Compensatory muscle activation in patients with glenohumeral cuff tears

Steenbrink, F.

Chapter 3

Arm load magnitude affects selective shoulder muscle activation

Frans Steenbrink[1,2], Carel G.M. Meskers[1,3], Bart van Vliet[1,4],
Jorrit Slaman[1,4], DirkJan (H.E.J.) Veeger[4,5], Jurriaan H. de Groot[1,3]

[1] *Laboratory for Kinematics and Neuromechanics, Leiden University Medical Center*
[2] *Department of Orthopaedics, Leiden University Medical Center*
[3] *Department of Rehabilitation Medicine, Leiden University Medical Center*
[4] *Department of Human Movement Sciences, MOVE, Vrije Universiteit Amsterdam*
[5] *Department of Biomechanical Engineering, Delft University of Technology*

Medical and Biological Engineering and Computing 2009; 47: 491-500.
Awarded with Young Investigator Best Fundamental Research Paper at the International Shoulder Group meeting, Bologna, Italy, July 2008.

Abstract

For isometric tasks, shoulder muscle forces are assumed to scale linearly with the external arm load magnitude, i.e. muscle force ratios are constant. Inverse dynamic modeling generally predicts such linear scaling behavior, with a critical role for the arbitrary load sharing criteria, i.e. the “cost function”. We tested the linearity of the relation between external load magnitude exerted on the humerus and shoulder muscle activation.

Six isometric force levels ranging from 17% to 100% of maximal arm force were exerted in 24 directions in a plane perpendicular to the longitudinal axis of the humeres. The direction of maximum muscle activation (EMG), the experimentally observed so called *principal action* (*PA*), was determined for each force magnitude in twelve healthy subjects. This experiment was also simulated with the Delft Shoulder and Elbow Model (DSEM) using two cost functions: 1) minimizing muscle stress and 2) a compound, energy related cost function. *Principal Action*, both experimental (PA_{exp}) and simulated (PA_{sim}), was expected not to change with arm forces magnitudes.

PA_{exp} of the trapezius pars descendens, deltoideus pars medialis and teres major changed substantially as a function of external force magnitude, indicating external load dependency of shoulder muscle activation. In DSEM simulations, using the stress cost function, small non linearities in the muscle force-external load dependency were observed, originating from gravitational forces working on clavicular and scapular bone masses. More pronounced non-linearities were introduced by using the compound energy related cost function, but no similarity was observed between PA_{exp} and PA_{sim}.

3.1 Introduction

Individual muscle forces change with armload direction. This load direction dependency was used to study muscle coordination in healthy subjects (Arwert et al., 1997; de Groot et al., 2004; Flanders and Soechting, 1990; Laursen et al., 1998; Meskers et al., 2004) and subjects with shoulder pathologies (de Groot et al., 2006; Steenbrink et al., 2006). The *principal action* (*PA*), which comprehends load direction dependent electromyography (*EMG*) parameters (de Groot et al., 2004; Laursen et al., 1998), is used as a descriptive parameter for muscle coordination. In practice, repeated measurements are performed before and after an intervention, while maximum force around the shoulder may be altered by these intervention, e.g. by pain reduction or muscle tendon transfers (Steenbrink et al., 2006). In the comparison of these experiments we assume that muscle forces scale linearly with external force magnitude. External forces could differ considerably in pre-post measurements (de Groot et al., 2006; Steenbrink et al., 2006) and inter-individually (de Groot et al., 2004; Meskers et al., 2004). So linearity is a pre-requisite, or should be predictable if muscle contraction patterns are to be compared under these relatively different loading conditions. In the jaw, linear scaling of muscle activity (*EMG*) and external load was indeed demonstrated (Blanksma et al., 1992; van Eijden et al., 1993). Non-linear muscle activation scaling with external arm load was however reported in the upper extremity (Happee and van der Helm, 1995).

In shoulder inverse dynamic modeling linearity is generally assumed and incorporated in the load sharing criteria that are needed to mathematically solve the redundancy problem in order to reach a unique muscle activation pattern (de Groot, 1998; Dul et al., 1984; Happee and van der Helm, 1995; Happee, 1994; Tsirakos et al., 1997). Praagman et al. (2006) introduced an energy related criteria with linear and non-linear terms, weighted by morphological parameters as fiber length and muscle mass. This criteria turned out to fit best with non-linear in vivo obtained muscle energy expenditure around the elbow using Near Infrared Spectroscopy. They stated that most cost functions are chosen rather arbitrary, mainly due to the fact that validation is difficult since muscle force cannot be measured accurately in-vivo. The *EMG* based *principal action* method offers an alternative method to compare in vivo with simulated muscle activation, in order to interpret the experimental results and to predict possible load dependencies of shoulder muscles activation patterns in future studies (de Groot et al., 2004; de Groot, 1998).

In the present study we experimentally test the assumption that relative shoulder muscle forces do not change with armload magnitude. The experiment was numerical simulated, using the Delft Shoulder and Elbow Model (DSEM) with both a linear and an energy related cost function (Praagman et al., 2006; van der Helm, 1994). We used the *principal action*, i.e. the direction of maximum muscle activation assessed by either EMG (experiment) or force (simulation), resp. PA_{exp} and PA_{sim}, as a parameter for muscle coordination.

3.2 Methods

3.2.1 Subjects

Twelve healthy subjects (five female; three left handed) with a mean age of 26 (SD 2.9 years) took part in the study. The local medical ethical committee granted permission and all subjects gave informed consent.

3.2.2 Experimental set-up

Subjects were seated with the dominant arm in a splint with the elbow in 90° of flexion (Fig. 3.1). The setup allowed for static, isometric contractions of shoulder muscles while loading the arm with a force of different magnitudes in different directions in a plane perpendicular to the humerus (de Groot et al., 2004; de Groot, 1998; Meskers et al., 2004). The humeral plane of elevation was approximately 60° rotated externally from the para-sagittal plane and the humerus was 60° abducted. The forearm was 45° externally rotated relative to the horizontal (see Fig. 3.1). The objective of the setup was to record only forces perpendicular to the longitudinal axis of the humerus. In rest, the arm was fully supported by means of a weight and pulley system to compensate for all gravitational forces and moments (de Groot et al., 2004; Meskers et al., 2004). The arm splint was attached to a 6DOF force transducer (AMTI-300, Advanced Mechanical Technology, Inc., Watertown MA, USA) by means of a low friction ball and a socket joint. The transducer was mounted on a low friction rail in line with the humerus. This construction allowed for movement of the arm along 4 degrees of freedom (three rotations and a translation), while translation along the axes perpendicular to the humerus long arm were constrained. These forces controlled the position of a cursor on a computer screen placed in front of the subjects (de Groot et al., 2004; Meskers et al., 2004) (Fig. 3.1).

EMG activity of twelve shoulder muscles was recorded (Table 3.1), and off-line post processed (de Groot et al., 2004; Meskers et al., 2004). Nine shoulder muscles were recorded with the use of bipolar silver bar surface electrodes (DelSys, Bagnoli-16, Boston MA, USA,

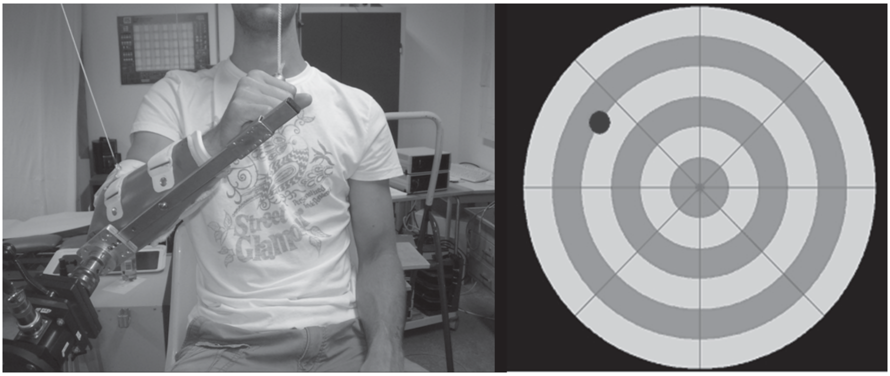

Figure 3.1: Experimental setup (left panel) and visual feedback (right panel); the subject had his arm in a splint, which is connected to a force transducers. Subjects are required to bring the arm force driven cursor (light grey small dot, centered in middle) into the target area (larger dark grey dot, upper left quadrant). The force, perpendicular to the longitudinal axis of the humerus, was recorded with a 6-dof force transducer (AMTI). The target indicated force direction (n=24) and force magnitude, i.e. radius (n=6), resulting in 144 combinations.

analog filter: 20Hz High pass, 450Hz Low pass, 10mm electrode length, inter-electrode distance of 10mm). Sample rate of analog filtered EMG and force data was 1000Hz. Before placement of the electrodes the skin was abraded, cleaned and a skin preparation gel (Skin Pure, Nihon Kohden) was used. The EMG of the three rotator-cuff muscles was recorded by means of bi-polar wire electrodes (Table 3.1). The wires were made of Teflon coated stainless steel with bare tips of 2mm length and were inserted with a sharp hollow needle. The electrode tips were bent in a sharp angle, so that after withdrawal of the needles, the wires would remain in situ. The wires for the subscapularis were inserted with a curved needle underneath the medial border of the scapula (Kadaba et al., 1992). Before insertion of the needles, the skin was anaesthetized with a 5% lidocaine solution. The needles for the subscapularis and infraspinatus were inserted until the scapular bone was touched.

3.2.3 Protocol

In the experimental set-up the force task existed of moving a cursor, driven by the forces exerted perpendicular to the longitudinal axis of the upper arm on the force transducer, into a target area (Fig. 3.1). Size of the target area was a predetermined area with a range of 3 times standard deviation (SD), determined from measurements on two subjects. Before

Table 3.1: Experimentally recorded shoulder muscles, localization of the electrodes and type of applied electrodes (similar to (de Groot et al., 2004, Meskers et al., 2004)) for comparison).

Muscle	Position	Electrode Type
Supraspinatus	2/3 line trigonum spinae-angulus acromialis 2 cm above spinal ridge	Wire
Infraspinatus	10cm below insertion site supraspinatus	Wire
Subscapularis	Halfway line angulus inferior trigonum spinae, underneath margo medialis	Wire
Trapezius (pars descendens)	2/3 on the line 7^{th} cervical vertebratrigonum spinae	Surface
Trapezius (pars ascendens)	Between the trigonum spinae and the eight thoracic dorsal spine, well above the caudal muscle ridge	Surface
Deltoid (pars anterior)	Middle of muscle belly, deltoideus anterior	Surface
Deltoid (pars medialis)	Middle of muscle belly, deltoideus medial	Surface
Deltoid (pars posterior)	Middle of muscle belly, deltoideus posterior	Surface
Serratus (anterior)	6^{th} head below angulus inferior scapulae	Surface
Teres major	Middle of muscle belly	Surface
Pectoralis major (pars clavicularis)	Middle of muscle belly, pectoralis major clavicular part	Surface
Latisimuss dorsi	6cm below angulus inferior scapulae	Surface

the experiment started the subjects maximum force target magnitude ($F_{\max}$) that could be maintained in all 24 directions was determined. Subsequently, six force levels were applied equidistantly, covering a range from 17% to 100% of $F_{\max}$. The force driven cursor was to be held within the target area for two seconds while the target randomly indicated 24 directions (angle) at 6 force magnitudes (radius), resulting in 144 combinations. Between the trials ample rest of at least five seconds was given in order to avoid too much fatigue effects. Subsequently the *principal action* at each force task could be determined off-line (de Groot et al., 2004; Meskers et al., 2004).

3.2.4 Data post-processing

EMG recordings were full-wave rectified and filtered for visual inspection (3rd order recursive Low Pass Butterworth at $10Hz$). The 2 seconds "in target" full-wave rectified *EMG* was averaged and rest level *EMG* was subtracted. For each of 6 force levels, the averaged rectified EMG was normalized with respect to the maximum EMG for the appropriate force level. Subsequently, a parameterized least squares curve was estimated through the 24 EMG values to obtain one direction of maximal EMG activity or *Principal Action* (PA_{exp}) for every muscle at force level (de Groot et al., 2004). Outliers and inaccurate estimations of the PA_{exp} were selected and removed by two investigators when consensus was achieved.

3.2.5 Statistical analysis

EMG data were collected for $n = 12$ subjects, $n_m = 12$ muscles, 24 force directions and $n_f = 6$ force levels. We tested the H0-hypothesis that muscle coordination did not change under different load magnitudes i.e. PA_{exp} of each muscle over the 6 force levels was constant. For each individual muscle a regression line, describing the *principal action* of that muscle as a function of force magnitude, was estimated. Subsequently the slope coefficient of this line (β) was tested not to differ from zero.

3.2.6 Model simulations

The experiment was simulated by inverse dynamic numeric modeling using the Delft Shoulder and Elbow Model (DSEM) (van der Helm, 1994). Kinematical input (arm position) was determined using 3D kinematical recording of one subject mounted in the experimental set-up using an electromagnetic tracking device (Meskers et al., 1998b). The ISG standardization protocol for the upper extremity including regression based GH-estimation (Meskers et al., 1998a; Wu et al., 2005) was used. A pointer was used to digitize 14 bony-landmarks with respect to sensors mounted on the thorax, the acromion (Karduna et al., 2001), the upper arm and the forearm. The subjects arm with the sensors attached was positioned into the splint and subsequently the position was recorded. All DSEM simulations were performed using this single position and an external force applied at the elbow in 24 directions at 6 force levels of the models $F_{\max}$, exactly simulating the experiment. In order to simulate the weight compensation on the arm in the experiments, gravity working on the humerus in the model was set to zero. By means of inverse dynamic simulation, muscle forces required to satisfy both the mechanical force-and moment equilibrium were calculated. Two different load sharing criteria were applied: a stress cost function, i.e. minimization of summed squared muscle stresses, and a compound linear and quadratic energy cost function (Praagman et al., 2006). Based on the estimated muscle forces the *Principal Actions* for the muscles in the DSEM were calculated (PA_{sim})(de Groot et al., 2004; de Groot, 1998).

Table 3.2: Average *Principal Action* PA_{exp} (SD) for 6 relative force levels and $n = 12$ subjects. Outliers were excluded resulting in different numbers of observations (N).

	Mean PA ± SD (deg)					
Muscle	**17 (%-Fmax)**	**33 (%-Fmax)**	**50 (%-Fmax)**	**67 (%-Fmax)**	**83 (%-Fmax)**	**100 (%-Fmax)**
Supraspinatus	35.03 (50.89) N=8	15.98 (32.11) N=9	35.91 (56.29) N=11	43.03 (56.00) N=9	42.14 (58.23) N=9	41.18 (50.35) N=10
Infraspinatus	6.12 (44.32) N=8	20.95 (24.74) N=9	17.81 (30.65) N=11	15.97 (25.75) N=11	20.56 (24.23) N=12	22.62 (28.02) N=12
Subscapularis	164.15 (71.34) N=8	147.63 (84.12) N=9	146.26 (76.61) N=10	152.36 (79.62) N=10	149.84 (75.11) N=11	154.10 (87.99) N=10
Trapezius descendens	16.05 (34.53) N=9	11.68 (35.73) N=11	22.62 (29.69) N=12	30.90 (32.77) N=12	36.00 (28.99) N=12	44.79 (26.82) N=12
Trapezius ascendens	93.70 (82.84) N=9	56.7 (54.82) N=11	79.76 (74.74) N=12	84.57 (55.73) N=12	65.51 (46.30) N=12	80.73 (69.66) N=12
Deltoid anterior	6.46 (49.70) N=8	-14.87 (7.76) N=10	-19.09 (12.93) N=12	-6.41 (16.20) N=12	-6.75 (18.27) N=12	-1.99 (25.04) N=12
Deltoid medialis	60.05 (23.95) N=10	62.93 (21.73) N=11	67.83 (22.44) N=12	68.71 (21.99) N=12	68.95 (18.89) N=12	73.02 (19.82) N=12
Deltoid posterior	92.52 (16.44) N=9	89.23 (14.24) N=10	91.17 (9.14) N=11	91.23 (9.80) N=11	91.82 (16.58) N=11	93.54 (11.97) N=11
Serratus anterior	300.52 (59.69) N=5	300.76 (49.82) N=9	306.48 (63.58) N=12	319.67 (68.83) N=12	316.23 (68.49) N=12	313.61 (61.88) N=12
Teres major	218.81 (54.20) N=8	203.97 (69.23) N=12	201.64 (66.61) N=12	175.08 (57.08) N=12	178.56 (56.70) N=12	172.62 (57.39) N=12
Pectoralis major clav.	265.81 (49.34) N=12	292.98 (26.85) N=11	277.37 (27.30) N=12	255.15 (67.93) N=12	253.37 (63.18) N=12	250.19 (66.34) N=12
Latissimus dorsi	158.71 (38.52) N=7	153.80 (18.34) N=10	151.68 (25.00) N=10	137.14 (18.38) N=11	155.69 (43.95) N=10	146.44 (22.45) N=9

Table 3.3: Linear regression slope parameters for the PA_{exp} to external load and their p values. Positive values represent a clock-wise shift of the PA_{exp}.

Muscle	**Linear component $\mathbf{PA_{exp}}$ (β)**	**P**
Supraspinatus	0.1995	.181
Infraspinatus	0.1362	.515
Subscapularis	0.1897	.322
Trapezius descendens	0.3857	.005*
Trapezius ascendens	-0.0283	.619
Deltoid anterior	0.1172	.156
Deltoid medialis	0.1436	.004*
Deltoid posterior	0.0222	.405
Derratus anterior	0.2143	.400
Teres major	-1.0804	.001*
Pectoralis major clav.	-0.3543	.230
Latissimus dorsi	-0.1204	.286

* Significant differences at $p < 0.05$.

3.3 Results

The average maximum force performed within the study population was 65 Newton (SD = 22.3). PA_{exp} for all muscles and loading conditions, as well as the number of observations after exclusion of outliers, is presented in Table 3.2. The trapezius descendens, deltoid medialis and teres major showed a significant shift of PA_{exp} as a function of external load. The maximum observed effect (teres major) of external loading on PA_{exp} was $-1.08°$ per % of $F_{\max}$. The PA_{exp} dependency was described by a linear regression model (Table 3.3).

In Figure 3.2 changes in *principal action* with respect to *principal action* at the first force level (*principal action* at 17% of $F_{\max} = 0°$) are presented. PA_{exp} are shown (circles), together with PA_{sim}, obtained using both a quadratic stress cost function (upward-pointing triangles) and a compound energy cost function (downward-pointing triangles). DSEM simulations with a quadratic stress cost function showed very small but noticeable non-linear scaling. In our model, we simulated gravity compensation of the humerus, but the observed non-linearities could still be introduced by gravity working on the clavicle and scapular bone, which was obviously not controlled for in the in vivo experiments. To make this effect more clearly visible, we performed model simulations including only one force direction, i.e. a force acting downwards on the arm, with two different magnitudes, i.e. $10N$ and $20N$. We subsequently compared estimated muscle forces in a model with gravity working on the clav-

Table 3.4: By DSEM simulations estimated muscle forces using the stress cost criteria, without (Fg-) and with (Fg+) taking mass of clavicula (0.156 kg) and scapula (0.705 kg)(van der Helm, 1994; Veeger et al., 1997) into account at a vertical downwards directed external load of 10 and 20 Newton respectively. Note that without gravity muscle forces scale linear (exact duplication of estimated muscle force with twice the external load), while non-linearities are introduced with gravity.

	Muscle forces (N)			
	Fg-		**Fg+**	
Muscle	**10N**	**20N**	**10N**	**20N**
Supraspinatus	1.72	3.44	1.70	3.32
Infraspinatus	23.81	47.62	24.69	48.66
Subscapularis	10.67	21.34	10.47	21.10
Trapezius descendens	8.06	16.12	13.21	21.48
Trapezius ascendens	2.71	5.42	3.41	6.04
Deltoid anterior	12.95	25.90	13.09	26.07
Deltoid posterior	0.87	1.74	2.08	3.17
Serratus anterior	15.53	31.06	19.42	34.80
Teres major	0	0	0	0
Pectoralis major clav.	10.28	20.56	11.93	22.50
Latissimus dorsi	0	0	0	0

icle and scapular bone masses, and a model without. Indeed, we found non-linear external load dependence introduced in the first model in contrast to the simulation with full gravity compensation (Table 3.4). The compound "energy cost function" appeared to result in a non-linear relation between PA_{sim} and external load, but except for the supraspinatus no similarity was observed between PA_{exp} and PA_{sim} (Fig. 3.2).

3.4 Discussion

Activation of three shoulder muscles appeared to be load dependent. This has consequences for the interpretation of muscle contraction patterns as measured in patients with shoulder disorders before and after intervention. In current shoulder model simulations (DSEM), non-linearities in the muscle force-external load relationships were not found using a quadratic stress cost function except when gravitational forces working on the clavicular and scapular bones were incorporated. More pronounced non-linearities were introduced using a compound energy related cost function, however not leading to a better resemblance of PA_{exp} to PA_{sim}.

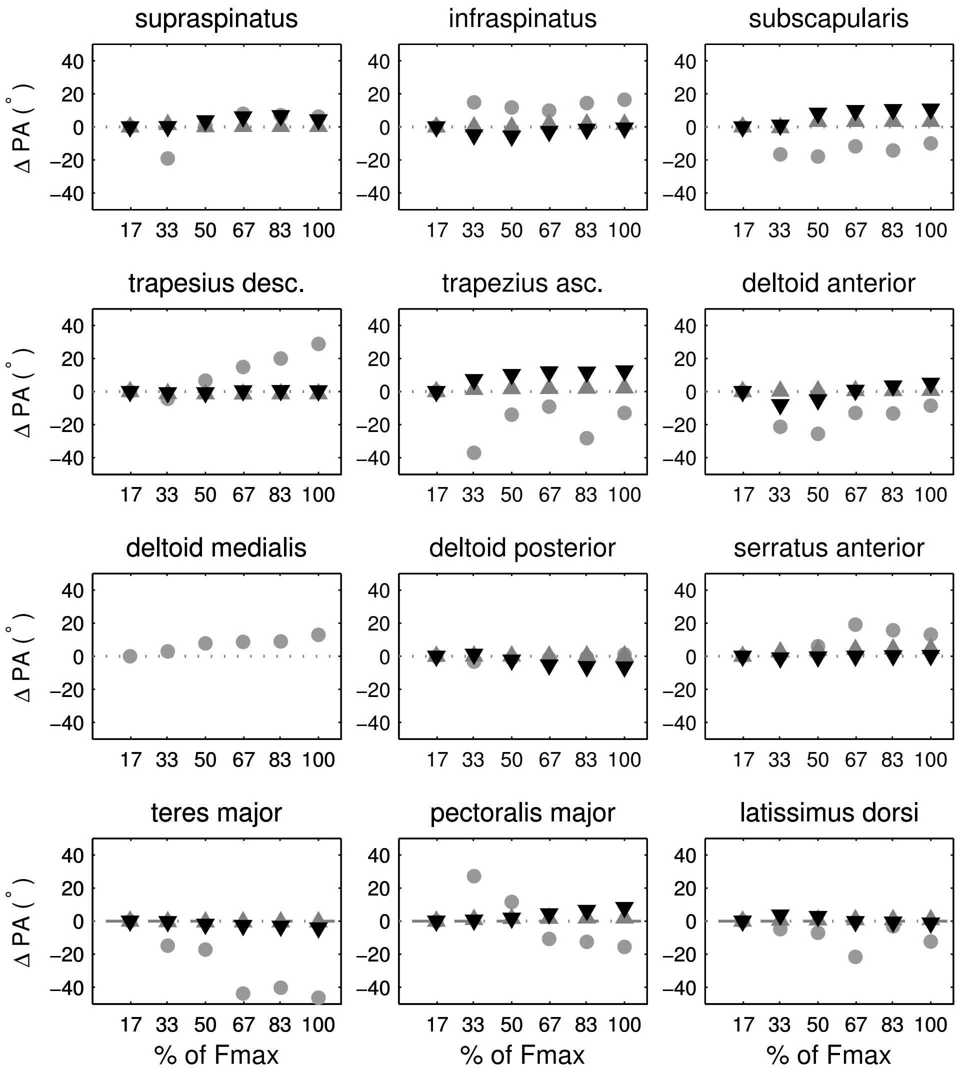

Figure 3.2: Changes in *principal action* with respect to the pricipal action at the first force level (*principal action* at 17% of $F_{\max} = 0°$); PA_{exp} (circles) and PA_{sim} with bone masses of the scapula and clavicle (stress cost function: upward-pointing triangles; energy cost function: downward-pointing triangles). PA_{exp} shows significant non linear relation to external loading for the trapezius descendens, deltoid anterior and teres major. PA_{sim} with the energy cost function and in lesser degree the stress cost function show a non-linear relation with external loading. PA_{sim} of deltoid medialis is lacking because the deltoids in the DSEM are divided in a clavicular part (presented with the deltoid anterior) and a scapular part (presented with the deltoid posterior).

3.4.1 Comparison with previous research

Only a few studies assessed load dependency of muscles in vivo. In a previous study by Meskers et al., external load dependency of shoulder muscle activation was found during a similar multi directional task using a similar *EMG* processing method (Meskers, 1998). In that study, clockwise shifts of deltoideus pars medialis (60°) and counter clockwise shifts of the serratus anterior (6°) and latissimus dorsi (20°) were found. However in contrast to the present study: 1) fixed force levels were used without normalizing, meaning that subjects were measured at different percentages of $F_{\max}$; 2) the external loads and force angles were not applied in randomized order, which might introduce muscle activation dependent recruitment bias and fatigue effects at the higher load tasks; 3) the positioning of the subjects in the present study was slightly different, i.e. the elevation angle was 15° lower.

Recruitment of muscles as a function of external load was studied on jaw muscles using a similar technique of relating *EMG* activity to increasing external forces (Blanksma et al., 1992; van Eijden et al., 1993). With increasing external forces, linear *EMG*-external force relationships where found for each jaw muscle (part). It was concluded that an increase in activity is achieved by the same, simultaneous increase in excitation activity. This would consequently imply a load independent *principal action* direction. Praagman et al. (2006) also reported linear scaling of muscle forces with external loading around the elbow by means of biomechanical model simulation using DSEM and muscle energy expenditure using Near Infrared Spectroscopy. Possible explanations of the discrepancy of the present study with previous work are that with 24 force directions in a full circle around the humerus, the resolution in the present study was considerable higher than in aforementioned studies.

3.4.2 Clinical consequences

In clinical settings, data are not acquired at different magnitudes of external force but at (near) maximum MVC (de Groot et al., 2006; Steenbrink et al., 2006). Thus influences of external loading, cross-talk and *principal action* estimation accuracy were presumed to be minimal. The maximum force a patient can exert will generally change as a result of therapeutic interventions. In patients it is therefore recommended to acquire *Principal Action* data at equal percentages of their $F_{\max}$.

The maximum effect of external loading on the *principal action* will not exceed 1.08° per percentage of MVC or Newton, resulting in 16° *principal action* shift for an external force change of 15N for the teres major. In pre-and post-intervention comparisons this is in the range of the inter-subject standard deviation and is substantially less than e.g. observed in patients with massive cuff tears where shifts for teres major increased 75° (de Groot et al., 2006; Steenbrink et al., 2006). These large *principal action* changes observed in patients cannot be explained by external force dependency but are obviously pathology dependent.

3.4.3 DSEM: load sharing criteria

The applied load-sharing functions either constrain or introduce non-linear scaling. The quadratic stress minimization allows synergy between agonist muscles more than linear criterions (Happee, 1994). The energy-related cost function with a linear and quadratic component was previously shown to lead to more realistic predictions of muscle activation (Praagman et al., 2006) for elbow-forearm external force tasks. Simulating the present experiment with the compound energy related criterion indeed predicted a non-linear external load-dependent muscle contraction, resulting in a better PA_{exp} to PA_{sim} resemblance for the supraspinatus and, at least for the contour also for the deltoid anterior. However, there was no resemblance for the remaining majority of muscles, implying that model simulations do not predict the observed effects in the experiment. In vivo we might apply alternative control strategies that are not caught adequately by the mechanical modeling and force distribution criteria. Additionally, force magnitude and direction induced changes of clavicle and scapula orientation may not be neglected, and should thus experimentally be controlled for, or incorporated in the simulations.

3.4.4 DSEM: gravitational loads

Introduction of gravitational forces resulted in non-linear muscle force-external load relations when the stress cost function was used, especially for the low loading conditions. Gravitation generates constant joint-torques that requires constant muscle force compensation. This baseline muscle loading interacts with the linear increasing external component, resulting in a non-linear appearance. Where bone masses will not be much of a factor, muscle masses probably will. Muscle masses and the application point of gravitational forces on the different muscle volumes are presently not adequately incorporated in the DSEM. Variations in the gravity forces - external load ratio could explain differences of the present findings with respect to the previous studies to some extent (Apreleva et al., 2000; Blanksma et al., 1992; Meskers, 1998; Praagman et al., 2006; van Eijden et al., 1993). It is recommended to take gravitational forces into account in model simulations, especially when the direction of the external force does not coincide with the direction of the vertical gravitational forces and the moment arms of external force directions are changing.

3.4.5 Possible error sources in the experiment

The validity of the *EMG* model as used in the present study is extensively discussed (de Groot et al., 2004; Meskers et al., 2004). When external force is increased, the signal over noise ratio will increase which will lead to optimal estimates of PA_{exp}. Therefore PA_{exp} estimations at low forces have reduced accuracy. However, it is unlikely that this phenomenon explains the present findings as shifts of *principal action* are not limited to the lower loading conditions.

Influence of cross-talk might also be external load dependent. However, the *principal action* is estimated at the peak of muscle activation and therefore the *principal action* method as such can be considered relatively insensitive to cross talk, even at the lower external loads.

During the experiments the gross position of the subjects was kept constant and special care was taken that subjects could not cheat to be able to meet the higher external forces. Small scapula positional changes could however not be ruled out and because external load direction dependent scapular positions were previously observed (de Groot et al., 2006), these changes are likely to increase with increasing external load magnitude influencing muscle moment arms around the acromioclavicular, sternoclavicular and glenohumeral joints, which affect the *principal action* direction. To what extent *principal actions* change as a function of scapular position changes requires further research.