BIOMECHANICAL ANALYSIS OF SHOULDER ARTHROPLASTY
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Proefschrift

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GENERAL INTRODUCTION
CHAPTER 1

1.1 BACKGROUND

1.1.1 Shoulder arthroplasty

A damaged glenohumeral joint caused by a humerus fracture through the plane of the joint or by a disease such as rheumatoid arthritis (Figure 1.1) usually results in severe pain. Pain can dramatically restrict functioning in daily life, because motions of the joint have become impossible. An effective procedure for pain relief is to replace the joint with an endoprosthesis, in clinical terms an arthroplasty. There are two types of shoulder arthroplasty. A replacement of both the humeral head and the glenoid is called a Total Shoulder Arthroplasty (TSA). When only the humeral head is replaced by an endoprosthesis, this is called a Hemi Shoulder Arthroplasty (HSA).

Despite the fact that approximately 90% of the patients report pain relief and the range of motion (ROM) improves, the functional capabilities after shoulder arthroplasty are limited. Tasks above shoulder level, like combing hair and reaching to a kitchen cupboard can only be performed by 55% of the patients with a shoulder endoprosthesis (Magermans et al. 2003). It is still unknown why the functional capabilities of patients after shoulder arthroplasty are limited.

1.1.2 DIPEX project

A large multi-disciplinary project called DIPEX (Development of Improved endoProsthesis for the Upper EXtremities) has been set up to address the problems involved with shoulder arthroplasty. The project is a collaboration between the Delft University of Technology and the university medical centers of Leiden and Rotterdam. One of the challenges of the DIPEX project is to integrate the medical and technical knowledge that subsequently will lead to a medically accepted solution.

DIPEX consists of six projects. Each project has its own objectives that are considered to be important for an improvement of shoulder arthroplasty. The first project aims to identify bottlenecks in the pre- and peroperative surgical process. This is achieved by an extensive task analysis of the surgical team. This project provides information to develop new protocols for the implantation of the prosthesis.
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The second project deals with the automated image processing of CT and MRI scans to allow pre- and peroperative 3D visualisation. On-line visualisation during surgery will lead to a better and more accurate implantation of the prosthesis. The third project involves a functional and biomechanical analysis of shoulder function after arthroplasty and is described in this thesis. The fourth project considers the fixation of the prosthesis. Fixation of the prosthesis is a very important aspect because 44% of the glenoid components show radiolucent loosening after approximately 10 years (Torchia et al., 1997). Existing fixation techniques will be described to find possible failure mechanisms and new fixation techniques using new shapes, new materials and new surfaces will be specified. The fifth project collects the results from the four projects and uses this information for the development of a new design of a shoulder endoprosthesis. The last project investigates if the accuracy of implantation of the glenoid component can be improved by means of guides and/or camera assisted implantation.

FIGURE 1.1: Left figure is an x-ray of a rheumatoid shoulder. It can be seen that the humeral head is almost coalesced to the glenoid. The middle picture is an example of a total shoulder arthroplasty, which is a replacement of humeral head and glenoid. The right figure is an example of a hemi shoulder arthroplasty, where only the humeral head is replaced.
1.1.3 Objectives

The main goal of the DIPEX programme is to develop an improved upper extremity endoprosthesis. Therefore an important objective of this thesis is to formulate specifications for the newly-designed endoprosthesis to improve functionality. To determine the specifications, the first objective of this thesis is to identify the discriminating factors in functional outcome after shoulder arthroplasty. The second objective is to find a possible treatment option to restore functioning in daily life after shoulder replacement. To identify the limitations in functional outcome, an objective is to perform a functional assessment. This means that the requirements for the performance of activities of daily living (ADL) must be specified.

Another important requirement for the new design and for the fixation of the prosthesis, is to determine the amount of load the prosthesis must be able to withstand. To accomplish this requirement, the last aim of this thesis is to determine a load spectrum of the glenohumeral joint during daily life.

1.2 KINEMATIC ANALYSIS

1.2.1 Quasi-static measurements

Shoulder motion analysis is complex, since the scapula moves underneath the skin. Therefore measuring scapula motions dynamically is difficult. Using an electromagnetic measuring device, the Flock of Birds (Ascension technology), the upper extremity motions can be measured quasi-statically (Meskers et al. 1998). Measuring quasi-statically means that each recorded motion is divided into a number of positions. Sensors are attached to thorax, humerus, forearm and to a scapulalocator (Figure 1.2) to determine position and orientation. The scapulalocator is a device to track motions of the scapula. To record scapula motions, the scapulalocator is placed on three palpated bony landmarks (Angulus Inferior, Trigonum Spinae, Angulus Acromialis) for each step. If instead of an electromagnetic system, opto electronic markers were attached to a scapulalocator to record scapula motions, the experimenter would always have to accommodate for the camera’s view. The experimenter might block the view of the markers. An electromagnetic tracking device is more suitable than conventional motion analysis systems like VICON or OPTOTRAK because the view of the sensors cannot be blocked. The Flock of Birds is thus able to continuously record position and orientation data of the sensors.

The only drawback of this measuring protocol is that since the scapula position must be palpated for each position, the motion can only be performed in a quasi-static way. These quasi-statically acquired motions represent the dynamically performed motions. Although there is a significant difference between these two types of performance, this difference is in the order of the inter-individual variance of $6^\circ$
and therefore assumed to be negligible and clinically irrelevant (de Groot et al., 1998).

Another method to obtain scapular motions is by means of regression. This method estimates scapular motions based on motions of the humerus. The regression method is only validated for healthy subjects and calculates healthy scapular motions, while patients with a limited functional outcome after arthroplasty will most likely show abnormal scapular motions. Using the scapulalocator, these abnormal motions can be identified when these motions are compared to the motions of a matched healthy control group (Vermeulen et al. 2002).

Because humeral and scapular motions are measured, this method provides information about glenohumeral function, which is defined as the motion of the humerus relative to the scapula. In the literature studies frequently report thoracohumeral function, which is defined as the motions of the humerus relative to the thorax. Ac-
ually, this is an unusual way of defining motion because the humerus and thorax do not form a joint. Due to practical aspects as mentioned above, most clinicians use the thoracohumeral values to describe upper extremity function. There are only two studies that analysed glenohumeral function during an abduction task of the shoulder after arthroplasty, using 2-D roentgen pictures (Boileau et al. 1992; Friedman, 1995). Both studies found discrepancies in glenohumeral function. The scapulothoracic rhythm, defined as the contribution of scapulothoracic motion to thoraco-humeral motion, was roughly 2:3 after arthroplasty which was different from the healthy rhythm of roughly 1:3. These results indicate that for identification of possible motion limiting factors, it is necessary to analyse 3-D glenohumeral motion for the entire range of motion of the shoulder.

The shoulder complex not only consists of the glenohumeral joint, but it includes the sternoclavicular joint, the acromioclavicular joint and the scapulothoracic gliding plane as well. Because these joints are all connected to each other, it can be stated that upper extremity function is dependent on the ability to use these joints. It might be possible that one of the joints is restricted due to the rheumatoid process. Using a kinematic analysis it is possible to identify possible compensating strategies when one of the joints is restricted.

1.2.2 Ambulatory measurements

An important requirement for the fixation and development of a shoulder prosthesis is the amount of load the prosthesis must be able to withstand. This requirement can be divided into two parts: First, the total amount of load during the life span of the prosthesis has to be determined and second, the frequency and amplitude of the peak loads must be determined. Using quasi-static measurements it is possible to calculate the glenohumeral load during one task, but only in a laboratory set-up. A 24h load spectrum is very difficult to estimate. By means of ambulatory measure-
ments it is possible to record motions for longer periods of time and during daily life.

The activities of the lower extremity and the trunk have been determined using ambulatory measurement systems. (Bussmann et al., 2001; Uiterwaal et al., 1998). However, ambulatory measurements of the upper extremity were never performed due to issues related to determining the coordinate systems of the involved segments. Moreover, the glenohumeral load spectrum has neither been determined.

Since it is not possible to track the positions of bony landmarks ambulatory, segment coordinate systems are determined using rotation axis. Instead of determining a segment axis on the basis of two bony landmarks, the segment axis is described by means of a rotation about an axis. For example, flexing and extending the elbow describes a rotation about one of the axes of the humerus and forearm. Defining the segment axes differently, will affect the position and orientation of the segment. A difference in segment orientation in combination with an external load will result in a difference in the calculated glenohumeral load. Therefore validation studies are necessary to investigate to what extent ambulatory measurements will provide accurate and useful information about the glenohumeral load spectrum.

1.3 DYNAMIC ANALYSIS

In addition to a kinematic analysis, a dynamic analysis is needed to find the contributions of the shoulder muscles to functional outcome since it is most likely that muscle status affects functional outcome after arthroplasty. Pain caused by the rheumatoid process will restrict the patient in using the arm before surgery. This immobilisation will cause atrophied shoulder muscles, which obviously will lead to a loss in muscle force. The effects of a decrease in the ability to produce force can be evaluated using a biomechanical model.

1.3.1 The Delft Shoulder and Elbow Model (DSEM)

The biomechanical model used in this thesis is the Delft Shoulder and Elbow Model (DSEM). This model is a finite element musculoskeletal model (Van der Helm, 1994) consisting of 31 shoulder and elbow muscles divided into 139 muscle elements. These 139 muscle elements include properties such as sarcomere length and Physiological Cross Sectional Area (PCSA), which are obtained from extensive cadaver studies (Veeger et al., 1991; Van der Helm et al., 1992; Klein-Breteler et al., 1999). The model can be used for multiple subjects or patients, but the individual morphologies of the simulated subjects are not taken into account. At this stage it is not feasible to acquire the musculoskeletal parameters of living subjects and therefore the morphology of one cadaver is used. Inter individual differences are accommodated for by means of measuring the upper extremity motions.
As can be seen in Figure 1.4 the model can run in two modes: inverse dynamically and forward dynamically. For inverse dynamic simulations, the input to model are recorded motions and external forces. The output of inverse dynamic simulations are muscle forces, reaction forces, joint moments, etc. When a forward dynamic approach is used, the muscles forces are used as input and the effect of these muscle force on the motions are determined. The desired mode is dependent on the objective of the study. When for example, the glenohumeral joint reaction force has to be determined, the most suitable mode is inverse dynamic because that is a direct output parameter of the inverse dynamic model. To identify the effect of a torn muscle (decrease in muscle force) on the upper extremity motions, it would be preferable to simulate forward dynamically because the effect on motion has to be studied. Simulating forward dynamically is however computationally expensive because each simulated time step is dependent on the previous time step. This means that to reach an optimal solution for one time step, the optimal solutions for all previous time steps must be calculated as well (Thelen et al. 2003). Therefore using forward dynamic simulations is not feasible at this stage. Inverse dynamic optimisation, on the other hand, is relatively fast. In this thesis an inverse dynamic approach is used to identify the mechanical role of muscles and the effect of pathological conditions, like rotator cuff tears. Furthermore, the influence of possible treatment options are investigated by changing musculoskeletal parameters. Despite the fact that pathological conditions are simulated, upper extremity motions of healthy subjects are used as input to the DSEM. If motions of patients were to be used for simulations, it would not be possible to investigate why ROM is limited, because no information is available in the area beyond the limitation. Furthermore, the musculoskeletal parameters of the DSEM can be adjusted to the pathological condition to find the influence on muscle forces. The model also generates information about joint moments, muscle moment arms and muscle lengths. In other words, the mechanical role of muscles can be analysed.

1.3.2 The Rotator Cuff Muscles

Due to the shape of the glenohumeral joint, the range of motion of the shoulder is large. However, because the glenohumeral joint is shallow and unconstrained, the humerus will dislocate without muscle activity. A subluxated humerus is a pathological condition that is also observed in spinal cord injured and CVA patients (Snels et al., 2002). These patients are also not able to activate the muscles that are required for glenohumeral stability. Although all glenohumeral muscles contribute to stability, the most important stabilisers are the rotator cuff muscles. Due to the small moment arms around the glenohumeral joint, the rotator cuff muscles can produce large compressive forces, while avoiding large antagonistic moments. The ro-
Rotator cuff muscles consist of the m. teres minor, m. infraspinatus, m. supraspinatus and the m. subscapularis (Figure 1.5). In addition to stabilising the glenohumeral joint, the rotator cuff muscles aid the prime movers of the shoulder to elevate and rotate the humerus. It is known that rotator cuff tears, in particular supraspinatus tears, restrict post-operative ROM (Gerber et al., 1988, Warner, 2001). The reason for this restriction has not been found. (Wirth & Rockwood, 1994).

Using the DSEM the mechanical role of the rotator cuff muscles can be quantified during activities of daily living, which might provide information about the aetiology of the limited shoulder function after arthroplasty (Chapter 5).

**Figure 1.4:** The two modes the Delft Shoulder and Elbow model can run in. The forward dynamic model has muscle forces as input and upper extremity motions as output. The inverse dynamic model calculates muscle forces required for the measured upper extremity motions.

**Figure 1.5:** The four rotator cuff muscles. Left figure is frontal view, right figure is dorsal view. 1 = m. teres minor; 2 = m. infraspinatus; 3 = m. supraspinatus; 4 = m. subscapularis
1.4 Treatment

Rotator cuff tears are often observed in shoulder arthroplasty patients (Edwards et al. 2002). As mentioned above, patients with massive rotator cuff tears usually have a poor prognostic outcome for function because these tears usually cannot be repaired. Small tears can be easily repaired, by means of suturing, without any negative effects on the functional outcome (Kronberg et al. 1997).

An alternative method to restore the post-operative function of patients with irreparable tears is a tendon transfer of latissimus dorsi (Figure 1.6) or teres major (Celli et al. 1998). The procedure currently in vogue is to release the latissimus dorsi tendon and to attach it to the superior part of the humerus. It has never been investigated whether this is mechanically the best option and why it is effective. It might be possible that a different muscle or attachment site is mechanically more favorable. An advantage of biomechanical modelling is the ability to adjust musculoskeletal
parameters to investigate the effects of alternative surgical procedures without using patients. The mechanical advantages and disadvantages of this procedure and adjustments to this procedure can be analysed using the DSEM. A tendon transfer procedure is simulated by attaching the tendon in the model to a new insertion. The results of the model study can subsequently be used to advise the orthopaedic surgeon about the most favorable procedure. Additionally, patients can be measured to validate the predictions of the DSEM.

1.5 OUTLINE OF THE THESIS

An extensive description of the literature with respect to possible discriminating factors for functional outcome is given in Chapter 2. Contributions of surgical, patient and design factors are discussed in this chapter.

The kinematic analysis is described in Chapter 3 and 4. The upper extremity motions of a healthy subject population and the method that is used to describe these motions can be found in Chapter 3. These motions are subsequently compared with the motions of patients with a shoulder endoprosthesis to identify the limitations in functional outcome, which is described in Chapter 4.

The dynamic analysis consists of an analysis of the mechanical role of the rotator cuff muscles during activities of daily living and a possible treatment option for the treatment of massive rotator cuff tears. The effect of the rotator cuff muscles with respect to stability and loading of the glenohumeral joint is described in Chapter 5. The treatment option that is simulated is the tendon transfer. Chapter 6 shows which tendon transfer results in the best functional outcome and Chapter 7 discusses why this transfer is the most effective in terms of mechanics.

Chapter 8 is a combination of the two analyses. Patients with rotator cuff tears are treated according to the results of the dynamic analysis. These patients have been measured pre- and post-operatively as described in the kinematic analysis to evaluate the predictions of the model.

Chapter 9 and 10 address the issues associated with the development of an ambulatory measurement system for the upper extremity. Using an ambulatory measurement system, the glenohumeral load spectrum can be determined. Chapter 9 shows the effects of different axis definitions on total segment motion and Chapter 10 shows how the glenohumeral load spectrum can be calculated and what the validity of this method is.

The last chapter will provide information about why functional outcome is limited in shoulder arthroplasty and it addresses specifications for a new endoprosthesis design.
DISCRIMINATING FACTORS IN THE OUTCOME AFTER SHOULDER ARTHROPLASTY
CHAPTER 2

2.1 INTRODUCTION

The main indication for shoulder arthroplasty is severe shoulder pain due to destruction of the glenohumeral joint, which can be caused by rheumatoid arthritis, osteoarthritis or humeral fractures. Among other things shoulder pain leads to a decreased range of motion and function (Aliabadi et al., 2000; Barrett et al., 1987; Barrett et al., 1989; Frich et al., 1988; Gill et al., 1999; Rozing and Brand, 1998). When conservative treatment does not result in pain reduction, a surgical procedure is usually performed. It is generally recognised that shoulder arthroplasty is an effective procedure in terms of pain relief. Although improvement in terms of range of motion (ROM) and function has been reported (Alund et al., 2000), it is not yet clear why optimal results are not always being reached, and what factors influence the functional results (Amstutz et al., 1988; Barrett et al., 1987; Barrett et al., 1989; Gill et al., 1999; Hawkins et al., 1989; Torchia et al., 1997; Worland and Arredondo, 1998).

Shoulder arthroplasty can be marred by complications. The incidence of complications was 14% in a study by Cofield (1984). The majority of complications involve implant failure or mechanical loosening caused by failure of the scapular fixation (Wirth and Rockwood, Jr., 1994). The combination of uncertain functional results and the risk of complications has led to the situation that some institutes favour arthrodesis, fusion of the glenohumeral joint, over arthroplasty. Although shoulder arthrodesis results in pain relief in approximately 75% of the patients, functional improvement is of course limited. Additionally, the positioning of the shoulder arthrodesis is critical. Excessive abduction and flexion causes a winging scapula, which results in shoulder pain (Clare et al., 2001).

Shoulder arthroplasty is often assumed to be a less successful procedure than hip arthroplasty. In hip arthroplasty, gain in function is higher (Christie et al., 1999). This difference in outcome between both procedures might be related to the difference in stability requirements. Due to its shallow socket and the orientation of the joint, stability in the shoulder is predominantly controlled by means of the rotator cuff muscles (Van der Helm, 1994). This is in contrast to the hip which is inherently...
stable due to a deeper and larger socket and a more favourable reaction force. The hip joint reaction force points almost vertically through the socket during standing and walking, whereas the direction of the glenohumeral joint reaction force has a larger functional range and is controlled by the rotator cuff muscles (Van der Helm, 1994).

To make shoulder arthroplasty as successful as hip arthroplasty, causes for the malfunctioning of the endoprostheses have to be identified. Roughly, there are three groups of possible factors that influence the functional outcome of arthroplasty: patient factors, design factors and surgical factors. Condition, joint status, muscle status, gender and age may be considered as patient factors. Design factors are the types of endoprostheses and fixation techniques. Surgical factors are the surgical approach and positioning of the prostheses.

It is expected that the outcome of shoulder arthroplasty will be highly dependent on the status of the rotator cuff. Since in the rheumatoid patient population the rotator cuff muscles are often compromised, it is expected that functional outcome will be unfavorable for this patient group in comparison to other patient groups irrespective of the choice of prostheses, which may differ between patient groups.

The objectives of this systematic literature review are to evaluate shoulder function after shoulder arthroplasty and to find discriminating factors related to patient, design and surgical technique in the outcome. If such factors can be defined, modifications to these factors might improve the outcome of shoulder arthroplasty.

2.2 METHODS

Several data bases, such as PubMed, Infotrieve and Web of Science, were searched with the following keywords: shoulder, arthroplasty, replacement, reconstruction, rotator cuff, rheumatoid arthritis, osteoarthritis and fractures. Only studies including functional assessment of patients were included in this review. This means that at least one functional score or one ROM test had to be included in the study. This resulted in forty-two studies (Aliabadi et al., 2000; Alund et al., 2000; Ambacher et al., 2000; Amstutz et al., 1988; Andrews et al., 2000; Arredondo and Worland, 1999; Barrett et al., 1987; Barrett et al., 1989; Baulot et al., 1995; Boileau et al., 1992; Boyd, Jr. et al., 1990; Brostrom et al., 1992; Cheng et al., 1997; Cofield, 1984; Cofield, 1994; Field et al., 1997; Frich et al., 1988; Gartsman et al., 1997; Gartsman et al., 2000; Gill et al., 1999; Godeneche et al., 1999; Hawkins et al., 1989; Kelly et al., 1987; Koch et al., 1997; Koorevaar et al., 1997; Levy and Copeland, 2001; McCoy et al., 1989; Movin et al., 1998; Neer et al., 1982; Norris and Lachiewicz, 1996; Nwakama et al., 2000; Sait and Scott, 2000; Sojbjerg et al., 1999; Sperling et al., 1998; Steinmann and Cofield, 2000; Stewart and Kelly, 1997; Stoffel et al., 2000; Torchia et al., 1997; Weiss et al., 1990; Williams, Jr. and Rock-
CHAPTER 2

Discriminating factors in the outcome after shoulder arthroplasty

wood, Jr., 1996; Worland and Arredondo, 1998; Zyto et al., 1998). The search was completed June 2001.

2.3 RESULTS

2.3.1 Quality of studies

Forty-two studies were found that met the inclusion criteria. However, these studies could not be used to conduct a formal statistical meta-analysis, since the methodological quality of the studies is insufficient. In order to establish cause and effect relationships an appropriate experimental design is necessary. A randomised clinical trial is nowadays seen as the gold standard for effect studies. Randomisation of patient and design factors is needed to identify differences in outcome related to these factors. The study of Gartsman et al (2000) is the only study that randomised hemi-prostheses and total shoulder prostheses.

Determination of the range of motion of the shoulder is often used to describe the effect of the surgical procedure. Post-operative forward flexion results were reported in 33 studies, while pre-operative forward flexion results were reported in 29 studies. Abduction was reported in 10 studies pre-operatively and in 16 studies post-operatively. External rotation was tested in 28 studies pre-operatively and in 34 studies post-operatively. The amount of internal rotation was usually determined by the level that was reached by the tip of the thumb on the spine.

ROM would be a suitable outcome measure if all studies measured ROM consistently. There are a few problems associated with ROM. Due to the complexity of shoulder movements, the latter are never purely planar. For example, forward flexion is always accompanied by external rotation. Most studies do not report how ROM was measured, which plane was used, etc. Forward flexion and abduction are considered as standardised movements. External rotation can be determined with and without humerus elevation. It would be preferable to have insight into the definitions used for describing the joint angles. The study of Hawkins et al (1989) is the only study that indicated which definitions of angles were used.

Another method to determine outcome is by functional assessment lists. A functional assessment list has additional value since it gives an indication of the level of functioning of the affected shoulder. Assessing shoulder function is very complicated due to the fact that most patients suffer from multiple-joint diseases which will influence the glenohumeral function as well. Every assessment method has to include four categories: pain, function, stability and strength. However the importance of each category depends on the individual requirements. The demands of a younger patient with recurrent dislocation differ markedly from those of an older patient with a rotator cuff tear. For some patients pain relief may thus produce sat-
isfaction out of proportion to functional improvement (Macdonald, 1993). This means that in addition to the four categories described above, the patient's perception of the outcome must be included as well (Kuhn and Blasier, 1998).

In this review 65% of the studies use a functional assessment list for evaluation. The Neer (1982) and ASES (Barrett et al., 1987) methods can be considered together because the ASES is based upon the Neer system. These assessment methods are also the most used (15 times) (Barrett et al., 1987; Barrett et al., 1989; Brostrom et al., 1992; Cheng et al., 1997; Field et al., 1997; Gartsman et al., 2000; Gill et al., 1999; Kelly et al., 1987; Koch et al., 1997; Neer et al., 1982; Sojbjerg et al., 1999; Sperling et al., 1998; Steinmann and Cofield, 2000; Stewart and Kelly, 1997; Williams, Jr. and Rockwood, Jr., 1996). The Constant score (Constant and Murley, 1987) is used in 10 studies (Alund et al., 2000; Ambacher et al., 2000; Baulot et al., 1995; Godeneche et al., 1999; Koorevaar et al., 1997; Levy and Copeland, 2001; Movin et al., 1998; Sait and Scott, 2000; Stoffel et al., 2000; Zyto et al., 1998). The Constant score (Constant and Murley, 1987) is used in 10 studies (Alund et al., 2000; Ambacher et al., 2000; Baulot et al., 1995; Godeneche et al., 1999; Koorevaar et al., 1997; Levy and Copeland, 2001; Movin et al., 1998; Sait and Scott, 2000; Stoffel et al., 2000; Zyto et al., 1998) with a mean postoperative score of 57.9 (±12.4) and the UCLA test (Amstutz et al., 1981) is used in 4 studies (Amstutz et al., 1988; Arredondo and Worland, 1999; Gartsman et al., 2000; Worland and Arredondo, 1998). In order to compare the different studies on functional outcome, all studies must score function consistently. However, there is no consensus about which assessment system should be used to evaluate the outcome of a shoulder arthroplasty. The American Shoulder and Elbow Surgeons developed the ASES, while the European Shoulder and Elbow Society adopted the Constant Score. The main difference between these scores is that the Constant score provides a total score and the Neer/ASES expresses the results in qualitative terms. A total score lends itself to comparison of multiple studies. However, in order to identify the improvements in the different categories, pain, function, ROM and stability, the subtotal scores have to be reported as well. This does not mean that qualitative terms are not advisable. Qualitative data, for example patient satisfaction and other pathology could be reported in addition to the quantitative data.

Eight studies partly satisfied these recommendations. Pain, although only postoperatively, ROM and function were reported. For the sub-score function, the number of patients that were able to perform activities of daily living were reported (Table 2.1). From Table 2.1 it can be seen that there is an improvement in function after shoulder arthroplasty, especially for the tasks below shoulder level, e.g. 82% of patients are able to wash the axilla post operatively. The tasks above shoulder level are still difficult to execute. Approximately 55% are able to perform tasks above
shoulder level. However the results have to be treated carefully since the ability to perform ADLs is measured by questionnaires, not by measurements. Results might be biased since a common problem with questionnaires is that the subjects tend to answer what the surgeon wants to hear, since the patient does not want to disappoint him. McGrory et al. (1996) confirmed that knee scores after replacement of the joint based on interviews with surgeons were significantly higher than knee scores based on interviews with independent observers. This is a common problem (Kuhn and Blasier, 1998), however only 6 studies reported that an independent observer was used to evaluate functional outcome (Alund et al., 2000; Barrett et al., 1987; Brostrom et al., 1992; Gartsman et al., 1997; Levy and Copeland, 2001; Movin et al., 1998).

2.3.2 Patient factors

Rodosky and Bigliani (1996) reported more than thirty different indications for shoulder replacement. The studies included in this review focused mainly on patients with rheumatoid arthritis (40%), osteoarthritis (30%) and fractures (10%).

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<td>28</td>
<td>4</td>
<td>27</td>
<td>2</td>
<td>24</td>
<td>3</td>
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</tr>
<tr>
<td>Barrett et al. (1987)</td>
<td>RA</td>
<td>11</td>
<td>3</td>
<td>9</td>
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<td>11</td>
<td>1</td>
<td>8</td>
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</tr>
<tr>
<td>Barrett et al. (1989)</td>
<td>RA</td>
<td>140</td>
<td>61</td>
<td>112</td>
<td>37</td>
<td>89</td>
<td>8</td>
<td>75</td>
<td></td>
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<td></td>
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</tr>
<tr>
<td>Cofield (1984)</td>
<td>ALL</td>
<td>65</td>
<td></td>
<td></td>
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<td></td>
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<tr>
<td>Movin et al. (1998)</td>
<td>FR</td>
<td>29</td>
<td>12</td>
<td>19</td>
<td>10</td>
<td></td>
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<td></td>
<td></td>
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</tr>
<tr>
<td>Nwakama et al. (2000)</td>
<td>OA</td>
<td>7</td>
<td>3</td>
<td>3</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td></td>
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</tr>
<tr>
<td>Kelly et al. (1987)</td>
<td>RA</td>
<td>40</td>
<td>17</td>
<td>34</td>
<td>12</td>
<td>39</td>
<td>5</td>
<td>22</td>
<td>5</td>
<td>29</td>
<td>10</td>
<td>10</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stewart &amp; Kelly (1997)</td>
<td>RA</td>
<td>37</td>
<td>19</td>
<td>28</td>
<td>15</td>
<td>27</td>
<td>4</td>
<td>18</td>
<td>1</td>
<td>26</td>
<td>3</td>
<td>23</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Percentages</td>
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<td></td>
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<td></td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>40% 75% 25% 82% 27% 55% 14% 57% 13% 73% 25% 85%</td>
</tr>
</tbody>
</table>
The patient population was mainly female (68.8%) and the mean age of the patients was 61.4 (± 7.7) years.

**Table 2.2:** Range of Motion of Rheumatoid Arthritis (RA) patients after shoulder arthroplasty. PreFF = Pre-operative Forward Flexion; PostFF = Post-operative Forward Flexion; PreEx = pre-operative External rotation; PostEx = post-operative External rotation. HSA = Hemi Shoulder Arthroplasty; TSA = Total Shoulder Arthroplasty.

<table>
<thead>
<tr>
<th>Procedure</th>
<th>Study</th>
<th># Shldrs</th>
<th>PreFF</th>
<th>PostFF</th>
<th>PreEx</th>
<th>PostEx</th>
</tr>
</thead>
<tbody>
<tr>
<td>HSA</td>
<td>Alund et al. (2000)</td>
<td>39</td>
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<td>83</td>
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</tr>
<tr>
<td></td>
<td>Koorevaar et al. (1987)</td>
<td>19</td>
<td>75</td>
<td>70</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Levy &amp; Copeland (1987)</td>
<td>14</td>
<td>50</td>
<td>106</td>
<td>5</td>
<td>45</td>
</tr>
<tr>
<td></td>
<td>Mean (SD)</td>
<td></td>
<td>64 (12.9)</td>
<td>86 (18.2)</td>
<td>13 (12.0)</td>
<td>36 (12.7)</td>
</tr>
<tr>
<td>TSA</td>
<td>Amstutz et al. 1988</td>
<td>18</td>
<td>40</td>
<td>80</td>
<td>38</td>
<td>48</td>
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<tr>
<td></td>
<td>Barrett et al. (1987)</td>
<td>11</td>
<td>34</td>
<td>100</td>
<td>9</td>
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</tr>
<tr>
<td></td>
<td>Barrett et al. (1989)</td>
<td>140</td>
<td>56</td>
<td>90</td>
<td>20</td>
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</tr>
<tr>
<td></td>
<td>Broström et al. 1992</td>
<td>22</td>
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<td>70</td>
<td>20</td>
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<tr>
<td></td>
<td>Cofield (1984)</td>
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<td></td>
<td></td>
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<td>49</td>
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<tr>
<td></td>
<td>Cofield (1994)</td>
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<td></td>
<td></td>
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</tr>
<tr>
<td></td>
<td>Frich et al. (1988)</td>
<td>35</td>
<td>48</td>
<td>78</td>
<td>5</td>
<td>20</td>
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<tr>
<td></td>
<td>Gill et al. (1999)</td>
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<td>58</td>
<td>100</td>
<td>17</td>
<td>36</td>
</tr>
<tr>
<td></td>
<td>Godeneche et al. (1999)</td>
<td>39</td>
<td>90</td>
<td>119</td>
<td>21</td>
<td>44</td>
</tr>
<tr>
<td></td>
<td>Hawkins et al. (1989)</td>
<td>34</td>
<td>58</td>
<td>100</td>
<td>22</td>
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</tr>
<tr>
<td></td>
<td>Kelly et al. (1987)</td>
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<td>75</td>
<td>11</td>
<td>40</td>
</tr>
<tr>
<td></td>
<td>Levy &amp; Copeland (2001)</td>
<td>27</td>
<td>47</td>
<td>104</td>
<td>6</td>
<td>44</td>
</tr>
<tr>
<td></td>
<td>McCoy et al. (1989)</td>
<td>29</td>
<td>61</td>
<td>76</td>
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<td></td>
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<tr>
<td></td>
<td>Norris et al (1996)</td>
<td>10</td>
<td>85</td>
<td>101</td>
<td>20</td>
<td>46</td>
</tr>
<tr>
<td></td>
<td>Sojbjerg et al. (1999)</td>
<td>62</td>
<td>44</td>
<td>75</td>
<td>5</td>
<td>38</td>
</tr>
<tr>
<td></td>
<td>Stewart &amp; Kelly (1997)</td>
<td>37</td>
<td>53</td>
<td>75</td>
<td>5</td>
<td>38</td>
</tr>
<tr>
<td></td>
<td>Mean (SD)</td>
<td></td>
<td>55 (16.3)</td>
<td>90 (15.0)</td>
<td>19 (11.6)</td>
<td>41 (8.2)</td>
</tr>
</tbody>
</table>
Almost all studies report pain relief. The mean percentage of patients that are free of pain post operatively is 90.4% (± 4.9). Because pain is the most important indication for surgery, almost all patients are satisfied with the result.

**Table 2.3:** Range of Motion of OsteoArthritis (OA) patients after shoulder arthroplasty. PreFF = Pre-operative Forward Flexion; PostFF = Post-operative Forward Flexion; PreEx = pre-operative External rotation; PostEx = post-operative External rotation. HSA = Hemi Shoulder Arthroplasty; TSA = Total Shoulder Arthroplasty.

<table>
<thead>
<tr>
<th>Procedure</th>
<th>Study</th>
<th># Shldrs</th>
<th>PreFF</th>
<th>PostFF</th>
<th>PreEx</th>
<th>PostEx</th>
</tr>
</thead>
<tbody>
<tr>
<td>HSA</td>
<td>Arredondo et al. (2000)</td>
<td>48</td>
<td>65</td>
<td>123</td>
<td>34</td>
<td>61</td>
</tr>
<tr>
<td></td>
<td>Gartsman et al. (2000)</td>
<td>24</td>
<td>89</td>
<td>127</td>
<td>12</td>
<td>47</td>
</tr>
<tr>
<td></td>
<td>Levy &amp; Copeland (1987)</td>
<td>5</td>
<td>72</td>
<td>130</td>
<td>10</td>
<td>44</td>
</tr>
<tr>
<td></td>
<td>Worland et al. (1998)</td>
<td>51</td>
<td>68</td>
<td>121</td>
<td>12</td>
<td>47</td>
</tr>
<tr>
<td></td>
<td><strong>Mean (SD)</strong></td>
<td><strong>74 (10.7)</strong></td>
<td><strong>125 (4.0)</strong></td>
<td><strong>19 (13.3)</strong></td>
<td><strong>51 (9.1)</strong></td>
<td></td>
</tr>
<tr>
<td>TSA</td>
<td>Amstutz et al. 1988</td>
<td>24</td>
<td>60</td>
<td>85</td>
<td>28</td>
<td>45</td>
</tr>
<tr>
<td></td>
<td>Barrett et al. (1987)</td>
<td>33</td>
<td>44</td>
<td>117</td>
<td>0</td>
<td>40</td>
</tr>
<tr>
<td></td>
<td>Baulot et al. (1995)</td>
<td>16</td>
<td>60</td>
<td>131</td>
<td>0</td>
<td>40</td>
</tr>
<tr>
<td></td>
<td>Cofield (1984)</td>
<td>31</td>
<td>86</td>
<td>141</td>
<td>14</td>
<td>49</td>
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<tr>
<td></td>
<td>Cofield (1994)</td>
<td>100</td>
<td></td>
<td>140</td>
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<td>60</td>
</tr>
<tr>
<td></td>
<td>Gartsman et al. (2000)</td>
<td>25</td>
<td>86</td>
<td>128</td>
<td>36</td>
<td>61</td>
</tr>
<tr>
<td></td>
<td>Godeneche et al. (1999)</td>
<td>148</td>
<td>103</td>
<td>152</td>
<td>9</td>
<td>44</td>
</tr>
<tr>
<td></td>
<td>Hawkins et al. (1989)</td>
<td>29</td>
<td>74</td>
<td>151</td>
<td>17</td>
<td>49</td>
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<td>Levy &amp; Copeland (2001)</td>
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<td>133</td>
<td>13</td>
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<td>Nwakama et al. (2000)</td>
<td>7</td>
<td>79</td>
<td>44</td>
<td>49</td>
<td>43</td>
</tr>
<tr>
<td></td>
<td>Torchia et al. (1997)</td>
<td>34</td>
<td>96</td>
<td>143</td>
<td>21</td>
<td>55</td>
</tr>
<tr>
<td></td>
<td><strong>Mean (SD)</strong></td>
<td><strong>75 (18.5)</strong></td>
<td><strong>124 (32.5)</strong></td>
<td><strong>21 (14.9)</strong></td>
<td><strong>50 (7.3)</strong></td>
<td></td>
</tr>
</tbody>
</table>

In Tables 2.2, 2.3 and 2.4 the active range of motion (ROM) per condition is displayed. Each condition is divided into two groups, the patients with a hemi-shoulder arthroplasty (HSA) and the patients with a total shoulder arthroplasty (TSA). It can be seen that all patient groups benefited from a shoulder arthroplasty in terms of ROM, in particular the osteoarthritis patients. An improvement of 30° in forward flexion was observed for the rheumatoid arthritis and fracture patients, in comparison to the osteoarthritis patients who gained 50°. The difference between pre-operative external rotation of the humerus and post-operative external rotation was about 20° for the rheumatoid arthritis and fracture patients and 30° for the osteoarthritis patients. The pre- and post-operative values for the osteoarthritis patients
were also higher, 125° vs 88° forward flexion, than the values of the rheumatoid arthritis and fracture patients. This improvement in ROM cannot be considered as the standard improvement since the standard deviation is large for all groups.

Table 2.4: Range of Motion of old fractures and traumatic fractures (FR) patients after shoulder arthroplasty. PreFF = Pre-operative Forward Flexion; PostFF = Post-operative Forward Flexion; PreEx = pre-operative External rotation; PostEx = post-operative External rotation. HSA = Hemi Shoulder Arthroplasty; TSA = Total Shoulder Arthroplasty.

<table>
<thead>
<tr>
<th>Procedure</th>
<th>Study</th>
<th># Shldrs</th>
<th>PreFF</th>
<th>PostFF</th>
<th>PreEx</th>
<th>PostEx</th>
</tr>
</thead>
<tbody>
<tr>
<td>HSA</td>
<td>Ambacher et al. (2000)</td>
<td>27</td>
<td>94</td>
<td>84</td>
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<td>Worland et al. (1998)</td>
<td>17</td>
<td>101</td>
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<tr>
<td></td>
<td>Zyto et al. (1998)</td>
<td>27</td>
<td>70</td>
<td>40</td>
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<td></td>
</tr>
<tr>
<td>Mean (SD)</td>
<td></td>
<td>88 (16.3)</td>
<td>58 (23.1)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TSA</td>
<td>Barrett et al. (1987)</td>
<td>6</td>
<td>12</td>
<td>79</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Cheng et al. (1997)</td>
<td>7</td>
<td>77</td>
<td>109</td>
<td>-4</td>
<td>11</td>
</tr>
<tr>
<td></td>
<td>Frich et al. (1988)</td>
<td>7</td>
<td>46</td>
<td>57</td>
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<tr>
<td></td>
<td>Torchia et al. (1997)</td>
<td>13</td>
<td>87</td>
<td>94</td>
<td>20</td>
<td>35</td>
</tr>
<tr>
<td>Mean (SD)</td>
<td></td>
<td>56 (33.8)</td>
<td>85 (22.2)</td>
<td>6 (12.5)</td>
<td>22 (12.1)</td>
<td></td>
</tr>
</tbody>
</table>

Rotator cuff function is an important aspect in the outcome of shoulder arthroplasty. Five studies reported the outcome of patients with and without rotator cuff pathology. These studies involved rheumatoid arthritis patients and osteoarthritis patients. Rotator cuff deficiencies are usually observed in rheumatoid arthritis patients. Rheumatoid arthritis patients are susceptible to rotator cuff problems since the inflammatory process also affects the surrounding tendons. This leads to pain and small ruptures of the tendons. In the study of Hawkins et al. (1989) rotator cuff problems were only present in rheumatoid arthritis patients and the average forward elevation was 88°, while osteoarthritis patients with intact rotator cuff muscles could elevate up to 150°. A decrease in active forward flexion from 79.3° pre-operatively to 44.3° post-operatively was reported by Nwakama et al. (2000). The rotator cuff muscles of all osteoarthritis patients were massively torn in this study. In a study by Cofield (1984) the patients with an intact rotator cuff had an abduction of 143°, in contrast to the patients with rotator cuff tears, who were able to abduct 63°. Torchia et al. (1997) reported that rotator cuff status was a predominant factor in the range of motion. Significant differences between intact rotator cuff, minor tear in cuff and major tear in cuff were found. The intact group was able to elevate to 136° while the major tear group averaged 68° of forward elevation. When considering the studies that score highly in ROM, the osteoarthritis patients with an intact cuff reach
123° of elevation in (Arredondo and Worland, 1999) and reach 121° of elevation in (Worland and Arredondo, 1998). However, rotator cuff status is not always a discriminating factor in active ROM. Williams and Rockwood (1996) reported a mean elevation of 120° post-operatively for 21 rheumatoid arthritis patients with rotator cuff deficiencies. Kelly et al. (1987) found less difference between the affected rotator cuff group and the non-affected group, a postoperative ROM of 64° and 77° respectively. In a study by Steinmann and Cofield (2000), the intact rotator cuff group reached 126° abduction, 7° less than the patients with cuff pathology before operation.

2.3.3 Design factors

Eleven different types of prostheses were used in the studies, although the Neer prosthesis (Neer et al., 1953; Neer et al., 1982) makes up the largest group of prostheses (67%). The Aequalis prosthesis was implanted in 9.9% of the arthroplasties and the Bipolar prosthesis (Swanson et al., 1986) in 5.7%. The other 17.4% were: the 'inverse' Delta (Grammont and Baulot, 1993), Gristina (Gristina et al., 1987), Isoelastic (Olsson et al., 1990; Sait and Scott, 2000), Scan Shoulder (Jonsson et al., 1986) and the Kessel constrained prosthesis (Brostrom et al., 1992). It is not possible to discriminate between the different prostheses on functional outcome due to the fact that most studies used multiple types of prostheses and did not report the results separately.

The humeral component was usually implanted uncemented (66.5%), while the glenoid component was implanted cemented in 87.6% of the cases. Because most studies did not report the results of the design factors separately, no distinction could be made between cemented and uncemented prostheses or between constrained and unconstrained prostheses on outcome measures.

In 16 studies (Alund et al., 2000; Ambacher et al., 2000; Andrews et al., 2000; Arredondo and Worland, 1999; Boyd, Jr. et al., 1990; Field et al., 1997; Gartsman et al., 2000; Koorevaar et al., 1997; Levy and Copeland, 2001; Olsson et al., 1990; Sait and Scott, 2000; Sperling et al., 1998; Stoffel et al., 2000; Williams, Jr. and Rockwood, Jr., 1996; Worland and Arredondo, 1998; Zyto et al., 1998) a hemi-arthroplasty was performed. The indication for a hemi-arthroplasty is not clear. It is used for all conditions (Tables 2.2, 2.3 and 2.4). It can be seen that there is no difference in outcome between the HSA and TSA groups for all conditions. There is only a difference in post-operative external rotation between the HSA and TSA group for the fracture patients. This can be explained by the fact that the HSA group are all traumatic fractures and the TSA group are old fractures.

These results are in agreement with the results of Boyd et al. (1990), Gartsman et al. (2000) and Sperling et al. (1998).
2.3.4 Surgical Factors

No distinction could be made between different approaches or between the positioning of the prostheses. Almost all studies use the delto-pectoral approach, however there is no standard approach since all surgeons have their own modifications to the standard procedure. In addition Neer (1982) states that a standard procedure is not possible because the operative technique varies with each diagnostic category. For example Brostrom et al. (1992) and Koorevaar et al. (1997) detach the deltoid from the clavicle, while most surgeons leave this intact. Another difference is leaving the coraco-acromial ligament intact (Kelly et al., 1987), Williams and Rockwood (1996). This makes it impossible to find differences in outcome between the different approaches.

It is not possible to determine the effect on outcome of the alignment of the prosthesis. The studies that report how the humeral component is implanted all use approximately the same amount of retroversion (20° - 40°). However there are some studies that evaluate the alignment of the prosthesis in cadaver material. According to Frankle et al. (2001) a more anatomic reconstruction of the humerus after fractures decreased the amount of torque required for 50° of external rotation compared to a non-anatomic reconstruction. Klages et al. (2001) demonstrated that less deltoid muscle force was needed to elevate the arm when the humeral head component
was inserted more medially. Williams et al (2001) varied the humeral head offset in the anterior-posterior direction and in the superior-inferior direction and tested the amount of internal and external rotation that was possible in an active model and in a passive model. It appears that ROM can be influenced by variations in humeral head offset, however compared to the intact joint no difference in ROM was found.

\[ \text{FIGURE 2.2: Deltopectoral approach. Deltoid muscle is split and the rotator cuff becomes visible} \]

\subsection*{2.4 Discussion}

Unfortunately most of the studies discussed here were of insufficient quality to allow for a formal meta-analysis. The intended statistical meta-analysis will therefore be limited to a systematic literature review in which published results are summarised. Without a randomised study design, no firm conclusions can be drawn from any individual study involved, because there are confounding factors. The most commonly occurring confounding factors are that the investigator was aware of the treatment used and the fact that the patient groups were not identical with respect to the measured variable. As a consequence, it might be possible that existing effects of factors on functional outcome have been missed. (Type II error). However, a randomised clinical trial is feasible for patients involved in joint arthroplasty (Barrack et al., 1997). Thus in shoulder arthroplasty this should be possible as well. In the future, as more effect studies will become available, a repetition of this study that does include a full meta-analysis is warranted. Since there are multiple factors that can affect the outcome, new studies should ideally categorise these factors. Since it
is difficult to have large homogeneous groups, multi-centre trials might offer a solution.

Results of this study indicate that shoulder arthroplasty can be classified as a successful procedure in terms of pain relief (90.4% free of pain). It appears, however, that functional improvement remains limited. Almost half of the patients still have problems using their arms at or above shoulder level (Table 2.1). It is possible that optimisation of the arthroplasty to patient factors may lead to better results.

Despite the less than ideal quality of the outcome studies it can be concluded that the status of the rotator cuff is the most important factor in functional outcome in terms of active ROM. Conforming to expectations, the functional results for the rheumatoid patient group were worse than in the osteoarthritis patient group, however not worse than in the fracture patient group.

The function of the rotator cuff is twofold: generating force and providing stability of the glenohumeral joint. Therefore both functions can affect the outcome. In order to provide more stability in the glenohumeral joint, other interventions have to take place. For example, inserting a humeral component and creating a new articulating surface with the acromial arch (Arntz et al., 1993) provided that an intact deltoid muscle is present. Another possible solution is tendon transfers of the intact muscles. In case of a torn supraspinatus tendon, Rozing and Brand (1998) detached the infraspinatus and the teres minor from the humerus and replaced these muscles more superiorly over the humeral head. In case of a large defect of the supraspinatus, the superior part of the subscapularis can also be moved superiorly. The quality of repair had a significant effect on the postoperative assessment.

Considering the fact that patient factors are the most important aspects in functional outcome, the question arises as to whether it is possible to compensate for these losses in function by means of developing or adjusting the design factors. The design should be primarily aimed at the restoration of function. Additionally, different designs are needed for different conditions. A possible solution is to insert a different type of prosthesis that compensates for the loss in rotator cuff force. A prosthesis that meets this criterion is the Delta prosthesis (Baulot et al., 1995). This is an inverse prosthesis, with the ball on the glenoid and the socket on the humerus. The glenohumeral rotation centre is lowered and medialised and therefore the lever arm of the deltoid muscles is increased by 25%. Sixteen osteoarthritis patients without a rotator cuff were treated with this prosthesis in our institution and the mean elevation angle was 131.3° postoperatively. However, long term results are not yet available, since the follow up period was 2.3 years.

More research is needed with respect to the surgical factors. It appears that modifications in placement and alignment of the humeral prosthesis in cadavers influences shoulder kinematics. However it is not known what the optimal alignment is
to reach the highest function. Furthermore, it is likely that the most suitable alignment is dependent on the status of the surrounding muscles, because moment arms will change along with changes in alignment.

On the basis of this review it can be concluded that shoulder arthroplasty can be classified as a successful procedure in terms of pain relief, but that functional improvement remains limited. Since possible discriminating factors in functional outcome are usually not categorised, it is impossible to draw conclusions with respect to design and surgical factors. It is therefore recommended that in future research multi-centre randomised clinical trials, as described by Barrack et al for the knee, is needed to get more insight into the less than ideal functioning of the shoulder after arthroplasty. More homogeneous patient groups, a detailed description of measuring the range of motion and function and independent observers are needed.
CHAPTER 2

Discriminating factors in the outcome after shoulder arthroplasty
3

Requirements for upper extremity motions during activities of daily living
CHAPTER 3

3.1 INTRODUCTION

The shoulder joint is a complex joint with many degrees of freedom. In daily life this mobility is used for a large number of different tasks. In rehabilitation practice, a number of key activities, Activities of Daily Living (ADL), are generally defined that describe the functional capacity of patients. If these tasks cannot be executed adequately due to a condition of the shoulder joint, like rheumatoid arthritis, osteoarthritis and humeral fractures, a surgical intervention like shoulder arthroplasty might be considered. Functional outcome scores like the Constant (Constant & Murley, 1987) and the ASES (Barett et al., 1987) are commonly used to evaluate the effect of a surgical intervention. These scores usually consist of ROM, pain and strength measurements and a subjective assessment of ADL. It would however be useful to also have an objective measure of functional outcome. In particular information about how ADL are performed is valuable information for evaluation and diagnosis. The performance of ADL is related to pain, strength and range of motion. Thus insight into how ADL are performed in combination with ROM, strength and pain measurements can subsequently be used to find why some patients are able to perform a task and others not.

To aid the clinician, information about how healthy subjects perform ADL and what the maximal joint angles of the upper extremity are, is needed. These data can also be used for comparison with the upper extremity motions of patients after a surgical intervention, such as shoulder arthroplasty.

At present it is still unclear what the required joint angles are to perform upper extremity ADL and how they are performed. The aim of this study is to obtain a detailed accurate 3-D description of preliminary reference data of ROM and a selection of ADL of the shoulder and elbow.
3.2 METHODS

3.2.1 Subjects

Twenty-four healthy female subjects without any shoulder complaints with a mean age of 36.8 (SD 11.8) years were included in this study. The protocol was approved by the review board of human experiments of the Delft University of Technology. All subjects gave written informed consent.

**Table 3.1:** Bony landmarks that are used to construct local coordinate systems. *Glenohumeral rotation center is calculated by means of linear regression

<table>
<thead>
<tr>
<th>Segment</th>
<th>Bony landmarks</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>Processus Xiphoideus</td>
</tr>
<tr>
<td></td>
<td>Incisura Jugularis</td>
</tr>
<tr>
<td></td>
<td>Proc Spinosus 7th cervical vertebrae</td>
</tr>
<tr>
<td></td>
<td>Proc Spinosus 8th thoracic vertebrae</td>
</tr>
<tr>
<td>Clavicle</td>
<td>Sternoclavicular joint</td>
</tr>
<tr>
<td></td>
<td>Acromioclavicular joint</td>
</tr>
<tr>
<td>Scapula</td>
<td>Processus Coracoideus</td>
</tr>
<tr>
<td></td>
<td>Acromioclavicular joint</td>
</tr>
<tr>
<td></td>
<td>Angulus Acromialis</td>
</tr>
<tr>
<td></td>
<td>Trigonum Spinae</td>
</tr>
<tr>
<td></td>
<td>Angulus Inferior</td>
</tr>
<tr>
<td>Humerus</td>
<td>Epicondylus Medialis</td>
</tr>
<tr>
<td></td>
<td>Epicondylus Lateralis</td>
</tr>
<tr>
<td></td>
<td>Glenohumeral Joint*</td>
</tr>
<tr>
<td>Forearm</td>
<td>Styloideus Ulnaris</td>
</tr>
<tr>
<td></td>
<td>Styloideus Radialis</td>
</tr>
</tbody>
</table>

3.2.2 Measurement device

A six degree-of-freedom electromagnetic tracking device, the Flock of Birds (Ascension Technology Inc., Burlington, Vermont, USA) was used for the recording of kinematic data. This device consists of one extended range transmitter that creates a 3D magnetic field. Following calibration of the measurement space as described in Meskers et al. (1998) the mean residual error was 2.3 mm for all three directions. Five sensors were used to measure simultaneously position and orientation of the upper extremity. The sensors were attached to a 6 cm long pointer, and to the sternum, humerus, forearm and a scapulalocator (Johnson et al., 1993) means of palpa-
tion of three bony landmarks on the scapula: the trigonum spinae, inferior angle and acromial angle. The pointer was used to measure bony landmarks (Table 3.1) of the upper extremity with respect to the sensor. The local vectors from bony landmarks to sensors were calculated, which were used to construct the local coordinate systems. From the local coordinate systems and the sensor motions, the bone and joint rotations were calculated. In a previous experiment of Meskers et al., (1998) the intra-subject, inter-subject and inter-observer variability were measured in 15 subjects. The intra-subject variability was approximately 2°, the inter-subject variability was about 7° and the inter-observer variability was approximately 3° for all three scapula rotations (protraction, laterorotation and spinal tilt).

3.2.3 Procedure

Seven range of motion (ROM) tasks and five activities of daily living (ADL) were measured (Table 3.2). Since dynamic tracking of the scapula is very difficult, measurements were performed in a quasi-static mode. All tasks were divided into small steps for which the positions and orientations of the sensors were recorded at each step. In a pilot study the motions of the humerus during quasi-static measurements and dynamic measurements were compared and it appeared that the motions were not different from each other.

<table>
<thead>
<tr>
<th>ROM</th>
<th>ADL</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward flexion</td>
<td>Perineal care</td>
</tr>
<tr>
<td>Retrollexion</td>
<td>Combing hair</td>
</tr>
<tr>
<td>Abduction</td>
<td>Eat with spoon</td>
</tr>
<tr>
<td>Adduction</td>
<td>Washing axilla</td>
</tr>
<tr>
<td>Internal rotation with</td>
<td>Lifting a bag (4 kg)</td>
</tr>
<tr>
<td>humerus in 90° scapular abduction</td>
<td></td>
</tr>
<tr>
<td>Elbow flexion</td>
<td>Reach above shoulder</td>
</tr>
<tr>
<td></td>
<td>level</td>
</tr>
</tbody>
</table>

3.2.4 ROM

For most ROM tasks the subjects were instructed to reach a maximal joint angle. This means that e.g. for a forward flexion and abduction task, the subject was instructed to elevate the humerus as high as possible. Internal rotation is defined as positive axial rotation of the humerus and external rotation as negative Axilla rota-
CHAPTER 3
Requirements for upper extremity motions during activities of daily living

The axial rotation task and the pronation task started differently than the other
tasks. The internal rotation with scapular abduction started with 90° of humeral el-
evation, the humerus making an angle of 30° with the frontal plane (scapular plane)and in maximum external rotation. The pronation task started with 90° of elbowflexion and the forearm maximally supinated. The step-size of most tasks was notdefined except for forward flexion and abduction. The step-size for these two taskswas indicated by marks on two semicircular pipes mounted beside the subject (Fig-
ure 3.1).

3.2.5 ADL

The selection of the ADL was made in consultation with local clinical staff. Animportant selection criteria was the importance for independent living. Besides,these tasks are also often used in evaluation scales (Constant & Murley, 1987). Sub-
jects were instructed to start in a neutral position with the arms hanging beside thebody, but were free to choose their way of performance. More specific instructionswere given for the lifting task. The 4 kg bag had to be lifted from the ground withboth arms in front of the body. Except for the lifting task, 16 subjects performed theADL without objects and 8 subjects held the specific object to perform the task.
During the experiment the kinematic data were analysed and it appeared that therewas no difference between the kinematics of the wash axilla task and the eating withspoon task. Therefore the wash axilla task was replaced by a reaching above shoul-
der level task.

FIGURE 3.1: Experimental set-up. Subject points to marks on the semicircular pipes in order to have
equal step sizes for each subject. Sensors are attached to thorax, humerus, forearm and
scapulalocator. Activities of Daily Living are also performed in this set-up.
3.2.6 Angle definitions

In this study only scapula, humerus and forearm angles were taken into account. Joint angles were expressed as Euler angles in other words: rotation about an axis. The axes are based on the local coordinate system of the bone. A local coordinate system is constructed by means of the coordinates of the measured bony landmarks. The definitions of local coordinate systems are displayed in Figure 3.2. To describe motions of a bone in 3D, at least three angles (rotations) are needed for each bone. It has to be taken into account that the order of rotation is essential (Karduna et al. 2000).

The rotation order that is used in the study was based on the International Society of Biomechanics standardisation proposal of the International Shoulder Group, where for the scapula the order was chosen as rotations about the y-z-x axes, and for the humerus as rotations about the y-z-y axes. Forearm motions were described as two independent rotations: flexion and pronation.

Scapula rotations were defined with respect to the thorax. The first rotation was defined as a rotation around the y-axis which can be defined as protraction/retraction of the scapula, where a negative rotation is a retraction motion. The second rotation was defined as a rotation around the rotated scapular z-axis where rotation is defined as mediorotation (negative sign, also known as downward rotation) and laterorotation (positive, also defined as upward rotation). The third rotation is defined as a rotation around the twice rotated scapular x-axis which can be defined as tipping forward (negative)/backward. In other words, tipping forward is when the angle changes inferior moves away from the thorax.

For example, the resting position of the scapula (with the arm hanging down) can be described as 30° to 40° protraction (positive rotation about the vertical y-axis), 0° to 10° lateral rotation around the rotated z-axis (noticable as the angle of the scapular spine with the transversal plane), and 10° to 20° tipping forward around the twice rotated (x-axis), which is aligned with the scapular spine.

The humerus angles were defined with respect to the thorax, following the globe system as described by Doorenbosch et al. (2003). The reference position is the arm in the vertical hanging position, with the longitudinal (y-axis) along the vertical axis and the x-axis (in the plane of GH, EM and EL) along the medial-lateral axis. The first rotation is defined as a rotation about the vertical axis, which was defined as plane of elevation. This rotation can best be visualised by looking from a top view to the different vertical planes around the shoulder, where 0° is when the humerus points laterally (abduction) and a positive elevation plane is when the humerus points ventrally. The second rotation is about the rotated z-axis and is defined as the humeral elevation angle, which can be interpreted as the angle between the long axis of the humerus and the long axis of the thorax. The third rotation is about the
rotated y-axis of the humerus and is defined as axial rotation of the humerus. In other words, internal (positive rotation) and external (negative) rotation of the humerus. To illustrate the used terminology: 90º abduction is defined as 0º elevation plane, 90º elevation angle and the axial rotation can vary. The difference with 90º forward flexion is 90º of elevation plane.

The forearm angles are rather straightforward: elbow flexion and pronation, where 0º coincides with the anatomical position.

3.2.7 Data Analysis

The maximal joint angles that were reached during each ROM task were calculated for all subjects. From these angles the following angles were selected for further analysis. The range of these maximal angles and the means and standard deviations were given for the following: laterorotation of the scapula (SL), the glenohumeral elevation plane (EP), glenohumeral elevation angle (EA) and glenohumeral axial rotation (AX) of the humerus, elbow flexion (EF) and pronation (PR).

To determine minimal required joint angles to perform ADL, it is assumed that the area in which problems might occur was near the goal position. For combing
hair this area is when the hand was near the head (the goal area). This does not mean
that a subject would be able to comb hair, but to reach near the head is a prerequisite
for performing the task. The maximal joint angles of the shoulder and elbow for
each subject in the goal area were used. To minimise the influence of outliers, the
5th and 95th percentile of these maximal angles were used and considered to be the
arc of motion for this population to perform ADL. It must be kept in mind, however,
that in principle, these angles cannot be considered independently. A sufficient
amount of humeral elevation without satisfactory elbow flexion can lead to the ina-
bility to perform a task.

3.3 RESULTS

3.3.1 ROM

During abduction and forward flexion the largest rotations occur scapulothoracic
and in the glenohumeral joint, as can be seen in Figure 3.3 and Figure 3.4. These
Figures show the mean motion of the scapula with respect to the thorax of 24 sub-
jects for abduction and forward flexion. It can be seen that the scapula laterorotates
50˚ during 150˚ of humeral elevation. It is also seen that there is about 80˚ of gleno-
humeral elevation, that is the motion of the humerus with respect to the scapula,
during abduction. The mean maximal elevations are similar for these two tasks (Ta-
ble 3.3).

**Table 3.3:** Maximal joint angles in degrees of 24 subjects for eight Range of Motion tasks. External rotation is negative axial rotation and internal rotation is positive axial rotation.

<table>
<thead>
<tr>
<th>ROM</th>
<th>Scapula laterorotation</th>
<th>Plane of elevation</th>
<th>Elevation angle</th>
<th>Axial rotation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Min</td>
<td>Max</td>
<td>Mean</td>
<td>SD</td>
</tr>
<tr>
<td>Forward flexion</td>
<td>25.7</td>
<td>67.1</td>
<td>52.1</td>
<td>9.5</td>
</tr>
<tr>
<td>Retroflexion</td>
<td>-3.3</td>
<td>24.3</td>
<td>11.8</td>
<td>6.9</td>
</tr>
<tr>
<td>Abduction</td>
<td>41.6</td>
<td>66.0</td>
<td>55.3</td>
<td>7.7</td>
</tr>
<tr>
<td>Adduction</td>
<td>3.9</td>
<td>45.1</td>
<td>22.3</td>
<td>10.0</td>
</tr>
<tr>
<td>Internal rot.</td>
<td>6.1</td>
<td>79.9</td>
<td>27.4</td>
<td>14.3</td>
</tr>
<tr>
<td>External rot.</td>
<td>-59.0</td>
<td>31.6</td>
<td>14.2</td>
<td>17.9</td>
</tr>
</tbody>
</table>

Mean elevation angles of the humerus are 131˚ for forward flexion vs. 132˚ for
abduction. The maximal elevation angles are 148˚ for forward flexion vs. 148˚ for
abduction. These values are not very high, which could be caused by the relatively
high mean elbow flexion of 46˚ and 33˚ in the last position of forward flexion and
CHAPTER 3
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A

B

**Figure 3.3:** Abduction kinematics. Displayed are the mean (solid lines) ± SEM (dotted lines) motion patterns of scapulo laterorotation (A) (with respect to thorax) and the glenohumeral elevation angle (B) against the thoracohumeral elevation angle of 24 healthy female subjects.

abduction respectively. The subjects probably flexed their arms to make pointing to the last mark easier. Additionally thorax rotations were observed during the eleva-
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A

FIGURE 3.4: Forward flexion kinematics. Displayed are the mean motion patterns of scapulo laterorotation (A) (with respect to thorax) and the glenohumeral elevation angle (B) against the thoracohumeral elevation angle of 24 healthy female subjects.

tion tasks. Thus an optical 180° of abduction and forward flexion is obtained. The scapula laterorotation values are also similar for both tasks. Mean values are 52° for
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forward flexion and 55° for abduction, maximum values are 67° for forward flexion and 66° for abduction. Table 3.3 also shows the results of the range of maximal joint angles. There is little variation between subjects in the maximum laterorotation, elevation angle, elbow flexion and pronation. The standard deviations for these angles are small for all tasks. However, the standard deviations for the Elevation Plane (y) and Axial rotation of the humerus (y2) are large. Because these angles are related to each other, the elevation plane will affect the axial rotation. Tasks that are performed near the 0° elevation plane, like abduction and the axial rotation tasks, can easily change sign. An abduction can be performed in -5 degrees elevation plane or in 5 degrees elevation plane. This causes the large standard deviations.

**TABLE 3.4:** Maximal joint angles of in degrees of 24 subjects used in the goal area of the activity of daily living. 5th and 95th percentiles are calculated. The 5th percentile is considered to be the minimal required angle. External rotation is negative axial rotation and internal rotation is positive axial rotation

<table>
<thead>
<tr>
<th>Task</th>
<th>Plane of elevation</th>
<th>Elevation angle</th>
<th>Axial rotation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean</td>
<td>SD</td>
<td>5th</td>
</tr>
<tr>
<td>Combing hair</td>
<td>58.5</td>
<td>14.3</td>
<td>35.7</td>
</tr>
<tr>
<td>Perineal care</td>
<td>-67.2</td>
<td>24.3</td>
<td>-27.5</td>
</tr>
<tr>
<td>Eat with spoon</td>
<td>60.0</td>
<td>14.4</td>
<td>36.3</td>
</tr>
<tr>
<td>Reaching</td>
<td>72.6</td>
<td>11.7</td>
<td>57.5</td>
</tr>
<tr>
<td>Washing</td>
<td>99.6</td>
<td>8.9</td>
<td>83.0</td>
</tr>
<tr>
<td>Lifting</td>
<td>79.2</td>
<td>18.8</td>
<td>41.4</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Task</th>
<th>Scapula laterorotation</th>
<th>Elbow flexion</th>
<th>Pronation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean</td>
<td>SD</td>
<td>5th</td>
</tr>
<tr>
<td>Combing hair</td>
<td>34.4</td>
<td>9.4</td>
<td>19.3</td>
</tr>
<tr>
<td>Perineal care</td>
<td>3.8</td>
<td>7.6</td>
<td>-7.5</td>
</tr>
<tr>
<td>Eat with spoon</td>
<td>25.9</td>
<td>8.8</td>
<td>13.3</td>
</tr>
<tr>
<td>Reaching</td>
<td>33.3</td>
<td>4.8</td>
<td>25.8</td>
</tr>
<tr>
<td>Washing</td>
<td>29.0</td>
<td>8.5</td>
<td>16.2</td>
</tr>
<tr>
<td>Lifting</td>
<td>22.6</td>
<td>13.2</td>
<td>3.3</td>
</tr>
</tbody>
</table>

3.3.2 ADL

Table 3.4 shows the results of the mean used joint angles during the selected ADL. The arcs of motion for each ADL are displayed in Figure 3.5.
Comb hair: The subjects in this experiment used at least \(73^\circ\) of humeral elevation to comb their hair. The range of humeral elevation angles from the 5th to the 95th percentile is \(29^\circ\). The subjects used large negative axial rotations (external rotation of the humerus), the range of used axial rotations goes from \(38^\circ\) to \(93^\circ\) as can be seen in Figure 3.6. It can also be seen that elbow flexion is very important (Table 3.4), in the target area the minimal elbow flexion that was used was \(112^\circ\). The highest used elbow flexion was \(157^\circ\).

Perineal care: Perineal care is the only task involving retroflexion of the humerus. The 5th percentile of humerus elevation is \(21^\circ\). The most important angle for performing the perineal care task is axial rotation. Large axial rotations (internal rotation) were observed. The 5th percentile of the axial rotation during this task is \(71^\circ\).

Eat with spoon: Eating does not require the full ROM of all joints. The most important joint angle is elbow flexion. Without elbow flexion, the subject will not be able to bring the spoon to the mouth. As seen in Table 3.4, the 5th percentile of elbow flexion during this task is \(117^\circ\).
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A

B

**FIGURE 3.6:** Combing hair kinematics. Displayed are the mean motion patterns of scapulo laterorotation (A) (with respect to thorax), the glenohumeral elevation angle (B), glenohumeral axial rotation (C) and elbow flexion (D), against the thoracohumeral elevation angle of 24 healthy female subjects.
Reaching: Reaching is the task that required the highest humeral elevation of 111°. For high humeral elevations, the scapula motion is also important. The lat-
errorotation of the scapula needed for this task is $26^\circ$. A subject usually reaches to grasp something. To grasp in the end phase of reaching pronation is needed. The 5th percentile of pronation for this task is $103^\circ$.

**Wash axilla:** As stated before, the wash axilla task is very comparable to the eating with a spoon task. The most important angles are the elevation plane and elbow flexion. An elevation plane of $83^\circ$ is used and the minimal used elbow flexion is $104^\circ$.

**Lift a 4kg bag:** The most challenging aspect of this task was the weight and not the movement.

### 3.3.3 ROM related to ADL

The ADL that require large humeral elevation angles like combing hair and reaching are comparable to the abduction and forward flexion motions. Similar amounts of glenohumeral motion are observed for these tasks. However, glenohumeral elevation is not the only prerequisite. For combing hair a combination of elevation and axial rotation is needed. The maximal axial rotation of the humerus is dependent on elevation plane and angle. This means that the maximal axial rotation without elevation is not an indicator of successful hair combing. This also applies to the perineal care task. The mean internal rotation during retroflexion ($88^\circ$) is larger than during the internal rotation task ($51^\circ$) at $90^\circ$ elevation in the scapular plane. With the humerus in retroflexion, probably more internal rotation is possible. This again shows that maximal ROM is dependent on position. A ROM task that can be used almost independent from other joint angles, to determine whether ADL are possible, is elbow flexion. Elbow ROM is in principle not dependent on humeral angles. As expected large elbow flexions were needed for most ADL.

### 3.4 Discussion

The goal of this study was to find the minimal requirements to perform ADL and how these ADL are performed. This study covers only a subset of tasks that are performed in daily living. Since only twenty-four subjects were measured, it might be possible that other combinations of joint angles to perform these tasks are possible. Motions of the head, in particular cervical spine motion, and wrist motions are not taken into account in this motion analysis. These two motions can also compensate for loss of motion in other joints. In particular patients with a shoulder endoprosthesis are expected to move differently. Patients will probably use compensatory movements to perform tasks. A future experiment following the same protocol for patients will demonstrate the necessity to compensate. However, it should be attempted to attain the normal motion patterns after surgery. In other words, in surgery it should be attempted to enable a normal way of performing ADL. A patient
study can also give more insight into the relationship between ROM and ADL. Differences in the ability to perform a task in a homogeneous patient group can indicate if the difference in the ability to perform a task is due to limiting compensation strategies or due to a limitation in ROM.

The results of this study can be compared to a study by Morrey et al. (1981) who determined the boundaries of motion for the elbow and forearm angles for a selection of ADL. Morrey et al. found 107° elbow flexion and 44° pronation for a combing hair position, which is comparable to the 112° elbow flexion and 54° pronation found in this study. Although differences are small, factors such as task definition and a different measurement device could account for these differences. Morrey et al. (1981) instructed the subjects to assume a position, like combing hair, which was then recorded. In the current study, all positions near the goal area are taken into account. A different measurement device, a triaxial goniometer attached to the subjects arm to determine the amount of elbow flexion and pronation is another possible explanation for the differences found. The Flock of Birds methodology is less likely to have measurement artefacts since the devices attached to the segments are not physically connected to each other.

To illustrate the use of the data, an indication of what the most limiting factor to perform most ADL would be for patients after shoulder arthroplasty, the measured ranges can be compared to post-operative ROM values found in the literature. Though it must be kept in mind that these post-operative ROM-values are measured using a goniometer. Magermans et al. (2003) concluded on the basis of a review study that for 735 rheumatoid arthritis patients the mean maximal elevation angle pre-operatively was 56° (SD 14) and 84° (SD 11) post-operatively, where ROM in healthy subjects is approximately 160°. This means that the elevation angle that patients have postoperatively would probably be sufficient to perform most ADL, except reaching. Since 55% of the patients are able to comb their hair, elevation is probably not the most limiting factor. Since large external rotations are observed during combing hair and the mean post-operative external rotation without elevation is 39° SD 9, external rotation might be a possible limiting factor for patients. This also shows that considering one ROM value, e.g. humeral elevation, is not a good indicator for the ability to perform an ADL. A combination of two or more angles is needed to perform ADL. In case of hair combing this is humeral elevation and external rotation.

Concerning the forearm, the post-operative ROM of the forearm is calculated by combining the results of Kudo et al (1994), Rozing (2000) and Wright et al (2000). These studies report almost the same amount of elbow flexion and pronation after implanting an elbow prosthesis. A mean elbow flexion angle of 131° and a mean pronation angle of 136° was possible post-operatively in the study of Kudo et al. Ro-
zing reported a range of 32 - 132° for elbow flexion and a mean pronation angle of 165°. In the study of Wright et al. a mean elbow flexion of 125° and a mean pronation angle of 162° were found. Elbow flexion and pronation thus do not appear to be limiting factors for performing this selection of ADL after implantation of an elbow prosthesis.

This study shows an extensive quantitative analysis of upper extremity motions during a selection of ADL. The minimal requirements for the performance of a subset of activities of daily living are determined. It can be stated that when these data will be compared to the motion patterns of patients with shoulder impairments, the limiting motion or combination of motions in the performance of ADL can be identified.
SHOULDER MOTIONS
AFTER ARTHROPLASTY
CHAPTER 4

4.1 INTRODUCTION

Shoulder arthroplasty is generally recognised as an effective procedure in terms of pain relief. Most studies report improvements in Range of Motion (ROM) and functional outcome. However, when compared to matched healthy individuals these improvements are still not near normal ROM. Patients with rheumatoid arthritis achieve an average post-operative thoracohumeral elevation angle of 100° (Gill et al. 1999; Levy & Copeland, 2001; Torchia et al. 1997) compared to an average of 160° in healthy individuals. After shoulder arthroplasty 75% of the patient population is able to perform perineal care, but only 55% are able to comb their hair and to reach shoulder level (Magermans et al. 2003). It is still not known what the determining factors are for this reduced functional outcome because of its multifactorial character (Levy & Copeland, 2001).

Motion analysis can give insight into some of the possible causes of poor functioning (Anglin & Weiss, 2000). There are, however, only two studies that have investigated 2D upper extremity motion patterns during a standardised scapular abduction after shoulder arthroplasty (Boileau et al. 1992; Friedman, 1995). Both studies found an increase in scapulothoracic motion contributing to arm elevation, or a 2:3 relationship instead of the healthy scapulohumeral rhythm of 1:3 (de Groot et al., 1998). However, it is not known to what extent a disturbed scapulohumeral rhythm is related to functional outcome or the ability to perform Activities of Daily Living (ADL) and what is causing the disturbed ratio.

The purpose of this study is to investigate to what extent upper extremity ROM is related to the ability to perform ADL after shoulder arthroplasty. In other words, how is post-operative glenohumeral ROM related to functional outcome? It is hypothesized that scapulothoracic motion of patients is not restricted, but that glenohumeral motion is disturbed after implantation of a prosthesis. This means that to compensate for this loss in glenohumeral motion, more scapulothoracic motion will be needed to be able to reach the same thoracohumeral elevation angle. Differences between patients that are able to perform tasks and patients that are not able to per-
form tasks will thus become apparent in differences in glenohumeral ROM or in the ability to compensate.

### 4.2 Materials and Methods

#### 4.2.1 Subjects

Thirteen patients (sixteen shoulders) have been measured. Patient characteristics are shown in Table 4.1. Mean age of the patients was 66.6 ± 16.2 years. In seven shoulders only the humeral head was replaced, a Hemi Shoulder Arthroplasty (HSA). In nine shoulders the humeral head and the glenoid were replaced, a Total Shoulder Arthroplasty (TSA). All subjects gave written informed consent prior to the experiment.

**Table 4.1:** Patient Characteristics. RA = Rheumatoid Arthritis; OA = OsteoArthritis; HSA = Hemi Shoulder Arthroplasty; TSA = Total Shoulder Arthroplasty

<table>
<thead>
<tr>
<th>Age</th>
<th>Gender</th>
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<td>F</td>
<td>RA</td>
<td>Eska/HSA</td>
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<td>RA</td>
<td>Biomed/TSA</td>
<td>L</td>
<td>95</td>
</tr>
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</table>

#### 4.2.2 Measurement Device

A six degree-of-freedom electromagnetic tracking device, the Flock of Birds (Ascension Technology Inc., Burlington, Vermont, USA), was used for the recording of kinematic data. Following calibration of the measurement space, the mean residual
error was 2.3 mm for all three directions. Five sensors were used to measure position and orientation of the upper extremity. The sensors were attached to a pointer, the sternum, humerus, forearm and a scapulalocator (Johnson et al. 1993). The latter device recorded the scapula positions by means of palpation of three bony landmarks on the scapula: the trigonum spinae, inferior angle and acromial angle. The pointer was used to measure bony landmarks of the upper extremity, which are used to construct the local coordinate systems. From the local coordinate systems and sensor motions the bone and joint rotations were calculated (Meskers et al. 1998).

4.2.3 Procedure

Four Range Of Motion (ROM) tasks, forward flexion, abduction, internal and external rotation with humeral elevation, and six Activities of Daily Living (ADL) were measured. The following tasks were measured: perineal care, combing hair, eating with a spoon, washing axilla, reaching above shoulder level, and lifting a 4 kg bag.

Since dynamic tracking of the scapula is very difficult, measurements were performed in a quasi-static mode. All tasks were divided into small steps for which the positions and orientations of the sensors were recorded at each step. In a pilot study the motions of the humerus during quasi-static measurements and dynamic measurements were compared and it was found that the motions were not significantly different from each other.

4.2.4 ROM

For most ROM tasks the subjects were instructed to reach a maximal joint angle. This means that e.g. for the forward flexion and abduction tasks, the subject was instructed to elevate the humerus as high as possible. Internal rotation is defined as positive axial rotation of the humerus and external rotation as negative axial rotation. The axial rotation task and the pronation task started differently than the other tasks. The internal rotation with scapular abduction started with 90° of humeral elevation, the humerus making an angle of 30° with the frontal plane (scapular plane) and in maximum external rotation. The pronation task started with 90° of elbow flexion and the forearm maximally supinated. The step-size of most tasks was not defined except for forward flexion and abduction. The step-size for these two tasks was indicated by marks on two semicircular pipes mounted beside the subject.

4.2.5 ADL

The selection of the ADL was made in consultation with local clinical staff. Selection criteria were the importance for independent living and the somewhat challenging character of the tasks. These tasks are also often used in evaluation scales (Barett et al. 1987; Constant & Murley, 1987). Subjects were instructed to start in a
neutral position with the arms hanging beside the body, but were free to choose their way of performance. More specific instructions were given for the lifting task. The 4 kg bag had to be lifted from the ground with both arms in front of the body. Except for the lifting task, 16 subjects performed the ADL without objects and 8 subjects held the specific object to perform the task. During the experiment the kinematic data were analysed and it appeared that there was no difference between the kinematics of the wash axilla task and the eating with spoon task. Therefore the wash axilla task was replaced by a reaching above shoulder level task.

4.2.6 Angle definitions

In this study the motions of clavicle, scapula and of the glenohumeral joint were taken into account. Joint angles were expressed as Euler angles in other words: rotation about an axis (Meskers et al. 1998). The axes are based on the local coordinate system of the bone. A local coordinate system is constructed by means of the coordinates of the measured bony landmarks. The rotation order that is used in the study was based on the International Society of Biomechanics standardisation proposal of the International Shoulder Group, where for the clavicle and scapula the order was chosen as rotations about the y-z-x axes, and for the glenohumeral joint as rotations about the y-z-y axes.

Clavicle and scapula rotations were defined with respect to the thorax. The first rotation of the clavicle and scapula was defined as a rotation around the y-axis which can be defined as protraction/retraction, where a negative rotation is a retraction motion. The second rotation of the clavicle was defined as a rotation around the rotated clavicular z-axis, where a positive rotation is defined as clavicular elevation and a negative rotation as depression. The second rotation of the scapula was defined as a rotation around the rotated scapular z-axis where rotation is defined as mediorotation (negative sign, also known as downward rotation) and laterorotation (positive, also defined as upward rotation). The third rotation of the clavicle is defined as a rotation around the twice rotated clavicular x-axis which can be defined as axial rotation (rotation about the long axis of the clavicle). The third rotation of the scapula is defined as a rotation around the twice rotated scapular x-axis which can be defined as tipping forward (negative)/backward. In other words, tipping forward is when the angulus inferior moves away from the thorax.

For example, the resting position of the scapula (with the arm hanging down) can be described as 30° to 40° protraction (positive rotation about the vertical y-axis), 0° to 10° lateral rotation around the rotated z-axis (noticeable as the angle of the scapular spine with the transversal plane), and 10° to 20° tipping forward around the twice rotated (x-axis), which is aligned with the scapular spine.
The glenohumeral joint angles were defined as motions of the humerus with respect to the scapula, following the globe system as described by Doorenbosch et al. (2003). The reference position is the arm in the vertical hanging position, with the longitudinal (y-axis) along the vertical axis and the x-axis (in the plane of GH, EM and EL) along the medial-lateral axis. The first rotation is defined as a rotation about the vertical axis, which was defined as plane of elevation. This rotation can best be visualised by looking from a top view to the different vertical planes around the shoulder, where 0º is when the humerus points laterally (abduction) and a positive elevation plane is when the humerus points ventrally. The second rotation is about the rotated z-axis and is defined as the glenohumeral elevation angle, which can be interpreted as the angle between the long axis of the humerus and the scapular spine. The third rotation is about the rotated y-axis of the humerus and is defined as axial rotation of the humerus. In other words, internal (positive rotation) and external (negative) rotation of the humerus. The forearm angles are rather straightforward: elbow flexion and pronation, where 0º coincides with the anatomical position.

4.2.7 Data analysis

On the basis of the ADL results, the patient population was divided into two groups: a patient group which was able to perform a task and a patient group which was not able to reach the target position. An ANOVA was applied to determine whether differences existed between the two patient groups. Using a Bonferroni post-hoc test, differences among the groups (able patients and unable patients) with respect to ROM could subsequently be identified.

A B-spline fitting method with penalties was used to calculate the mean motion pattern. This fitting method can also be used as an exploratory method to find statistical differences in the motion patterns of different groups. The 95% confidence intervals of the mean motion patterns were calculated and when these intervals did not intersect each other, it could be concluded that the motion patterns differ significantly (Eilers & Marx, 1996).

4.3 RESULTS

4.3.1 Ability to perform ADL

The tasks that required a high humeral elevation were the most difficult tasks to perform. Post-operatively 81.3% of the patients were not able to reach to shoulder level and 37.5% patients were not able to comb their hair. One patient (6.3%) was unable to perform the perineal care task. None of the patients was able to lift a 4kg bag, however 2 kg could be lifted by most patients. The patients did not experience
any problems during the eating with a spoon and washing the axilla task. These two tasks could be performed by all patients.

**Figure 4.1:** Glenohumeral ROM for the patient group that was able to comb hair and for the patient group that was not able to comb hair. AB = abduction, FF = forward flexion, IR = internal rotation, ER = external rotation. Only glenohumeral ER was significantly different.

### 4.3.2 Successful patients versus non-successful patients

**ROM** A difference in ROM between two patient groups might be an explanation as to why some patients were successful and other patients were not in performing a task. With respect to the ability to perform a combing hair or reaching high task, this might be related to a difference in glenohumeral ROM between both patient groups. Concerning glenohumeral ROM (Figure 4.1), it appeared that there was no significant difference between the two patient groups for abduction (F = .12, p>.05), forward flexion (F =.53, p>.05) and internal rotation at a high humeral elevation angle (F = 1.27, p>.05). The only significant difference can be found in glenohumeral external rotation (F = 4.86, p <.05).

**ADL** Combing hair and reaching above shoulder level are both tasks that require high (between 60° and 80°) glenohumeral elevation angles. Less glenohumeral elevation was observed during combing hair for the patients that were able to perform the task and for the patients that were not able to comb their hair (Figure...
Figure 4.2: Mean motion patterns with 95% CI of glenohumeral elevation (A), glenohumeral axial rotation (B), scapulothoracic protraction (C), scapulothoracic laterorotation (D) scapulothoracic posterior tilt (E) and SC retraction (F) during combing hair of patients who were less able to comb their hair and patients who were less able to comb their hair.
**Figure 4.2:** Mean motion patterns with 95% CI of glenohumeral elevation (A), glenohumeral axial rotation (B), scapulothoracic protraction (C), scapulothoracic laterorotation (D), scapulothoracic posterior tilt (E) and SC retraction (F) during combing hair of patients who were less able to comb their hair and patients who were less able to comb their hair.
**FIGURE 4.2:** Mean motion patterns with 95% CI of glenohumeral elevation (A), glenohumeral axial rotation (B), scapulothoracic protraction (C), scapulothoracic laterorotation (D) scapulothoracic posterior tilt (E) and SC retraction (F) during combing hair of patients who were less able to comb their hair and patients who were less able to comb their hair.
4.2A & Figure 4.2B). A possible discriminating factor between these two groups is the ability to compensate for loss of glenohumeral motion. A compensating strategy might be the use of more scapulothoracic motion which could result in higher humeral elevation angles. This could mean that successful hair combers are better scapulothoracic compensators than non-successful patients. In Figure 4.2C to E it can be seen that there was hardly any difference between the scapulothoracic motions of both patient groups. The able patients showed slightly more posterior scapular tilt and slightly more scapular laterorotation. Most compensation was found in the sternoclavicular joint, as displayed in Figure 4.2F. Able patients appeared to compensate in the sternoclavicular joint by retracting the clavicle 15º more than the less able patients. Similar compensatory motions were also observed for the reaching task.

The ability to perform perineal care was determined mainly by the glenohumeral elevation angle. One patient was not able to perform the perineal care task. This patient was able to sufficiently rotate the humerus internally, however only 10º of glenohumeral elevation was observed (Figure 4.3). In other words, the patient was not able to bring the humerus behind the body.

4.4 DISCUSSION

The ADL which were most difficult to perform were hair combing and reaching above shoulder level. 38% Of the patients were not able to comb their hair and 81% of the patients were not able to reach above shoulder level. These are both tasks that require high glenohumeral elevation angles. In contrast to what was expected, it appeared that there was no significant difference in glenohumeral ROM between both patient groups.

Patients that were able to perform the difficult tasks like combing hair showed more compensatory motions. Slightly more scapulothoracic motions were observed in the successful hair combing patient group, but the differences became particularly visible in the sternoclavicular joint. The successful patient group showed significantly more clavicular retraction than the patient group that was unable to comb their hair. This compensatory motion was needed to make an additional thoraco-humeral external rotation possible. It appeared that in this mechanism, the clavicle retracts which causes the scapula to tilt posteriorly and since there is almost no glenohumeral motion possible, the humerus will move with the scapula. This results in an additional thoraco-humeral external rotation. This additional external rotation is also observed in the external rotation ROM of the able patient group. The compensation mechanism thus functions as follows: the clavicle retracts more to produce more external rotation of the humerus. The additional external rotation of the humerus is thus required for hair combing. In conclusion, compensatory motions
Shoulder motions after arthroplasty are very important to achieve better function after implantation of a shoulder endoprosthesis.

**FIGURE 4.3:** Mean motion patterns with 95% CI of glenohumeral elevation (A) and scapulothoracal laterorotation (B) during perineal care of patients who were able to perform the task and of a patient who was not able to perform the task.
A possible explanation for the observed limitation in glenohumeral ROM might be that rotator cuff muscles are atrophied to such an extent that glenohumeral motion is not possible any more. Passive ROM measurements would have been useful to obtain insight into the maximal achievable glenohumeral motion. If passive ROM had been higher than active ROM this would mean that more glenohumeral motion is in principle possible. A lack of rotator cuff force could result in reduced stability of the joint. Stabilising the glenohumeral joint with insufficient rotator cuff force will probably cause co-contraction of alternative muscles. This co-contraction would subsequently limit glenohumeral motion. It is of course also possible that the prime shoulder movers were in fact not able to lift the weight of the arm. To investigate which of these two mechanisms were responsible for the limited glenohumeral motion, a future EMG study will indicate to what extent muscles are active.

Another possible explanation for why glenohumeral ROM might be limited, is that due to the implantation of the prosthesis the glenohumeral rotation centre with respect to the humeral shaft was changed after implantation. According to de Leest et al. (1996) the retroversion angle is an important aspect in the positioning of the rotation centre. A misoriented humeral head prosthesis could cause a change in the moment arms of muscles that cross the glenohumeral joint. However it is not known to what extent a change in moment arms would affect functioning or how accurate the positioning of the prosthesis should be in order to achieve a maximal functional outcome. The effect of an implanted prosthesis on muscle moment arms will in the future be investigated using a biomechanical model of the upper extremity.

In conclusion, a sufficient amount of glenohumeral external rotation is needed for activities of daily living that require high glenohumeral elevation angles. After shoulder arthroplasty the amount of glenohumeral motion is usually restricted, but it appears to be possible to compensate for this limitation by means of clavicular retraction. It must be taken into account that compensating strategies could of course cause secondary problems in other joints, which will ultimately affect total motion. Therefore, improving glenohumeral range of motion should remain the most important aspect in shoulder arthroplasty and focusing on compensating strategies is not the best solution. To assist the clinician in diagnosis and evaluation, it is advisable to measure glenohumeral ROM instead of thoracohumeral range of motion, because this will give more insight into joint status and possible functional outcome. Future research must indicate why glenohumeral range of motion is limited and how it can be improved. Important aspects that need attention are the effect of muscle status and accuracy of prosthesis implantation on functional outcome. Whether shoulder arthroplasty results can be improved in the future, will be dependent on the availability of this information.
FUNCTIONING OF THE ROTATOR CUFF MUSCLES DURING ACTIVITIES OF DAILY LIVING
CHAPTER 5

5.1 INTRODUCTION

Shoulder joint involvement in rheumatoid arthritis often leads to pain and loss of function of the upper extremity. To reduce pain and improve function, a shoulder endoprosthesis can be implanted. Although results in terms of pain reduction have been quite satisfactory, several studies have indicated that functional results, characterised in terms of Activities of Daily Living (ADL), were less than optimal (Barrett et al., 1987; Barrett et al., 1989; Cofield, 1984; Stewart & Kelly, 1997).

On the basis of studies by Torchia et al. (1997), Cofield (1984, Hawkins et al. (1989) it could be concluded that poor rotator cuff function was a predominant factor in the outcome of shoulder replacement surgery (Magermans et al. 2003). Torchia et al. (1997) found a significant difference in post-operative range of motion between the ‘intact rotator cuff’ group and the ‘major tear in rotator cuff’ group. All patient groups were included. The intact group was able to forward elevate the humerus 136° post-operatively, while the major tear group averaged 68° of forward elevation. In the study by Cofield (1984), the patients with an intact rotator cuff were able to abduct 143°, in contrast to patients with rotator cuff tears, who were able to abduct 63°. In a study by Hawkins et al. (1989) 28 rotator cuff problems were present in rheumatoid arthritis patients and the average forward elevation after shoulder replacement was 88°, while osteoarthritis patients with intact rotator cuff muscles could elevate up to 150°.

Despite the importance of rotator cuff muscles in the outcome after implantation of an endoprosthesis, at this stage it is not known how the muscles contribute to functional outcome. Stability is thought to be an important function of the rotator cuff. The rotator cuff muscles have relatively small moment arms around the gleno-humeral joint allowing them to generate large compressive forces in almost every position of the humerus, while avoiding large antagonistic moments. This prevents the humeral head from luxating. The other function of the rotator cuff muscles is synergistic, which means that the rotator cuff muscles assist the prime shoulder movers in moving the humerus. At this stage, it is unknown which of these two
functions can be considered as critical in the functional outcome after shoulder arthroplasty.

Using cadaver models it is possible to obtain insight into the function of the rotator cuff muscles. The amount of external force that is needed to subluxate the gleno-humeral joint when the rotator cuff muscles are producing a certain amount of force has been used to determine the mechanical role of the rotator cuff muscles (Blasier et al., 1992, Soslowsky et al., 1997, Wuelker et al., 1998). Halder et al. (2001) applied force to cadaver tendons and measured the humeral translation in the glenoid in order to determine the contributions of rotator cuff muscles to the prevention of glenohumeral luxation. It is, however, difficult to consider the entire shoulder complex with cadaver models. In addition it is difficult to take the function of other shoulder muscles into account while only a limited range of positions can be studied.

The aim of this study is to evaluate the mechanical effect of the rotator cuff during normal activities of daily living using an inverse dynamic biomechanical shoulder and elbow model. It is hypothesised that a decrease in rotator cuff force will lead to a poor functional outcome. This poor functional outcome can either be the result of a lack of glenohumeral stability or a lack of synergistic function, i.e. the required force to balance the external moments.

5.2 METHODS

5.2.1 Subjects

Twenty-four healthy female subjects (36.8 ± 11.8 yr.) with no history of shoulder complaints participated. Written informed consent was obtained from all subjects.

5.2.2 Measurement device

A six degree-of-freedom electromagnetic tracking device, the Flock of Birds (Ascension Technology Inc., Burlington, Vermont, USA) was used for the recording of kinematic data. This device consists of one extended range transmitter that creates a 3D magnetic field and a number of receivers whose positions and orientations were recorded. Following calibration of the measurement space, the mean residual error was 2.3 mm for all three directions. Five sensors were used to measure the position and orientation of the upper extremity. The sensors were attached to a pointer, the sternum, humerus, forearm and a scapulalocator (Johnson et al. 1993). The latter device tracks the scapula motions by means of palpation of three bony landmarks on the scapula: the trigonum spinae, inferior angle and acromial angle. The pointer was used to measure bony landmarks of the upper extremity. The local vectors from bony landmarks to sensors were calculated and were used to construct local coordi-
nate systems. From the local coordinate systems the bone and joint rotations were calculated (Meskers et al., 1998).

5.2.3 Procedure

Three Range Of Motion (ROM) tasks and six Activities of Daily Living (ADL) were measured. The following tasks were measured: forward flexion, backward flexion, abduction, perineal care, combing hair, washing axilla, reaching above shoulder level, and lifting a 4 kg bag.

Since dynamic tracking of the scapula is impossible due to extensive movement of the bone under the skin, measurements were performed in a quasi-static way. All tasks were divided into small steps for which the positions and orientations of the sensors were recorded at each step. For the ROM tasks the subjects were instructed to reach a maximal joint angle.

The selection of the ADL was made in consultation with local clinical staff. The ADL were selected because they are important for independent living and are somewhat challenging. These tasks are also often used in upper extremity evaluation scales (Barrett et al., 1987; Constant and Murley, 1987). The subjects were instructed to start in a neutral position with the arms hanging beside the body, but were free to choose their way of performance. More specific instructions were given for the lifting task. The bag had to be lifted from the ground with both arms in front of the body. Except for the lifting task, 16 subjects performed the ADL without objects and 8 subjects held the specific object to perform the task. During the experiment the kinematic data were analysed and it appeared that the motions of the wash axilla task and the eating with a spoon task were highly comparable. Therefore the wash axilla task was replaced by a reaching above shoulder level task.

5.2.4 The Delft Shoulder and Elbow Model

The Delft Shoulder and Elbow Model (Van der Helm, 1994) is a finite element musculoskeletal model consisting of 31 muscles divided into 139 muscle elements. The rotations of thorax, clavicle, scapula, humerus and forearm are the input variables for the model. The model calculates the muscle forces required for each measured position to satisfy equilibrium and stability constraints. The constraint that the external moments must be balanced by the muscle forces is termed the moment constraint and the requirement for the stability in the glenohumeral joint (that the reaction force vector must be directed into the glenoid cavity) is termed the stability constraint. The ‘load sharing problem’ (the distribution of muscle forces) is solved using an optimisation routine in which the criterion is the minimisation of the sum of squared muscle stresses. Forces were also bounded by the inclusion of muscle length-tension relations where maximum muscle force is function of maximum
muscle stress, PCSA and optimal muscle length. The musculo-skeletal parameters were obtained from a previous cadaver study (Klein Breteler et al., 1999).

The effect of passive structures, like the shoulder capsule, are not taken into account in the model. The reason for this is that passive structures are difficult to model in an inverse dynamic analysis due to the very high stiffness and because the zero-force length and stress-strain relationship are not precisely known. These two aspects make the loads calculated in these structures very sensitive to errors in kinematics. Another aspect that has not been taken into account is pain. Pain is a very complex variable and therefore very difficult to model.

5.2.5 Simulations

All measured activities were simulated with the model. This means that the shoulder and forearm kinematics of all subjects and were used as input to the model and the muscle forces necessary to produce these kinematics were calculated. To investigate the effect of different musculo-skeletal parameters, the maximal possible force of each rotator cuff muscle was restricted. In general, the maximum force the rotator cuff muscles are able to generate is defined at the optimum length, by the PCSA of the rotator cuff muscles and a maximum muscle stress, which is assumed to be 100 N · cm⁻². In the DSEM this resulted in a maximum rotator cuff force of 3981 N, where the infraspinatus can generate 1432 N, the teres minor can generate 497 N, the supraspinatus can generate 621 N and the subscapularis can generate 1431 N maximally.

Because the relation between rotator cuff pathology and force reduction is unknown, the measured activities were simulated with a maximal rotator cuff force limited to 10% - 50% (398N - 1990N) of the original rotator cuff force. In addition, all activities were simulated with the omission of a single rotator cuff muscle, that is: without either infraspinatus, supraspinatus, subscapularis or teres minor, respectively.

5.2.6 Data analysis

All movements for all subjects (192 cases) were used as input. For all cases simulations were run with the original parameter set of the model (normal situation), with limited rotator cuff force (0 - 50%) and without individual rotator cuff muscles.

The simulations were classified as either successful or not successful. When the model was not able to find a solution, i.e. not able to meet its constraints, this was defined as non-successful. Non-successful simulations occurred either because of the inability to generate a moment balance, or the inability to produce a gleno-humeral joint reaction force that was directed into the glenoid. These two options refer to the function of the rotator cuff as either an essential contributor to the mo-
ment balance, or as an essential stabiliser of the glenohumeral joint. As such, the results can be interpreted as an indication of the main function of the rotator cuff muscles.

In addition to the functional analysis, for the successful simulations the magnitude and location of the glenohumeral joint reaction force was calculated for each measured position and for each subject.

**Table 5.1:** Number of simulations of which constraint could not be met. There are two constraints. The stability constraint, the glenohumeral joint reaction force vector must be directed into the glenoid and the moment constraint, which means that the model must be able to generate a moment balance.

<table>
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<th>Task</th>
<th>Without RC</th>
<th>With 10% RC</th>
<th>With 20% RC</th>
<th>With 30% RC</th>
<th>With 40% RC</th>
<th>With 50% RC</th>
<th>Without SSce</th>
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CHAPTER 5
Functioning of the rotator cuff muscles during activities of daily living

5.3 RESULTS

5.3.1 Normal Situation

A total of 192 simulations were performed of which 190 were considered as successful (Table 5.1). Two lifting tasks could not be simulated due to measurement errors. The glenohumeral joint reaction forces for each activity of daily living are plotted in Figure 5.1. In this figure the glenoid is represented as an ellipse and the location and magnitude of the joint reaction forces are shown. The highest joint reaction forces (2100 N) are located in the middle of the glenoid and obviously occur during the highly loaded task of lifting a 4 kg bag. During eating with a utensil the highest joint reaction forces are located in the middle-inferior part, in the anterior-superior part during Perineal care, in the middle of the glenoid during combing hair and in the posterior-inferior part during reaching and washing the axilla. The highest joint reaction forces of these ADL are between 200 and 600 N.

5.3.2 Without Rotator Cuff

In Table 5.1 the results of the simulations are displayed. In 77% of the simulations, the model could not meet the constraints when movements were simulated without the rotator cuff. Of these 77% the model was not able to meet the moment constraints in 36% of the cases and stability constraints in 64% of the cases.
moment constraint could not be met in particular in the tasks that require higher humeral elevation angles and the stability constraint could not be met in the tasks that require retroflexion or adduction of the humerus (Figure 5.4).

5.3.3 Elevation tasks without rotator cuff force

In 50% of the abduction, 58% of the forward flexion, 50% of the reaching and 67% of the combing hair tasks the moment constraint could not be satisfied. A similar pattern for why the moment constraint was not met occurred for all humeral elevation tasks. The constraint is usually not met when the humerus is elevated approximately 90 degrees. The scapular part of the deltoid must balance the moment that is generated by the weight of the arm. The external moment is maximal at 90 degrees of humeral elevation, which means that the force of the deltoids at 90 degrees of humeral elevation is maximal as well. The infraspinatus assists the deltoids in a forward flexion of the humerus and the subscapularis assists the deltoid in an abduction task. The deltoid is able to compensate for the loss in muscle force of

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**Figure 5.2:** Absolute number of non-successful simulations for each fraction of muscle force. The non-successful simulations were divided for type constraint that was not: RC0 = Rotator cuff muscles could not produce force; RC10 = Rotator cuff could produce 10% of maximal force, etc; nosub = subscapularis could not produce force; nosup = supraspinatus could not produce force; noinf = infraspinatus could not produce force; notm = teres minor could not produce force.
these muscles, however the additionally generated force of the deltoid must be compensated for by another muscle, which is the coracobrachialis. This muscle is not able to produce a sufficient amount of force, because it is maximally contracted, to compensate for the additional deltoid force and therefore the model will not be able to meet the moment constraint.

**5.3.4 Retroflexion and adduction tasks without rotator cuff force**

In 67% of the retroflexion tasks, 42% of the perineal care tasks, 86% of the lifting tasks and 100% of the washing axilla tasks the model was not able to meet the requirements for glenohumeral stability. The result of glenohumeral instability is a displacement of the joint reaction force in the anterior-superior direction of the glenoid during perineal care and retroflexion. During eating with a spoon the joint reaction force displaces superiorly, during washing the axilla the joint reaction force vector displaces posteriorly and superiorly, and during lifting a 4kg bag the joint reaction force displaces anteriorly and inferiorly. Besides displacement, there is also an increase in total joint reaction force after removal of the rotator cuff for most ADL. The peak glenohumeral joint reaction forces increase from about 500N to 1500 N (Figure 5.3).

**Figure 5.3:** Location and magnitude of glenohumeral joint reaction force vectors in the glenoid plane of all subjects given for each measured position for each activity of daily living when all four rotator cuff muscles are dysfunctional.
5.3.5 ADL with submaximal rotator cuff force

The results of these simulations are displayed in Figure 5.2. It can be seen that an increase in rotator cuff force particularly increases the ability to satisfy the stability constraint. With 10% of the maximal possible rotator cuff force only 9% of all simulations cannot meet the stability constraint, which is a 41% increase compared to the 0% rotator cuff condition. With 50% rotator cuff force, 96% of all simulations could be performed successfully. The constraints were not met in only 4% of the simulations.

![Figure 5.2: Percentage of type constraint that was not met per task when the rotator cuff could not produce any force.](image)

5.3.6 The omission of the individual rotator cuff muscles

From Table 5.1 it can be seen that a loss of subscapularis, infraspinatus, supraspinatus or teres minor force is mechanically a relatively mild problem in the execution of activities of daily living in the model. The surrounding muscles, in particular the deltoids and the pectoralis major, are able to stabilise the glenohumeral joint and to generate sufficient force to create a moment balance. These muscles have large moment arms, which means that not much additional force is needed to generate a moment balance. As shown in Figure 5.3, the percentage of simulations that could
meet the stability and moment requirements varied between 4% (without supraspinatus) and 10% (without teres minor).

5.4 DISCUSSION

Pain, lack of strength and a lack of motion in the glenohumeral joint are potential causes for the moderate function after implantation of a shoulder prosthesis. However, all studies report pain relief post-operatively, which means that pain does not restrict the post-operative outcome.

Atrophied rotator cuff muscles or ruptures are common problems for rheumatoid arthritis patients (Wirth & Rockwood, 1994). A dysfunctional rotator cuff might be the cause of the limited functioning that is frequently seen in rheumatoid arthritis patients, which is the largest group of patients who receive an endoprosthesis. The current study investigated the mechanical role of the rotator cuff during the performance of a selection of ADL and ROM.

Due to a variation in motion patterns, the motions of all 24 healthy subjects were simulated. Using this variety in the performance of ADL is indirectly a way to accommodate for the variety in morphology. A small difference in humeral position results in a change in the length and moment arm of a muscle which can be decisive in the success of a simulation. This simulation procedure can thus be seen as Monte Carlo simulation. It has to be taken into account that the kinematics used in this study are upper extremity motions of healthy subjects. Patients will probably adjust their movements after shoulder arthroplasty, which means that the individual muscle contributions will change. In a future experiment patients after shoulder arthroplasty will be measured in order to identify differences in kinematics and dynamics.

From the results of the simulation procedure it appears that the function of the rotator cuff depends on the task that is performed, in other words it depends on the position of the humerus. It is also possible that during the first phase of motion the rotator cuff functions as a stabiliser, while in a later phase the rotator cuff muscles will function as prime movers. The function of the individual rotator cuff muscles can also change during a certain motion. This is in agreement with the results from Lee et al. (2000), who showed that the contribution of each muscle to stability is highly dependent on the position of the humerus. They held a cadaver humerus in 60° abduction and rotated the humerus from 0° to 90° external rotation. In the mid-range of motion the subscapularis and supraspinatus contributed most to stability, while in end range of motion all rotator cuff muscles except for the supraspinatus were important stabilisers.

The stabilising function of the rotator cuff was especially visible at higher retroflexion angles and at higher adduction angles, as in washing the axilla and eating with a spoon. Apparently not much force was required to stabilise the glenohumeral
joint, because with 20% of the maximal possible rotator cuff force the model was able to meet the stability constraints in 98% of the cases. It was also demonstrated that when the rotator cuff does not produce any force and the model was not able to stabilise the joint, the humerus will luxate in the superior direction. These results are also in agreement with clinical findings since proximal migration is an often reported complication in patients with rotator cuff disorders (Wirth & Rockwood, 1994).

The rotator cuff muscles were also required to support the deltoids to achieve higher humeral elevation angles. Tasks that required humeral elevation angles higher than 90° were often problematic with a dysfunctional rotator cuff. These results were in agreement with patient data found in the literature. Rheumatoid arthritis patients who have cuff tears were usually not able to elevate above 90° after implantation of a shoulder prosthesis (Hawkins et al. 1989, Kelly et al. 1987, Torchia et al. 1997). With 10% of the maximal possible rotator cuff force the number of unsuccessful simulations decreased from 77% to 26%. This means that in most cases loss of rotator cuff force could be compensated for by other muscles. The main compensating muscle is the scapular part of the deltoid.

Innervation and vascularisation aspects have not been modelled, but these effects are assumed to be small. Another point that has not been included in the model, is the dynamic stability of the glenohumeral joint. The analysis was a quasi-static analysis of ADL and ROM, which means that velocity and acceleration effects were not taken into account. A stable joint, in this case, means that the DSEM was able to point the joint reaction vector inside the glenoid. This does not mean that the DSEM was able to stabilise the joint after perturbations.

It could be concluded that the rotator cuff functions as a prime mover in tasks that required higher humeral elevation angles. The rotator cuff stabilises the glenohumeral joint in tasks that require retroflexion and adduction of the humerus. In particular, the rotator cuff muscles prevent the humerus from migrating proximally. The mechanical effect is thus dependent on the task that is being performed. To improve functioning after shoulder arthroplasty for patients with cuff tears, these two functions must be compensated for. Possible solutions are designing a prosthesis that will compensate for the rotator cuff force, like the Delta prosthesis (Grammont et al., 1993). Another solution might be to transfer a muscle, for example latissimus dorsi or teres major or developing a rehabilitation program that will focus on strengthening the shoulder.
CHAPTER 5
Functioning of the rotator cuff muscles during activities of daily living
EFFECTIVENESS OF TENDON TRANSFERS FOR MASSIVE ROTATOR CUFF TEARS
CHAPTER 6

6.1 INTRODUCTION

Repair of small or incomplete rotator cuff tears can be performed conventionally by suturing the tendons. Massive rotator cuff tears are not easily repaired (Handelberg, 2001, Postacchini & Gumina, 2001). To compensate for possible loss of rotator cuff function, other techniques like muscle transfers have been developed (Warner, 2001). To restore the external rotation function of the humerus, which is important for elevation of the humerus, Gerber (1988) transferred the latissimus dorsi to the superolateral humeral head, where the external rotators, infraspinatus and supraspinatus, are attached. For tears in the supraspinatus, three transfer options are reported: the latissimus dorsi transfer, the teres major transfer and the transfer of both muscles. The results of these transfers are satisfactory, since pain relief and improvements in range of motion (RoM) are accomplished (Aoki et al. 1996; Gerber, 1988; Gerber, 1992; Miniaci & MacLeod, 1999, Warner & Parsons, 2001). However, it is not known if there are better transfer options.

The purpose of this study is twofold: firstly, to find which muscle will be the most suitable muscle for transfer (latissimus dorsi, teres major or both) in the case of a dysfunctional supraspinatus and secondly, to find what the best possible attachment site is to obtain the best function. Since it is impossible to test the procedures on patients, the effects on functional outcome of the transfers will be quantified using a biomechanical musculoskeletal model of the upper extremity. This biomechanical shoulder model includes all joints, muscles and ligaments of the shoulder. The different tendon transfer procedures were simulated in the model and subsequently the ability to perform functional tasks was tested. However, the same task performed by two different subjects does not produce the same results. To account for this variation in the performance of activities of daily living (ADL) a Monte Carlo-like simulation procedure is used. Based on the number of successful simulations, it is determined what the best tendon transfer is in the case of a dysfunctional supraspinatus muscle.
6.2 METHODS

6.2.1 Input motions

Input to the model were motions of the thorax, humerus, clavicle, scapula and forearm of twenty-four healthy female subjects (mean 36.8 (SD 11.8) yr.) with no history of shoulder complaints during six ADL tasks (combing hair, perineal care, lifting a 4kg bag, washing the axilla, eating with a spoon and reaching above shoulder level) and three RoM tasks (forward flexion, abduction and retroflexion). The input motions were defined as orientations and positions of the thorax, orientations of the clavicle with respect to the thorax, orientations of the scapula with respect to the thorax, humerus orientations with respect to the thorax and the orientation of the forearm with respect to the humerus. Kinematics were recorded using a six degree-of-freedom electromagnetic tracking device, the Flock of Birds (Ascension Technology Inc., Burlington, Vermont, USA). Since the scapula moves underneath the skin, the scapula cannot be tracked dynamically. To obtain information about scapular motions a scapulalocator was used (Johnson et al., 1993). Each activity of daily living and each RoM task was divided into small steps and at each step the scapula orientation was measured using the scapulalocator. In other words, the tasks were performed in a quasi-static way. In a pilot experiment it appeared that quasi-static motions of thorax, humerus and forearm were not significantly different from dynamic motions.

6.2.2 The Delft Shoulder and Elbow Model

The Delft Shoulder (Van der Helm, 1994) and Elbow Model (DSEM) is a finite element musculoskeletal model consisting of 31 muscles divided into 139 muscle elements. The rotations of the thorax, clavicle, scapula, humerus and forearm are the input variables for the model. The model calculates the muscle forces required for each measured position to satisfy equilibrium and stability constraints. The constraint that the external moments must be balanced by the muscle forces is termed ‘the moment constraint’. The requirement for stability in the glenohumeral joint (i.e. the joint reaction force vector must be directed into the glenoid cavity) is termed here ‘the stability constraint’. The cost function used in this application of the model was the sum of squared muscle stresses, modified for muscle length-tension relations, i.e. maximum muscle force is a function of maximum muscle stress, physiological cross sectional area (PCSA) and muscle length. The musculo-skeletal parameters were obtained from a cadaver study (Klein Breteler et al., 1999). For general purposes, a maximum muscle stress of 100 has been used (Van der Helm, 1994). In pilot studies on the capacity of patients with rotator cuff tears it became clear, however, that these patients were able to apply approximately 40% of the
maximum force of the model during lifting. It was decided to downscale the maximum stress to 40.

6.2.3 Reduction of rotator cuff force

Except for the rotator cuff muscles, all structures, including the tendon of the long head of the biceps, were left intact. The effect of a massive tear of the supraspinatus obviously is a large reduction in muscle force. It is not known what the exact effect of a tear is on the muscle force. In this study it was assumed that due to a massive tear, the muscle is eventually not able to generate force. As a consequence, all simulations were performed without any supraspinatus force. Since it is likely that the other rotator cuff muscles are also affected in the case of a tear and because it is unknown what the reduction in force is, 100 different combinations of maximal possible force (5 levels of infraspinatus (10% - 50%) x 5 levels of teres minor (10% - 50%) x 4 levels of subscapularis (10% - 40%)), with respect to a muscle stress of 40, were used to find the relationship between muscle force decrease and functional outcome. Since simulation results indicated that the magnitude of the subscapularis force only marginally influenced the number of successful simulations, it was decided to drop this variable from the set of tendon transfer simulations to be performed.

6.2.4 Simulation of tendon transfers

To test which muscle was most suitable for transfer and which insertion was most suitable, the latissimus dorsi, teres major or both were transferred in the model. The location of the insertion of the transferred muscles was also varied. In clinical practice the tendon is usually transferred to the supraspinatus insertion in the case of a supraspinatus tear to close the defect and restore the anatomy. From a mechanical point of view, the tendon must be transferred to an attachment site where the muscle will have favorable moment arms, muscle length and glenohumeral stability (which are all implemented in the DSEM). Four locations were chosen: the teres minor insertion, the infraspinatus insertion, the supraspinatus insertion and the upper subscapularis insertion. Each muscle transfer was simulated with 25 different combinations of infraspinatus and teres minor force. The above combinations led to a total of 300 muscle parameter sets (3 transfers x 4 insertions x 25 reductions in force) that were used to simulate 8 tasks of 24 subjects (totalling 57600 simulations = 60 days on a 2GHz processor).

6.2.5 Data analysis

The simulations were classified as either successful or not successful. When the model was not able to find a solution, i.e. not able to meet its constraints, this was defined as non-successful. Non-successful simulations failed either because of the
inability to generate a moment balance, or the inability to produce a glenohumeral joint reaction force that was directed into the glenoid. To test which tendon transfer procedure is the most successful, a Friedman two-way ANOVA by ranks is used. A two factor (task and transfer option as a factor) multivariate analysis is performed to find differences between the transfers per task. Transfer option is the combination of the transferred muscle and location of insertion.

**Figure 6.1:** Percentage of successful simulations for all tasks expressed as a function of rotator cuff force level.

### 6.3 Results

**6.3.1 Supraspinatus Tear, Other Rotator Cuff Muscles Intact**

Without a functional supraspinatus, the model predicted that the other muscles were able to compensate for the loss in supraspinatus force. For abduction, in the healthy situation the supraspinatus aids the deltoids in the first phase of humeral elevation. In proportion to the deltoids, the supraspinatus generates about 5% (0.5 Nm) of the abduction moment at 90° of humeral elevation. This 0.5 Nm can easily be compensated for by the deltoids, which are strong enough to balance the external moment when supraspinatus is not able to produce force.
6.3.2 Supraspinatus tear, other rotator cuff muscles vary in maximal force

In Figure 6.1 the percentage of successful simulations for all 8 tasks are plotted for each combination of reduced rotator cuff force. It can be seen that the amount of infraspinatus force is an important factor, because when the maximal possible infraspinatus force decreases the number of successful simulations decreases. With 10% maximal possible infraspinatus and teres minor force, 34% of the simulations were successful. With 50% of maximal possible infraspinatus and teres minor force 72% of all functional tasks could be simulated successfully.

6.3.3 Latissimus dorsi transfers

Figure 6.2 shows the percentage of successful simulations after a latissimus dorsi transfer. Figure 6.3 shows the mean improvements per task for each muscle for all attachment sites. As shown in Table 1, there are significant differences in improvement which are dependent on the attachment site. The Friedman test showed that there were significant differences between all the transfers (df = 12, Chi Square = 58.8, P =.00). From the latissimus dorsi transfers, the transfer to the infraspinatus
CHAPTER 6
Effectiveness of tendon transfers for massive rotator cuff tears

Insertion has the highest ranking and a 5th place overall ranking. The number of successful simulations improved significantly for all tasks (Table 1), except for the perineal care task where the number decreased significantly (P = 0.00) and the number of successful simulations for abduction did not change (P = 1.0). The most often performed procedure is to transfer the latissimus dorsi to the supraspinatus insertion, which showed significantly lower improvements than a transfer to the infraspinatus on the washing the axilla task (P = 0.03), the reaching above shoulder level task (P = 0.00) and the lifting task (P = 0.00). However a transfer to the supraspinatus insertions showed a significant higher improvement in the number of successful simulations for the retroflexion task (P = 0.00). The transfers to teres minor and sub-

**FIGURE 6.3:** Mean percentage improvement in successful simulations for all rotator cuff levels after tendon transfer of latissimus dorsi, teres major or both to the supraspinatus, infraspinatus, subscapularis or teres minor insertion. Improvements are displayed for each task and in proportion to the rotator cuff tear situation.
scapularis show a significantly lower number of successful simulations for most tasks than the transfers to supraspinatus and infraspinatus insertion.

**Table 6.1:** For each transfer the P values are shown per task. P < .05 means a significant difference in the number of successful simulations between the situation before transfer and after transfer. A + sign means a significant improvement and a minus sign means a significant decrease in successful simulations with respect to the situation before the tear. Last two columns represent the mean rank for each transfer. Highest mean rank means that it is the mean most successful transfer.

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### 6.3.4 Teres Major transfers

Figure 6.4 shows the results of the tendon transfer of the teres major to all four insertions. In accordance with the latissimus dorsi tendon transfers, the improvement after transfer is dependent on attachment site. The transfer to the supraspinatus insertion was the best option for the teres major tendon transfers and a second overall best option according to the Friedman test (Table 1). Significant improvements in functional outcome on all tasks are observed (Table 1), except for the perineal care task (P = .67) and abduction (P = 1.0). In Figure 6.2 the improvements are dis-
played and it can be seen that in particular large improvements occur in the tasks that require high humeral elevation angles, like washing the axilla (130%), reaching (1300%) and forward flexion (90%). In contrast to the transfer options to other attachment sites, not a single task shows a decrease in the number of successful simulations.

6.3.5 Combined latissimus dorsi and teres major transfers

Like the other two transfer options, the effect of this procedure was that the model was able to simulate significantly more tasks successfully. Except for the perineal care task ($P = .67$) and the abduction ($P = 1.0$), all other tasks could be simulated more successfully after the transfer. The result of the Friedman test is that the transfer to the supraspinatus insertion showed the highest ranking overall (Table 1), however there is no significant difference in the number of successful simulations on all tasks with the transfer of the teres major to the supraspinatus insertion. In the cases when both muscles are transferred to either the supraspinatus or the infraspinatus
insertion the amount of possible rotator cuff force does not really affect the percentage of successful simulations as can be seen in Figure 6.5.

**Figure 6.5:** Percentage of successful simulations for all tasks after tendon transfer of the latissimus dorsi and teres major to the supraspinatus, infraspinatus, teres minor and subscapularis, expressed as a function of rotator cuff force level

### 6.4 Discussion

A massive tear in the supraspinatus tendon is accompanied by a decrease in muscle force and an imbalance in rotator cuff force. When such a tear causes shoulder pain, use of the arm will be restricted, which will result in atrophied rotator cuff muscles. The relationship between tears in the rotator cuff and rotator cuff force decrease is still not known. Therefore this study used combinations of restrictions in rotator cuff forces to assess the effect of force on functional outcome. According to the DSEM, with a dysfunctional supraspinatus and a 50% restriction in force of the remaining rotator cuff muscles, 72% of the tasks could be simulated. This implies that when patients with a rotator cuff tear are severely impaired in daily life, probably more than 50% of the rotator cuff force is restricted.

Since validation of the calculated forces is difficult because muscle force cannot be measured, some studies determined the effect of a supraspinatus tear (Itoi et al.,
1997) or supraspinatus block by means of anaesthetics (Howell et al., 1986) on abduction moment. Although these studies find a 15% to 20% higher contribution of supraspinatus to total abduction moment, the reported moment arm of 2.5 cm of the supraspinatus at 90° of abduction (Howell et al., 1986) is the same as found in the current study. In addition to this finding, Howell et al. report that supraspinatus, infraspinatus and deltoid muscles contribute to almost 100% of the abduction moment, which is also comparable to the current study. The reason why the contributions of supraspinatus to total moment differ from the current study can be found in the different method to calculate muscle force. Howell et al. calculate muscle force on the basis of EMG and PCSA and Itoi et al. calculated the loss in abduction moment compared to the contralateral side as a consequence of a tear. Another aspect that has to be taken into account are dynamical effects. Both studies used a dynamometer, which means that dynamical effects are incorporated as well. As Itoi et al. indicated, the contribution of supraspinatus is highly dependent on angular velocity. It is most likely that the contribution of supraspinatus will be higher when a dynamical analysis could be performed, because during a dynamical movement, the dynamical stability has to be guaranteed as well. This will probably lead to more rotator cuff activity. Unfortunately the scapula cannot be measured dynamically. It can thus be stated that measuring patients with the same method and evaluating the post-operative results is probably the best method to find out how trustworthy the model is.

A non-functioning rotator cuff can cause a less stable glenohumeral joint or superior migration of the humeral head which can result in impingement. If the tear is irreparable and cannot be treated by suturing the tendons, an alternative procedure is to transfer the tendon of the latissimus dorsi. Besides the latissimus dorsi (Gerber, 1988), the teres major (Celli et al. 1998) or both muscle tendons can be transferred.

The current study simulated these three surgical procedures in the DSEM to find the best transfer option, which uses anatomical data from one cadaver (Klein Breteler, 1999) and therefore interindividual differences in morphology cannot be accommodated. To account for interindividual differences to some extent, motion data of many subjects were input to the DSEM. In this manner the variation in the performance of a task is introduced into the model, making this simulation procedure a Monte Carlo-like procedure. The use of healthy motions as input to the model is necessary to investigate why the healthy motion cannot be achieved. When the morphology of the model is adapted to the pathological condition, the effect of a dysfunctional rotator cuff on healthy kinematics becomes visible. Post-operatively, the aim is to restore a healthy motion pattern of the upper extremity. By means of this simulation procedure the effect of the tendon transfer can easily be evaluated.
Effectiveness of tendon transfers for massive rotator cuff tears

If the motion can be simulated successfully with the adapted morphology the procedure can be considered effective.

Unfortunately there are some limitations of the DSEM. Innervation and vascularisation differences after transfer have not been taken into account, but these effects are assumed to be small. It is assumed that the transferred muscle will adapt to its new function, however it is not known to what extent the transferred muscle is able to do so.

Another point that has not been accounted for, is the dynamical stability of the glenohumeral joint. The analysis is a quasi-static analysis of ADL and RoM, which means that velocity and acceleration effects are not taken into account. A stable joint, in this case, means that the DSEM is able to point the joint reaction vector inside the glenoid. This does not mean that the DSEM is able to stabilise the joint after perturbations.

On the basis of the results of the current study, a muscle tendon transfer of both muscles or teres major to the supraspinatus insertion is mechanically speaking the most effective procedure to compensate for the loss in rotator cuff force. Except for the perineal care task and abduction, considerable improvements are seen in the ability to perform ADL and RoM. The DSEM was able to simulate significantly more activities of daily living and RoM tasks after these muscle tendon transfers. These two transfer options are the only transfers where no significant decrease in the number of successful simulations for the perineal care task is observed. In the cases of the more restricted rotator cuff forces, the gains in functional outcome are relatively higher.

The improvements seen in RoM and functional outcome were in agreement with experimental results of several authors. Since these authors only measured a forward flexion or external rotation task and tested if the latissimus dorsi was active a direct comparison is however not possible. All publications report pain relief and improvement in RoM. Aoki et al (1996) found a gain in post-operative forward flexion of 36º after latissimus dorsi tendon transfer, Gerber (1992) found an even higher gain of 52º in post-operative forward flexion and Warner & Parsons (2001) reported gains in forward flexion and external rotation of respectively, 60º and 38º. Celli et al (1998) reported gains in the Constant score and a gain of 35º external rotation in abduction after teres major tendon transfer.

In conclusion, according to the simulation procedure used in the current study, a tendon transfer of teres major and latissimus dorsi or teres major alone to the supraspinatus insertion appears to be the most effective procedure in the case of large irreparable rotator cuff tears and strongly atrophied dysfunctional rotator cuff muscles. Practical factors, like subacromial space, volume of the muscles and tendons, tensile properties and the ability to split the muscles, will finally determine which...
is the preferred transfer option. The predicted improvement in functional outcome will be evaluated in the future by measuring patients that have undergone this procedure.
BIOMECHANICAL ANALYSIS OF TENDON TRANSFERS FOR MASSIVE ROTATOR CUFF TEARS
CHAPTER 7

7.1 INTRODUCTION

Massive rotator cuff tears impair function in daily living. The common surgical treatment of suturing a tendon often results in a suboptimal functional recovery (Postacchini and Gumina, 2002). To improve post-operative function new surgical techniques have been developed such as a tendon transfer of the latissimus dorsi, (Gerber, 1988, Aoki et al. 1996, Miniaci & MacLeod, 1999, Warner et al. 2001) or a transfer using the teres major (Celli et al. 1988). It appears that a tendon transfer of latissimus dorsi or teres major to the site of the tear is a satisfactory procedure for reducing pain and improving postoperative range of motion. In a previous simulation study Magermans et al. (2003) concluded on the basis of an evaluation of their success when following movement trajectories that a teres major tendon transfer to the supraspinatus insertion was the most effective procedure for improving functional outcome in patients with a dysfunctional rotator cuff. That study did not evaluate why a teres major tendon transfer would be more effective than a latissimus dorsi transfer or a combined transfer or why a transfer to this attachment would be the best option.

The purpose of this study is to evaluate the mechanical effects of these tendon transfer procedures, based on their consequences for moment arms, muscle forces and muscle lengths. An adjusted biomechanical model of the upper extremity, with a transferred tendon of latissimus dorsi, teres major or both to one of the four attachment sites will be used to investigate the mechanical effects.

7.2 METHODS

7.2.1 Input motions

Input to the biomechanical model were three standard ROM profiles (forward flexion, abduction and retroflexion) that cover almost the entire range of movement of the shoulder. These profiles were based on the averaged movement trajectories of twenty-four healthy female subjects (36.8 ± 11.8 yr.).
7.2.2 The Delft Shoulder and Elbow Model (DSEM)

The Delft Shoulder (Van der Helm, 1994) and Elbow Model is a finite element musculoskeletal model consisting of 31 muscles divided into 139 muscle elements. Input variables to the model are the average rotations of the thorax, clavicle, scapula, humerus and forearm. The model calculates the muscle forces required for each measured position to satisfy moment and stability constraints. Two important constraints are included in the model. First, the constraint that the external moments must be balanced by the muscle forces (‘the moment constraint’). Second the model requires stability in the glenohumeral joint (i.e. the joint reaction force vector must be directed into the glenoid cavity) which is termed here ‘the stability constraint’. Muscle forces are distributed, based on the following cost function:

\[ J = \sum_{i=1}^{n} \left( \frac{F_i}{PCSA_i} \right)^2 \]

where \( n \) is the number of muscle elements, \( F_i \) is the force and \( PCSA_i \) the physiological cross-sectional area. Maximum muscle forces were limited on the basis of the inclusion of muscle length-tension relations, where the maximum muscle force is a function of maximum muscle stress, \( \sigma_{max} \), \( PCSA_i \) and muscle fibre length, \( l_{fi} \) (Klein Breteler et al., 1999):

\[ 0 \leq F_i \leq f(l_{fi}) \cdot PCSA_i \cdot \sigma_{max} \]

Van der Helm (1994) proposed a \( \sigma_{max} \) of 100 \( N \cdot cm^{-2} \) for general use. In pilot studies on the capacity of patients with rotator cuff tears it became clear, however, that these patients were only able to apply approximately 40% of the maximum force of the model during lifting. It was therefore decided to downscale the \( \sigma_{max} \) to 40 \( N \cdot cm^{-2} \).

7.2.3 Simulation of supraspinatus tear

The obvious effect of a massive tear of the supraspinatus is a large reduction in force. In this study it was assumed that due to a massive tear, the supraspinatus was not able to generate force at all and all simulations were performed without supraspinatus force. Since it is likely that the other rotator cuff muscles are also affected in the case of a tear, a previous study (Magermans et al. 2004) simulated combinations of force reductions in infraspinatus, teres minor and subscapularis. This resulted in the finding that after a simulated tendon transfer the largest improvements were observed when the rotator cuff muscles were restricted to 10% of their maximal possible force. Therefore, in this study, all simulations were performed using a restriction of 90% in rotator cuff force.
7.2.4 Simulation of tendon transfers

To test which muscle was the most suitable for transfer and what insertion was the most suitable, the latissimus dorsi, teres major or both were transferred in the model. The location of the insertion of the transferred muscles was also varied. Four locations were chosen: the insertions of the teres minor, the infraspinatus, the supraspinatus and the proximal insertion of the subscapularis.

7.2.5 Data analysis

When the average motion profiles were simulated with the modified DSEM representing a rotator cuff tear, the model indicated that all three tasks could not fully be performed. During the forward flexion task, the model could not produce a sufficient moment at a humeral elevation angle of 77° and an external rotation of 41° (Figure 1a). For abduction the moment constraint could not be met at 126° of elevation and an externally rotated humerus of 48° (Figure 1b).

The DSEM could not perform a retroflexion task, because The DSEM could not meet the requirements for stability. The glenohumeral joint reaction force vector was directed outside the glenoid at a humeral elevation angle of 58° and an internally rotated humerus of 79° (Figure 1c).

The above instants in average motion profiles were subsequently chosen for further evaluation, focusing on the mechanical effect of transfer procedures. Due to the evaluated transfers, the moment arms and the lengths of the transferred muscles change, which will result in a change in the ability to produce force and joint moments. The moment arms are defined in two ways due to interpretation reasons. The moment arms are described with respect to the thorax and with respect to the local humeral coordinate system. The moment arm around the thoracic x-axis is the forward flexion/retroflexion axis. When a muscle has a positive moment arm around this axis, it means that the muscle can produce an forward flexion moment. The moment around the y-axis is around the local humeral coordinate system, where the y-axis is defined as the long axis of the humerus. A positive moment arm means that a muscle can produce an internal rotation moment. The z-axis is the thoracic z-axis and is defined as the ad/abduction axis of the humerus, where a positive moment arm means that a muscle can produce an abduction moment.

7.3 Results

7.3.1 Effect of supraspinatus tear and reduced rotator cuff force

As a consequence of the reduced maximal force conditions and the dysfunctional supraspinatus, used in this simulation, the three ROM tasks could not be performed by the DSEM due to the inability to meet the constraints. As mentioned above, dur-
ing forward flexion the moment constraint could not be met at a humeral elevation angle of 77° and an external rotation of 41°. At this position the supraspinatus has a moment arm around the forward flexion axis (x-axis) of 1.3 cm and a moment arm of 0.8 cm around the y-axis (internal rotation) and the infraspinatus has a moment
arm of 0.5 cm around the x-axis and a moment arm of -1 cm around the y-axis (external rotation). The moment arms of these muscles make that in the normal situation, without any restrictions, the infraspinatus aids the deltoid muscles in elevating and rotating the humerus. Both muscles produce an forward flexion (x-axis) moment (2 Nm for infraspinatus and 5.5 Nm for deltoids) and an external rotation (y-axis) moment (infraspinatus produces 2.2 Nm and the deltoids produce 2 Nm). When supraspinatus is not able to produce any force and infraspinatus is restricted to 10% of its maximal force, the infraspinatus can only produce a moment of 0.6 Nm around the x-axis and y-axis (Figure 2 & 3). The deltoids are able to compensate this moment, but the additionally produced moment of 1.1 Nm around the y-axis is unwanted and must be counterbalanced by the coracobrachialis (Figure 3), which reaches its maximum force (Figure 4). As a consequence and in this case, the model could not meet the constraint during forward flexion.

The inability to meet the moment constraint during abduction occurred at a humeral elevation of 126° and an externally rotated humerus of 48°. The moment constraint could not be met due to insufficient force in the other rotator cuff muscles, in particular the subscapularis and infraspinatus.

**FIGURE 7.2:** A moment balance of an forward flexion of the glenohumeral joint around the retro/forward flexion axis. The constraint could not be met at a humeral elevation angle of 77°. At each measured position the humerus is in equilibrium and therefore the sum of all moments around the glenohumeral joint must equal zero.
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Biomechanical analysis of tendon transfers for massive rotator cuff tears

**Figure 7.3:** Moment balance of an forward flexion of the glenohumeral joint around the vertical axis

**Figure 7.4:** Relative muscle force of coracobrachialis during forward flexion
7.3.2 Function of latissimus dorsi during forward flexion after transfer

The effect of the transfer of latissimus dorsi to all four insertions is that the muscle becomes an anteflexor of the humerus, while in the situation before transfer the muscle was a retroflexor. The moment arm around the x-axis (forward flexion/retroflexion axis) changes from -3.3 to 1.8 cm when transferred to the supraspinatus insertion. In the case of a transfer to the teres minor or subscapularis insertion, the latissimus dorsi can hardly produce an forward flexion moment because the moment arms are small, i.e. 0.2 cm in the case of a transfer to the teres minor and 0.5 cm when transferred to the subscapularis (Figure 5). The moment arm around the y-axis of the humerus (long axis of the humerus) changes from 1.2 cm, which means that the latissimus dorsi is an internal rotator, to -0.4 cm in the case of a transfer to the supraspinatus and to -1.6 cm in the case of a transfer to the infraspinatus, which means that the latissimus dorsi can produce an external rotation moment. The transfers to the teres minor and subscapularis do not change the internal rotation function of the latissimus dorsi.

![Figure 7.5: Moment arms of the latissimus dorsi (row 1) and teres major (row 2) after transfer to the four rotator cuff insertions during forward flexion. The first column represents the moment arm about the global x-axis, which is the forward flexion/retroflexion axis. The second column represents the moment arm about the local y-axis, which is the internal/external rotation axis of the humerus. The third column represents the moment arm about the global z-axis, which is the abduction/adduction axis.](image-url)
However the moment constraint could not be met for the forward flexion tasks when the latissimus dorsi was transferred to the supraspinatus, subscapularis and teres minor. The only successful forward flexion simulation was when the latissimus dorsi tendon was transferred to the infraspinatus. The transferred muscle must compensate for the loss in rotator cuff force, in particular the infraspinatus which aids the deltoids in elevation and external rotation of the humerus. As mentioned above, after transfer, the latissimus dorsi was able to produce an elevation and external rotation moment, but the external rotation moment arm of the latissimus dorsi is only -0.4 cm in the case of a transfer to the supraspinatus. Although an external rotation moment arm is required, a moment arm of -0.4 cm is not sufficient, because the latissimus dorsi must produce an amount of force that is beyond its maximal force. A larger moment arm is needed to decrease the amount of force that is needed to produce the required moment. This is obtained by a transfer to the infraspinatus insertion (moment arm of -1.6 cm). A tendon transfer to this insertion makes it possible for the latissimus dorsi to produce the required external rotation moment.

7.3.3 Function of latissimus dorsi during abduction after transfer

As seen in Figure 6, at 126° of abduction the function of the latissimus dorsi is abduction in the normal situation (moment arm of 0.3 cm around the z-axis). Except when transferred to the infraspinatus (moment arm of -0.4), the function of latissimus dorsi does not change. The moment arms around the z-axis vary between 0.9 cm (supraspinatus insertion) and 2.5 cm (teres minor insertion). The latissimus dorsi also becomes an external rotator when transferred to the supraspinatus insertion (moment arm of -0.1 cm) or to the infraspinatus insertion (moment arm of -0.8 cm).

Similar to forward flexion, the only successful simulation in abduction is when the muscle is transferred to the infraspinatus. To compensate for the loss in rotator cuff force, the transferred muscle must produce a sufficient amount of external rotation moment, which is normally produced by the infraspinatus. The deltoids are easily able to produce the desired external rotation moment when the infraspinatus force is restricted. The additional force produced by the deltoids is counterbalanced by the coracobrachialis, but the coracobrachialis reaches its maximal force. In the case of a transfer to the infraspinatus insertion, the latissimus dorsi is able to aid the deltoids to compensate for the loss in infraspinatus force and therefore the coracobrachialis does not have to counterbalance for the additional produced deltoid force. The moment arm around the y-axis of the latissimus dorsi is only -0.1 cm when transferred to the supraspinatus insertion and therefore the latissimus dorsi would have to generate more force than possible to produce the required external rotation moment.
Function of latissimus dorsi during retroflexion after transfer

At 58° of humeral retroflexion, which is the position where the DSEM could not meet the stability requirement, the moment arm around the x-axis (forward flexion/retroflexion axis) varies between -0.5 cm in the case of a transfer to the teres minor insertion and -2.0 cm in the case of a transfer to the supraspinatus insertion. The latissimus remains an internal rotator during retroflexion, the moment arms changing only a few mm, to between 1.0 cm (transfer to infraspinatus) and 1.8 cm (transfer to supraspinatus). The moment arm was 1.3 cm in the normal situation. The main function of the transferred muscle is to compensate for the loss in subscapularis force, because in the normal situation the subscapularis functions as an internal rotator, retroflexor and stabiliser of the glenohumeral joint. Due to the restriction in subscapularis force, the muscle cannot function properly and therefore the stability requirement cannot be met. After tendon transfer of the latissimus dorsi, except for the transfer to the teres minor insertion, all constraints could be met.
7.3.5 Function of teres major during forward flexion after transfer

After transfer the moment arm of the teres major changes from a retroflexor (moment arm of -4.8 cm) to an anteflexor (Figure 5). The moment arm of the teres major is 1.7 cm when attached to the infraspinatus, 0.7 cm when attached to the teres minor, 2.0 cm when attached to the supraspinatus and 1.2 cm when attached to the subscapularis. An external rotation function is obtained when transferred to supraspinatus and in particular when transferred to the infraspinatus. The moment arm around the y-axis changes from 0.6 cm (normal) to -1.9 cm in the case of a transfer to the infraspinatus insertion (Figure 5). The moment arm after transfer to the subscapularis and teres minor insertions remains positive; in other words, the muscle remains an internal rotator. Due to the difference in moment arm, the teres major is required to generate 10 N less force when transferred to the infraspinatus insertion than when transferred to the supraspinatus insertion.

The DSEM was able to successfully simulate the forward flexion task after a tendon transfer to the infraspinatus and supraspinatus insertion. The reason why a transfer of the teres major tendon to the supraspinatus is possible in contrast to a lat-
issimus dorsi transfer is that the latissimus dorsi reaches its maximal force, while teres major reaches 50% of its maximal force. This difference is a consequence of the difference in PCSA (5.62 cm² for the latissimus dorsi versus 6.08 cm² for the teres major), relative muscle length (80.5% versus 86.4%) and moment arm (1.8 cm versus 2.0 cm). The DSEM could not meet the moment constraints in the cases of a transfer to the subscapularis or teres minor due to the same reason as described in the case of a latissimus dorsi tendon transfer.

7.3.6 Function of teres major during abduction after transfer

The function of teres major changes from adductor to abductor in all cases except for a transfer to the infraspinatus insertion, where the moment arm is -0.6 cm. With respect to the transfer to the other insertions, the moment arm around the z-axis varies between 0.6 cm for a transfer to the supraspinatus and 2.2 cm for a transfer to the subscapularis insertion (Figure 6). The remaining moment arms of the teres major are only a few mm different than during an forward flexion. The amount of force that is generated by the teres major when the tendon is transferred to the infraspinatus is 58 N, which is 13 N less than when transferred to the supraspinatus.

As in the forward flexion task, the moment constraint was not met when the tendon was transferred to the subscapularis or teres minor insertion.

7.3.7 Function of teres major during retroflexion after transfer

As can be seen in Figure 7, during retroflexion, the moment arm of the teres major around the x-axis changes from positive (0.6 cm) to negative in the case of a transfer to the supraspinatus (-1.7 cm), subscapularis (-2.1 cm) and teres minor insertion (-1.0 cm), which means that the teres major can produce a retroflexion moment. In the case of a transfer to the infraspinatus the moment arm becomes 0.4 cm smaller, which means that the teres major remains an anteflexor. For all transfer options the positive moment arm around the y-axis is maintained, making the teres major an internal rotator during retroflexion. Except for a transfer to the infraspinatus insertion, where the moment arm about the y-axis decreases to 0.2 cm, the moment arm increases by at least 0.9 cm, which means that an internal rotation moment would require less force.

The stability constraint that could not be met in the normal situation could be met after transfer in all transfer options. This is mainly due to the compensation of the subscapularis muscle.

7.3.8 Effect of tendon transfer of both muscles

Moment arms and lengths of both muscles are equal to the individual tendon transfers. The results are similar to those described in the transfer of the teres major muscle. Regarding the muscle forces, there are also no differences. The teres major
is the most active and shows the same activation pattern in the combined transfer as in the teres major transfer. The activity of the latissimus dorsi that is seen in the case of a combined transfer is negligible. The latissimus dorsi is less favorable than the teres major, because, as described above, the PCSA, relative length and moment arms of the latissimus dorsi are smaller, which means that to produce the desired moment, more force is needed than when the DSEM activates the teres major.

7.4 DISCUSSION

Loss of rotator cuff force will be the result of massive rotator cuff tears. The loss of force is not only due to a deterioration of the muscle but also due to the associated pain. The effect of pain on mobility is likely to lead to a worsening of the condition of the rotator cuff due to associated atrophy. A tendon transfer appears to lead to a reduction in pain (Gerber, 1988) and will thus help to break this vicious cycle.

In addition, a tendon transfer of either latissimus dorsi or teres major or a combination of these two tendons appears to be a very effective procedure in mechanical terms. In particular, the tendon transfer of teres major to the supraspinatus insertion appears to be the most effective (Magermans et al. 2003).

The two main functions of the rotator cuff muscles, stabilising the glenohumeral joint and aiding the prime movers of the humerus can be completely fulfilled by the teres major. After transfer the change of moment arms of the teres major turns the muscle into an external rotator and elevator of the humerus during forward flexion and abduction. In particular the external rotation function is very important. During forward flexion and abduction approximately 50º of humeral external rotation was used. The main cause for not meeting the constraints is that in the healthy situation there are only a few muscles (infraspinatus, teres minor and a small part of the supraspinatus) that are able to produce an external rotation moment around the glenohumeral joint. One of the main goals of the tendon transfer procedure is not to close the defect, but to create a muscle that is able to produce an external rotation moment. It has to be taken into account that muscles usually have moments arms around more than one axis. It might be possible that due to a small moment arm around the external rotation axis, the transferred muscle will generate a high force to produce the desired moment. This might result in an unwanted moment around another axis. A tendon transfer of the latissimus dorsi to the supraspinatus insertion was therefore not a satisfactory option because the moment arm around the y-axis was too small for the muscle to produce the required external rotation moment. This also shows that the attachment site is an important aspect that has to be taken into account by the surgeon. If the attachment of the tendon is nearer the subscapularis insertion the moment arm of the transferred muscle is not an external rotation moment arm but an internal rotation moment arm. Hence, high humeral elevation will
be difficult to perform. When the muscle is transferred to the teres minor insertion, the muscle remains an adductor and internal rotator. Therefore a transfer of the tendon to the superior posterior part of the humerus appears to be the best solution. According to this study it would be preferable to transfer the tendon to the supraspinatus insertion because after this transfer, the retroflexion function is also restored.

With respect to the muscle used, a transfer of the teres major is the best option because it is a stronger muscle than the latissimus dorsi which means that less relative muscle force is needed to stabilise and move the humerus. It is also a better procedure than a transfer of both muscles because if both muscles are transferred, a strong adductor is lost. In this study no adduction tasks are simulated. In activities of daily living not many tasks involve active adduction as this movement is achieved passively by gravitational force. However, there are some tasks such as rising from a chair that require active adduction. When armrests are used to rise from a chair, the moment around the hip decreases 50%, which means that a considerable amount of adduction is being generated during this task (Janssen et al. 2002). These tasks might become a problem when both muscles are transferred and will then lead to a considerable limitation in mobility. Additionally, it appears that after transfer, the latissimus dorsi does not contribute much to overall function because teres major is the most active muscle due to its more favorable characteristics. With respect to practical issues, a transfer of both muscles might cause subacromial impingement because the two muscles are too large to fit in the subacromial space.

The DSEM has been validated by checking the muscle activities predicted by the DSEM with measured EMG for healthy subjects without any shoulder complaints (Van der Helm, 1994). It might be possible that patients with a dysfunctional rotator cuff show different muscle activation patterns. In other words, it is likely that these patients will use compensatory motions which means that the muscles will adapt. In the near future, the upper extremity motions of patients will be used pre- and post-operatively and these motions will be measured to evaluate the predicted outcome. It is expected that the post-operative functional outcome is comparable to the predicted outcome by the DSEM. However, it is likely that the predicted pre-operative ROM will not be equal because patients will use compensatory motions, which will result in a higher pre-operative ROM. To account for this difference, these compensatory motions will be input to the DSEM as well. It is expected that the DSEM will be able to simulate these compensatory motions. The results of the simulated pre-operative motions gives insight into the underlying mechanisms which limit pre-operative functioning. Besides motions, electromyography will also be measured to ascertain whether the transferred muscle is indeed active. One drawback remains and that is the knowledge about the effect of atrophy or tear size
on rotator cuff force. In this study the results of a worst case scenario (a dysfunctional supraspinatus and only 10% rotator cuff force) were presented.

In conclusion, this biomechanical analysis indicated that a tendon transfer of teres major to the supraspinatus insertion is effective on the basis of the mechanical parameters: moment arms, muscle length and force. A tendon transfer of teres major to the supraspinatus insertion is mechanically the best procedure for the correction of the effect of loss of rotator cuff force. The teres major becomes an abductor, forward flexor and external rotator during forward flexion and abduction and the function of the teres major during retroflexion is retroflexion, internal rotation and adduction. This new function of the teres major approximated the original function of the lost structures and is therefore likely to lead to the best restoration of upper extremity functioning in daily life.
COMPARISON OF MODEL PREDICTIONS TO THE FUNCTIONAL OUTCOME OF A PATIENT
CHAPTER 8

8.1 INTRODUCTION

A tendon transfer of either latissimus dorsi or teres major for the treatment of a dysfunctional rotator cuff or massive tears in the supraspinatus muscle usually result in a satisfactory functional outcome (Gerber, 1992, Celli et al. 1998). Although it is not exactly known which tendon transfer procedure will lead to the best functional outcome, pain relief is often accomplished and range of motion improves. Recently, using a biomechanical model of the shoulder, the Delft shoulder and elbow model (DSEM), the mechanical effectiveness of tendon transfers was evaluated (Magermans et al., 2004). In a previous simulation study of the mechanical effect of twelve different tendon transfer procedures was simulated. To determine the effect on the ability to perform functional tasks, in the model the latissimus dorsi, teres major and both muscles were transferred to the four insertions of the rotator cuff muscles. Transfers to the supraspinatus and infraspinatus insertion resulted in a 25% better outcome than a transfer to the teres minor or subscapularis insertion. Concerning the transferred tendon, it appeared that a transfer of either teres major or a combination of teres major and latissimus dorsi resulted in the highest functional outcome. 98% of the tasks could be simulated after a transfer. It was concluded that the most effective tendon transfer procedure would be a transfer of the teres major muscle to the insertion of the supraspinatus. To be more precise, the model predicted that a healthy motion pattern could be achieved after transfer.

If the model is to be used for prediction of tendon transfers, a comparison between the model predictions and post-operative gain is required. For an optimal comparison, patient specific information is required. Information about the patients morphology, muscle status and muscle strength might affect the outcome of the predictions. Obtaining all this detailed information is however practically impossible for living subjects.

Another aspect that will affect the outcome is pain. The experience of pain might restrict motion while the muscles are strong enough to move the arm. Since modeling pain adequately is very complex, the comparison of the model predictions and the patient measures might differ due to the experience of pain.
The objective of the current study was to compare the predictions of the model to the functional result of a patient with a massive tear in the supraspinatus muscle that has undergone a teres major tendon transfer. By using a more or less generic model for individual purposes, the mechanical integrity of the model is also investigated.

8.2 METHODS

8.2.1 Subject

One male patient (age 56 years) with a massive supraspinatus tear was included in the study. The tear in the supraspinatus was a result of trauma that occurred a year before.

8.2.2 Measurement device

A six degree-of-freedom electromagnetic tracking device, the Flock of Birds (Ascension Technology Inc., Burlington, Vermont, USA), was used for the recording of kinematic data. Following calibration of the measurement space, the mean residual error was 2.3 mm for all three directions. Five sensors were used to measure position and orientation of the upper extremity. The sensors were attached to a pointer, the sternum, humerus, forearm and a scapulalocator (Johnson et al. 1993). The latter device recorded the scapula positions by means of palpation of three bony landmarks on the scapula: the trigonum spinae, inferior angle and acromial angle. The pointer was used to measure bony landmarks of the upper extremity, which are used to construct the local coordinate systems. From the local coordinate systems and sensor motions the bone and joint rotations were calculated (Meskers et al. 1998).

8.2.3 Procedure

Pre- and 4 months post-operatively, five Range Of Motion (ROM) tasks were measured: forward flexion, retroflexion, abduction, internal and external rotation with and without humeral elevation. The following Activities of Daily Living (ADL) were measured: perineal care, combing hair, eating with a spoon, washing axilla, reaching above shoulder level, and lifting a 4 kg bag. The current study only used three ROM tasks (forward flexion, retroflexion, abduction) for comparison.

Since dynamic tracking of the scapula is very difficult, measurements were performed in a quasi-static mode. All tasks were divided into small steps for which the positions and orientations of the sensors were recorded at each step. In a pilot study the motions of the humerus during quasi-static measurements and dynamic measurements were compared and it appeared that the motions were not significantly different from each other.
The patient that was included in the current study was a strong man with a su-
prasinatus tear caused by trauma, which did not necessitate adjustment of the com-
monly used maximal muscle stress for the DSEM of 100 N cm\(^{-2}\). As was
previously necessary (Magermans et al. 2004), the maximal force of the supraspin-
atus force was set to zero. Healthy motion patterns of the three ROM tasks were
simulated with the DSEM. It was expected that the lack of supraspinatus force
would not affect performance because the deltoids will probably compensate for
this loss in force. Without the ability to produce force with the supraspinatus mus-
cle, local muscle stresses and the direction of the glenohumeral joint reaction force
vector will probably change.

With respect to the post-operative simulation, the teres major muscle was trans-
ferred to the prescribed insertion of the supraspinatus, which was the same proce-
dure the patient had undergone. The same ROM tasks were then used as input to the
model with the transferred teres major muscle. It was expected that the performance
will not be affected, but that local muscle stresses will decrease and that the position
of the glenohumeral joint reaction force vector will become more favorable.

These predictions were compared to the pre- and post-operative results of the
measured patient. Due to the fact that pain affects the ability to move 5 ml (1%)
Lidocain was injected into the subacromial bursa pre-operatively.

8.2.4 Angle definitions

Joint angles were expressed as Euler angles in other words: rotation about an axis.
The axes are based on the local coordinate system of the bone. A local coordinate
system is constructed by means of the coordinates of the measured bony landmarks.
The rotation order that is used in the study was based on the International Society
of Biomechanics standardisation proposal of the International Shoulder Group.

Glenohumeral joint angles were defined as motions of the humerus with respect
to the scapula, following the globe system as described by Doorenbosch et al.
(2003). The reference position is the arm in the vertical hanging position, with the
longitudinal (y-axis) along the vertical axis and the x-axis (in the plane of GH, EM
and EL) along the medial -lateral axis. The first rotation is defined as a rotation
about the vertical axis, which was defined as plane of elevation. This rotation can
best be visualised by looking from a top view to the different vertical planes around
the shoulder, where 0º is when the humerus points laterally (abduction) and a posi-
tive elevation plane is when the humerus points ventrally. The second rotation is
about the rotated z-axis and is defined as the glenohumeral elevation angle, which
can be interpreted as the angle between the long axis of the humerus and the scapu-
lar spine. The third rotation is about the rotated y-axis of the humerus and is defined
as axial rotation of the humerus. In other words, internal (positive rotation) and external (negative) rotation of the humerus.

![Graphs of Scap y, Scap z, Scap x, ghy, ghz, ghya](image)

**Figure 8.1:** Motion patterns of the scapula and GH-joint during a forward flexion task. Model = the mean motion pattern of 24 healthy subjects, pre = motion pattern after lidocain injection before tendon transfer, post = motion pattern after tendon transfer.

### 8.3 Results

#### 8.3.1 Simulation of the pre-operative status

With a maximal muscle stress of $100 \, N \cdot cm^{-2}$ and without the ability to produce force with the supraspinatus muscle, healthy motion patterns could be simulated. A thoracohumeral elevation angle of $148^\circ$ was achieved during forward flexion, $149^\circ$ during abduction and $65^\circ$ during retroflexion (Figure 8.3). Before the lidocain injection the pain experience of the patient restricted the motions to such an extent that there was hardly any motion visible. After lidocain injection the patient was able to achieve near normal ROM (Figure 8.1). During forward flexion a thoracohumeral elevation angle of $129^\circ$ was recorded, during abduction $142^\circ$ was recorded and dur-
ing retroflexion 56° was recorded. From the pre-operative motion patterns it can be seen that the patient demonstrated more scapulothoracic motion. Roughly 20° more protraction (Scap y) and 15° more laterorotation (Scap z) is demonstrated by the patient. The measured glenohumeral elevation angle (ghz) of the patient is approximately 35° lower than the mean healthy glenohumeral elevation angle.

As a result of healthy motion patterns combined with a torn supraspinatus muscle, the pre-operative glenohumeral joint reaction force vector during forward flexion, was directed in the superior part of the glenoid during the first 60° of humeral elevation and turned more inferior-posterior for the subsequent 90° of elevation (Figure 8.4). To calculate the trajectory of the glenohumeral joint reaction force vector as a result of the motions of the patient, the motions of the patient also had to be simulated using the model. The reaction force vector as a result of the patient’s motion also demonstrated a superior to posterior-inferior trajectory.
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8.3.2 Post-Operative

The healthy motion patterns of all ROM tasks could also be simulated after teres major tendon transfer. Compared to the pre-operative model simulations, there was no change in performance because a maximal performance was achieved pre-operatively.

Post-operative measurements of the patient resulted in an improvement in both thoracohumeral and glenohumeral ROM. As seen in Figure 8.3, the thoracohumeral elevation angle increased to 142° during forward flexion, to 151° during abduction and to 64° during retroflexion. The glenohumeral elevation angle increased as well and was 89° during forward flexion, 97° during abduction and 55° during retroflexion. These measured post-operative ROM values were almost similar to the predicted ROM. The motion patterns of the patient after transfer were also more similar to the simulated healthy motion patterns during forward flexion (Figure 8.1) and abduction (Figure 8.2). In particular the scapula motions (Scap y and Scap z) and the glenohumeral elevation (ghz) were comparable to the healthy motions.

With respect to the glenohumeral joint reaction force after the tendon transfer, the vector was located more inferiorly than pre-operatively using healthy motions as input. When the motions of the patient were used as input combined with a teres major tendon transfer, the trajectory of the reaction force vector was also located more inferiorly (Figure 8.5). Additionally, it appeared that the sum of the local muscle stresses was lower after tendon transfer in both cases.

**Figure 8.3:** Comparison of the pre-operative thoracohumeral elevation angle, predicted by the model and measured in the patient.
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FIGURE 8.4: Location of glenohumeral joint reaction force vector in the glenoid based on healthy motion patterns. Model 1 = pre-transfer and model 2 = post-transfer.

FIGURE 8.5: Location of glenohumeral joint reaction force vector in the glenoid based on the motion patterns of patients. Model 1 = pre-transfer and model 2 = post-transfer.
8.4 DISCUSSION

The objective of the current study was to compare the model predictions pre- and post- teres major tendon transfer to the pre- and post-operative status of a patient that has undergone the same procedure.

It appeared that the predictions of the model were almost similar to the functional results of the measured patient. Pre-operatively, the model predicted that with a torn supraspinatus muscle there would be almost no change in performance. These results were in agreement with the pre-operative measurements of the patient after a subacromial lidocain injection. The scapular motion pattern after injection was however different from the healthy motion pattern, despite the almost similar thoracohumeral ROM. Since glenohumeral ROM pre-operatively was almost 35° lower than normal, the observed additional scapulothoracal motion is probably required for the scapula to lift the humerus to the same thoracohumeral angle. It can be concluded that the patient tries to compensate for the lack of glenohumeral motion by using more scapulothoracal motion.

The experienced pain before the injection was most-likely caused by a proximal migrated humeral head that exerted high forces on the bursa. Without a functioning supraspinatus muscle, the model calculated that the glenohumeral joint reaction force vector was to be located more superior. After simulating the motions of the patient, it was demonstrated that a similar reaction force pattern was visible. Reasons for the more superior located pattern of the reaction force in the glenoid might be a dysfunctional coordination pattern caused by the torn supraspinatus muscle. Another explanation might be that due to the pain and associated immobility, the other rotator cuff muscles and in particular the infraspinatus muscle, will atrophy to such an extent that humeral depression will become very difficult.

Post-operatively, the maximal thoracohumeral difference with the ROM predicted by the model and observed in the patient was only 14º. After the teres major tendon transfer to the insertion of the supraspinatus, the patient did not experience any pain during the measurements and the patient was able to achieve near normal ROM. Additionally, the glenohumeral joint reaction force vector was located more inferior after tendon transfer which is in agreement with the results of the measured patient. It can thus be stated that the teres major will function as a humeral head depressor as well.

From the results of this patient, it appears that pain relief accomplishes an improvement in function to such an extent, that perhaps removal of the bursa would have been sufficient. However, without a tendon transfer the glenohumeral reaction force will most likely remain in the upper part of the glenoid which likely will result in returning pain. It can thus be concluded that the teres major tendon transfer is a successful treatment option for a massive supraspinatus tear, because besides pre-
venting proximal migration of the humerus, the transfer accomplished pain relief and improvements in ROM.

In conclusion, the DSEM was capable to predict the range of motion pre- and post teres major tendon transfer. Additionally, the model was able to identify the probable mechanical causes for pain. These results are the first steps towards a patient specific use of the model. To draw conclusions about the use of the model as a pre-operative planning tool, more patients that have undergone this procedure have to be measured.
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COMPARISON OF UPPER EXTREMITY MOTIONS USING OPTICAL AND AMBULANT SYSTEMS
CHAPTER 9

9.1 INTRODUCTION

A basic requirement for the development of a shoulder endoprosthesis is the load on the prosthesis during activities of daily living (ADL). The prosthesis must be able to withstand long lasting loads as well as peak loads. Estimating a load spectrum of the upper extremity for a longer period of time and during daily life is a very complex procedure. In the laboratory setup, net moments and muscle forces around the glenohumeral joint can be estimated using an inverse dynamic model of the upper extremity which uses the motions of body segments and the external load as input (Van der Helm et al., 1994). Van der Helm et al. (1996b) estimated joint load during wheelchair activity in the laboratory setup using a stereophotogrammetric system. The 3D positions of upper extremity bony landmarks were used to construct the local coordinate systems of the body segments. An overall glenohumeral load spectrum during daily living has never been determined. A measurement system that is able to provide accurately estimated joint load outside the laboratory for longer periods would be a solution.

An ambulatory measurement system has been used by Baten et al. (2003) to estimate net lumbosacral moments, as a measure for mechanical load on the back. With this ambulatory measurement system (Amber) the kinematics of the trunk were measured with accurate inertial sensors attached to the body segments at an arbitrary position (Baten et al. 2000; Luinge et al. 1999). The body motions and the EMG of the back muscles, representing the unknown load at the hand, were subsequently used for the training of a neural network. This neural network was used for the estimation of the net moments around L5S1.

The ambulatory system contains 3D accelerometers, 3D gyroscopes and 3D magnetometers (rms < 5°). The output of these sensors are respectively, acceleration, angular velocity and heading with respect to the magnetic north of the sensor expressed in its local coordinate system. These physical parameters are used as input to a Kalman filter which compensates for the integration drift and which estimates orientations of the sensor with respect to a global coordinate system defined by the direction of gravitation and magnetic north (Luinge et al. 1999). The sensor coordi-
nate system can subsequently be used to determine the coordinate system of a segment. The segment axes are defined by means of reference movements and the gravity vector. (Baten et al. 2003). This method assumed that the relationship between the coordinate systems of sensor and segment is constant. The purpose of this study was to validate a method for estimation of upper extremity motions with Amber using the gravity vector and rotation axes. The repeatability of the method and the comparison between the segment coordinate systems determined with Amber and Optotrak, according to the ISB standardisation proposal based on bony landmarks (van der Helm et al. 1996a), was used for validation.

9.2 METHODS

9.2.1 Subjects and tasks

Five healthy subjects (age: 26 (SD 3) years, weight: 67 (SD 9) kg, length: 1.75 (SD 0.1) m) participated in a validation experiment.

The performed tasks can be divided into segment calibration tasks and validation tasks. Each task was performed two times. The following validation tasks were performed: Moving the hand over the head, perineal care, lifting a 1,75 kg bag, washing axilla and bringing a weight of 0.75 kg to the mouth (eating movement).
9.2.2 Experimental setup

The Amber setup included three MT9 inertial sensors (100Hz), each consisting of 3D gyroscopes, 3D accelerometers and 3D magnetometers. The sensors were calibrated according to the procedure as described in Table 9.1.

As can be seen in Figure 9.1, inertial sensors were mounted on thorax, humerus and forearm. The sensor on the thorax was attached to the sternum, the humerus sensor was attached to the dorsal side at the distal end and the forearm sensor was attached to the dorsal side between the styloids. The relative movement of the sensor with respect to the segment caused by muscles, tendons or skin is relatively small at these positions. The humerus sensor was attached to a rigid plate that was placed between the skin and the sensor to minimize skin movement. All sensors were attached with double adhesive tape. Fixomull tape was placed over the sensor for a more rigid attachment to the segment.

### Table 9.1: Sensor calibration procedure and accuracy for each sensor.

<table>
<thead>
<tr>
<th>Sensor</th>
<th>Procedure</th>
<th>Accuracy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Accelerometers</td>
<td>Determination of the magnitude of the acceleration vector after aligning of the sensor with the gravity.</td>
<td>9.81 ± 0.2 ms⁻²</td>
</tr>
<tr>
<td>Gyroscopes</td>
<td>Integration of the angular velocity for a rotation of the sensor over 90º</td>
<td>90 ± 1º</td>
</tr>
<tr>
<td>Magnetometers</td>
<td>Determination of the magnitude of the heading vector with the sensor successively placed on 6 sides.</td>
<td>1 ± 0.1 a.u.</td>
</tr>
</tbody>
</table>

Each trial started with five seconds without movement, which is needed for the sensor orientation to converge correctly. The sensors use kalman filters to determine the gravity vector and the magnetic north (Luinge et al. 1999).

The Optotrak LEDs were attached with double adhesive tape to the bony landmarks that are required to construct the segment coordinate systems as defined by van der Helm et al. (1996a). In addition, three LEDs were placed on each Amber sensor which is required for the determination of the rotation between the coordinate systems of both measurement systems.

9.2.3 Segment coordinate system of AMBER

For the determination of the segment coordinate system of AMBER, the relative orientation between the coordinate systems of the sensor and the segment has to be estimated (segment calibration). The segment is calibrated using rotation axes and
the gravity vector (Table 9.2). This procedure is described in more detail by Baten et al. (2003).

**TABLE 9.2:** The segment calibration methods using rotation axes and/or the gravity vector for the thorax (Th), humerus (Hum) and forearm (Fa). LS = coordinate system of sensor.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Segment axis 1</th>
<th>Segment axis 3</th>
<th>Segment axis 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>$Y_{Th}^{LS} = \frac{g_{y}^{Th}}{\omega_{x}^{Th}}$</td>
<td>$Z_{Th}^{LS} = \frac{\omega_{x}^{Th} \cdot g_{y}^{Th}}{\omega_{x}^{Th} \cdot g_{y}^{Th}}$</td>
<td>$X_{Th}^{LS} = Y_{Th}^{LS} \times Z_{Th}^{LS}$</td>
</tr>
<tr>
<td>Humerus</td>
<td>$Y_{Hum}^{LS} = \frac{\omega_{x}^{Hum}}{\omega_{y}^{Hum}}$</td>
<td>$Z_{Hum}^{LS} = \frac{\omega_{x}^{Hum} \cdot \omega_{y}^{Hum}}{\omega_{x}^{Hum} \cdot \omega_{y}^{Hum}}$</td>
<td>$X_{Hum}^{LS} = Y_{Hum}^{LS} \times Z_{Hum}^{LS}$</td>
</tr>
<tr>
<td>Forearm</td>
<td>$Y_{Fa}^{LS} = \frac{\omega_{y}^{Fa}}{\omega_{y}^{Fa}}$</td>
<td>$Z_{Fa}^{LS} = \frac{\omega_{x}^{Fa} \cdot \omega_{y}^{Fa}}{\omega_{x}^{Fa} \cdot \omega_{y}^{Fa}}$</td>
<td>$X_{Fa}^{LS} = Y_{Fa}^{LS} \times Z_{Fa}^{LS}$</td>
</tr>
</tbody>
</table>

The rotation axes, used for estimation of segment axes, are described in Table 9.3. The choice for the movements is based on the ability of the subject to make a movement in one plane around an axis and the repeatability of the movement by the subject.

**TABLE 9.3:** Segment calibration procedure that is used for AMBER

<table>
<thead>
<tr>
<th>Segment</th>
<th>Motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>$\omega_{x}^{Th}$ is determined when the subject is making a back flexion and extension. $g_{y}^{Th}$ is determined when a subject is standing in the anatomical position. In this posture the y-axis is aligned with the gravity vector.</td>
</tr>
<tr>
<td>Humerus</td>
<td>$\omega_{x}^{Hum}$ is determined by flexing and extending the elbow joint. This movement is made symmetrically by holding a bar with both arms at right angles to the bar. The elbow are flexed, making a square angle. The thumbs are laterally placed on the bar. The elbows are pressed against the body while making a flexion of about 90°. $\omega_{y}^{Hum}$ is determined by rotating the humerus internally and externally 45. The elbow is flexed 90°. The olecranon is supported.</td>
</tr>
<tr>
<td>Forearm</td>
<td>$\omega_{x}^{Fa}$ is determined by flexing and extending the elbow as described for the humerus. $\omega_{y}^{Fa}$ is determined by pronating and supinating with the elbow placed on the hip and the forearm in the frontal direction. The distal end of the rotation axis is fixed taking a pin with a fixed position between the ring finger and the little finger. The pin is the axis of rotational movement.</td>
</tr>
</tbody>
</table>
The three segment axes are normalised and form the columns of the rotation matrix $R_{LSLB}$ that defines the relative orientation between the sensor $O^G_{LS}$ and the body segment $O^G_{LB}$. The segment coordinate system of the body segment can now be calculated as follows: $O^G_{LB} = R_{LSLB} \cdot O^G_{LS}$.

### 9.2.4 Segment coordinate system of Optotrak

Using Optotrak, the segment coordinate systems are derived from the 3D position data of the bony landmarks as proposed by van der Helm et al. (1996).

### 9.2.5 Segment coordinate system of Amber in comparison with Optotrak

It is necessary that the obtained segment coordinate systems of both measurement systems are expressed in the same global coordinate system. Because Optotrak is considered to be the golden standard, the global system of Optotrak was used. The difference in segment coordinate systems was determined while the subject was standing upright.

The two systems were synchronised afterwards by cross correlating the magnitude of the angular velocities of a segment obtained with both systems. The angular velocities were determined by differentiating (Berme et al. 1990) and filtering (2nd order Butterworth, 50 Hz) the segment coordinate systems.

For the comparison of the segment coordinate systems obtained from Amber and Optotrak, the relative orientation between the two segment coordinate systems was calculated (eq. 1) which was used to express the segment coordinate system of Amber in the global coordinate system of Optotrak.

$$O^G_{LB,\text{Opto}} = R_{A2O} \cdot O^G_{LB,\text{Amber}} \Rightarrow R_{A2O} = O^G_{LB,\text{Opto}} \cdot (O^G_{LB,\text{Amber}})^{-1}$$

### Table 9.4: Definitions of segment coordinate systems based on bony landmarks

<table>
<thead>
<tr>
<th>Segment</th>
<th>Rotation 1</th>
<th>Rotation 2</th>
<th>Rotation 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>$X^G_{Th} = \text{Perpendicular to the plane fitted to the points } L^G, C^G$ \text{and} ( \frac{(P^G + T^G)}{2} )</td>
<td>$Y^G_{Th} = \frac{(L^G + C^G)}{2} - \frac{(P^G + T^G)}{2}$</td>
<td>$Z^G_{Th} = X^G_{Th} \times Y^G_{Th}$</td>
</tr>
<tr>
<td>Humerus</td>
<td>$Y^G_{Hum} = \frac{\left(\begin{array}{c} EM^G + EL^G \ EM^G - EL^G \end{array}\right)}{2}$</td>
<td>$Z^G_{Hum} = \text{Perpendicular to } Y^G_{Hum}$ \text{and} $EM^G - EL^G$</td>
<td>$X^G_{Hum} = Y^G_{Hum} \times Z^G_{Hum}$</td>
</tr>
<tr>
<td>Forearm</td>
<td>$Y^G_{Fa} = \frac{\left(\begin{array}{c} EM^G + EL^G \ EM^G + EL^G \end{array}\right)}{2}$</td>
<td>$Z^G_{Fa} = \frac{\left(\begin{array}{c} SR^G + SU^G \ SR^G + SU^G \end{array}\right)}{2}$</td>
<td>$X^G_{Fa} = Y^G_{Fa} \times Z^G_{Fa}$</td>
</tr>
</tbody>
</table>
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Comparison of upper extremity motions using optical and ambulant systems

Where $O_{LB, Amber}^G$ is the orientation of the segment obtained from Amber expressed in the global coordinate system, $O_{LB, Opto}^G$ is the orientation of the segment obtained from Optotrak expressed in the global coordinate system and $R_{12O}$ is the relative orientation between both coordinate systems.

The calculated orientation is subsequently converted into a quaternion representation. The quaternion vector represents a minimal rotation of an angle $\theta$ between the two segment orientations obtained from Optotrak and Amber around a unit vector $u = (ux, uy, uz)$ and can be written in the following format:

$$q = \begin{bmatrix} q_0 \\ q_1 \\ q_2 \\ q_3 \end{bmatrix} = \begin{bmatrix} \cos \left( \frac{\theta}{2} \right) \\ \sin \left( \frac{\theta}{2} \right) ux \\ \sin \left( \frac{\theta}{2} \right) uy \\ \sin \left( \frac{\theta}{2} \right) uz \end{bmatrix}$$

The vector $u$ can be written as helical angles: $\tilde{\theta} = \theta \cdot \tilde{u}$, $\alpha$ and $\tilde{\beta}$ are used for the comparison of Amber with Optotrak, since these variables describe the minimal rotation between the two segment coordinate systems, $O_{LB, Amber}^G$ and $O_{LB, Opto}^G$. The mean $\theta$ consists of two errors, a error that is caused by the difference in the definition of the long axis and an error that is caused by the difference in one of the orthogonal axes.

9.2.6 Data analysis

To determine the segment axes using Amber, the procedure was to rotate around a joint axis that was used to define the segment coordinate system. This method was called the segment calibration method. The mean angle $\bar{\alpha}$ between the direction of the angular velocity vector $\omega$ and the mean direction of the angular velocity $\bar{\omega}$ was determined for all trials and was used as a measure for accuracy of the estimation of the rotation axis. The accuracy of the estimation of a rotation axis depends on the accuracy of performance of the subject and on the rigidity of the relation between the sensor and the segment, which can be distorted by wobbling of the sensor.

Each segment calibration movement was repeated five times, which resulted in five matrices representing the relative orientation between sensor and segment $R_{12LB}$. The relative orientation of the five matrices with respect to the matrix calculated with the whole data set is translated into a quaternion representation. This resulted in five rotation angles $\theta$. The mean of these rotation angles $\bar{\beta}$ was used as a measure for repeatability.
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9.3 RESULTS

9.3.1 Accuracy and repeatability of segment orientation using Amber

Mean $\tilde{\alpha}$, representing the accuracy of estimation of the joint rotation axis, has a maximum value of 3.6 (SD 0.4) for elbow flexion (Table 9.5).

**Table 9.5:** Mean $\tilde{\alpha}$ (SD) for each motion, showing the accuracy of the estimation of the rotation axes for five subjects

<table>
<thead>
<tr>
<th>Motion</th>
<th>Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Elbow flexion</td>
<td>3.6 (0.6)</td>
</tr>
<tr>
<td>IR/ER</td>
<td>0.9 (0.1)</td>
</tr>
<tr>
<td>Thorax flexion</td>
<td>2.6 (0.1)</td>
</tr>
<tr>
<td>Pro/supination</td>
<td>1.3 (0.2)</td>
</tr>
</tbody>
</table>

Regarding the repeatability of the segment calibration method, a mean $\beta$ of 0.7° was found for the thorax, a $\beta$ of 1.6° for the humerus and a $\beta$ of 2.6° (SD 1.4) was found for the forearm (Table 9.6).

**Table 9.6:** Mean $\tilde{\beta}$ (SD) per segment representing the repeatability of the segment calibration.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thorax</td>
<td>0.7 (0.6)</td>
</tr>
<tr>
<td>Humerus</td>
<td>1.6 (0.8)</td>
</tr>
<tr>
<td>Forearm</td>
<td>2.6 (1.4)</td>
</tr>
</tbody>
</table>

9.3.2 Comparison of amber with Optotrak

The mean $\theta$ and standard deviation were determined for the three segments of each subject for the performed validation trials (Table 9.7). The highest error that was found, was 50.2° (SD 6.2) for the forearm segment.

**Table 9.7:** Mean difference (SD) per segment between segment coordinate system determined with Amber and Optotrak.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Thorax</th>
<th>Humerus</th>
<th>Forearm</th>
</tr>
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<td>17.7 (6.2)</td>
<td>22.9 (5.1)</td>
</tr>
<tr>
<td>2</td>
<td>9.0 (1.3)</td>
<td>12.3 (4.5)</td>
<td>19.7 (8.7)</td>
</tr>
<tr>
<td>3</td>
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<td>36.4 (7.3)</td>
<td>50.2 (6.2)</td>
</tr>
<tr>
<td>4</td>
<td>10.9 (3.1)</td>
<td>23.3 (6.4)</td>
<td>45.0 (5.7)</td>
</tr>
<tr>
<td>5</td>
<td>3.7 (1.8)</td>
<td>25.7 (4.9)</td>
<td>43.5 (7.1)</td>
</tr>
</tbody>
</table>
9.4 DISCUSSION

Using Amber, the direction of the rotation axes can be estimated accurately. The maximal deviation in the direction of an angular velocity vector during a movement is found for the elbow flexion. During this movement the mean $\alpha$ was $3.6^\circ$ when calculated with Amber. Additionally, when the motion was repeated five times during the segment calibration method, the intra-individual repeatability varied between $0.7^\circ$ and $2.6^\circ$. These values can be compared to the repeatability that can be achieved with a method based on bony landmarks measured with a six-degree-of-freedom measurement device (Meskers et al. 1998a).

Although the flexion axis is determined accurately, wrist rotations are not accommodated for. An alternative rotation axis for $\alpha$ might thus be a wrist flexion. By placing the hands on a table the forearm rotates around the wrist axis, which is anatomically pointing laterally.

9.4.1 Difference between amber and Optotrak

The difference between the orientation of the segments, described by $\theta$, can be divided in a structural difference and a variational difference.

To investigate if there were no errors in procedure, the obtained orientations based on motions around segment axes and on the gravity vector were also applied using Optotrak. It appeared that the differences in orientation of the segments between the two methods were small ($< 3^\circ$). Additionally, the rotation axis measured with Amber and Optotrak can be determined accurately and show good repeatable values. This means that the variation in segment orientation is caused by the definition differences. When bony landmarks are used, the axes of segment coordinate system is estimated based on two points. A disadvantage of this method is that the axes do not represent the rotation axes or the gravity vector. It can thus be concluded that the differences in definition will cause the relatively high $\theta$. However these differences can be considered as structural because they will remain constant for each subject.

With respect to variational difference, a source of variation might be skin movements or segment deformation. For Amber this might result in a variation in the relative orientation between the sensor coordinate system and the segment coordinate system. A relation between the difference in segment coordinate systems and acceleration, angular velocity and euler angles of the segments separately was not found. It is possible that the large variation is caused by the variation in a combination of the different variables. Regarding Optotrak, the position of bony landmarks with respect to the local coordinate system might change during a movement due to the skin movement. To which extent these variational errors will lead to the relatively large differences in $\theta$ is unknown.
The goal of the current study was to validate a method that is used for the recording of upper extremity motions. The method used rotation axes and the gravity vector to calculate the segment coordinate systems. It can be concluded that this method is able to produce accurate and repeatable segment coordinate systems. However, compared to the segment coordinate systems calculated on the basis of bony landmarks, the differences between the segment orientations were large. This is particularly caused by the difference in definition of segment coordinate systems. Since segment axes defined by rotations are a better estimation of the anatomical rotation axis than segment axes defined on the basis of bony landmarks, it might be preferable to use rotation axes and the gravity vector to determine the segment coordinate systems of thorax, humerus and forearm.

Using the Delft Shoulder and Elbow model, further research will determine the effect of the differences in segment orientation on the glenohumeral load spectrum.
CHAPTER 9
Comparison of upper extremity motions using optical and ambulant systems
10

ESTIMATION OF SHOULDER LOAD USING AN AMBULATORY MONITORING SYSTEM
CHAPTER 10

10.1 INTRODUCTION

A basic requirement for the development of a shoulder endoprosthesis is the load on the prosthesis during activities of daily living (ADL). The prosthesis must be able to withstand long lasting loads as well as peak loads. Estimating a load spectrum of the upper extremity for a longer period of time and during daily life is a very complex procedure. In the laboratory setup, net moments and muscle forces around the glenohumeral joint can be estimated using an inverse dynamic model of the upper extremity which uses the motions of body segments and the external load as input (Van der Helm et al., 1994). Upper extremity motions can be recorded using palpation of bony landmarks (Meskers et al. 1998) or using optical measurement systems in combination with a regression analysis to determine scapula orientations (de Groot et al. 1998; Pascoal, 2002). Van der Helm et al. (1996b) estimated joint load during wheelchair activity in the laboratory setup using a stereophotogrammetric system. An overall glenohumeral load spectrum during daily living has never been determined. A measurement system that is able to provide accurately estimated joint load outside the laboratory for longer periods would be a solution. Such an ambulatory system must accurately measure upper extremity motions and external load.

An ambulatory measurement system has been used by Baten et al. (2003) to estimate net lumbosacral moments, as a measure for mechanical load on the back. With this ambulatory measurement system (Amber) the kinematics of the trunk were measured with accurate inertial sensors attached to the body segments at an arbitrary position (Baten et al. 2000; Luinge et al. 1999). The body motions and the EMG of the back muscles were recorded. The amplitudes of the EMG were used since these signals represent the unknown external load. The EMG and motion signals were used for the training of a neural network. This neural network was used for the estimation of the net moments around L5S1.

The ambulatory system consists of inertial sensors that each contains 3D accelerometers, 3D gyroscopes and 3D magnetometers. The segment coordinate system can be calculated using rotation axes and the direction of the gravity vector. This
procedure is described in more detail by Baten et al. (2003). From a previous experiment it appeared that there were differences of 3-50° in segment orientations when compared to the segment orientations that were derived from bony landmarks using an optical system (Optotrak). To what extent these differences in segment orientation determination affect glenohumeral loading is unknown. Moreover, it is not known how accurate the glenohumeral load can be estimated using this method.

The purpose of this study was to validate the method for estimation of the net glenohumeral load during activities of daily living using an ambulatory measurement system. The glenohumeral load is estimated by using the upper extremity motions from Amber combined with EMG as input to a neural network. The neural network was trained to estimate the glenohumeral net moments. These net moments were then used as input to a large scale musculoskeletal model to calculate the glenohumeral load. The obtained glenohumeral load was finally compared to the load that was derived when Optotrak measurements in combinations with a known external load were used as input to the same musculoskeletal model.

10.2 METHODS

10.2.1 Subjects

Five healthy subjects (age: 26 (SD 3) years, weight: 67 (SD 9) kg, length: 1.75 (SD 0.1) m) participated in a validation experiment. All subjects gave written informed consent prior to the experiment.

10.2.2 Procedure

The experiment is divided into two parts, the training part and the validation part. In the training part, four complex motions of the subjects were recorded using an optical measurement system (Optotrak) and an ambulatory system (Amber) combined with an EMG system. During these complex motion tasks the subjects were asked to try to include all possible motion combinations of the shoulder. The duration of the tasks was between 45 and 60 seconds. The first task was performed without the subject holding a weight in the hand, while the other tasks were performed with weights of 0.75 kg, 1.5 kg and 3 kg. These complex motions recorded with Amber, in combination with the known external load were then used as input to a musculoskeletal model (Van der Helm, 1994) to calculate the net moments. These net moments were used to train a neural network. The neural network required the motions recorded with Amber and simultaneously measured EMG as input.

In the validation part of the experiment, the subjects performed four activities of daily living (ADL). These ADL were also recorded using Optotrak and Amber. Each task was performed two times. The following tasks were performed: Combing
Chapter 10

Estimation of shoulder load using an ambulatory monitoring system

Hair (CH), perineal care (PC), washing axilla (WA) and bringing a weight of 0.75 kg to the mouth (eat). First, the glenohumeral load was calculated for the motions recorded with Optotrak combined with a known external load, using the musculoskeletal model. The obtained glenohumeral load spectrum is assumed to be the golden standard. For validation of the ambulatory system, the motions recorded with Amber and the EMG were used as input to the trained neural network to estimate the net moments. These estimated net moments were subsequently used inside the musculoskeletal model to estimate the external load. The estimated external load in combination with the measured motions were finally used to calculate the glenohumeral load spectrum (Figure 10.1).

10.2.3 Experimental setup

The Amber setup included three MT9 inertial sensors (100Hz), each consisting of 3D gyroscopes, 3D accelerometers and 3D magnetometers (RMS < 5°). As can be seen in Figure 9.1, inertial sensors were mounted on thorax, humerus and forearm. The sensor on the thorax was attached to the sternum, the humerus sensor was attached to the dorsal side at the distal end and the forearm sensor was attached to the dorsal side between the styloids. The relative movement of the sensor with respect to the segment caused by muscles, tendons or skin is assumed to be small at these positions. The humerus sensor was attached to a rigid plate that was placed between the skin and the sensor to minimize skin movement. All sensors were attached with double adhesive tape. Each trial started with five seconds without movement, which is needed for the sensor orientation to converge correctly. The signals of the sensors were integrated to determine the gravity vector and the magnetic north. To prevent the integration drift that is often associated with these signals a Kalman filter is used (Luinge et al. 1999).

The Optotrak LEDs were attached with double adhesive tape to the bony landmarks that were required to construct the segment coordinate systems as defined by van der Helm et al. (1996a).

All segment motions were described according to the coordinate system of Optotrak. Therefore three LEDs were placed on each Amber sensor for the determination of the rotation between the coordinate systems of both measurement systems.

10.2.4 Neural network

For proper training of the neural network the input and output must be known. This meant that the net moments during the complex training tasks first had to be calculated. The net moments were calculated by means of the Delft Shoulder and Elbow Model (DSEM), which is a musculoskeletal inverse dynamic model of the upper extremity (Van der Helm, 1994). The calculated net moments were defined as
deformations of the nodes that form the acromioclavicular joint, the trigonum
spinae, the glenohumeral joint and the forearm. There were eight nodes and for each
deformation or net moment, a neural network was constructed.

The architecture of each neural network with backpropagation consisted of one
14 nodes input layer, two hidden layers each consisting of 30 nodes and one output
layer containing the net moment. A tansig transfer function was used for the
first and second hidden layer and a purelin function was used for the output layer.

A stepwise regression analysis determined the most sensitive inputs for the neural
network. The regression analysis used the EMG recordings of pectoralis major sternal part, pectoralis major clavicular part, trapezius descendens, trapezius ascen-
dens, biceps, triceps, brachioradialis, brachialis and the motions, velocities and
accelerations of all segments as input. All EMG signals were sampled with 1000Hz
and filtered with an analog filter that used a Highpass filter of 5Hz with a cutoff fre-

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**Figure 10.1:** Procedure to calculate the load spectrum of the glenohumeral joint. Motions obtained from Amber (AMotions) and the known external load are used to calculate the net moments using the Delft Shoulder and Elbow Model (1). The AMotions and EMG are used to train the neural network (NNt). The output of (1) is used for training (2) Activities of daily living measured with Amber (AMotions) are used for validation of the neural network (NNv). The net moments obtained from the neural network can finally be used to calculate the load on the glenohumeral joint using the DSEM. (3). The glenohumeral load is also calculated using the motions derived from Optotrak (Omotions) with the known external load (4) and compared to the output of (3).
quency of 400Hz. All EMG signals were rectified and subsequently digitally filtered with a 2nd order zero lag highpass butterworth filter (cutoff frequency = 100Hz) to filter out heart beat and then filtered with a 2nd order lowpass Butterworth filter with a cutoff frequency of 2.18 Hz. From this regression analysis it was concluded that the EMG recordings of the deltoid muscles, the velocities and accelerations of the segments did not contribute significantly to the output. Therefore the neural network used the recordings of the upper extremity motions obtained from Amber and the EMG of the other nine muscles as input. The upper extremity motions included orientations of thorax, humerus and forearm.

10.2.5 Delft Shoulder and Elbow Model

The shoulder model is a 3-D large scale musculoskeletal model of the upper extremity, consisting of the thorax, clavicle, scapula, humerus and forearm. In addition to the segments, all muscles and the musculoskeletal parameters are included. These parameters were all obtained from extensive cadaver measurements (Van der Helm et al., 1992, Veeger et al., 1991, Klein-Breteler et al. 1999). Muscle forces, joint reaction forces, moments and moment arms can be calculated using a cost function. A cost function is required to solve the load-sharing problem. The DSEM uses a cost function that minimises the squared muscle stresses.

To calculate the load on the glenohumeral joint, the measured motions and external loads of the segments can be used as input. However, when the motions will be recorded using an ambulatory measurement system, the external loads are not known. Using the neural network, the net moments can be estimated which is an indirect measure for the external load. The recorded motions of the ambulatory measurement system in combination with the estimated net moments of the neural network can be used as input to model to calculate glenohumeral load. It has to be taken into account that the motions obtained from Amber do not include scapular and clavicular motions, these motions were calculated on the basis of a regression analysis (Pascoal et al. 2002).

10.2.6 Data analysis

The glenohumeral load spectrum was determined for each subject and task. The load was calculated based on three procedures. The first procedure used a combination of a known external load and with segment motions calculated using bony landmarks (Optotrak). The second procedure used a known external load and Amber to calculate the segment motions. The last procedure used the estimated net moments from the neural network and the segment motions of Amber. Now, the effect of a difference in calculated segment motions (Amber vs Optotrak) on load can be determined by comparing procedure 1 and 2. The effect of a difference in the known
and estimated external load on glenohumeral load can be determined by comparing procedure 2 and 3. The combined effect of differences in segment motion definition and external load estimation was determined by comparing procedure 1 and 3.

The root mean square error (RMSE) was calculated between the procedures for each task and for each subject. The curves of the net glenohumeral reaction force were compared.

Table 10.1: RMSE between the estimated net moment of the neural network and the calculated net moment by the model. Fac = force on acromioclavicular joint, Fts = force on trapezius spinae, Fgh = force on glenohumeral joint, Fel = force on elbow and Fps = force on pro/supination axis. CH11 = combing hair by subject 1, repetition 1, PC = perineal care, WA = washing axilla and Eat = Eating.

<table>
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<th>Task and Subject</th>
<th>Facx (N)</th>
<th>Facy (N)</th>
<th>Ftsx (N)</th>
<th>Fghx (N/m)</th>
<th>Fghy (N/m)</th>
<th>Fghz (N/m)</th>
<th>Felx (N/m)</th>
<th>Fpsy (N/m)</th>
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<td>0.04</td>
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<td>1.97</td>
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<td>0.03</td>
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<td>0.07</td>
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<td>2.03</td>
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<tr>
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<td>7.38</td>
<td>7.18</td>
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<td>0.13</td>
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<td>3.69</td>
<td>2.65</td>
<td>2.45</td>
<td>0.32</td>
</tr>
</tbody>
</table>

10.3 RESULTS

10.3.1 Effect of segment axis definition on glenohumeral load

Due to the fact that the view of some Optotrak markers was blocked by other segments, it was not possible to calculate the segment orientations during almost all tasks of the last two subjects and some tasks of the other three subjects. Thirteen tasks were included for further analysis.

At first, a comparison between both measurement systems is performed. As described in Chapter 9, the segment orientations were calculated using two methods. Using Optotrak, the orientations were calculated using bony landmarks. The ambu-
FIGURE 10.2: Glenohumeral reaction force (N) produced by an optical system with a known external load (Opto), by an ambulant system with a known external load (Amber) and by an ambulant system combined with an estimated net moment from a neural network (NN).
CHAPTER 10
Estimation of shoulder load using an ambulatory monitoring system

Figure 10.2: Glenohumeral reaction force (N) produced by an optical system with a known external load (Opto), by an ambulant system with a known external load (Amber) and by an ambulant system combined with an estimated net moment from a neural network (NN)
CHAPTER 10

Estimation of shoulder load using an ambulatory monitoring system

Figure 10.2: Glenohumeral reaction force (N) produced by an optical system with a known external load (Opto), by an ambulant system with a known external load (Amber) and by an ambulant system combined with an estimated net moment from a neural network (NN).
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A multiaxial measurement system used rotation axes to determine the segment orientations. Differences in orientation between 5° and 50° were found for both methods. The smallest differences in orientation were observed for the thorax and the largest differences in orientation were observed for the forearm. The effects of these differences in orientation on the glenohumeral load are shown in Figure 10.2. It can be seen that the glenohumeral load pattern is comparable for both methods. The RMSE between these two methods varies between 50N and 180N (Figure 10.3).

10.3.2 Effect of difference in external load on glenohumeral load

The reaction forces obtained from the motions of Amber with the known external load were compared to the reaction forces obtained from the motions of Amber in combination with the estimated external load. The differences in reaction force between Amber and the Neural network were larger than the difference between the two measurement systems. As seen in Figure 10.2, the patterns are comparable, only the amplitude differs. The RMSE between the two procedures varies between 60N and 300N. The large difference between the two procedures is related to the RMSE of the estimated net moments of the neural network and the net moments calculated by the DSEM (Table 10.1).

**Figure 10.2:** Glenohumeral reaction force (N) produced by an optical system with a known external load (Opto), by an ambulant system with a known external load (Amber) and by an ambulant system combined with an estimated net moment from a neural network (NN).
10.3.3 Effect of segment axes definition and external load on glenohumeral load

The reaction forces obtained from the motions of Optotrak with the known external load were compared to the reaction forces obtained from the motions of Amber in combination with the estimated external load. Although a similar pattern (Figure 10.2) for the reaction force is demonstrated, the RMSE between the two procedures was slightly higher than between Amber and the neural network (Figure 10.3). The RMSE varied between 100N and 350N.

10.4 DISCUSSION

The aim of the current study was to validate a method that is able to predict the glenohumeral loading pattern with an ambulant measurement system. The glenohumeral load predicted by the ambulant system was compared to the load spectrum that was calculated on the basis of an optical measurement system. The differences in glenohumeral loading calculated with both systems can be caused by the differ-
ence in segment axes definition, the difference in known and estimated external
force or a combination of these two.

The difference in segment axes definition affected glenohumeral loading, howev-
er not to great extent. The RMSE for the hair combing tasks varied between 50 and
65N for the hair combing tasks, which is a 10% deviation with respect to the peak
loads. A 10% difference can be considered as acceptable. The eating task also dem-
onstrated similar reaction force patterns. Since both measurement systems demon-
strated accurate results, it can be concluded that with respect to record segment
motions, both systems can be used. The glenohumeral reaction force has not been
measured yet and therefore a conclusion about which method is superior cannot be
drawn. In the future an experiment with an instrumental prosthesis that is able to
calculate joint reaction forces can provide this information. A similar experiment
has already been performed for the hip (Bergman, 2001).

The aim of the ambulant system is to use it in daily living where external loads
are not known and thus have to be derived differently. A neural network was con-
structed to predict the net shoulder moments on the basis of segment orientations
and EMG of upper extremity muscles. At this moment the predictions of the neural
network are not satisfactory. The neural network overestimated the net shoulder
moments between 0.1 and 22 Nm. These net moment were used to estimate the ex-
ternal force that was used as input to the DSEM. It appeared that the external force
estimated from the estimated net moments, was different from the actual external
force. The larger the difference in estimated external force, the larger the difference
in glenohumeral loading. One of the main explanations why the glenohumeral re-
action force differs between the two systems is that despite very satisfactory results
for the training of the neural network (MSE < .01), it might be possible that the
training set was not complete. An optimal training set for the neural network in-
cludes a sufficient amount of combinations of upper extremity motions in combina-
tion with velocities, accelerations and external loads. Since the shoulder joint is a
ball- and socket-joint, the joint is extremely mobile. In the current study a “com-
plex” motion was performed by the subjects. Since the neural network is capable of
interpolating, it was assumed that this would be sufficient. It might be possible that
measuring for a longer period of time would have provided better results, but one
of the requirements for the definition of an optimal training set is time. It is practi-
cally not desirable to let a subject perform a standardised training task including all
combinations, for 30 minutes before ambulant measuring is possible. Future re-
search must define the best training set.

In conclusion, the ambulatory measurement system can be used in the future to
estimate glenohumeral loading during activities of daily living when net moments
and thus external forces can be estimated more accurately. In that case, the loading
requirements for a shoulder endoprosthesis can probably be defined. With respect to segment motions based on an optical system and on an ambulant system, the differences for glenohumeral loading are in the order of 10%, which is acceptable. However, when the external load was not known and the glenohumeral reaction force was estimated using a neural network, the force was overestimated, creating larger differences (50%). Future research must focus on improving the estimation of glenohumeral loading by defining a different, more extensive training set for the neural network.
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Estimation of shoulder load using an ambulatory monitoring system
11

GENERAL DISCUSSION
CHAPTER 11

11.1 IMPROVING SHOULDER ARTHROPLASTY

Shoulder arthroplasty is a complex surgical procedure for the orthopaedic surgeon. Despite the fact that surgeons are able to establish pain relief and an improvement in range of motion, the functional outcome after shoulder arthroplasty remains limited. The main objective of this thesis was to identify why this limitation in functional outcome exists and how it is possible to improve the functional outcome after arthroplasty.

Another objective was to identify the minimal requirements for an adequate performance of upper extremity activities of daily living. The requirements are acquired by means of an accurate 3-D motion analysis of healthy subjects during a subset of functional tasks. Subsequently, patients after shoulder arthroplasty were measured and compared with these requirements. From the 3-D motion analysis it appeared that some patients could not meet the requirements for glenohumeral motion, particularly the requirements for glenohumeral external rotation. In other words, glenohumeral motion is restricted. The demonstrated restriction can be caused by two mechanisms: A lack of glenohumeral stability or a lack of force to produce glenohumeral motion. A biomechanical analysis of the rotator cuff muscles showed that stabilising and moving the glenohumeral joint, in particular externally rotating the glenohumeral joint are the two main functions of these muscles. From these analyses it can be stated that the found restriction in glenohumeral motion is most likely caused by an insufficient amount of rotator cuff force. The results are in agreement with the literature, where it was found that the status of the rotator cuff muscles is a predominant factor in the post-operative outcome.

To improve shoulder arthroplasty the focus should be on replacing the functions of the rotator cuff. On the basis of both a biomechanical analysis and a patient study it can be concluded that the best option to replace rotator cuff functioning is a tendon transfer of the teres major to the insertion of the supraspinatus. This surgical procedure is able to provide glenohumeral stabilisation and glenohumeral external rotation when the rotator cuff is massively torn.
The challenge in shoulder arthroplasty is to design a new endoprosthesis that is able to provide glenohumeral stability and motion without relying on the rotator cuff. In particular, glenohumeral external rotation is required for an optimal result, hence shoulder arthroplasty must always be accompanied by a tendon transfer of the teres major if the rotator cuff is malfunctioning.

11.2 THE CLINICAL CHALLENGE

Objectives of surgery are to relieve pain and to restore function. This is often achieved by trying to restore the original anatomy, which is very difficult because the state of the bones is usually very poor due to the rheumatoid process. Despite the fact that surgeons are able to relieve the pain, the functional outcome after shoulder arthroplasty remains limited. As mentioned above, one of the goals of this thesis was to identify discriminating factors in functional outcome after shoulder arthroplasty.

By means of a formal meta-analysis an attempt was made to find these discriminating factors. As described in Chapter 2, factors that might contribute to functional outcome were obtained from forty-two studies. The factors were divided into three categories: design factors, patient factors and surgical factors. Unfortunately, due to the insufficient quality of the studies the inclusion criteria for a formal meta-analysis were not met. Factors that contributed to the poor quality were: heterogeneous patient groups, no randomly assigned treatment and awareness of the treatment used. Additionally, there were no comparative studies available.

The unsatisfactorily quality of the included studies resulted in the inability to draw conclusions about design factors and surgical factors. According to the systematic literature review the most important discriminating factors in functional outcome were pathology and in particular rotator cuff status. Patients with osteoarthritis showed larger improvements (50° in forward flexion) in range of motion (ROM) than patients with rheumatoid arthritis (35°) or complex humeral fractures (30°). Osteoarthritis patients were also able to elevate the humerus higher than 120°, which is a rather satisfactory result compared to 86° and 88° of humerus elevation of the rheumatoid arthritis or fracture patients. This difference in outcome might be a result of rotator cuff dysfunction, which is usually more observed in rheumatoid patients than in osteoarthritis patients. In contrast to osteoarthritis patients, rheumatoid arthritis patients are usually seen by the orthopaedic surgeon in a late stage when the inflammatory process has affected the soft tissue. Osteoarthritis mainly affects the joint due to osteophytes and does not have such a large influence on muscular status as rheumatoid arthritis. This is also demonstrated by the pre-operative results of both patient groups. The pre-operative forward flexion of these patients was approximately 55°, which is 20° lower than osteoarthritis patients. Additionally,
the studies of Hawkins et al. (1989), Nwakama et al. (2000), Cofield (1984) and Torchia et al. (1997) showed that patients with rotator cuff pathology have a significantly lower functional outcome than patients with an intact rotator cuff. Condition of the patient and muscular status, or more specifically, rotator cuff status, are the only factors that discriminate in the functional outcome after shoulder arthroplasty.

For the identification of discriminating factors, one of the recommendations of the literature study was to conduct multi-center trials, thereby focusing on one of the possible factors. Edwards et al. (2003) recently reported the results of such a multi-center trial. The purpose of the authors was to compare the functional results of total shoulder arthroplasty (TSA) and hemi shoulder arthroplasty (HSA) for a primary osteoarthritis patient population with intact rotator cuff muscles. The authors concluded that the results of TSA are superior to HSA with respect to functional outcome. The study of Edwards et al. (2003) demonstrated that for the identification of discriminating factors in shoulder arthroplasty, large multi-center trials are an excellent study design.

11.3 USING BIOMECHANICAL TOOLS

11.3.1 Upper extremity motion analysis

Motion analysis has proven to be a valuable tool for gaining insight into pathological motions. With respect to glenohumeral motion after shoulder arthroplasty there are two studies available (Boileau et al. 1992, Friedman, 1995). Furthermore, these studies only measured abduction in 2D using roentgen and not glenohumeral motion in daily life. Although upper extremity motions are measured quasi-statically due to the use of the scapulalocator, the methodology used in this thesis provides detailed and accurate information about upper extremity functioning. The rotations of thorax, clavicle, scapula, humerus and forearm can be analysed and compared in 3D. It remains a challenge to find an accurate tool that is able to record in vivo scapula motions dynamically in 3-D.

In addition to the quasi-static measurements that were performed to obtain detailed information of glenohumeral functioning, a first step towards an upper extremity ambulant measuring system was made. The objective of the use of an ambulant measuring system was to determine the glenohumeral load spectrum during daily living. This load spectrum was to be used as the load requirements for the design and fixation of the new endoprosthesis. Although the ambulant measuring system used different methods (rotation axes and the gravity vector) for the calculation of the local segment coordinate systems, the results were accurate and repeatable. The differences between segment orientation derived from the ambulant
system and the optical system did not seem to have a large effect on the gleno-humeral reaction force (<10%).

11.3.2 Upper extremity motion description

The description of the motions differs from the clinical nomenclature. The motions in this thesis are described according to standardisation proposal by the International Shoulder Group to the International Society for Biomechanics. The motions of thorax, clavicle, scapula, humerus and forearm are described as rotations about the axes of their local coordinate systems. The local coordinate systems are constructed on the basis of the coordinates of bony landmarks. Each motion is described as a joint motion. This means that motions of clavicle and scapula are described with respect to the thorax, motions of the humerus with respect to the scapula and motions of the forearm are described with respect to the humerus.

There are multiple reasons for choosing definitions that are different from the current clinical definitions. First of all, the clinical definitions do not cover the entire range of motion of the upper extremity. For example, when the hand is placed flat on the head and the elbow points laterally, is the humerus then externally rotated or internally rotated? This example is known as Codman’s paradox.

Another aspect is that the clinical terminology might cause confusion. For example, scapular abduction is defined as an abduction in the scapular plane, but the definition of scapular plane varies between 30° and 45°.

The general clinical description also does not directly contain information about the glenohumeral joint, but actually describes thoracohumeral motion. An forward flexion is a description of a thoracohumeral angle, which is actually a rotation about a non-existing joint. It is thus strongly recommended to use the methodology and terminology as used in this thesis due to the fact that it provides an accurate, complete and useful tool to compare and analyse the 3D upper extremity motions.

11.3.3 Inverse dynamic simulations

The DSEM is a large scale musculoskeletal model of the shoulder and elbow. The model is used inverse dynamically. This means that for calculation of muscle forces an optimisation criterion or cost function is needed. To solve the load-sharing problem, the model uses a cost function that minimises the sum of squared muscle stresses. Two important constraints are included in the model. The first constraint that has to be met is termed the moment constraint, which means that the model must balance the external moments by means of muscle forces. When a combination of muscle forces cannot be found to maintain moment equilibrium, the moment constraint is not satisfied and the model will report this. When the moment constraint cannot be met, the required muscle force is higher than the muscles can pro-
duce. When the muscle that is not able to produce the required force is identified, the importance of the muscle for the simulated motion is determined. To satisfy the second constraint, the stability constraint, the glenohumeral joint reaction force vector must point inside the glenoid cavity. If there is a combination of muscle forces available that is able to balance the external moments, but not to direct the glenohumeral joint reaction force vector inside the glenoid, the model will report that the second constraint cannot be met. When the muscle is identified that causes the instability, the stabilising role of the muscle for that particular motion is determined. In conclusion, the ability to satisfy the constraints of the model directly provides information about the mechanical role of muscles.

The DSEM is a representation of the complete upper extremity. Because it is a representation, there are of course some limitations. The model contains very detailed information about the anatomical characteristics of the upper extremity. The anatomical characteristics are obtained from extensive and accurate cadaver measurements. The currently used characteristics are of one cadaver, which means that interindividual morphological differences are not taken into account. Differences in muscle lengths, PCSA, ligament lengths and bone geometry will affect the output of the model. Due to the insufficient resolution and contrast of current visualisation techniques like CT and MRI it is not yet feasible to obtain patient-specific information.

Interindividual differences are partly accommodated for by means of measuring the variation in upper extremity motions. Twenty-four healthy subjects were measured for this purpose and subsequently used as input to the model.

Although the DSEM is used inverse dynamically, the model is evolving and currently it can run forward dynamically as well. As described in Chapter 1, forward dynamic simulations are computationally very expensive and therefore not ready to be applied in clinical practice. However, direct effects of adjustments in morphology or surgical procedures on kinematical output can now be established using a forward dynamic model.

11.4 APPLICATION OF BIOMECHANICAL TOOLS

11.4.1 Limitation in glenohumeral motion

Detailed information about why the functional outcome after shoulder arthroplasty is limited was obtained from an accurate and extensive motion analysis. Upper extremity motions of thirteen patients (16 shoulders) during six activities of daily living and six ROM tasks were measured and compared to the motions of twenty-four healthy subjects. As expected, the ability to move the glenohumeral joint was significantly reduced after shoulder arthroplasty. Differences between
CHAPTER 11
General Discussion

healthy glenohumeral motions and patient motions of 30° and 40° were found. In contrast to what is thought, the restriction in glenohumeral motions was not related to the ability to perform ADL. Except for glenohumeral external rotation, glenohumeral ROM was not significantly different between patients that were able to perform tasks above shoulder level and patients that were not able to perform these tasks. It appeared that patients who were able to perform tasks had the ability to perform compensatory motions in the sternoclavicular joint. The additional clavicular retraction resulted in more glenohumeral external rotation, which subsequently led to more glenohumeral elevation (Chapter 4). In conclusion, after shoulder arthroplasty a sufficient amount of glenohumeral external rotation is required for upper extremity functioning above shoulder level. It is possible to compensate for a lack of glenohumeral external rotation by means of an additional clavicular retraction. Caution is needed, because when joints are used to a great extent, overloading of this joint might become a secondary problem. These aspects have to be accommodated for in the rehabilitation process of the patient.

11.4.2 The mechanical role of rotator cuff muscles

The question that automatically arises from the fact that glenohumeral motion is restricted after shoulder arthroplasty is: why is glenohumeral ROM limited and how can it be improved? Since 90% of all patients report pain relief post-operatively, the restriction in motion is probably caused by a mechanical factor, like a lack of muscle force. A dysfunctional muscle that is not able to generate a sufficient amount of force can result in the inability to move the arm or to stabilise the glenohumeral joint. The mechanical role of glenohumeral muscles like the rotator cuff can be investigated using the Delft Shoulder and Elbow Model (DSEM).

In order to investigate the role of the rotator cuff muscles the maximal amount of muscle force the rotator cuff muscles could produce was varied and the individual rotator cuff muscles were omitted. Subsequently, the ability to satisfy the constraints when simulating the measured ADL and ROM was tested. It appeared that the main function of the rotator cuff muscles during humeral elevation tasks, like forward flexion, abduction and combing hair, is to assist the deltoids in moving the humerus. Elevation tasks require approximately 40° of external rotation. This required external rotation is normally fulfilled by the rotator cuff muscles, in particular the infraspinatus. Although the deltoids are strong enough to compensate for the lack of rotator cuff force, the additional produced deltoid force will result in undesired moments due to the relatively large moment arms of the deltoids. To balance these undesired moments, other muscles must become active to counteract the deltoids. Due to the fact that the glenohumeral joint must be stabilised as well, one advantage of the rotator cuff muscles is that due to the relatively small moment arms
around the glenohumeral joint, large compressive forces can be produced while avoiding antagonistic moments. In tasks, like washing the axilla and perineal care the main function of the rotator cuff muscles is to stabilise the glenohumeral joint (Chapter 5). It appeared that without rotator cuff activity, the humeral head will dislocate in posterior-superior direction. These results are in agreement with the literature, where proximal migration is often observed in patients with a deficient rotator cuff. With respect to the individual rotator cuff muscles, there is not one muscle that is most discriminating in the functional outcome. However, a combination of loss in infraspinatus and supraspinatus force will have a detrimental effect on the ability to meet the moment and stability constraints.

The motions are measured quasi-statically and therefore dynamical aspects have not been taken into account. From this analysis it cannot be stated what the role of the rotator cuff muscles is in dynamical stability. In other words, how do the rotator cuff muscles react to external perturbations? It might be possible that to compensate for perturbations, all rotator cuff muscles will co-contract as an attempt to stabilise the joint. This will result in a disturbed contraction pattern of the glenohumeral muscles, which might restrict motion as well. In the future an intra-muscular EMG study will provide more information about the role of the rotator cuff in dynamical situations.

In conclusion, a dysfunctional rotator cuff, or in particular tears in infraspinatus and supraspinatus will lead to an impaired functional outcome. The loss of force in these muscles results in a loss in the ability to externally rotate the humerus and in the ability to prevent the humeral head from luxating in the superior direction. Shoulder arthroplasty should focus on restoring the main rotator cuff functions. One of the options to restore rotator cuff function might be to transfer other muscles.

11.4.3 Tendon transfers

A dysfunctional rotator cuff is an indication for a poor functional outcome. Of all rotator cuff tears 80% of the tears develop in the supraspinatus/infraspinatus area (Warner, 2001). This means that the abduction and external rotation moment that is normally produced by the rotator cuff muscles is lost. Loss of abduction is rather easily compensated for by means of the deltoids. External rotation, however, is difficult to compensate for since only infraspinatus and teres minor have an external rotation moment arm. This can also be seen in patients with massive tears, where external rotation is often severely limited. As a consequence, since no single motion is pure, the loss of external rotation affects humeral elevation tasks, like combing hair as well. Without external rotation the hand will never reach the head. These functions can be restored by means of tendon transfers.
Tendon transfers for the treatment of massive tears demonstrated very positive results in the literature. The studies from Gerber (1992) and Aoki et al. (1997) showed an improvement in post-operative forward flexion. However, it is not exactly known which tendon transfer is the most effective in obtaining the best functional outcome. The DSEM has proven to be a very useful tool to identify the mechanical role of muscles, which is very convenient when a tendon has been transferred. On the basis of important mechanical properties such as muscle moment arms and muscle length, the mechanically best transfer can be studied. To evaluate the mechanical effects of the tendon transfers an extensive analysis including six ADL and three ROM tasks measured from twenty-four subjects was performed. For varying degrees of loss in rotator cuff force, the effect on the ability to perform these tasks is investigated. Subsequently the latissimus dorsi, teres major or both muscles are transferred to the insertions of the rotator cuff muscles and the same tasks were simulated. It can be concluded that a tendon transfer of teres major to the insertion of the supraspinatus resulted in the best functional outcome and was mechanically the most effective.

Ideally, tendon transfers should be simulated forward dynamically to directly find the effects on the kinematics, however as explained before, forward dynamic simulations are computationally very expensive and therefore not suitable yet for use in clinical practice. Although based on the results of one patient, as seen in Chapter 8, the predictions of the DSEM are comparable to both the pre-operative situation and the predicted outcome. Additionally, the model showed a probable mechanical cause for the experienced pain pre-operatively. The model is thus on its way to become a useful tool in the clinical setting, because the predictions of the outcome after a surgical procedure, such as a tendon transfer, are valid.

11.4.4 Glenohumeral load spectrum

An important design consideration is the load that is experienced on the prosthesis. A method to determine a load spectrum for the endoprosthesis has been developed using an ambulant measuring system in combination with a neural network and the shoulder model. The ambulant system is able to derive segment orientations from sensors that contain accelerometers, gyroscopes and magnetometers. The system also monitors EMG. The combined information of EMG and segment motion was used as input to a neural network that estimated net moments of the shoulder. These net moments were used as input to the DSEM to calculate the glenohumeral reaction force. However, it appeared that the use of a neural network to estimate net moments requires more research. The results of the neural network are insufficiently accurate and therefore produced an overestimation of the glenohumeral reaction force. If the neural network can be adjusted and tuned, the use of an ambulatory
measuring system will be very valuable to determine load requirements of the prosthesis.

11.5 **FUTURE IMPLICATIONS**

11.5.1 *Towards a new design*

The main goal of the DIPEX project is to develop a new endoprosthesis for the upper extremity which improves functional outcome after shoulder arthroplasty. As mentioned in the introduction, other groups in the DIPEX project are designing new prostheses. From the perspective of this thesis it can be concluded that the design process should focus on a prosthesis that is able to function without rotator cuff muscles. This means that the new design must be able to provide glenohumeral motion and it must be self stabilising. At this moment the prostheses that meets one of these criteria are: the constrained prostheses and the delta-prosthesis (Grammont, 1993) The constrained prostheses are intrinsically stable, however the reaction force points rather superiorly which results in loosening of the glenoid component. The Delta prosthesis is a reversed prosthesis. This means that the shoulder is still a ball and socket joint, only reversed. In other words, the glenoid component is now ball-shaped and the humeral head does not exist any more. Because the glenohumeral rotation center is displaced more medially, the deltoid muscles are able to produce glenohumeral elevation without having to produce extra force. Short-term functional results for the patient group that previously almost always scored poor on functional scales now appear to be performing satisfactorily. However, the Delta prosthesis is difficult to fixate into the bone and the glenohumeral reaction force is rather high. Furthermore, the long-term loosening results are not known yet.

11.5.2 *Tendon transfers combined with shoulder arthroplasty*

It must be taken into account that when the rotator cuff muscles are completely dysfunctional or torn, there are not many scapulohumeral muscles left that are able to produce glenohumeral motion and stability. In addition to the deltoid muscles, the thoracohumeral muscles, like pectoralis major, teres major and latissimus dorsi must be able to compensate for rotator cuff function and thus have to produce glenohumeral motion. Since a range of 40° of external rotation of the humerus is required for many ADL, a tendon transfer subsequent to arthroplasty might offer a solution. As explained before, rotator cuff function is usually impaired for arthroplasty patients, hence in this case a tendon transfer probably will be a valuable addition to shoulder arthroplasty. A teres major tendon transfer to the insertion of the supraspinatus has proven to be an excellent procedure in the case of massive rotator cuff tears. Instead of being able to produce an adduction and internal rotation moment,
the teres major muscle generates an abduction and external rotation moment, which are exactly the required moments for this group of patients.

11.5.3 Recommendations

To obtain more information about muscle status in patients, future experiments should include passive motion measurements and EMG of the upper extremity. Differences in passive and active motion measurements demonstrates the inability of the muscles to move the joint. In combination with intra-muscular EMG measurements of the rotator cuff, it will become visible if these muscles are massively co-contracting which will restrict motion or if the muscles are inactive or are operating on a very low level. In addition, a very valuable experiment would be to identify the mechanical role of the rotator cuff muscles of healthy subjects in dynamic situations and how these muscles react to external perturbations. It might be possible that the reflex activity of the rotator cuff muscles is disturbed after shoulder arthroplasty, which will lead to the inability to stabilise the glenohumeral joint after perturbations. Although this probably will place a burden on the patient, the reflex activity of the rotator cuff muscles after shoulder arthroplasty is important information.

Apart from the conclusions with respect to the improvement of shoulder arthroplasty, this thesis demonstrated that cooperation between engineers and physicians is very valuable, in particular for rehabilitation practice and orthopaedics. Tools such as motion analysis and biomechanical musculoskeletal models can be used for an accurate diagnosis of upper extremity pathologies. It directly shows why the patient is unable to move and using the models, treatment options can be investigated and optimised. Furthermore, the same tools can be used to evaluate surgical procedures. Despite the fact that the models are improving and will most likely be used in the operative theatre, the engineer is still needed to act as an advisor. The ultimate goal would be to use the DSEM as a stand-alone package which can be adopted by the surgeon. To achieve this, the DSEM must have a user-friendly, easy to interpret interface and the model must come to a solution fast because surgeons do not have much time to wait for the computer to come to a solution. When this is the case, the DSEM is ready to be used as a pre-operative planning tool for the orthopaedic surgeon to evaluate, for example, which prosthesis will result in the mechanically most feasible solution.
REFERENCES

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References


REFERENCES

S


T


U


V


W

REFERENCES


Z

LIST OF PUBLICATIONS

This thesis is based on the following articles and proceedings:


Magermans, D.J., Nagels, J., Chadwick, E.K., Veeger, H.E.J., Van der Helm, F.C.T (2002). Motion patterns of the shoulder for patients with a shoulder endoprosthesis during combing hair. Proceedings of the 4th meeting of the international shoulder group, Cleveland, USA


LIST OF PUBLICATIONS
SUMMARY

Damage to the glenohumeral joint due to fractures or diseases such as rheumatoid arthritis usually results in pain and a limitation in every day functioning. Replacement of the glenohumeral joint (shoulder arthroplasty) is an effective surgical procedure with respect to pain relief and range of motion, but not for the performance of functional tasks. The cause of the restricted functionality after shoulder arthroplasty is unknown. The main goal of the multidisciplinary DIPEX project is to improve the functional outcome after shoulder arthroplasty, possibly through the design of a new endoprosthesis. One of the objectives of this thesis is to formulate the specifications for this endoprosthesis. These specifications are divided into motion requirements for the performance of activities of daily living and loading requirements for the glenohumeral joint. A second objective is to identify discriminating factors in functional outcome. The third objective is to find a possible treatment option to restore the post-operative functionality.

The main conclusion of this thesis is that improvements in shoulder function after replacement of the glenohumeral joint can be achieved by replacing or improving the function of the rotator cuff muscles. The rotator cuff muscles are able to provide glenohumeral stability and to produce a sufficient amount of glenohumeral motion, in particular external rotation. The results from the studies included in this thesis indicate that these two functions are the most important requirements for a satisfactory functional outcome.

The indication of the importance of the rotator cuff muscles was found in a large literature review that was performed to identify discriminating factors in shoulder arthroplasty. Design, surgical and patient factors all contribute to the post-operative outcome, but firm conclusions about which factors were discriminating could not be drawn due to the methodologically insufficient quality of most studies. Confounding factors were multiple conditions were included in one patient group, non random allocation of patients to groups, unclear descriptions of the definitions of range of motion and non-blinded experimental designs. Despite the poor quality of the included studies, the status of the rotator cuff muscles appeared to be a dominant factor in the post-operative results.
To determine requirements for a satisfactory post-operative functional result, the upper extremity motions of twenty-four healthy subjects during activities of daily living were measured. It must be taken into account that the shoulder is a very complex joint to measure since the scapula moves underneath the skin. Using a scapulalocator the joint motions of the scapula can be recorded quasi-statically by means of palpation. Sensors of an electromagnetic tracking device (Flock of Birds) were attached to the thorax, humerus, scapulalocator and forearm to measure the upper extremity motions. Tasks like combing hair, washing axilla, lift a bag, perineal care and eating were performed. From the performed joint motion patterns, it appeared that for the tasks that require high glenohumeral angles, large external rotation values were observed.

To identify the joint motion that limits functional outcome after shoulder arthroplasty, the same protocol was repeated for thirteen patients with an endoprosthesis. By dividing the patients into a group that was for example able to comb hair and into a group that was not able to comb hair, the differences in motion could be found. This task required large glenohumeral elevation angles and was difficult to perform for most patients. Successful patients showed more glenohumeral elevation during hair combing than unsuccessful patients. However, in contrast to what was expected, the restriction in glenohumeral motion was not related to the ability to perform this functional tasks. Except for glenohumeral external rotation, both patient groups did not differ significantly in active glenohumeral elevation angles. During hair combing successful patients were able to achieve higher glenohumeral angles due to compensation strategies. The successful patients demonstrated more scapulothoracic motion and in particular more sternoclavicular retraction. This additional sternoclavicular retraction is required to produce additional glenohumeral external rotation which results in more function. Hence, it can be concluded that an important motion requirement is glenohumeral external rotation. Despite the fact that compensatory motions can improve functioning, it must be taken into account that this might lead to secondary problems in other joints such as for instance the sternoclavicular joint and acromioclavicular joint.

The observed glenohumeral motion restriction is most likely caused by dysfunctional rotator cuff muscles. An inverse dynamic upper extremity model (Delft Shoulder and Elbow Model) was used to analyse the functions of the rotator cuff muscles during activities of daily living. It appeared that two mechanisms might be responsible for the observed limitation. First, loss in supraspinatus and infraspinatus force will result in a decreased ability to external rotate the humerus. As mentioned before, a decrease in the ability to produce external rotation will lead to a decrease in glenohumeral elevation. Tasks like combing hair and reaching require large elevation and external rotation angles and will thus be difficult to perform.
Second, a decrease in rotator cuff force will lead to a decrease in the ability to stabilise the glenohumeral joint, in particular during perineal care and washing the axilla. Although dynamical aspects have not been taken into account, the performed analysis demonstrated that the rotator cuff muscles have very important functions which are required for a satisfactorily functional outcome after shoulder arthroplasty.

Transferring the tendon of latissimus dorsi or teres major muscle is a surgical procedure for the treatment of massive irreparable rotator cuff tears. By means of a biomechanical analysis the mechanical effect of this procedure was investigated. The Delft Shoulder and Elbow Model was adjusted to mimic the pre-operative status of a patient with a massive rotator cuff tear. Since it is not exactly known to what extent a tear in the supraspinatus muscle affects the maximal force of the other rotator cuff muscles, the maximal force of these muscles was varied. Regarding the tendon transfers, in the model the teres major, latissimus dorsi and both muscles were transferred to the four insertions of the rotator cuff muscles. Subsequently, the ability to simulate the different motions was evaluated. Additionally, important mechanical parameters such as muscle moment arms and muscle lengths were investigated. It was concluded that a transfer of the teres major muscle to the insertion of the supraspinatus muscle resulted in the mechanically most favourable option.

To validate the conclusions that a tendon transfer of teres major to the insertion of the supraspinatus muscle results in the predicted functional outcome, the model predictions were compared to the functional outcome of a patient that underwent the same procedure. Since pain is a very complex mechanical parameter, which is difficult to model, the effect of pain on motion was not taken into account. For comparison, the patient received a subacromial lidocain injection to reduce the effect of pain. It appeared that after lidocain injection the patient showed near normal motion patterns. These results were comparable to the model predictions when the supraspinatus could not produce any force and the other rotator cuff muscles were intact. The motions of the patient were also analysed and it appeared that the modelled glenohumeral reaction force was located more superiorly, which might be an indication for the subacromially experienced pain. After transfer the functional outcome of the patient improved. Pain relief, improvements in glenohumeral elevation and external rotation were demonstrated. The post-operative outcome of the patient was very similar to the predicted functional outcome. Since glenohumeral motion and glenohumeral stability was improved it appears that this procedure is able to restore rotator cuff functioning.

The last important design consideration for the development of the new endoprosthesis is the determination of the load on the glenohumeral joint. A durable and successful life of the endoprosthesis is required for a satisfactory outcome. Estimating
a load spectrum of the glenohumeral joint during daily life is a complex procedure. Using an ambulatory measurement system the load spectrum can be calculated. The ambulatory measurement system consisted of sensors that were attached to thorax, humerus and forearm. Each sensor contained accelerometers, a gyroscope and a magnetometer.

In a laboratory set-up activities of daily living were measured with the ambulatory system and with an optical system. The orientation of the segments were calculated by means of constructing local coordinate system based on the 3-D coordinates of bony landmarks (optical system) and by means of reference movements and the gravity vector (ambulatory system). It appeared that the difference in axes definition causes differences in segment orientation varying between 3° for the thorax and 50° for the forearm. Although there is a difference, it can be concluded that both methods produce accurate and reproducible results. Since kinematically determined axes might be a better estimation of the segment rotation axes, a combination of the two axes definitions might be preferred in the future.

The segment orientations can now be accurately determined using an ambulant measurement system. The next step was to validate the calculation of the glenohumeral load. The glenohumeral load was calculated by using the ambulatory measured segment motions in combination with measured EMG as input to a neural network. This neural network estimated the net moments of the shoulder complex. The estimated net moments were then used as input to the Delft Shoulder and Elbow Model to calculate the glenohumeral reaction forces. The obtained glenohumeral reaction forces were then compared to the reaction forces that were obtained from calculations based on the related experimental data. It appeared that the output of the neural network causes differences in the glenohumeral reaction forces that were more than 10%. It was concluded that, although the method appeared promising and might be very valuable in the future, more research will be needed to improve accuracy.

In conclusion, the most important requirements for the new endoprosthesis were formulated in this thesis. The new endoprosthesis should be able to function without the rotator cuff muscles. In other words it should be able to replace the rotator cuff functions. The prosthesis must allow for a sufficient amount of external rotation and it should be self-stabilising. Another potential solution to improve the functional outcome might be to transfer the teres major muscle additional to shoulder arthroplasty. With respect to the load requirements a conclusion cannot be drawn yet. An ambulant measurement system will be the solution in the future, but more research is needed to estimate the glenohumeral load.
SAMENVATTING

Een beschadigd glenohumeraal gewricht als gevolg van fracturen of aandoeningen zoals reumatoïde artritis leiden regelmatig tot pijn en beperkingen in het dagelijks functioneren. Een vervanging van het glenohumeraal gewricht (schouder arthroplastiek) is een effectieve methode om pijn te verminderen en range of motion te verbeteren. Echter, de mogelijkheid tot het uitvoeren van dagelijkse taken blijft beperkt. De oorzaak van deze beperkte functionele resultaten zijn tot op heden onbekend. Het multidisciplinaire DIPEX project heeft als doel gesteld om de functionele uitkomst na schouder arthroplastiek te verbeteren, mogelijk door het ontwikkelen van een nieuwe endoprothese. Een van de doelen van dit proefschrift was het formuleren van specificaties voor deze nieuw te ontwikkelen endoprothese. Deze specificaties zijn onderverdeeld in bewegingseisen om zoveel mogelijk taken van het dagelijks leven uit te kunnen voeren en eisen met betrekking tot de belasting op het glenohumerale gewricht. Een tweede doel van dit proefschrift was het identificeren van discriminerende factoren voor de functionele uitkomst. Het vinden van een mogelijke behandelmethode die de postoperatieve functionaliteit herstelt was het derde doel.

De resultaten van deze dissertatie leiden tot de conclusie dat na vervanging van het glenohumerale gewricht er verbeteringen in de schouderfunctie bewerkstelligd worden als de functies van de rotator cuff vervangen worden. De rotator cuff is in staat om het glenohumerale gewricht te stabiliseren en om voldoende glenohumerale beweging te produceren, met name externe rotatie. De resultaten van de individuele studies van dit proefschrift demonstreren dat deze twee functies het meest bepalend zijn voor een goede functie.

Het belang van de rotator cuff werd duidelijk in een literatuurstudie die was uitgevoerd om de discriminerende factoren in schouder arthroplastiek te identificeren. Chirurgische, ontwerp en patiënt factoren dragen allemaal bij aan het postoperatieve resultaat, echter conclusies met betrekking tot welke factoren discrimineren konden niet getrokken worden vanwege de beperkte methodologische kwaliteit van de geïncludeerde studies. Ondanks de beperkte kwaliteit leek de status van de rotator cuff een bepalende factor voor de postoperatieve resultaten.
Om eisen te formuleren voor een voldoende postoperatieve resultaat zijn de bewegingen van de bovenste extremiteit van vierentwintig gezonde proefpersonen gemeten gedurende activiteiten van het dagelijks leven. Er moest rekening gehouden worden met het feit dat de schouder een complex gewricht is om te meten omdat de scapula onder de huid beweegt. Door middel van palpatie kunnen de bewegingen van de scapula met behulp van een scapulalocator quasi-statisch gemeten worden. Sensoren van een electromagnetisch meetsysteem (Flock of Birds) werden bevestigd aan de thorax, humerus, scapulalocator en aan de onderarm om de bewegingen van de bovenste extremiteit te meten. Taken zoals haren kammen, oksel wassen, eten, tillen, en billen afvegen werden door de proefpersonen uitgevoerd. Het bleek dat er tijdens de taken die veel glenohumerale elevatie nodig hadden om uitgevoerd te worden tevens veel externe rotatie geobserveerd werd.

Hetzelfde meetprotocol werd herhaald voor dertien patiënten met een prothese om de bewegingen te identifieren die de functionele uitkomst na schouder arthroplastiek beperkt. Verschillen in bewegingen konden gevonden worden door de patiënten onder te verdelen in een groep die in staat was om hun haar te kammen en een groep die niet in staat was om hun hair te kammen. Tijdens haren kammen werden hoge glenohumerale hoeken geobserveerd en het bleek tevens een lastige taak om uit te voeren. Succesvolle patiënten bleken in staat om meer glenohumeraal te eleveren tijdens haar kammen dan niet succesvolle patiënten. Echter, in tegenstelling tot wat wordt verwacht bleek de beperking in glenohumerale beweging niet gerelateerd te zijn aan de mogelijkheid om de taak uit te voeren. Op totale actieve glenohumerale externe rotatie na, was er geen significant verschil in actieve glenohumerale elevatie hoeken. Tijdens haar kammen zijn succesvolle patiënten in staat hogere elevatiehoeken te halen vanwege de mogelijkheid om te compenseren. De succesvolle patiënten waren in staat meer scapulothoracal bewegen, maar met name waren ze in staat meer sternoclaviculair te retracteren. Deze additionele sternoclaviculaire retractie is nodig om extra glenohumerale externe rotatie te produceren die op zijn beurt weer nodig is voor meer glenohumerale functie. Uit deze resultaten kan geconcludeerd worden dat glenohumerale externe rotatie een belangrijke bewegingseis is voor een goede functionele uitkomst.

De geobserveerde glenohumerale bewegingsbeperking is waarschijnlijk het gevolg van een dysfunctionerende rotator cuff. Met behulp van een invers dynamisch model (Delft Schouder and Elleboog Model) zijn de functies van de rotator cuff geanalyseerd tijdens gemeten taken van dagelijks leven. Het bleek dat twee mechanismen waarschijnlijk verantwoordelijk zijn voor de geobserveerde limitatie. Ten eerste, krachtsverlies van de supraspinatus en infraspinatus leidt tot een afname in de externe rotatie. Zoals reeds besproken leidt een afname in de externe rotatie tot een afname in de glenohumerale elevatie. Taken zoals haar kammen en reiken
vereisen hoge elevatie en externe rotatie hoeken en worden dus moeilijk uitvoerbaar. Ten tweede leidt een afname in de kracht van de rotator cuff tot een afname in glenohumerale stabiliteit, met name tijdens het billen afvegen en tijdens het wassen van de oksel. Ondanks het feit dat dynamische aspecten niet zijn meegenomen demonsstreert de uitgevoerde analyse dat de rotator cuff zeer belangrijke functies bezit die nodig zijn voor een goede functionele uitkomst na schouder arthroplastiek.

Een mogelijke chirurgische procedure voor de behandeling van onherstelbare scheuren van de rotator cuff is een peestranspositie van de latissimus dorsi of van de teres major. Door middel van een biomechanische analyse werd het mechanische effect van deze procedure onderzocht. Het Delft Schouder en Elleboog Model werd aangepast om de preoperatieve status van de patiënt met een onherstelbare rotator cuff scheur te benaderen. Omdat het effect van een scheur in de supraspinatus op de spierkracht van de overige rotator cuff spieren onbekend is, werd de maximale spierkracht van de overige rotator cuff spieren gevarieerd in het model. Vervolgens werden de aanhechtingen van de latissimus dorsi, teres major of beide spieren verplaatst in het model naar de inserties van de vier rotator cuff spieren. De mogelijkheid om bewegingen te simuleren werd nu bepaald met het anatomische aangepaste model. Bovendien werd het effect op belangrijke mechanische parameters zoals spierlengte en momentarmen van spieren onderzocht. Na deze analyse kon geconcludeerd worden dat vanuit mechanische oogpunt een transpositie van de teres major naar de insertie van de supraspinatus de beste optie is.

Om te kunnen concluderen dat een peestranspositie van de teres major naar de insertie van de supraspinatus werkelijk resulteert in de voorspelde functionele uitkomst, werden de modelvoorspellingen vergeleken met de functionele resultaten van een patiënt die behandeld werd met deze procedure. Het effect van pijn is niet meegenomen omdat pijn een zeer complexe mechanische parameter is om te modelleren. Om het effect van pijn teniet te doen kreeg de patiënt voordat de metingen plaatsvonden, een subacromiale lidocaïne injectie. Na deze injectie liet de patiënt een nagenoeg gezond bewegingspatroon zien, wat overeenkomt met voorspellingen van het model wanneer alleen de supraspinatus geen kracht meer kon produceren. Na analyse van de bewegingen van de patiënt bleek dat de locatie van de gemodelleerde glenohumerale gewrichtsreactiekraft meer superior was. De superior gerichte reactiekraft kan een indicatie zijn van de preoperatieve subacromiale pijnervaring. Na de peestranspositie verbeterde de functionele resultaten. Vermindering van pijn, verbeteringen in glenohumerale elevatie en externe rotatie werd door de patiënt gedemonstreerd. De postoperatieve resultaten waren zeer vergelijkbaar met de voorspelde functionele resultaten. Vanwege het feit dat glenohumerale beweeglijkheid en glenohumerale stabiliteit toenamen, kan gesteld worden dat deze procedure de functies van de rotator cuff tot op zekere hoogte kan herstellen.
Het laatste belangrijke ontwerpcriterium voor de ontwikkeling van de nieuwe endoprothese is de bepaling van de belasting op het glenohumerale gewricht. Een duurzaam en succesvol leven is vereist voor een goed resultaat. Het bepalen van een glenohumeraal belastingspectrum gedurende het dagelijks leven is een complexe procedure. Met behulp van een ambulant meetsysteem kan het belastingspectrum worden uitgerekend. Het ambulante meetsysteem bestaat uit sensoren die worden bevestigd aan de thorax, humerus en onderarm. Iedere sensor bestaat uit versnellingsopnemers, een gyroscop en een magnetometer.

In een laboratorium opstelling werden vervolgens activiteiten van het dagelijks leven gemeten met een ambulant en optisch meetsysteem. De oriëntatie van de segmenten werd berekend door assenstelsels te definiëren op basis van de 3D coördinaten van botpunten (optisch systeem) en op basis van referentiebewegingen in combinatie met de zwaartekrachtvector (ambulant systeem). Het bleek dat het verschil in asdefinitie verschillen in segment oriëntatie veroorzaakt variërend tussen 3° voor de thorax en 50° voor de onderarm. Ondanks dit gevonden verschil kan er geconcludeerd worden dat beide methoden accurate en reproduceerbare resultaten produceren. Omdat kinematisch bepaalde assen een betere schatting zijn van de rotatieassen van een segment is er een voorkeur voor het gebruik van een combinatie van deze twee methoden voor de bepaling van de segment oriëntatie.

Nu de segment oriëntaties nauwkeurig bepaald konden worden met een ambulant meetsysteem, was de volgende stap om de berekening van de glenohumeraal belasting te valideren. De glenohumeraal belasting werd berekend door de ambulant gemeten bewegingen in combinatie met gemeten EMG te gebruiken als input voor een neurale netwerk. Dit neurale netwerk schat de netto momenten van het schouder mechanisme. De geschatte netto momenten dienden vervolgens als input voor het Delft Schouder and Elleboog Model om de glenohumeraal reactiekrachten te berekenen. De verkregen reactiekrachten werden vergeleken met de reactiekrachten gebaseerd op gerelateerde experimentele data. Het bleek dat de output van het neurale netwerk verschillen in de glenohumeraal reactiekrachten veroorzaakte die groter waren dan 10%. Ondanks het feit dat de methode in de toekomst waarschijnlijk zeer waardevol is, is er nog veel onderzoek nodig om de nauwkeurigheid van het neurale netwerk te verbeteren.

Concluderend, de belangrijkste vereisten voor een nieuwe endoprothese zijn geformuleerd in dit proefschrift. De nieuwe prothese dient te functioneren zonder hulp van de rotator cuff. Met andere woorden, de prothese dient de functies van de rotator cuff te vervangen. De endoprothese moet voldoende externe rotatie toestaan en bovendien moet de prothese zelfstabiliserend zijn. Een potentiële oplossing om de functionele resultaten te verbeteren is een transpositie van de teres major in combinatie met schouder arthroplastiek. Met betrekking tot de belastingeisen voor de en-
Doproteese kunnen er nog geen conclusies worden getrokken. Een ambulante meetsysteem biedt in ieder geval de mogelijkheid om het glenohumerale belasting-spectrum te bepalen, maar meer onderzoek is noodzakelijk om de nauwkeurigheid van dit systeem te vergroten.
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