crucial to arrive at an acceptable final solution. Although these have little moment potential to equilibrate the major virtual external moment component, they are indispensable for equilibrating the remaining two components of external moment.

All recruited muscles, from both groups A and B, oppose the largest external moment component. If the three initial external moment components are of the same amplitude, the programme will continuously switch its focus from one component to another, ensuring convergence of all three components alternately. Also, because the focus is on the largest external moment component, the largest component will generally be balanced first, so that in later loops, the largest component may be another one than the initially largest component (Figure 26). Muscles that may be antagonists for the initial external moment will therefore be recruited to balance this new largest virtual external component, simulating antagonistic co-contraction.

In the next step, forces are attributed to the selected muscles from group A and B.

Figure 26: Change of largest component leading to simulation of co-contraction. In the initial loops, the largest component (second component, indicated by the arrow) of the error (or virtual external moment) is balanced faster than the other two components. This may induce a change in the largest component after several loops (first component, at loop n+1).
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Equilibrium equations (Figure 24D)

To obtain equilibrium, a system of three equations must be satisfied, one each per motion component (Equ. 9).

Elevation
\[
\left( \sum_{i=1}^{27} \text{muscle moment}(i) \right) + \text{virtual external moment} = 0
\]

Plane of elevation
\[
\left( \sum_{i=1}^{27} \text{muscle moment}(i) \right) + \text{virtual external moment} = 0 \quad \text{Equ. 9}
\]

Axial rotation
\[
\left( \sum_{i=1}^{27} \text{muscle moment}(i) \right) + \text{virtual external moment} = 0
\]

Where
\[
\sum_{i=1}^{27} \text{muscle moment} (i) = \sum_{i=1}^{27} \text{Force}(i) \times \text{moment arm}(i) \quad \text{Equ. 10}
\]

The index (i) stands for each of the 27 muscle segments. There are therefore 27 unknown “Force(i)” for each of the 3 components.

To overcome the problem of indeterminacy, we hypothesize that muscles exert a force proportional to their maximum potential force. The maximal force a muscle can develop is limited by several parameters such as its optimum length, the contraction velocity, the fibre type composition and especially the physiological cross-sectional area (Fick, 1910; Karlsson and Peterson, 1992). In this study, an average of the PCSA values found in Wood et al. (1989), Veeger et al. (1991), and Van der Helm (1994) has been used, this being \( k \times \text{PCSA} \), in which \( k \) is a constant factor of 0.8 MNm\(^{-2}\) (Table 4).
Table 4: Physiological cross-sectional area (mm$^2$) multiplied by the constant factor of 0.8 to estimate the potential maximal force $F_{max}$ (N) for the 27 muscles segments.

Assuming that all protagonists would exert force simultaneously in proportion to their PCSA, one arbitrary muscle “u” is chosen and all other muscle forces related to it. The PCSA values link all muscle forces together and thus decrease the number of unknowns from 27 to 1 per equation (or per rotational degree of freedom):

$$R(i) = \frac{PCSA(i)}{PCSA(u)}$$  \text{Equ. 11}
where $R(i)$ is the ratio between the PCSA of the muscle “$i$” and the PCSA of the chosen reference muscle $(u)$.

The sum of all internal muscle moments can then be written as:

$$
\sum_{i=1}^{27} \text{muscle moment} = \sum_{i=1}^{27} \text{Force}(i) \times \text{moment arm}(i) \quad \text{Equ. 12}
$$

$$
\Rightarrow

\sum_{i=1}^{27} \text{muscle moment} = F(u) \times [R(1) \times M(1) + R(2) \times M(2) + \cdots + R(27) \times M(27)] \quad \text{Equ. 13}
$$

with $M(i)$ standing for the moment arm of muscle $(i)$.

Inserting Equ. 13 in Equ. 9 and isolating $F(u)$, we obtain:

\[
\begin{align*}
\text{Elevation} & : F(u) = -\frac{\text{virtual external moment (plane of elev.)}}{R(1) \times M(1) + R(2) \times M(2) + \cdots + R(27) \times M(27)} \\
\text{Plane of elevation} & : F(u) = -\frac{\text{virtual external moment (elevation)}}{R(1) \times M(1) + R(2) \times M(2) + \cdots + R(27) \times M(27)} \quad \text{Eqn. 14} \\
\text{Axial rotation} & : F(u) = -\frac{\text{virtual external moment (axial rotation)}}{R(1) \times M(1) + R(2) \times M(2) + \cdots + R(27) \times M(27)}
\end{align*}
\]

**Force averaging (Figure 24 E)**

Equ. 14 delivers three values of force for each muscle, one for each plane. These values are averaged, constituting a compromise that would not satisfy any of the single given equations but is the best approximation on the whole.

Knowing $F(u)$ and from the hypothesis that all active muscles develop a force proportional to their PCSA, each muscle force $F_i$ then follows from Equ. 15:

$$
F(i) = R(i) \times F(u) \quad \text{Equ. 15}
$$
Force redistribution to prevent negative force values and antagonistic action (Figure 24F)

Since a muscle can only pull, the force it exerts can only be positive or null (Equ. 16).

\[ F(i) \geq 0 \quad \text{Equ. 16} \]

Negative muscle force values that issue from the equations are not real, although muscle moments can still be positive or negative. The total moment of all selected muscles will always oppose the greatest virtual external moment component. However, the muscles of category A may introduce undesirable antagonistic effects in the other two directions. If the antagonists exert together a greater moment than that of the protagonists in at least one of the two remaining directions, this would also lead to an increase of misbalance in the corresponding direction. In the case of such antagonistic action, the solution of the equations may even lead to negative muscle force values. Consider one equation from the system shown in Equ. 13:

\[ F(u) = \frac{-\text{virtual external moment}}{[R(1) \times M(1) + R(2) \times M(2) + \ldots + R(27) \times M(27)]} \quad \text{Equ. 17} \]

For the largest component, all activated muscles are protagonists, so we always have

\[ \text{sign} \ (M(i)) = -\text{sign} \ (\text{virtual external moment}), \forall \ i \quad \text{Equ. 18} \]

This exclusively delivers positive values in Equ. 17.

However, for the other two components, muscles from Group A can be antagonists.

\[ \exists i, | \text{sign} \ (M(i)) = \text{sign} \ (\text{virtual external moment}) \quad \text{Equ. 19} \]

If the moment exerted by the antagonists is larger than that of the protagonists, Equ. 17 yields negative values for \( F(u) \). The protagonists must therefore always exert a greater moment than that of the antagonists to avoid disagreement with Equ. 16. If this is not the case, an inner loop redistributes the forces attributed from Equ. 15 by reducing the forces of the antagonists until the turning moment they exert is inferior to that of the protagonists. In
this way, solely positive muscle force values are obtained and overall protagonistic action is ensured in all three directions.

**Force limitation (Figure 24 G)**

A muscle cannot exert more force than a given maximum value (Table 4). Therefore

$$ F(i) \leq F_{max}(i) $$  \hspace{1cm} \text{Equ. 20}

If the sum of iterative contributions of any single muscle reaches its maximal force value, the muscle is turned off for the rest of the simulation. Other muscles that have not reached their limit must take over to resist the remaining virtual external moment.

**Prevention of overshooting (Figure 24H)**

The convergence of the ASFE is ensured as long as the virtual external moment is reduced after each iteration loop (which was controlled in the Force redistribution for prevention of negative force values and antagonistic action, Figure 24F). The overall internal moment is therefore controlled to always oppose the momentary virtual external moment so that the error always decreases.

Because only the signs of the muscle moment components are controlled to oppose that of the virtual external moment components and no special considerations have been made to regulate the amplitude of the muscle components, the attributed muscle forces can reach excessively high values and might result in overshooting the initial external moment. This can become problematic for convergence and also represents a waste of energy by demanding over-activity from muscles that are less optimal. To overcome this problem, all forces are multiplied in each loop with a constant shrinking factor of 0.05.

$$ F(i) > 0.05 \times F(i) $$  \hspace{1cm} \text{Equ. 21}
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This makes the incremental steps very small, and leads to a smoother iteration process (Figure 27).

![Diagram](image)

**Figure 27: Effect of force increment reduction to prevent overshooting. With force increment reduction (continuous line) and without (dashed line)**

**Error calculation (Figure 24 I)**

At the end of the main loop, the overall three orthogonal muscle moment components are calculated and the difference between these and the corresponding virtual external moment components determined. These differences (error(1), error(2), error(3)) are an indication of the remaining amount of imbalance. The total error is quantified by an amplitude as follows:

$$Amplitude = \sqrt{Error(1)^2 + Error(2)^2 + Error(3)^2}$$  \hspace{1cm} \text{Equ. 22}

The closer the muscle and external moments approach each other, the smaller is this amplitude. The loop stops when the amplitude of the error reaches a chosen set value.

**Resulting force and rotator cuff ground level activation for joint stability (Figure 24J)**

Once all muscle forces have been obtained, the joint resultant is calculated as the sum of all muscle forces plus the external force and its spatial direction computed (Equ.23).
In the first version of the ASFE, stability was computed using a geometrical criterion (van der Helm, 1994b). The GH joint does not dislocate as long as the humeral joint reaction force (the resultant) acts within the boundaries of the glenoid (Figure 28). If the joint reaction force points out of the glenoid boundary, stability is gained by increasing the general rotator cuff muscle force (to provide the so-called “concavity compression”). This entails recruiting such muscles that might be primarily antagonistic the initial external moment. Consequently, in such cases more protagonistic activity and therefore more overall muscular energy is required. This superimposed force exerted simultaneously by all the rotator cuff muscles is in some cases indispensable for joint stability. If stability is not reached at the end of the process described before, ground level activation is introduced, or raised, and the algorithm restarted.

\[ \text{Resultant} = \sum_{i=1}^{27} \text{muscle force}(i) + \text{External force} \quad \text{Equ. 23} \]

Figure 28: Geometric criterion of joint stability.
If the intersection of the glenoid plane and the resultant force lies within the glenoid boundary (black star), the joint is stable. If the intersection lies outside of the glenoid (grey star), the joint is instable.
In order to include the deformation of the joint structures to have a more reliable simulation of stability, this geometrical stability criterion is later replaced by the use of the stability ratio. If the ratio of the shear over the compressive components of the resultant force is larger than a given value, the joint is considered unstable. The limit value for the stability ratio is computed in the FE representation of the joint, as described later in the following Chapter.

To test the validity and capabilities of the algorithm to balance an external force acting in any direction several examples were simulated. These involved muscular moments of positive and negative elevation, as well as positive and negative plane of elevation with the humerus elevated 30° in the scapular plane and in neutral rotation. It must be noted that these examples are given for anatomical data measured on the epoxy shoulder model (Figure 6) and using the geometrical stability criterion (Figure 28). Results using anatomical data measured from the wrapping method (Chapter 3) and using the stability ratio as stability criterion will be shown in the next chapter. The external moment to be overcome in all these cases was 9Nm, which, in the following examples, would correspond to a 15 N force acting tangentially, at a distance of 0.6 m from the centre of the humeral head, i.e. at the end of an outstretched arm. No weight was attributed to the arm itself. These particular directions of the external force were chosen to compare the issuing results with some of the estimates reported in literature. However, a force in any arbitrary direction and spatial position could have been used.

Furthermore, for the same humerus position as described above, two cases of pure torque of 9Nm about the humeral axis were examined, one of positive and one of negative axial rotation. In this manner, all rotational degrees of freedom were tested.

Finally, abduction was simulated as a sequence of for static positions at 0, 45, 90 and 120° elevation with respect to the torso in the scapular plane. The arm weight (35N) was applied at the middle of the arm (0.3m from humeral head centre). Because the ASFE considers a fixed scapula, the rotation of the scapula was taken into account by rotating the external force vector according to a scapulo-humeral rhythm ratio of 1:2.
4.3. RESULTS

Muscular positive elevation (abduction, Figure 29)
This corresponds to an external force that gives rise to adduction about the GH joint. The anterior and especially the middle part of the deltoid (DEL3, DEL4) are particularly active, representing about 1/2 of the joint resultant, followed by the biceps (long head), supraspinatus and infraspinatus. The GH resultant force amounts to 450 N.

Muscular negative elevation (adduction, Figure 29)
The external force in this case tends to abduct the arm. Teres major (TMAJ) takes the greater part in load sharing, with the triceps, posterior part of the deltoid, latissimus dorsi, caudal parts of the pectoralis major and infraspinatus, biceps (short head), coracobrachialis and teres minor contributing. The magnitude of the resultant in this case amounts to 204 N.

Figure 29: Muscle forces for elevation.
Forces exerted by the 27 muscle segments for the elevation cases of loading with the humerus in 30° elevation in the scapular plane and in neutral rotation. The inserted field shows the intersection of the resultant with the glenoid boundary, the right side corresponding to the anterior direction and the left side to the posterior direction.
Muscular positive plane of elevation (anterior flexion, Figure 30)
The external force to be balanced acts posteriorly. The anterior part of the deltoid (DEL1) becomes the major contributor while the whole pectoralis major, the middle deltoid segments DEL2 and DEL3, teres minor, the cranial part of infraspinatus, supraspinatus, subscapularis, biceps (short head), and coracobrachialis assist. In this particular case, the co-activation of the rotator cuff had to be raised to 1.6% of each rotator cuff muscle maximal force in order to increase “concavity compression” and prevent dislocation of the joint. The joint resultant in this case amounted to 316 N.

Muscular negative plane of elevation (posterior flexion, Figure 30)
The external force to be balanced acts anteriorly. In this case the force exerted by the middle portion of the deltoid (DEL3, DEL4) reaches a very high value, amounting to about 2/3 of the total muscle force. The other muscles involved are the teres major, latissimus dorsi, the posterior part of the deltoid, triceps and infraspinatus. The joint resultant in this case exhibits a value of 466N.

Figure 30: Muscle forces for plane of elevation.
Forces exerted by the 27 muscle segments for the plane of elevation cases of loading with the humerus in 30° elevation in the scapular plane and in neutral rotation.
Muscular positive axial rotation (internal axial rotation, Figure 31)
An external turning moment of 9Nm was applied to externally rotate the humerus. The main muscle which equilibrates this is the subscapularis (SSC1, SSC2, SSC3) with a force amounting to about ¾ of the joint resultant. The middle part of the deltoid (DEL3, DEL4) and the latissimus dorsi assist. The joint resultant amounts to 642 N.

Muscular negative axial rotation (external axial rotation, Figure 31)
An external turning moment of 9Nm was applied to internally rotate the humerus. This is equilibrated mainly by the infraspinatus (ISP2, ISP3) and teres minor. Coracobrachialis, biceps short head and supraspinatus assist a little. The joint resultant reached 478N.

Figure 31: Muscle forces for axial rotation.
Forces exerted by the 27 muscle segments for the axial rotation cases of loading with the humerus in 30° elevation in the scapular plane and in neutral rotation.

In order to validate the joint reaction force, the performance of the ASFE and other optimization criteria in comparison to the instrumented shoulder implant (Bergmann et al., 2007) is described in the section below for a classic example.

Positive elevation (abduction) in the scapular plane
The joint reaction force predicted by the ASFE (Figure 32) for 0°, 45° and 60° elevation fell within the ranges reported by other numerical methods (Poppen and Walker, 1978; Terrier
et al., 2008; Van der Helm, 1994b), and with in vivo measurements using instrumented shoulder prostheses (Nikooyan et al., 2010). For 120° elevation, the estimated joint reaction force was much higher than the values obtained with other models, but followed the trend measured in vivo (Nikooyan et al., 2010).

![Joint reaction force](image)

**Figure 32:** Joint reaction force during positive elevation in the scapular plane. The resultant obtained by other models (Poppen and Walker, 1978; Terrier et al., 2008; Van der Helm, 1994b) (light grey). The dark grey curves show the measurements obtained with the instrumented shoulder implant in two patients (Nikooyan et al., 2010).

### 4.4. DISCUSSION

We have not found any method described in the literature which would allow readily estimating the forces acting across the GH joint when an arbitrary combination of external loading conditions occur. It has been shown that with this algorithm any external force (within reasonable limits of magnitude) can be considered, leading to an estimation of muscle forces by searching for a solution in which these are a minimum, resulting in a joint resultant of lowest possible value.

The ASFE offers the advantage of very short calculation time (few seconds), with a commonly available PC when run with the well-known and widespread Matlab software.

The ASFE is not a mathematically exact method of arriving at a final solution but, like any iterative method, is a search for an acceptable possibility within given constraints: tensile muscular activity, direction of muscle action, maximum attributable muscle force, minimal energy requirements and position of joint resultant with respect to glenoid boundaries.
A major simplification of the model is that scapula motion in relation to the thorax has not been taken into account, the former remaining in its anatomically neutral position. This would certainly affect the estimated action of the thoraco-humeral muscles when the humerus is elevated beyond about 30° relative to the scapula, but this was considered acceptable for the present purpose, since only latissimus dorsi and pectoralis major are involved. All other muscles link the humerus to the scapula, so that movement of the latter does not change the directions of action of the muscles as defined in the chosen scapular coordinate system.

Estimated muscular maximal potential turning moments are central to the recruitment process. These are function of the maximal potential muscle force (factor k = 0.8MN/m² that multiplies the PCSA) and the moment arm. However, these parameters are subject to uncertainties. First, the factor k has been chosen to lie within the considerable range of 0.4-1 MN/m² found in the literature (Crowninshield and Brand, 1981). However, this factor needs to be more precisely determined in further studies, as it limits the maximal external force that can be balanced. Second, PCSA values are generally underestimated, as these were obtained on generally old donors where cadaveric muscles may have lost some volume. However, the lost volume of cadaveric shoulders and the factor k affect all muscles in the same way. The relative values between muscles, and consequently the recruitment of muscles in group A are therefore not affected. Third, as mentioned in Chapter 3, the muscle string representation used in both the physical epoxy model and the numerical wrapping model to compute the muscle path represents the action of muscle fibres closest to the joint. However, the corresponding large PCSA of these muscles compensates and warrants that such muscles become recruited. This is confirmed by the frequent activation of the middle deltoid in the studied examples. Muscular force is also influenced by other factors than PCSA, such as optimum muscle length, contraction velocity and fibre type composition. These factors are not considered in the model and should be added in future work.

Recruitment and dosage of muscular force to balance an external force is the goal of the algorithm; however, two major problems are encountered: a) there are more unknown muscle forces than equilibrium equations, and b) each muscle (or muscle segment) most often exerts forces in all three orthogonal directions simultaneously, not all of which are conducive to equilibrating the external force about a spherical joint. Segmentation of
muscles increases the number of unknown muscle forces, making condition a) more difficult to fulfil. On the other hand this was found to facilitate equilibration of the external force by providing more selective variables and allowing for finer regulation, thus making b) simpler to comply with. The individual role played by each muscle segment becomes obvious and this also vividly illustrates the necessity to break down muscles, especially those with broad insertions, into component parts. As already observed by (Jössel, 1880), muscular units, although anatomically designated as one, must not be functionally considered as one single motor unit, but made up of several, and this is probably how fine control of the neuromuscular system indeed functions (Wickham, 1998).

The worked-out examples illustrate the possibility of application of the ASFE clearly. A comparison of the results obtained for the 6 different loading cases shows immediately which external loads are more difficult to cope with than others. For the same external load amplitude, but in different directions, the joint resultant can vary between 204 (negative elevation) and 642 N (positive axial rotation).

Direct comparison of the estimates of muscle activity obtained through the ASFE with those obtained by others is difficult because of methodological and geometric differences. Nevertheless, the middle and anterior deltoid and supraspinatus muscles, known to be primary movers in positive elevation/abduction (Kronberg et al., 1990), are shown to play a predominant role for this activity in our model, too. The biceps (long head) also appears to play a significant role, an observation shared by Karlsson and Peterson (1992), Van der Helm (1994) and David et al. (2000). Laursen et al. (1998) and Kronberg et al. (1990) measured EMG activity from the infraspinatus during abduction and considered it a stabilizer.

In the case of negative elevation/adduction, others have attributed the latissimus dorsi and pectoralis major a leading role in this function. Although all these muscles do indeed come into play, we observe a very important role taken by teres major, corresponding to the findings of Pearl et al. (1992), together with high activity on the part of the triceps. The lower magnitude of the joint resultant (204N) in the worked-out example for adduction indicates that the external force applied is far below what can be coped with.

Negative plane of elevation/posterior flexion surprisingly called upon the middle part of the deltoid (DEL3 and DEL4) to play a dominant role. The middle and the posterior deltoids were
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described as prime movers for posterior flexion in the EMG study by Kronberg et al. (1990). Actually in our model, these muscle segments bring a modest contribution to posterior flexion directly, but they are of major value because of their high positive elevation component which is required to neutralise the negative elevation moments of all other important posterior flexors involved (teres major, latissimus dorsi, DEL5, DEL6, and triceps).

Positive plane of elevation/ anterior flexion also incorporates some interesting features. While the anterior part of the deltoid (DEL1) plays a leading role, as would be expected (Kronberg et al., 1990), and pectoralis major (Pearl, 1992) and coracobrachialis together with the biceps (Van der Helm, 1994b) are also known to be anterior flexors, the substantial activity of teres minor has not been described in this context before. Although teres minor does not exhibit any flexion potential in this case, it is called upon to compensate the unwanted internal rotation side-effect of the anterior flexors. Interestingly, anterior subluxation of the GH joint in this case necessitates an increase of the general rotator cuff activity which, however, entails simultaneous action of antagonists that somewhat increase the magnitude of the joint resultant.

Positive/internal axial rotation was seen not to incorporate the action of pectoralis major, a muscle described as an important medial rotator by Kronberg et al. (1990). This is because the very high anterior flexion activity of this muscle makes it unsuitable. Subscapularis on the other hand is much more efficient, in combination with latissimus dorsi (both cited as prime movers by Kronberg et al., 1990) and the middle part of the deltoid. The latter can, in our study, more easily compensate the unwanted anterior flexion effect of subscapularis.

Negative/external axial rotation required the action of infraspinatus and teres minor, as also reported in the literature (Basmajian, 1985; MacConaill, 1977). In this humerus position, together with the posterior segment of supraspinatus, they require some compensation from coracobrachialis and the short-head biceps to neutralise a posterior flexion tendency.

The good agreement concerning muscle recruitment between the results of our study and other simulations studies as well as EMG measurements show that the recruitment criteria described in this study (group A and B) lead to meaningful results.
The examples show that in most situations, rotator cuff co-activity is not required for stability. The resultant mainly falls within the glenoid boundaries without necessitating any co-contraction. This also ensures that the magnitude of the resultant is kept as low as possible.

Deviations in predicted joint resultant forces between the ASFE and other models during simulated elevation below 90° (Figure 32) can be explained by differences in model geometry (muscle segmentation, humeral head size, muscle size, muscle origin and insertion site) and recruitment strategy. The large deviation at 120° abduction also owes to the ASFE incorporation of muscle co-contraction that is not considered in the other models. Muscles of the rotator cuff and the deltoid, which represents the totality of muscles included by Poppen et al. and Terrier et al., also showed a decreased activity past 90° in the ASFE, except for the posterior deltoid segment. The rise in joint resultant force above 90° is due to other muscles. At 120° abduction, the majority of abductors externally rotate the humerus. Only the cranial segments of subscapularis and pectoralis major can counterbalance external rotation, but they are also strong forward flexors. Here, other muscles such as teres major are activated to help balance these components, even if such a muscle is a very strong adductor. Interestingly, in vivo measurements using instrumented shoulder prostheses indicated that joint reaction forces increase as abduction exceeds 90° (Nikooyan et al., 2010a). Although the ASFE is able to replicate this trend, it does not necessarily mean that the muscle forces found here correspond to in vivo muscular contribution during abduction, as many physiologically plausible muscle force patterns can lead to a similar joint reaction force. Experimental measurements of muscular activity in postures above 90° of abduction are now required. However, it may indicate that co-contraction is a very important aspect of shoulder biomechanics that has thus far been widely neglected. It must again be emphasised that as in any model which endeavours to predict muscle forces, the solution arrived at is strongly determined by the given boundary conditions. This is confirmed in Chapter 5, where this trend is not reproduced on another anatomy and with a different implementation of joint stability.

In summary, a useful algorithm has been devised which is capable of predicting muscle forces for any given position of the humerus in equilibrating an external force acting in an arbitrary direction. Good agreement between some worked-out examples using this
algorithm and other simulation studies as well as EMG measurements has been found. The strategy of force sharing among the muscles opens up the possibility to reflect on the outcome of muscle deficiencies and on causes of joint instability as encountered in clinical practice.

REFERENCES


CHAPTER 5
5. AN INTEGRATED MODEL OF ACTIVE GLENOHUMERAL STABILITY

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ABSTRACT

While muscular stabilization of the glenohumeral joint remains poorly understood, the complexity of glenohumeral stabilization critically limits the utility of simplified modeling approaches. We present the first model of the glenohumeral joint implementing active muscle driven humeral positioning and stabilization without a priori constraints on glenohumeral kinematics. The method established in chapter 3 was used to predetermine the muscle path, and the approach described in chapter 4 was used to evaluate recruitment and resultant force contribution of 27 individual muscle segments at any given joint position. Artificial boundary conditions were applied in an anatomically precise 3D finite element model of the joint and progressively released until the humeral head was completely free to rotate and translate within the fixed glenoid according to the compressive component of the predetermined resultant force. The shear component was then added, applying a load case with no boundary conditions other than muscular force. The framework was exploited to simulate elevation as a composite of instantaneous positions and demonstrate that joint stability can theoretically be achieved exclusively through muscular activity. Predicted muscle moment arms, muscle on-off activity pattern, humeral head translations, joint contact forces and stability ratios were successfully validated against existing experimental or in vivo values. This is the first model able to cope with the wide glenohumeral joint range of motion in all degrees of freedom and the considerable variability in external loading conditions as encountered in daily life activities. This framework can be used to address clinical hypotheses related to shoulder joint stability that cannot be pursued using simplified modeling approaches.

Keywords: Glenohumeral stability, dislocation, active/muscular stabilization, labrum, stability ratio.
5.1. INTRODUCTION

Stability of the GH joint is primarily ensured by active muscular contraction, and the sheer complexity of the joint has restricted elucidation of the stabilization mechanisms. Most numerical models simplify the GH joint as a ball-and-socket with a fixed center of rotation, prohibiting humeral head translations within the glenoid and limiting their use in the study of stability. In addition, constrained humeral head translations preclude analysis of tissue deformations, contact area or pressure (Buchler et al., 2002; Hopkins et al., 2006) and may lead to underestimated muscular forces, joint reaction forces and stresses (Terrier et al., 2008; Veeger and van der Helm, 2007). Although kinematically unconstrained models of the GH joint are clearly required to investigate stabilization and motion (Favre et al., 2009; Hill et al., 2008; Veeger and van der Helm, 2007), validated six DOF models are lacking from the literature.

The purpose of this study was to develop (to our knowledge) the first 3D FE model of the GH joint to implement active muscle driven humeral positioning and stabilization at any joint position, for any loading condition, without *a priori* defined kinematic constraints. We then test the hypothesis that GH joint stability can be achieved during elevation exclusively through muscular action (without the ligamentous capsule).

5.2. METHODS

A 3D deformable model of the GH joint was combined with the methods developed previously regarding muscle wrapping (Chapters 3) and recruitment (Chapter 4) to simulate active joint balance and stabilization (Figure 33). Each step can be automatically performed for any instantaneous (treated as static) joint position. First, the muscle path is computed for a position of interest (Figure 33A). Second, this anatomical information is used to estimate the forces required from each muscle segment to balance a given external moment and stabilize the joint (Figure 33B). The resultant force is finally applied in a 3D FE model for simulation of joint contact and humeral head translation (Figure 33C).
Figure 33: Schematic of three components integrated within the modeling framework. Muscle wrapping delivers muscle moment arm and line of action data (A). Muscle forces are algorithmically estimated (B) and stability is assessed according to deformable joint contacts (C). These three steps must each be performed at each instantaneous position.

For the implementation of joint contact characteristics (Figure 33C), an anatomically precise numerical model of the scapula was created based on CT scans of 400 micrometers isotropic resolution (Philips Brilliance 40 CT scanner, Philips Healthcare, Best, NL) of a fresh frozen human scapula that was free of visible deformities. The bony contour was outlined using a global threshold (ImageJ, Bethesda, MD, USA) and 3D reconstruction was performed in Matlab (MathWorks, Natick, MD, USA). The full joint being unavailable, the geometry of the humerus was imported from the Bel repository (Van Sint Jan, 2004). The size of the humeral head diameter (48mm) was controlled to match the height of the glenoid (33mm) according to established anthropometric relationships (McPherson et al., 1997).

The humeral and scapular geometries were meshed with quadratic tetrahedral elements in ANSYS (Workbench version 12.0, Ansys Inc., Canonsburg, PA, USA) and imported into Marc Mentat 2008r1 (MSC.Software, Santa Ana, CA, USA). In this study focusing on active GH stabilization, the capsule was not modeled. The cartilage and labrum were modeled, with geometry based on published anatomical data (Figure 34). The humeral head cartilage surface was defined as a sphere (Boileau and Walch, 1997; Soslowsky et al., 1992) with a thickness of approximately 1mm (Fox et al., 2008). The humerus was placed in the rest...
position (Wu et al., 2005) so as to create a 1.3mm clearance in the glenoid cavity center, corresponding to the glenoid cartilage thickness in this location (Yeh et al., 1998). By virtue of congruency of both cartilage surfaces (Soslowsky et al., 1992), the lateral border of the glenoid cartilage was made to coincide with the humeral cartilage surface. The labrum thickness was designated according to published data (Howell and Galinat, 1989).

![Joint contact surfaces.](image)

**Figure 34: Joint contact surfaces.**

Isotropic linear-elastic material properties (Young’s Modulus of 18 GPa and poisson’s ratio of 0.3) were assigned to the bone (Currey et al., 2001). Hyperelastic neo-Hookean material properties were defined for the cartilage and the labrum, with the strain-energy density function (Buchler et al., 2002) defined as

\[
W = C_{10}(I_1-3) \quad \text{Equ. 24}
\]

with \( C_{10} = E/4(1+v) \) \quad \text{Equ. 25}

where \( E \) represents the elastic modulus, \( v \) the Poisson’s ratio and \( I_1 \) the first invariants of the Cauchy–Green stress tensor.

Values of \( C_{10}=1.79 \) and \( C_{10}=12.5 \) were attributed to the cartilage (\( E = 10\)MPa and \( v = 0.4 \)) (Buchler et al., 2002) and labrum respectively (\( E = 70\) MPa, \( v = 0.4 \)) (Carey et al., 2000; Smith et al., 2009). Contact between GH cartilage surfaces was modeled as frictionless (Terrier et al., 2008).

A series of successive step was defined to progressively release the constraints applied on the humerus (Table 5). The medial part of the glenoid was always kept fixed in translation and rotation. The humerus was initially fixed in translation, but rotations were always
permitted. First, a medial displacement toward the glenoid was imposed to the humerus until an initial touching contact was automatically detected by the FE software, while keeping the other two translational degrees of freedom fixed. Second, the medio-lateral translations were also fixed (to later be gradually released) and a medio-lateral compressive force representing the compressive joint reaction force component was applied to keep the two bones touching in the later steps. Third, the medio-lateral translational DOF on the humerus was progressively released (using the “gradual release” function of Marc Mentat). Fourth, the remaining inferior-superior and antero-posterior translational DOF of the humeral head were successively gradually released, leaving the head fully free to translate, rotate and center within the concave glenoid by the compressive force. Finally, the shear joint reaction force component was applied and the model was equilibrated to assume a self-aligned, “physiological” configuration.

<table>
<thead>
<tr>
<th>Step</th>
<th>Boundary condition humerus</th>
<th># increments</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Move</td>
<td>M-L displacement towards glenoid</td>
<td>10</td>
</tr>
<tr>
<td></td>
<td>Fixed AP, I-S translations</td>
<td></td>
</tr>
<tr>
<td>2. Freeze all</td>
<td>M-L compression</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Fixed M-L, AP, I-S translations</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Fixed AP, I-S translations</td>
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</tr>
<tr>
<td></td>
<td>RELEASE M-L translations</td>
<td></td>
</tr>
<tr>
<td>4. Release AP, I-S</td>
<td>M-L compression</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td>RELEASE AP, I-S translations</td>
<td></td>
</tr>
<tr>
<td>5. Shear</td>
<td>M-L compression</td>
<td>10</td>
</tr>
<tr>
<td></td>
<td>Shear</td>
<td></td>
</tr>
</tbody>
</table>

Table 5: Series of steps defined to gradually release the fixed medio-lateral (M-L), antero-posterior (A-P), inferior-superior (I-S) translation constraints on the humerus. By the end of step 5, the humerus was completely free to rotate and translate and kept within the glenoid by the concavity compression effect of the joint resultant force. All degrees of freedom of the scapula remain fixed during the full simulation.

The implementation of the labrum and cartilage was validated against experiments reporting the stability ratios (shear forces required to dislocate the joint with dissected capsule in 8 different directions with a 50N compressive load) (Halder et al., 2001; Lazarus et al., 1996; Lippitt and Matsen, 1993). To simulate the experiments, the glenoid was
oriented to lie in a plane perpendicular to the compressive force. Dislocation was defined as the point where the displacement curve of the humeral center increased in an abruptly non-linear manner.

The deformable model was then combined with two previously established methods to apply physiologically relevant GH joint resultant forces. The muscle paths for 27 muscle segments were computed using the current bone geometries within the automated wrapping paradigm developed in Chapter 3 (Figure 33A). The resulting muscle moment arms and lines of action were then input to the algorithm established in Chapter 4 for distributing forces among the muscles in accordance to their relative mechanical advantage for counteracting components of the external moment (Figure 33B). If joint stability was less than a critical threshold, supplementary rotator cuff activity was incrementally applied until the joint was stabilized. In its original form, the algorithm considered the joint to be stable if the resultant force fell within the glenoid boundaries (see Chapter 4). Here, to account for the introduced deformability of the glenoid boundaries, the joint was considered to be stable if the stability ratio did not exceed the values computed in the FE model as described above.

To demonstrate comprehensive validity of the three integrated components, GH elevation in scapular plane from 0 to 80° was simulated in steps of 10°, and the outputs were validated against experimental evidence. The muscle moment arms were validated by direct comparison with published experiments (Favre et al., 2005; Hughes et al., 1998; Kuechle et al., 1997; Liu et al., 1997; Nyffeler et al., 2004; Otis et al., 1994; Poppen and Walker, 1978). This anatomical data was used to estimate the muscle forces required to balance the arm weight (35 N) applied at 30cm from the humeral head center in each GH position. The external force vector was rotated with a 2:1 proportion to account for the scapulohumeral rhythm (Poppen and Walker, 1976). The on-off patterns of muscle activity were compared with available electromyographic data (Habermeyer et al., 1987; Inman et al., 1944; Yanagawa et al., 2008). The joint reaction forces were compared with previously published simulations (Oizumi et al., 2006; Poppen and Walker, 1978; Terrier et al., 2008; van der Helm, 1994; Yanagawa et al., 2008) and in vivo telemetric implant data (Nikooyan et al., 2010). The joint reaction forces were applied to the FE model oriented in the respective GH position. Superior-inferior and antero-posterior joint translations were compared with
reported in vivo measurements on x-ray (Chopp et al., 2010; Bey et al., 2008), open MRI (Graichen et al., 2005; Sahara et al., 2007) and fluoroscopy (Teyhen et al., 2010).

5.3. RESULTS

The integrated model solved in three hours for one instantaneous position on a standard desktop PC (Intel Pentium 4, 3 GHz, 1 GB RAM). The wrapping process required 4 minutes, the muscle force estimation less than 1 minute and the FE simulation 3 hours on average. The modeling implementation of the passive intra-articular stabilizing structures (labrum and cartilage) was shown to be appropriate, with all simulated stability ratios falling within published experimental ranges (Figure 35). Quantitative comparison of the integrated model output indicated that predictions of muscular abduction moment arms (Figure 36), muscular activation patterns (Figure 37), joint reaction forces (Figure 38) and humeral head translations (Figure 39) during scapular plane elevation were all in agreement with published data.

Figure 35: Quantitative validation of stability ratios. Simulated (red stars) and reported experimental (grey areas) values. The corresponding von Mises stress distributions at the time just prior to dislocation are displayed to demonstrate that the humerus translates in the direction of the shear force.
Figure 36: Quantitative validation of moment arms.
Elevation moment arms of all muscle segments compared to published in vitro data (individual studies shown in grey, area of experimental values is shaded) during scapular plane GH elevation. The joint angles are given for a fixed scapula. The horizontal axis description is shown for the first panel only.
Integrated modeling of active glenohumeral stability

Figure 37: Qualitative validation of muscle activity.
Muscle forces required for arm elevation in the scapular plane and in vivo measured range of muscle activity by means of EMG (Habermeyer et al., 1987; Inman et al., 1944; Yanagawa et al., 2008) shown with the shaded grey area.
Figure 38: Quantitative validation of joint resultant. Simulated joint reaction force (red) during GH elevation in the scapular plane. The grey curves represent the reaction forces obtained in other modeling studies (Apreleva et al., 2000; Oizumi et al., 2006; Poppen and Walker, 1978; Terrier et al., 2008; van der Helm, 1994), and the black curves represent in vivo values recorded by means of a telemetric implant in two patients (Nikooyan et al., 2010). The corresponding contact normal stresses obtained for each elevation degree are shown below the x-axis.

Figure 39: Quantitative validation of joint translations. Humeral inferior-superior (top) and antero-posterior (bottom) translations during scapular plane elevation. The shaded gray area spans the plus minus standard deviation range of experimental values, and the grey curves show the corresponding mean values. To allow comparison between the different methods, all translations were reported with respect to the inferior-superior position of the humers at 0° GH elevation (Chopp et al., 2010; Bey et al., 2008; Sahara et al., 2007; Teyhen et al., 2010). Due to the lack of anterior-posterior translations data at low elevation angle (Graichen et al., 2005; Sahara et al., 2007), the reference position for anterior-posterior translations was set to 20° GH elevation.
Little supplementary rotator cuff activity was required to provide stability between 30 and 50° GH elevation (Figure 40), while muscle forces were generally sufficient to automatically center the joint for the rest of motion.

![Figure 40: Supplementary necessary rotator cuff action for stability. Percentage of the potential maximal force of the rotator cuff muscles necessary to stabilize the joint during GH scapular plane elevation.](image)

### 5.4. DISCUSSION

Current numerical models of the GH joint rely on *a priori* kinematic constraints that simplify the joint using ball-and-socket assumptions (Holzbaur et al., 2005; Karlsson and Peterson, 1992; van der Helm, 1994; Yanagawa et al., 2008), precluding abduction and flexion and restricting humeral vertical translations by a non-physiological distal spring (Buchler et al., 2002), or limiting muscle activity and motion to considerations in a single dimension (Terrier et al., 2007; Terrier et al., 2008). We present here an alternative model that overcomes these limitations. The model integrates our earlier work automatically predicting muscle paths, with an algorithm to estimate GH muscle activity to balance the joint in six DOF, with a newly developed deformable model of GH contact to successfully predict realistic 3D joint behavior for GH elevation. Muscle path, muscle forces and joint contact characteristics can now be computed automatically at any joint position, with a composite of instantaneous positions enabling simulation of dynamic motions. Moreover, by virtue of the muscle force estimation algorithm (see Chapter 4), external forces acting in any arbitrary direction can be...
simulated (Favre et al, 2005). This is therefore the first model that can cope with the wide GH joint range of motion in all degrees of freedom and the considerable variability in external loading conditions as encountered in daily life activities. This is especially important for clinical applications, where one wishes to have a realistic representation in more than one single special case of loading or for more than one unique motion.

A remarkable aspect of this model was its ability to replicate a wide range of experimental data. Predicted muscle moment arms, muscle on-off activity patterns, humeral head translations, joint contact forces and stability ratios delivered values and trends that compared favorably with existing experimental, modeling or in vivo data.

Previous experimental reports of an initial upward humeral head migration, followed by inferior translation throughout the rest of abduction (Bey et al., 2008; Chopp et al., 2010; Poppen and Walker, 1976) could be simulated with this model. As expected, the most important elevators were the supraspinatus (at the beginning of elevation) and deltoid (towards the end of elevation) (Figure 37). From 30 to 50° elevation, the moment arm of the external force increases (as the limb center of gravity moves away from the torso), demanding substantial middle deltoid activity. With its upward directed line of action, the middle deltoid pulls the humerus superiorly (Ackland and Pandy, 2009). In our six DOF model, this large shear force must be counteracted by additional rotator cuff activity to provide increased concavity compression. This is in accordance with a previous model for which the joint reaction force fell close to the superior border of the glenoid from 30 to 45° of GH abduction (Poppen and Walker, 1978). It should be noted that resting muscle tone (Debski et al., 1999; Schiffen et al., 2002), intraarticular pressure (Itoi et al., 1993) or subacromial spacer effect of the supraspinatus (Werner et al., 2006) might already be sufficient to confine this level of humeral head translations. After 60° of elevation, the deltoid then becomes a stabilizer, owing to a line of action that approaches a perpendicular line to the glenoid surface (Ackland and Pandy, 2009).

The muscle force estimation algorithm used in the present study previously delivered a remarkable different joint reaction pattern during active GH elevation (see chapter 4). The joint reaction force continued to increase over 60° GH elevation, mimicking the trend measured in vivo on two patients by means of telemetric implants (Nikooyan et al., 2010).
This pattern was not reproduced with the present anatomy, highlighting the influence of anatomical muscle arrangement. We could also verify the hypothesis (Terrier et al., 2008; Veeger and van der Helm, 2007) that the ball-and-socket assumption neglects muscular stabilization and leads to underestimated muscular and joint reaction forces (Figure 38).

The hypothesis that joint stability during GH elevation can theoretically be achieved exclusively through muscular activity was confirmed. For most simulated positions, the predicted muscle forces automatically centered the joint even without substantial rotator cuff activity. Moreover, at the levels for which the rotator cuff activity was required (midrange of motion), the capsule is not expected to carry load and provide stability (Veeger and van der Helm, 2007). First, this suggests that an optimal coordination of muscular activity can guarantee joint stability, while sensorimotor alterations may contribute to deficits in functional stability (Warner et al., 1996). Second, it provides a theoretical background for the hypothesis that the ligaments and capsule are secondary mechanical stabilizers of the GH joint (Lee et al., 2000; Matsen, 2002; Myers and Lephart, 2002; Schiffern et al., 2002).

Despite the progress the model makes toward a comprehensive model of GH function, several limitations should be noted. First, the contribution of the ligamentous capsule to stability, and the potential influence this may have on muscle recruitment, remains a key point for future investigation. We neglected the capsule to isolate the effects of active joint stabilization. While the capsule has minimal effect over the large majority of the modeled joint excursion, it becomes functionally relevant at range of motion extremes (Bigliani et al., 1996). The capsule is currently being added to the FE model. Second, the cartilage and the labrum anatomies were estimated according to the bony structures (not image based). This approach nonetheless has the advantage to avoid inclusion of age related soft tissue degeneration often found in these structures and validation of the implementation of these structures against experiments demonstrated that this approach was adequate. Third, the moment arms were computed with a fixed joint of rotation and were not updated with joint translations, but translations remained small (less than 1mm). Finally, muscle forces may be underestimated because dynamic effects were neglected and the contribution of multi-articular muscles for adjacent joints was not considered.
In summary, we present a robustly validated integrative framework that bridges algorithms for muscle wrapping and muscle recruitment, with a deformable model of the GH joint. This platform can be used to investigate important unsolved questions in shoulder biomechanics and surgery. While the lack of a modeled capsule may limit predictive capacity at extreme ranges of motion (remaining grounds for future work), in moderate joint positions, influences of individual shoulder anatomy, defects of the articular surfaces, muscle dysfunction, or prostheses can already be systematically and quantitatively investigated.

REFERENCES


Integrated modeling of active glenohumeral stability


CHAPTER 6
6. SYNTHESIS

REDUCING A DISLOCATED SHOULDER,
APOLLONIUS OF KITIUM, 9TH CENTURY
6.1. CONCLUSIONS

Clinical complications related to GH dislocations represent a significant health problem. A deeper understanding of the mechanisms involved in shoulder stability is needed to improve treatment. This is especially true with the growing physical activity of the aging population. An increasing number of seniors perform sports, such as tennis or golf, which put a high demand on the shoulder, eventually rising the likelihood of joint overuse.

Previously available modeling approaches to investigate GH stability were substantially limited by their inability to actively stabilize the joint, incapacity to simulate all degrees of freedoms, and were restricted to very few joint positions and loading conditions.

In this thesis, it was possible for the first time to alleviate all of these limitations. This is the first model that can cope with the wide GH joint range of motion in all degrees of freedom and the considerable variability in external loading conditions as encountered in daily life activities. This is especially important for clinical applications, where one wishes to test more than one single special case of loading, for one unique motion. Moreover, no existing model of the shoulder shows the level of validation (when existing) reached in this thesis. Muscle moment arms, muscle length, muscle force, muscle recruitment, stability ratios, joint translations and joint reaction force were compared against experimental measurements, demonstrating that the model delivered reliable results. Another unique feature of the integrated model lies in its applicability and speed. Any physiologically sound joint position and loading condition can be simulated automatically in about 3 hours only (4 minutes for the wrapping process, 10 seconds for the muscle force estimation, between 2 and 3 hours for the FE contact simulation) on a standard desktop PC (Intel Pentium 4, 3GHz, 1 GB RAM). These qualities make it a very functional and powerful tool for studying various aspects of shoulder biomechanics.

In the first part of this thesis, a novel method was established to automatically compute the muscle path in any joint position. This technique was based on the built-in contact detection offered by commercial FE solvers. This provided the capital advantage to avoid special algorithms or underlying surfaces for the definition of contacts, allowing the method to function in any joint position without preprocessing. The muscles were represented as
simplified deformable strings and bones as rigid surfaces, which kept the required calculation time considerably low in comparison to volumetric muscle representations. The validation demonstrated that realistic muscle moment arm, muscle length and line of action data could be extracted from this model.

In the second part, an algorithm was established to estimate the muscle forces required for first balancing and then stabilize the joint in the position of interest. This algorithm implemented an innovative recruitment strategy that focused on selecting muscles with a relative mechanical advantage, and a corresponding set of muscles that counterbalanced secondary joint moments. Applying this process in small iterative steps led to the unique feature to allow recruitment of antagonistic muscles. At the end of the muscle force distribution for joint equilibrium, additional incremental muscle force was attributed to the rotator cuff muscles until joint stability was satisfied. This procedure can be used to automatically estimate realistic muscle forces for any combination of physiological joint positions and external loads.

In a final part, the resultant force was applied in an anatomically precise FE model of the GH joint. A series of loading steps allowed the gradual release of the constrained humeral degrees of freedom without being confronted with rigid body motions. This deformable model was the first to allow six degrees of freedom, simulating validated joint contact characteristics for the whole elevation motion. This model, combined with the two previous methods, demonstrated that the joint can be stabilized by the muscles alone. This suggests that an optimal coordination of muscular activity can guarantee joint stability, while sensorimotor alterations may contribute to deficits in functional stability. Also, it provides a theoretical background for the hypothesis that the ligaments and capsule are secondary mechanical stabilizers of the GH joint.
6.2. VISION FOR THE FUTURE

A realistic model of the GH joint was successfully implemented. This model was originally developed to investigate pressing clinical issues of GH joint stability. However, as an integrated model, it can also be exploited to study other important aspects of shoulder surgery that are not directly connected to joint stability. Here is a non-exhaustive description of a few clinical applications (on joint stability or other issues) for which the model could be used, or toward which it could be further developed.

**Structural defects (tears of rotator cuff tendon or labrum, glenoid fracture)**

Explicit incorporation of tissue deformations and failure allows investigations into how these factors can affect muscle force activity or joint stability. Tears of the rotator cuff tendons are common, and have been shown to be present in over 50% of asymptomatic people above the age of 60 years (Sher et al., 1995), reflecting accumulated damage over the course of a lifetime. These tears may progress to massive, symptomatic tears, eventually leading to irreversible changes in the physiology of the muscle and tendon (Gerber et al., 2004), making surgical repair difficult. Torn cuff tendons can lead to pain, instability, dysfunction, osteoarthritis and in the worst cases can degenerate to cuff-tear arthropathy, in which the humeral head collapses (Neer et al., 1983). The pathogenesis of rotator cuff tears remains unclear, and the best treatment for this common disorder remains a source of debate (Williams et al., 2004). The evolution to a symptomatic tear is not well understood. Interestingly, the functional deficits associated with rotator cuff tears are highly variable among patients, varying from normal range of motion to pseudoparalysis, indicating that some patients are able to cope with the torn muscles, while others cannot. Here, our model could help understand how compensation involving the remaining viable muscles occurs and define new rehabilitation strategies. This can be simulated very easily by assigning the torn muscles a zero cross-sectional area in the ASFE. By doing so, no force is attributed to these muscles and the remaining muscles must compensate for the lost function.

The effect of a glenoid defect or a torn labrum can be simulated by removing the relevant anatomical part from the FE model. Surgical reconstructions such as the Latarjet procedure (Latarjet, 1958) or the Bankart repair (Bankart, 1938) can then be tested and improved.
Interindividual variations

Large anatomical variations can be seen between individuals (Figure 41). Our method could be transferred to patient specific models to study particular skeletal deformities, altered bone or cartilage material properties (as seen in osteoarthritis). With semi-automatic extraction of anatomy from CT and MRI images, models of individual patients are now plausible (Young et al., 2008).

In a more straightforward fashion, the influence of changes in anatomy can be investigated by parametrically changing the bony geometry or muscle arrangement in the existing model. Stability being greatly influenced by the direction of the resultant force with respect to the glenoid, the result of alteration in this relationship can be investigated. The response to changes in glenoid version, glenoid dysplasia, glenoid bone loss or scapular dyskenesia can be examined by changing the orientation of the glenoid with respect to the humerus.

Figure 41: Large interindividual variations in bone anatomy. These scapula specimens were scanned at our hospital.

Shoulder arthroplasty

The number of partial and total shoulder replacements remains relatively small in comparison to the hip or knee, but has tripled in just nearly 10 years in the US, reaching
29'000 occurrences in 2004 (Kozak et al., 2006). This growth rate is similar to that of knee arthroplasties, but with an increased revision burden (Day et al., 2010), with the two major complications being loosening of the glenoid component and joint instability (Franklin et al., 1988; Wirth and Rockwood, 1996).

Although the etiology of glenoid loosening is not yet fully understood, it is generally believed to have several potential (and possibly interrelated) sources, notably implant design, implant wear and particle formation, surgical technique, poor bone quality, and eccentric implant loading. Eccentric loading can be due to migration of the humeral head (Franklin et al., 1988) or disadvantageous implant positioning (Nyffeler et al., 2006). In its ability to explore mechanical phenomena, our FE model offers potential in investigating these issues. For studies on loosening of the glenoid component in shoulder arthroplasty, a more realistic representation of the underlying bone micro-architecture may improve the representation of the interface mechanics and force transmission in the bone, enhancing failure risk prediction (Pistoia et al., 2002; Ulrich et al., 1997). Also, long term predictions of implant fixation could be improved by including bone mechanical adaptation laws into the FE models (Beaupre et al., 1990; Cowin et al., 1992; Huiskes et al., 2000; Prendergast and Taylor, 1994). Such FE models for the shoulder do exist but have been limited to 2D analyses of the glenoid (Andreykov et al., 2005; Sharma et al., 2007). Simulation of fibrous tissue interposition between the implant and bone could also be added (Weinans et al., 1990).

As with the normal GH joint, instability of the prosthetic joint is also a primary concern. Because stability of the shoulder depends so heavily on the active musculature, our model may improve our understanding of the stability changes induced by an implant. Constrained devices such as the reverse prosthesis (in which the ball is transposed to the scapula and the socket to the humerus) provide increased intrinsic stability, but they have been associated with higher rates of surgical revision (Matsen et al., 2007). Moreover, the geometric and kinematic changes induced by the reverse design present many questions to be investigated (Boileau et al., 2005). Our model may provide a basis for studying the consequences of changes in size, type and position of the prosthesis on implant longevity, soft tissue tension and force balance.
Tendon transfer

Muscle transfer represents a viable treatment option to restore the lost function in a variety of difficult shoulder issues. The anatomical insertion of a functioning muscle is detached and transferred to a different location to fulfill a new role and restore a lost function. For instance, the latissimus dorsi transfer can restore external rotation in case of irreparable rotator cuff tears (Gerber, 1992) or be further combined with arthroplasty (Gerber et al., 2007). However, the clinical outcome of muscle transfers is still difficult to predict, and the biomechanical implications of such procedures are only beginning to be understood (Favre et al., 2008a; Magermans et al., 2004a; Magermans et al., 2004b). Our model simulating muscle wrapping can provide first information on the new muscle path and moment arm of the transferred muscle. The ASFE can then help understanding how the changes induced by the transferred muscle influence the distribution of muscle forces over the whole joint. Finally, the influence on joint reaction force can be assessed with the FE model of the joint. This would provide guidelines (choice of muscle to transfer, optimal insertion site, muscle training) to maximize the functional potential.

REFERENCES


CURRICULUM VITAE

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