Compensatory muscle activation in patients with glenohumeral cuff tears

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Chapter 4

Glenohumeral stability in simulated rotator cuff tears

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Abstract

Rotator cuff tears disrupt the force balance in the shoulder and the glenohumeral joint in particular, resulting in compromised arm elevation moments. The trade-off between glenohumeral moment and glenohumeral stability is not yet understood. We hypothesize that compensation of lost *ab*duction moment will lead to a superior redirection of the reaction force vector onto the glenoid surface, which will require additional muscle forces to maintain glenohumeral stability.

Muscle forces in a single arm position for five combinations of simulated cuff tears were estimated by inverse dynamic simulation (Delft Shoulder and Elbow Model) and compared with muscle forces in the non-injured condition. Each cuff tear condition was simulated both without and with an active modeling constraint for glenohumeral stability, which was defined as the condition in which the glenohumeral reaction force intersects the glenoid surface.

For the simulated position an isolated tear of the supraspinatus only increased the effort of the other muscles with 8%, and did not introduce instability. For massive cuff tears beyond the supraspinatus, instability became a prominent factor: the deltoids were not able to fully compensate lost net *ab*duction moment without introducing destabilizing forces; unfavorable *ab*ductor muscles (i.e. in the simulated position the subscapularis and the biceps longum) remain to compensate the necessary *ab*duction moment; the teres minor appeared to be of vital importance to maintain glenohumeral stability. Adverse *ad*ductor muscle co-contraction is essential in order to preserve glenohumeral stability.

4.1 Introduction

Massive rotator cuff tears disrupt the force and moment balance in the shoulder and the glenohumeral joint in particular. This generally coincides with severe pain and disabilities (Iannotti et al., 2006, Jost et al., 2000). Severity of cuff afflictions range from isolated supraspinatus tearing to partial/full tearing of supraspinatus, infraspinatus, teres minor and subscapularis, the so-called massive tears. The biceps longum is known to have a stabilizing effect on the humeral head, but is frequently affected in patients with cuff tears (Warner and McMahon, 1995, Kempf et al., 1999, Murthi et al., 2000).

In patients with massive rotator cuff tears, pathological co-activation of large muscles with an adducting component (teres major, latissimus dorsi and pectoralis major) was observed during an isometric *ab*duction moment task in a single position (de Groot et al., 2006, Steenbrink et al., 2006). This position is considered critical for several shoulder pathologies, i.e. cuff tears, impingement syndrome, arthritis and habitual shoulder instability and/or subluxation. The alteration in muscle activation patterns in these patients was assumed to be the compensatory response for stabilization of the glenohumeral joint. Proximal migration of the humeral head during *ab*duction moments observed in patients with massive rotator cuff tears (Deutsch et al., 1996, Paletta, Jr. et al., 1997, Yamaguchi et al., 2000, Bezer et al., 2005) is assumed to be related to increased deltoid activity (McCully et al., 2006). The deltoids generate an increased force, in order to compensate lost *ab*duction moment of e.g. the supraspinatus which results in an increased upward directed force component on the humeral head, resulting in a proximal migration and risk of compressing the subacromial tissues (Graichen et al., 1999).

We previously postulated that, in order to prevent proximal migration, patients co-activate their *ad*ductor muscles to pull down the humeral head during arm elevation (de Groot et al., 2006, Steenbrink et al., 2006). This coordinative change would restore glenohumeral stability at the cost of arm *ab*duction moment. The objective of this study was to determine, by means of model simulation, the mechanical effect of rotator cuff tears on both muscle force balance and glenohumeral stability. We hypothesize that rotator cuff tears will lead to an upward rotated joint reaction force vector piercing through the glenoid surface. The glenohumeral joint is considered unstable if the joint reaction force vector directs outside the glenoid rim. To redirect the joint reaction force vector through the glenoid surface and restore glenohumeral stability additional muscle forces are required.

4.2 Methods

Massive cuff tears were simulated using kinematic and force data from a previous experiment. Position data were obtained from 15 experimental patient recordings in which the injured arm was secured in a splint in a standardized position (de Groot et al., 2006, Steenbrink et al., 2006).

4.2.1 Simulation design

Inverse dynamic simulations were performed using the Delft Shoulder and Elbow Model (DSEM). Muscle forces in five combinations of simulated cuff tears were estimated and compared with muscle forces in the original condition. Each simulation was performed without and with a constraint for glenohumeral stability, respectively (see paragraph 4.2.3).

4.2.2 Delft Shoulder and Elbow Model

In the Delft Shoulder and Elbow Model (van der Helm, 1994) anatomical structures are represented by appropriate elements (van der Helm et al., 1992; Veeger et al., 1991). The model contains 31 muscles, divided in 139 muscle elements. Musculoskeletal parameters were obtained from extensive cadaver studies (Klein Breteler et al., 1999, Veeger et al., 1991). Input variables for the model are the orientations of the model elements (thorax, clavicle, scapula, humerus, radius and ulna) and direction and magnitude of the external arm load (applied at the olecranon of the humerus). The model calculates muscle forces required to satisfy mechanical force and moment equilibrium. The load sharing criterion J minimized the sum of squared muscle stresses, Eq. 4.1.

$$J = \sum_{i=1}^{n} \left(\frac{F_i}{\mathrm{PCSA}_i} \right)^2 \tag{4.1}$$

Where n is the number of muscle elements. Fi is muscle force produced by muscle element i. $PCSA_i$ is the physiological cross-sectional area of this muscle element.

Maximum muscle element force is depended on the physiological cross-sectional area, $PCSA_i$, the maximum muscle stress ($\sigma = 100Ncm^{-2}$, van der Helm, 1994), the fraction coefficient and a relative force-length function ($f(l_i, l_{fi}) : 0 \leq f(l_i, l_{fi}) \leq 1$), where l_{fi} is the actual element length, and $f(l_{fi})$ the optimal muscle length. If $f(l_i = l_{fi})$ then $f(l_i, l_{fi}) = 0$ (Klein-Breteler et al., 1999).

The fraction coefficient $c_i : 0 \leq c_i \leq 1$ is used to eliminate cuff muscles forces, Eq. 4.2.

$$0 \leq F_i \leq f(l_i, l_{fi}) \cdot \mathrm{PCSA}_i \cdot \sigma_{\max} \cdot c_i \tag{4.2}$$

Where F_i is force of muscle element i. $f(l_i, l_{fi})$ is the force-length function. PCSA_i is the physiological cross sectional area. σ_{max} is the maximum muscle stress ($= 100N \cdot cm^{-2}$) and c_i is the fraction coefficient of muscle element i, used to eliminate the cuff muscles. When a complete tear is simulated, $c_i = 0$.

4.2.3 The glenohumeral stability constraint

The model allows exclusion or inclusion of a glenohumeral stability constraint. The glenohumeral joint is considered stable when the resultant force vector is aimed within the glenoid surface. If this vector points outside the glenoid surface it cannot be fully counteracted by the joint reaction force vector and a dislocating force component results in glenohumeral instability. The glenohumeral stability constraint requires that the joint reaction force has a piercing point onto the glenoid surface at all times. In cases where this condition is not met, the model calculates the additionally required muscle forces to redirect the resultant vector onto the glenoid rim (van der Helm, 1994).

4.2.4 Model input

The average position for simulations was derived from patients with massive cuff tears (De Groot et al., 2006, Steenbrink et al., 2006). Because of inaccuracies in positioning and morphological variances the recorded average plane of elevation (Ry) was: 79° (SD 11°)), arm elevation (Rx) was: 46° (SD 10.7), and external rotation (Ry') was: 31° (SD 18.9°) with the elbow in 90° flexion (Fig. 4.1) according to the definitions of the International Society of Biomechanics for the shoulder in the local coordinate system of the thorax (Wu et al., 2005). The variances of observed arm positions in each of the three humeral angles σ_y^2 ($122^{\circ\cdot\circ}$), σ_x^2($114^{\circ\cdot\circ}$) and $\sigma_{y'}^2$($358^{\circ\cdot\circ}$) were used to estimate variance (or sensitivity) of the calculated muscle forces, σ_{fi}^2. Because the weight of the arm was counterbalanced in the experiments gravity working on the arm in the model was set to zero. An external force of 25 Newton (average patients' ability, Steenbrink et al. 2006) was applied to the olecranon and equaled a glenohumeral elevation moment of 7.3Nm.

4.2.5 Simulated cuff pathologies

Cuff tears were simulated by canceling force production of the "torn" muscle(s), by setting fraction coefficient c_i to zero (Eq. 4.2). In the common order of progressive rotator cuff tears the following cuff tears were simulated: 1. *supraspinatus*; 2. *supraspinatus and infraspinatus*; 3. *supraspinatus, infraspinatus and the teres minor*; it appeared that teres minor elimination did not result in a successful simulation for the constrained condition. Therefore the teres minor was preserved in further simulations; 4. *supraspinatus, infraspinatus and subscapularis*; 5. *supraspinatus, infraspinatus, subscapularis, and biceps longum*. Cuff tears were simulated without and with the constraint of glenohumeral stability.

4.2.6 Data analysis

Kinematic results of the simulations were presented as potential moment vectors (PMV), i.e. the moment resulting from a 1 Newton muscle force, and expressed in the global coordinate system (Veeger and van der Helm, 2007). The potential moment vector of each muscle was obtained by averaging the PMV's of the representing muscle elements (Fig. 4.2). Some muscle may have muscle elements with antagonist function, e.g. the subscapularis consists of 11 independent elements of which 9 elements have an *ab*duction PMV-component and 2 elements have an adducting PMV-component.

For the five simulations of rotator cuff tears, in combination with unconstrained and constrained glenohumeral stability, the magnitude of muscle forces, i.e. total force of representing muscle elements, that exerted a moment around the glenohumeral joint were determined and subsequently compared. The additional muscle activity required for glenohumeral joint stability was defined as the differences between forces estimated in the unconstrained and constrained conditions. Sensitivity of the calculated muscle forces, expressed by their variance, was estimated by the inner product of the variance in observed arm positions $[\sigma_y^2, \sigma_x^2, \sigma_{y'}^2]$ and the squared (numerical) partial derivatives of estimated muscle forces for each of three glenohumeral joint angles, Eq. 4.3.

$$\sigma_{F_i}^2 = \begin{bmatrix} \sigma_y^2 \\ \sigma_x^2 \\ \sigma_{y'}^2 \end{bmatrix}^T \begin{bmatrix} (\partial F_i/\partial y)^2 \\ (\partial F_i/\partial x)^2 \\ (\partial F_i/\partial y')^2 \end{bmatrix}^T \tag{4.3}$$

Where $\sigma_{F_i}^2$ is the variance in estimated muscle force for muscle i. $\sigma_y^2, \sigma_x^2, \sigma_{y'}^2$ are the observed variances in arm orientations Ry, Rx and Ry' (obtained from Steenbrink et al.,

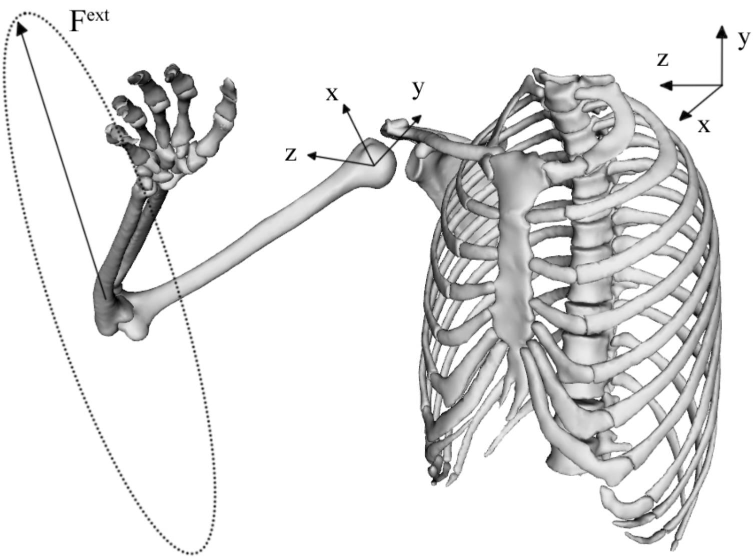

Figure 4.1: Representation of kinematic model input obtained from experimental set-up. An average arm position of 79° plane of elevation, 46° elevation and -31° axial rotation was used. An external force of 25 Newton (the average patient's ability) was applied to the olecranon, directed upward in the plane of elevation and perpendicular to the longitudinal axis of the humerus (y).

2006). $(\partial F_i/\partial y)^2, (\partial F_i/\partial x)^2, (\partial F_i/\partial y')^2$ are the partial derivatives of the muscle forces for glenohumeral orientations Ry, Rx and Ry'.

For the five simulations of rotator cuff tears, in combination with unconstrained and constrained glenohumeral stability, the magnitude of muscle forces, i.e. total force of representing muscle elements, that exerted a moment around the glenohumeral joint were determined and subsequently compared. The additional muscle activity required for glenohumeral joint stability was defined as the differences between forces estimated in the unconstrained and constrained conditions. Sensitivity of the calculated muscle forces, expressed by their variance, was estimated by the inner product of the variance in observed arm positions $[\sigma_y^2, \sigma_x^2, \sigma_{y'}^2]$ and the squared (numerical) partial derivatives of estimated muscle forces for each of three glenohumeral joint angles, Eq. 4.3. For every simulation the effort, quantified by the criterion value J, was compared with the effort in the normal condition.

Table 4.1: Function overview of muscles crossing the glenohumeral joint as measured in the simulated arm position (Figure 4.1).

Muscle	Abduction	Adduction	Retroflexion	Forward flexion	Endorotatie	Exorotatie
DE	+			+		+
SS	+		+			+
IS	+			+		+
SSc	+		+		+	
Tmi		+	+			+
BL	+			+	+	
LD		+	+		+	
PM		+	+		+	
TM		+	+		+	

4.3 Results

The *PMV*'s of muscles are constant for the simulated arm position (Fig. 4.2). The required moment vector of the external force around the glenohumeral joint is located outside the axes of the figure at [X = -20cm, Y = 11cm, Z = -17cm]. The deltoids, supraspinatus, infraspinatus, subscapularis and biceps longum include an abducting component and are primarily appropriate for the simulated *ab*duction/forward flexion task (Fig 4.3). The teres minor, pectoralis major, latissimus dorsi and teres major include an antagonistic *ad*duction moment which counteracts the force task (Table 4.1).

4.3.1 Supraspinatus tear

Unconstrained stability: Deltoid force and subscapularis *ab*ductor force increased 14% and 61% and the reaction force piercing point rotated in posterior-superior direction. The glenohumeral joint was stable. The predicted muscle forces were sensitive for arm position, as indicated by the standard deviation of the forces, but did not address other muscles than currently active (Fig.4.3A). The muscular effort, i.e. costs function J (Equation 4.1), increased 8% with respect to the normal condition (Fig. 4.4). For moment equilibrium also the endo-/exorotation moments of principal moment actuators/generators need to be compensated. The glenohumeral contact force intersects the glenoid surface, indicating that glenohumeral stability is preserved.

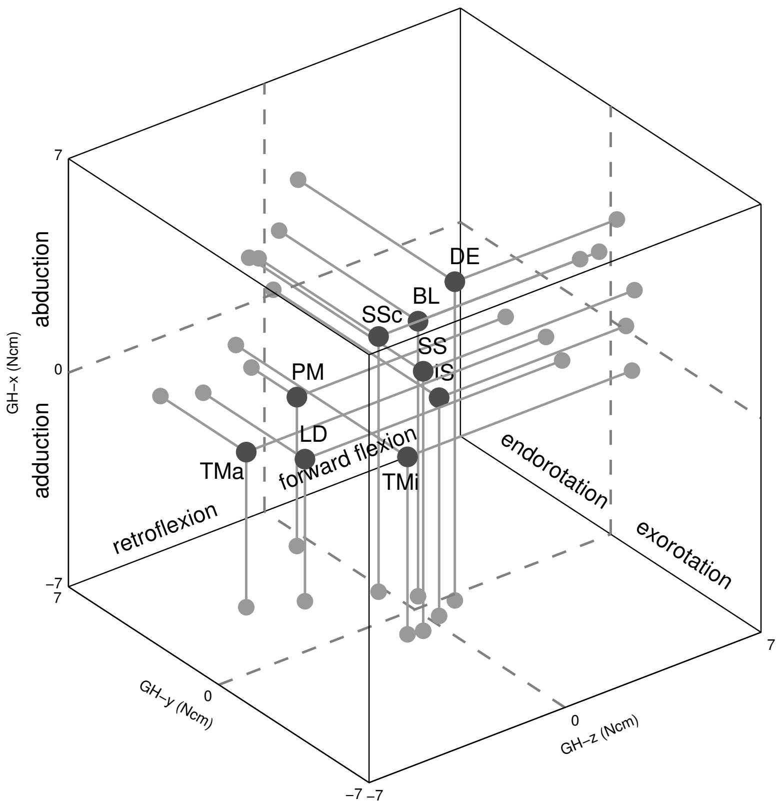

Figure 4.2: Potential Moment Vector plot, obtained by model simulation for the experimental arm position; the projections on the three axes of rotation indicate the muscles' potential contribution for the represented directions of movement. Muscles with potential contributions around the glenohumeral joint which were found to be active in our simulations are the deltoids (DE), supraspinatus (SS), infraspinatus (IS), subscapularis (SSc), teres minor (TMi), biceps longum (BL), pectoralis major (PM) and the latissimus dorsi (LD). The teres major (TMa) is presented for reference with patient observations (Steenbrink et al., 2006).

4.3.2 Supraspinatus and infraspinatus tear

Unconstrained stability: Deltoid force increased 35%; the subscapularis force decreased because its endorotation moment could not be compensated for by the infraspinatus (Fig.4.3B). A posterior-superior glenohumeral destabilizing force originated.

Constrained stability: Deltoid forces decreased, subscapularis force increased and substantial teres minor forces were required. The muscular effort, without and with stability constraint, increased 28% and 43% respectively (Fig.4.4).

4.3.3 Supraspinatus, infraspinatus and teres minor tear

Unconstrained stability: The teres minor was not active in combination with supraspinatus and infraspinatus and the model converged to the latter solution.

Constrained stability: The model did not converge to a solution. This indicates that glenohumeral integrity is not provided by the remaining muscles. The stabilizing action of the teres minor seems unique and cannot be compensated for. This tear conditions was not illustrated.

4.3.4 Supraspinatus, infraspinatus and subscapularis tear

Unconstrained stability: Deltoid and biceps longum forces increased and introduced posterior-superior glenohumeral instability (Fig.4.3C). Constrained stability: Further increase of biceps longum forces in combination with substantial teres minor forces and position sensitive introduction of latissimus dorsi and pectoralis major forces were required. The muscular effort without and with stability constraint increased 37% and 111% respectively (Fig. 4.4).

4.3.5 Supraspinatus, infraspinatus, subscapularis and biceps longum tear

Unconstrained stability: The most extended cuff tear resulted in the largest deltoid forces in combination with teres minor forces and introduced maximum posterior-superior glenohumeral instability (Fig.4.3D).

Constrained stability: Additional teres minor and deltoid forces in combination with pectoralis major and latissimus dorsi forces were required. The latter muscles generated a large adduction moment. The muscular effort in this simulation increased 46% for the unconstrained glenohumeral joint and 163% when glenohumeral stability was preserved (Fig. 4.4).

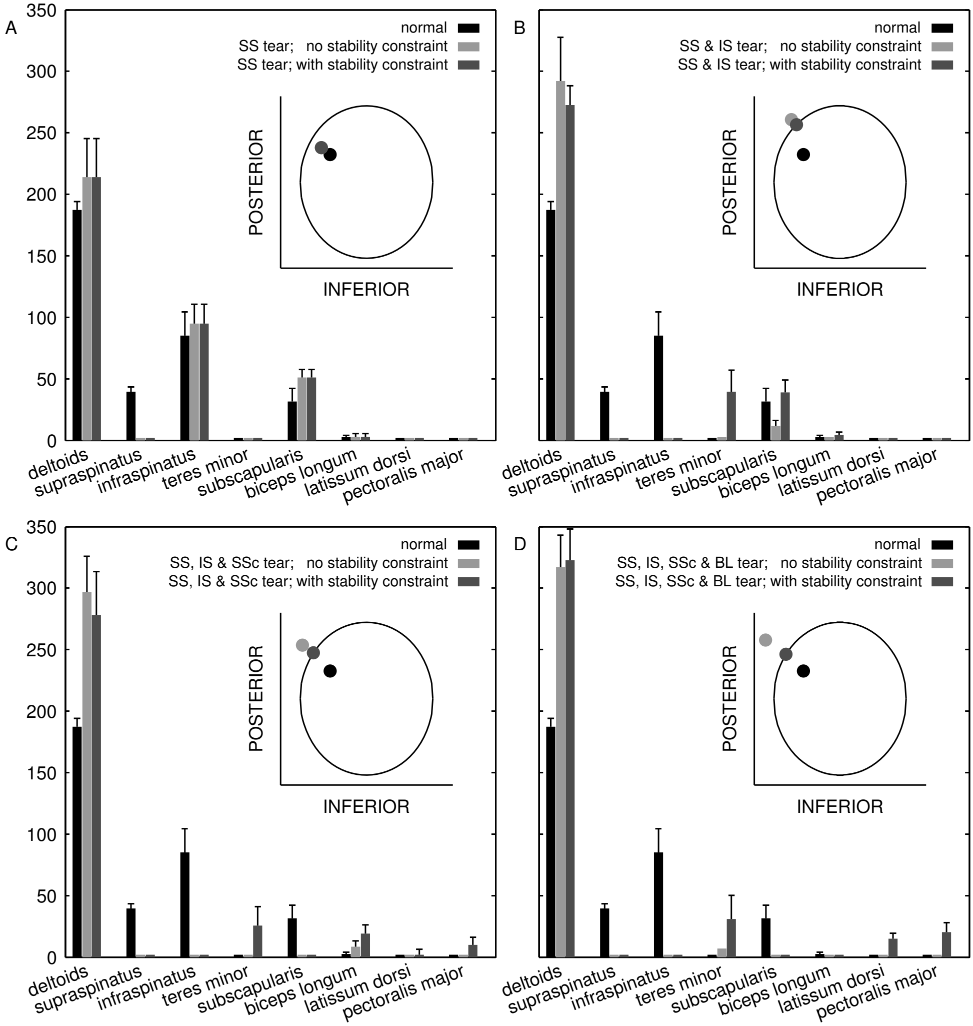

Figure 4.3: By the DSEM predicted muscle forces (Newton), and subsequent application point of the glenohumeral joint reaction force on the glenoid surface (inlay), as a result of the simulated conditions. Rotator cuff tears are respectively supraspinatus tear (A), supraspinatus and infraspinatus tear (B), supraspinatus, infraspinatus and subscapularis tear (C), supraspinatus, infraspinatus and subscapularis and biceps longum tear (D).

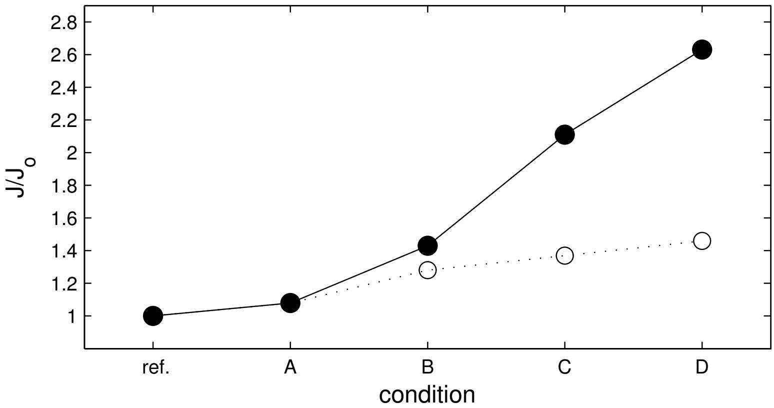

Figure 4.4: Relative increases of optimization criterion J (i.e muscular effort) for glenohumeral moment and additional glenohumeral stability with increasing tear. (For description of conditions see Fig. 4.3)

4.4 Discussion

The objective of this study was to analyze the muscular compensation for rotator cuff tears of varying magnitude and to identify additional muscle forces required for glenohumeral stability.

4.4.1 *Ab*duction compensation

Extending cuff tears result in increased deltoid muscle forces and confirms previous simulation (Magermans et al., 2004), cadaver (Apreleva et al., 2000, Hsu et al., 1997, Parsons et al., 2002) and experimental nerve blocking studies (McCully et al., 2007). The consequence of increased deltoid force is the posterior-superior shift of the reaction force vector piercing point. An isolated supraspinatus tearing does not necessarily result in an unstable glenohumeral joint, which may mechanically explain a-symptomatic cuff tears (Kelly et al., 2005, Yamaguchi et al., 2001).

In accordance with cadaver studies, the extent of a rotator cuff tear beyond the supraspinatus into the infraspinatus tendon induces glenohumeral instability (Apreleva et al., 2000, Hsu et al., 1997, Parsons et al., 2002) and may explain the relationship between fatty degeneration of the combined infraspinatus and teres minor and proximal migration in rheumatoid arthritis (van de Sande et al, 2007). We conclude that the deltoid muscle is an efficient *ab*ductor muscle. For the simulated arm position in the DSEM, the upper 9 elements of the subscapularis (11 in total) are abducting synergists. In case of minor tears, the subscapularis compensated supraspinatus losses in combination with the infraspinatus.

Biceps longum *ab*duction moment contribution on the condition of glenohumeral stability (Warner and McMahon, 1995) and specifically in massive cuff tears (Kido et al., 2000; Beall at al., 2003) was indeed observed in our simulation (condition C). From a mechanical point of view, tenotomy of the long head of the biceps, used to reduce symptoms of pain and inflammation in the follow up of patients with cuff tears (Boileau et al., 2007, Walch et al., 2005), induces increased co-contraction of muscles with large *ad*ductor components (pectoralis major/latissimus dorsi) and thus a serious additional muscular effort (Fig. 4.4D).

4.4.2 Glenohumeral stability

Deltoid forces efficiently substituted lost cuff *ab*duction moments at the cost of glenohumeral stability. This was evidently illustrated by deltoid reduction upon required glenohumeral stability in combination with additional abducting cuff muscle forces (subscapularis/biceps longum) and the consequent increase of the optimization criterion J, Fig. 4.4. Remarkably, teres minor co-contraction forces are vital for glenohumeral stabilization if the infraspinatus ceased function. Because of its relative small moment arm and vertically directed line of action the teres minor seems extremely useful to compensate the extra-glenoidal force component and stabilize the glenohumeral joint, with minimal interference with the intended elevation moment. Recent clinical observations also claim the importance of the teres minor for glenohumeral stability (Costouros et al., 2007, Simovitch et al., 2007).

If all *ab*ductor synergists were set to zero (condition 5, Fig. 4.3D), the deltoid muscle was the only muscle left to generate the required *ab*duction moment. Muscles with large *ad*ductor components (pectoralis major/latissimus dorsi) were required for glenohumeral stability. This 'expensive' co-contraction seems to be the only solution left to generate net *ab*duction moment. This is in line with publications by Hinterwimmer et al. (2003) and Graichen et al. (2005) and our own experimental findings (Steenbrink et al., 2006) where *ad*ductor activation of latissimus dorsi, pectoralis major and teres major was observed in patients with massive cuff injury. Experimentally observed teres major co-activation (de Groot et al., 2006; Steenbrink et al., 2006), was however absent in this simulation study. This may be the result of subject specific anthropometry on the observed combination of muscle activation. *ad*ductor muscle co-activation is a possible cause of observed limitation in maximal arm elevation in patients with cuff injury (Iannotti et al., 2006, Jost et al., 2000).

4.4.3 Limitations of this study

The outcome of this study is only valid for the specified position and only reflects mechanical considerations. Other somatic symptoms of cuff pathology, such as pain, were not included in this study. Massive cuff tears may result in kinematic changes of scapulo-thoracic and scapulo-humeral positions, as illustrated by a supraspinate nerve block experiment (McCully et al., 2006). The kinematic changes will affect the *PMV*'s of muscles and thus the force and moment balance. We partially overcame this shortcoming by approximation of the sensitivity of muscle forces for arm position. The shape of the glenoid, its relative position and the presence of a labrum (absent in the DSEM) may slightly affect the absolute magnitude of muscle forces presented but not the relative muscle forces.

4.4.4 Functional/clinical implications

Cuff injuries of the supraspinatus and infraspinatus required adduction forces of the teres minor whereas tears extending these muscles required forces by larger *ad*ductor muscles, i.e. pectoralis major and latissimus dorsi. On conditional teres minor preservation, patients with massive cuff injuries are theoretically able to generate *ab*duction forces with sufficient glenohumeral stability. This may explain a symptomatic rotator cuff tears (Kelly et al., 2005, Yamaguchi et al., 2001). Symptomatic rotator cuff tears with proximal migration are common and indicate that patients fail to fully compensate the lost stabilizing forces. The cause of this failure is unknown, but might involve unrecognized teres minor failure or disturbed proprioceptive or nociceptive sensory feedback, as e.g. subacromial pain suppression increased maximal arm force and arm mobility (Ben Yishay et al., 1994, de Groot et al. 2006) and restored activation patterns (Steenbrink et al., 2006). Simulation indicated the teres minor to be the solely indispensable cuff *ad*ductor in case of a complete infraspinatus deficiency. Post-hoc simulation of artificial (mathematical) elimination of teres minor moments around all three axes (Fig. 4.2) with maintenance of its force contribution resulted in a 121% increase of teres minor force. We concluded that teres minor is primarily required for glenohumeral stability and not humeral endorotation moment compensation.

Pathological *ad*ductor co-contraction during arm elevation load is the general mechanical finding of this study. This coincides with our experimental observations in patients with massive cuff tears (de Groot et al. 2006, Steenbrink et al. 2006) and can therefore be regarded as an indication for cuff disease. In order to understand subacromial pathologies the challenge is to develop an experiment which addresses the causal relation between muscle activity and glenohumeral (in)stability. Experimental research should focus on identifying the causal relations between compensating muscle activity by loading the arm with various moments and constant forces in patients with cuff tears.

4.5 Conclusion

An isolated tear of the supraspinatus does not necessarily lead to glenohumeral instability. For massive cuff tears beyond the supraspinatus, instability became a prominent factor. Moments efficient deltoids introduced a large destabilizing force component and alternative *ab*ductor muscles (i.e. subscapularis and biceps longum) required 'costly' co-contraction. The teres minor appeared to be of vital importance in glenohumeral stability because of its stabilizing force vector.