Isometric shoulder muscle activation patterns for 3-D planar forces: A methodology for musculo-skeletal model validation

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Abstract

Objective. To present an isometric method for validation of a shoulder model simulation by means of experimentally obtained electromyography and addressing all muscles active around the shoulder joints.

Background. Analysis of muscle force distribution in the shoulder by means of electromyography during motion tasks is hampered by artificial and non-linear amplitude modulation and is often limited to downward directed external forces. This application of EMG is therefore inadequate and insufficient for the validation of shoulder model simulations. We suggest an isometric method including multi-directional forces to overcome these problems.

Methods. A force with constant magnitude is actively rotated stepwise in 20 directions perpendicular around the arm while kept in one position. The isometric muscle activation (EMG) is a function of the clockwise-rotated force angle, characterized by baseline activation, and a section of increased muscle activation characterized by baseline interceptions and direction and magnitude of maximum muscle activation. Comparison of the parameterised muscle activation with predicted muscle forces from model simulation illustrates the applicability for musculo-skeletal model validation.

Results. All recorded shoulder muscles were active over a section of force angles of at least $180^\circ$. Some muscles demonstrated two activation sections. The estimated model sensitivity for the baseline interceptions was $SD = 5^\circ-10^\circ$. The Principal Action was the most reliable parameter ($SD = 4^\circ$). A correlation of 0.778 was observed between model simulations and EMG recordings.

Conclusions. The methodology addresses all shoulder muscles over a substantial section of planar force directions. This enables the comparison of experimentally determined direction of activation on- and offset and direction of maximum activation with equivalent muscle forces, predicted from model simulation.

Relevance

The proposed method is a promising tool for both comparing EMG recordings and model simulations and can also be used as a stand-alone method for follow-up of clinical interventions.

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1. Introduction

In the understanding of shoulder problems, muscle force distribution in the shoulder is a prime theme, however difficult to analyse. In vivo recording of muscle forces (An et al., 1990) in the shoulder is not possible yet and muscle forces are generally predicted by means of musculo-skeletal modelling (Veeger et al., 1991; Hogfors et al., 1991; Karlsson and Peterson, 1992; Karlsson,
coincides optimally to the moment arm of the external moment because the moment arm of the muscle force maximally contributes to the desired maximum muscle activation is generated when the muscle activation is linearly related to muscle force magnitude (Flanders and Soechting, 1990; De Groot et al., 1992; Arwert et al., 1997; De Groot, 1998) and maximum muscle activation is generated when the muscle force and the principal action, and (3) when the task-force direction rotates 360°, a single muscle contributes to the external moment in a section of 180°, starting 90° counter-clockwise from the principal action and ending 90° clockwise from the principal action (Fig. 1). In the opposite section, the muscle is an antagonist of the external moment task and the activation will be minimal.

In theory, muscle activation is cosine-shaped and the pattern does not account for muscle co-contraction (Flanders and Soechting, 1990). Multiple cosine-shaped activation patterns are expected for muscles, active over more than one degree of freedom (1) within one joint, e.g. the m. deltoideus affecting 3-degrees of freedom of the glenohumeral joint or (2) multiple joints, e.g. the m. trapezius affecting both the sternoclavicular and the acromioclavicular joints. The interaction of the shoulder and arm muscles will superimpose a variety of possible activation patterns. We developed a description of muscle activation that, in contrast with the ideal case described above, can describe a flexible range of muscle activation with asymmetrical activation patterns, based on a limited number of only five parameters.

We assume that for isometric and isotonic tasks, EMG represents muscle activation and we assume that both muscle force and muscle activation are linearly proportional for these tasks. The five parameters estimated from EMG recordings and the equivalent parameters derived from inverse dynamic model simulation, allow for comparison of model predicted muscle forces and experimentally determined muscle activation.

The EMG, in contrast with the formalized inverse dynamic model simulation, is a noisy signal. The accuracy of the estimated parameters depends on the noise in the signal and therefore we determined the sensitivity of the outcome parameters. The application of isometrical EMG recording of varying force directions is a promising method for clinical evaluation and follow-up of shoulder muscle function, e.g. after surgical intervention. The intra- and inter-individual variability of the parameters is not the subject of this paper, and is discussed elsewhere (Meskers, 1998; Meskers et al., 2004).

2. Methods

2.1. Data acquisition

The subjects were standing with the head and hips fixed in an external frame. The subject’s right arm was
elevated to 90° in the sagittal plane and the elbow flexed to 90° with the lower arm in the plane of elevation (Fig. 1a). A cuff that fitted on the upper arm and forearm was connected to a 6-degrees of freedom force transducer (AMTI-300, Advanced Mechanical Technology, Inc., Wavertown MA, USA) by means of a low friction ball and socket joint (De Groot and Brand, 2001). The transducer was mounted on a rail in line with the humerus and the subject could only exert a force perpendicular to the humerus (Fig. 1a). Forces parallel to the humerus, i.e. push and pull, and moments around the three (perpendicular) humeral axes would result in movements of the arm. The subject generated a force of constant magnitude in 20 directions at equidistant angles of 18° in a single plane perpendicular to the (right) humerus. F(i) represents a force generating an abduction moment around the horizontal frontal body axis and a lateral moment around the vertical frontal body axis. For each force direction, this results in the grey dots. The maximum activation is found at an external force of PA = 0°; the clockwise initiation of muscle activation is located at I₁ = 270°; the clockwise termination of muscle activation is located at I₂ = 90°. The muscle is only activated in the upper sections of the external force, generating an abduction moment. (d) The theoretical muscle activation is a cosine-shaped function of the angle \( \alpha \) of force direction with a peak activation \( A_{p_0} \) at the optimal force direction PA, and a threshold activation \( A_t \) allowing only positive activation. The intercepts of the activation peak with the activation threshold characterize the clockwise initiation \( I_1 \) and termination \( I_2 \) of muscle activation.

Bi-polar surface EMG (Ag–AgCl Medi-Trace® pellet electrodes) of 12 shoulder muscle(-part)s (Table 1) was recorded during 3 s of sustained force in each of the 20 force directions. The EMG was on-line band pass filtered [third order Butterworth: 25–200 Hz], and sampled at 500 Hz and subsequently off-line processed.

The positions of the humerus, the clavicle, the scapula and the thorax, needed for inverse model simulation with the Dutch Shoulder and Elbow Model (DSEM) (Van der Helm et al., 1992; Van der Helm, 1994a, Veeger et al., 1991) were recorded by means of palpation and subsequent digitization (De Groot, 1997; De Groot and Brand, 2001).

2.2. Off-line signal processing

(For symbols and definitions, see also Table 2). The 3 s EMG recording was off-line rectified and averaged for each force direction. The angular trace of 20 average EMG, (Fig. 1d) was normalized between the minimal and maximal recorded EMG within the trace:
The electrodes were positioned with the arm in a 90° anteflexion.

Table 1

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Abbr.</th>
<th>Position</th>
</tr>
</thead>
<tbody>
<tr>
<td>m. trapezius (pars descendens)</td>
<td>TD</td>
<td>Cranial from the trigonum spineae, 2 cm below the muscle ridge</td>
</tr>
<tr>
<td>m. trapezius (pars transversus)</td>
<td>TT</td>
<td>Between the spine and the medial ridge of the scapula, about 4 cm cranial from the trigonum spineae</td>
</tr>
<tr>
<td>m. trapezius (pars ascendens)</td>
<td>TA</td>
<td>Between the trigonum spineae and the eighth thoracic dorsal spine, well above the caudal muscle ridge</td>
</tr>
<tr>
<td>m. deltoideus (pars anterior)</td>
<td>DA</td>
<td>Middle of the muscle belly</td>
</tr>
<tr>
<td>m. deltoideus (pars medialis)</td>
<td>DM</td>
<td>Middle of the muscle belly</td>
</tr>
<tr>
<td>m. deltoideus (pars posterior)</td>
<td>DP</td>
<td>Middle of the muscle belly</td>
</tr>
<tr>
<td>m. infraspinatus</td>
<td>IS</td>
<td>Between the scapular spine and the angulus inferior</td>
</tr>
<tr>
<td>m. serratus anterior</td>
<td>SA</td>
<td>Sixth head below the frontal axillary fold</td>
</tr>
<tr>
<td>m. latissimus dorsi</td>
<td>LD</td>
<td>About 6 cm below the angulus inferior</td>
</tr>
<tr>
<td>m. pectoralis major (pars clavicularis)</td>
<td>PC</td>
<td>Middle of the muscle belly of the clavicular part</td>
</tr>
<tr>
<td>m. pectoralis major (pars sternalis)</td>
<td>PS</td>
<td>Middle of the muscle belly of the sternal part</td>
</tr>
<tr>
<td>m. teres major</td>
<td>TM</td>
<td>Middle of the muscle belly</td>
</tr>
</tbody>
</table>

The estimated muscle activation, \( A(\varphi) \), was described with the vertical upward direction \( \varphi = 0 \). The active portion was identified by the intercept force angles \( I_1 \) and \( I_2 \). The shape of the activity amplitude of maximum activity \( A_{pa} \) and the amplitude of maximum activity \( A_{pa} \). The width of the active portion was variable; the shape of the activity peak allowed for asymmetry and the amplitude of the estimated EMG was not to become smaller than \( A_0 \).

\[
\text{aEMG}_i = \frac{(\text{EMG}_i - \min(\text{EMG}_i))}{\max(\text{EMG}_i)} \quad \text{for } i = 1, 2, \ldots , 20
\]

2.3. Parameter estimation

The estimated muscle activation, \( A(\varphi) \) was described by means of the five parameters (see Fig. 1) i.e. \( A_0 \): baseline muscle activation; \( A_{pa} \): maximum muscle activation; PA: angle of maximum muscle activation; \( I_1 \): clockwise onset of muscle activation; \( I_2 \): clockwise offset of muscle activation.

One section of aEMG-muscle force directions represents the activated muscle; the remaining section represents the muscle activation rest level. The inactive portion was identified by the intercept force angles \( I_1 \) and \( I_2 \) and the baseline muscle activation \( A_0 \). The active section was identified by the intercept force angles \( I_1 \) and \( I_2 \), the force angle of maximum activity PA and the amplitude of maximum activity \( A_{pa} \). The width of the active portion was variable; the shape of the activity peak allowed for asymmetry and the amplitude of the estimated EMG was not to become smaller than \( A_0 \).

2.4. The algorithm

The estimated EMG amplitude \( A(\varphi) \) is the amplitude of baseline activity \( [A_0] \) and the superimposed square \( [A_2] \) and cubic \( [A_3] \) contributions in the aEMG activity peak.

For \( \varphi \leq I_1 \) and \( \varphi \geq I_2 \), \( A(\varphi) \) is equal to \( A_0 \):

\[
A(\varphi) = A_0 \quad \text{for } \varphi \leq I_1 \quad \text{and} \quad \varphi \geq I_2
\]

The \( A(\varphi) \) interval between the intercepts \( I_1 \) and \( I_2 \) is transferred to a normalized interval \( x \in [-1, 1] \) where \( x(I_1) = -1 \) and \( x(I_2) = 1 \):
The estimated EMG amplitude is calculated for the new domain \([-1, 1]\) and is the summation of the baseline activity \(A_0\), a square component \(A_2\) describing the activity peak and a cubic component \(A_3\) introducing the asymmetry of the estimated EMG:
\[
A(x) = A_0 + A_2(x) + A_3(x) \quad \text{for } x \in [-1, 1]
\]  
(4)

The square component \(A_2(x)\) is a predefined concave function of the parameter \(k_2\). The function results in positive amplitudes if \(k_2\) is constrained to positive values only. For \(x = -1\) and \(x = 1\) the function equals 0, for \(x = 0\) the function reaches its maximum amplitude (Eq. (5)).
\[
A_2(x) = k_2(1 - x^2) \quad \text{for } x \in [-1, 1] \quad \text{and} \quad k_2 \geq 0
\]  
(5)

The cubic component \(A_3(x)\) introduces an asymmetrical amplitude component by means of parameter \(k_3\). When positive, \(k_3\) results in the clockwise shift of the principal action, when negative, \(k_3\) results in the anti-clockwise shift of the principal action (Eq. (6)).
\[
A_3(x) = k_3(x - x^3) \quad \text{for } x \in [-1, 1] \quad \text{and} \quad |k_3| \leq k_2
\]  
(6)

The estimated EMG peak amplitude \(A_2(x) + A_3(x)\) should always be constrained to positive values and \(A_3(x)\) should always be smaller than the amplitude of \(A_2(x)\) in order to prevent \(A_2(x) + A_3(x)\) from becoming negative. Therefore, the absolute value of \(k_3\) should not exceed \(k_2\).

For computational reasons the constraints are defined by means of continuous functions: \(k_2\) is a positive exponential function of parameter \(k_2\) (Eq. (7)); \(k_3\) is a sinusoidal function of parameters \(k_2\) and \(k_3\) (Eq. (8)). For any arbitrary value of \(k_2\) and \(k_3\), the required constraints are fulfilled.
\[
k_2 = e^{k_2}
\]  
(7)
\[
k_3 = k_3 \sin(k_3)
\]  
(8)

Summarized, the shape of the estimated EMG is described by Eq. (2) and the supplementary equation (Eq. (9)), based on five parameters, \(A_0, I_1, I_2, k_2\) and \(k_3\).
\[
A(\phi) = A_0 + e^{k_2}(1 - x^2) + e^{k_3} \sin(k_3)(x - x^3)
\]  
(9)

The best fit of the estimated EMG amplitude \(A(\phi)\) on the recorded aEMG observations was found by means of the Levenberg–Marquardt least squares optimisation algorithm (Ljung, 1987), minimising the object function \(C\):
\[
C = \sqrt{\sum_{i=1}^{n} (\text{aEMG}(\phi_i) - A(\phi_i))^2} \quad \text{for } n = 20
\]  
(10)

The two shape factors of the EMG activity, \(k_2\) and \(k_3\), are difficult to interpret and therefore transformed to the corresponding principal action \(PA\) and corresponding amplitude \(A_{pa}\). The normalized angle of the maximum estimated EMG, \(x_{pa}\) was obtained from the derivative of \(A(\phi)\) (Eq. 9):
\[
-2k_2x_{pa} + k_3(1 - 3x_{pa}^2) = 0
\]  
(11)

The principal action \(PA\) of the muscle and its corresponding amplitude were obtained by means of substitution of \(x_{pa}\) into Eqs. (3) and (4) respectively. The optimally estimated curve \(A(\phi)\) from the recorded aEMG data is now described by the two amplitude parameters \(A_0\) and \(A_{pa}\), and the three directional parameters \(I_1, PA\) and \(I_2\).

2.5. Model simulation

In this paper we will illustrate the isometrical comparison between the estimated muscle activation obtained from a musculo-skeletal model of the shoulder (Van der Helm, 1994a) and the recorded shoulder muscle activations. The positions of the humerus, clavicle, scapula and thorax, needed for the kinematic input of the shoulder model, were obtained from palpation and subsequent digitization of bony landmarks of the specific bones (De Groot, 1997; De Groot and Brand, 2001). The external force was applied at the elbow. The five parameters described above were estimated for both recorded muscle activation and simulated muscle forces. The comparison between recorded and simulated muscle activation will be presented as an illustration of the application of the method for model validation.

3. Results

The aEMG of \(m.\ trapezius\ pars\ descendens\) is used to illustrate the estimation of the muscle activity as a function of force direction, \(A(\phi)\) (Fig. 2). The clockwise onset of \(A(\phi)\) is estimated at \(I_1 = 293^\circ\) (or \(-67^\circ\)) and the offset is estimated at \(I_2 = 188^\circ\). The (grey) section of muscle activation is thus \(188^\circ - (-67^\circ) = 255^\circ\) wide. The muscle is not activated for the remaining section of \(105^\circ\), with a base-line activation level of \(A_0 = 0.07\) (Fig. 2f). The square contribution \((k_2 = 0.64)\) defines the symmetrical peak activation between the two intercepts (Fig. 2e); asymmetry is introduced by the cubic contribution \((k_3 = -0.64, \text{Fig. 2d})\). The negative cubic factor results in a counter-clockwise shift of the square peak activation of \(42.4^\circ\), from \(60.5^\circ\) to the resultant \(PA = 18^\circ\) (Fig. 2e).

The estimated muscle activation \(A(\phi)\), i.e. \(A_0 + A_2 + A_3\), and the recorded aEMG signals are presented in Fig. 2a, (polar mode) and Fig. 2b, (linear mode). The observed fit between the recorded data and estimated curve was good. The linear mode facilitates visual inspection of the curve fitting and the visual determination...
A principal action (PA) of the m. trapezius pars descendens major and dom controlled by the specific muscle. Both of muscle activity and the number of degrees of freedom. There seems to be no relation between the range which is extreme, but in concordance with the recorded between the muscles differs about 180° and glenohumeral joints, while the range of activity same joints, i.e. the sternoclavicular, acromioclavicular.

Fig. 2. The recorded aEMG and the estimated $A(\theta)$ for m. trapezius pars descendens and the individual estimated components. (a) The polar angle representation of the recorded aEMG (dots) and the estimated activation curve $A(\theta)$ coincides with the force display (Fig. 1 a and b). The estimated principal action (PA) of the m. trapezius pars descendens is directed upward and a little lateral. The section of muscle activation (gray), ranges from clockwise initiation $I_1 = 293°$ to clockwise termination $I_2 = 188°$ and almost covers the upper and lateral quarters of force directions. (b) In the linear representation the force direction of muscle on- and offset, and the direction of maximum activation (PA) can easily be identified (in contrast to panel a). (c) The estimated muscle activation $A(\theta)$ on a shifted angle base from $-90°$ (i.e. equal to 270°) to 270° composed of the individual (d) the cubic component $A_3(\theta)$: $k_3 = -0.64$, (e) the square component $A_2(\theta)$: $k_2 = 0.64$ and (f) the rest or baseline component $A_0(\theta) = 0.069$.

of activity on- and offset and direction of maximum activity. The linear mode is therefore easy to interpret when comparing different recordings, e.g. for intra- and inter-individual comparison. The polar mode is a task related presentation of the direction of activity. The origin (centre) of the plot coincides with the longitudinal axis of the humerus pointing into the paper. The 0° force direction is an abduction force in the sagittal plane directed from the elbow towards the hand. For the right arm the m. trapezius pars descendens is active in three quadrants of the external force with a maximum activity in the upper right quadrant, i.e. a lateral and upwards directed arm force.

Fig. 3 shows the recorded aEMG and the estimated $A(\theta)$ for all 12 EMG-recorded muscle(-parts). The estimated range of activity and the direction of principal action generally coincide with the recorded data. All estimated activity ranges are equal (m. deltoideus pars posterior, m. pectoralis pars clavicularis and m. pectoralis pars sternalis) or greater than 180°. The estimated range of activity for m. latissimus dorsi is 234°, which is extreme, but in concordance with the recorded data. There seems to be no relation between the range of muscle activity and the number of degrees of freedom controlled by the specific muscle. Both m. pectoralis major and m. latissimus dorsi are active over the same joints, i.e. the sternoclavicular, acromioclavicular and glenohumeral joints, while the range of activity between the muscles differs about 180°. The curve fitting method seems applicable for all recorded shoulder muscles.

The accuracy of each of the estimated parameters was determined by a method described by Jenkins and Watts (1968). The variance of each of the parameters was calculated by multiplication of the squared partial derivatives of the specific parameter and the variance of the aEMG. The variance of the recorded EMG data, $\sigma_{aEMG}^2$, is determined from the residual error, i.e. the difference between the estimated $A(\theta)$ and the aEMG, and assumed normally distributed. The partial derivative for each parameter, e.g. $\frac{\partial (PA)}{\partial (aEMG_i)}$, for the principal action PA, was numerically approximated. E.g. Eq. (12) was applied to determine the variance of the PA. The result of this numerical approximation and averaged for all 12 muscles, is summarized in Table 3.

$$\sigma_{PA}^2 = \sum_{i=1}^{n} \left( \frac{\partial (PA)}{\partial (aEMG_i)} \right)^2 \sigma_{aEMG}^2$$ for $n = 209$

The average standard deviation of the principal action, $\sigma_{pa}$, was less than 4° and about equal for each of the recorded muscles. The standard deviation for the intercepts, $\sigma_{I_1}$ and $\sigma_{I_2}$, was 5° and 10° respectively, and showed a higher variation between the different muscles. The accuracy of the estimation seems to be related to the slope of the muscle activity peak. The clockwise activation offset $I_2$ was generally steeper and could therefore
be estimated more accurate. The minimum and maximum activation levels, $A_0$ and $A_{pa}$, were well estimated, but as the activation amplitude is normalized, these parameters do not contain relevant information.

The goal of the presented method is the application for comparing experimentally determined muscle activation with simulated muscle force (i.e. equivalent to muscle activation for isometric and isotonic tasks). An example of comparison between recorded muscle activation (parameters: $I_1 = 13^\circ$; $I_2 = 205^\circ$; $PA = 108^\circ$; $A_0 = 0.02$; $A_{pa} = 0.83$) and simulated muscle activation (parameters: $I_1 = 6^\circ$; $I_2 = 200^\circ$; $PA = 131^\circ$; $A_0 = 0$; $A_{pa} = 1$) for *m. trapezius pars transversus* is shown (Fig. 4). The recorded and simulated muscle activation on- and offset parameters for *m. trapezius pars medialis* coincide within the variance of the intercepts; the principal action is clockwise rotated in the musculo-skeletal model of the shoulder.

The overall congruence of experimentally determined muscle activation parameters and simulated muscle activation parameters is illustrated in Fig. 4c. Most of the parameters, indicated by white dots (intercepts) and black dots (principal action) are found within $45^\circ$ of the optimal fit (grey diagonal field). A significant Pearsons
correlation coefficient of 0.778 ($p < 0.01$) between recorded and estimated muscle activity was determined. Only five parameters differ more than 45\(^\circ\) between recorded and simulated muscle activation. Three of these parameters belong to the m. infraspinatus. This muscle shows about 160\(^\circ\) anti-clockwise rotation from recorded to simulated activation parameters.

4. Discussion

The objective of this paper was to present a method of parameterisation of muscle activation that could be applied for the validation of muscle force predictions by inverse dynamic model simulation of the shoulder. The estimated muscle activation was to be described by a minimal number of parameters. By means of the applied experimental methodology, the modulation of EMG due to electrode displacement, muscle length changes, activation and contraction dynamics, fatigue was to be minimal and all shoulder muscles should be addressed. We will first discuss each of these objectives. Secondly we will discuss the functional results from our EMG recordings and at last we will comment on the application of model validation.

4.1. Methodology of principal action estimation

From the results presented here, we have shown that each of the selected muscles was activated during 50\% or more of the force directions. Although we did not apply a full load of humerus by adding an axial torque we showed, as expected, that all muscles within the kinematic chain of the thorax, clavicle, scapula and humerus were loaded somewhere within the task. We illustrate by this ‘unnatural loading’, that we can address the function of all muscles by end-point force generation (at the elbow) in absence of an external axial moment generation e.g. the rotator cuff muscles.

In contrast to EMG recording while moving the arms in vertical (abduction) planes (e.g. Van der Helm, 1994b), the muscle length and electrode positions in the applied method above did not change due to the isometric character of the task. Short periods of EMG sampling (3 s), alternating force directions and applying the sub-maximal force magnitude minimised the effects of fatigue. During the 3 s of sampling the force magnitude and force direction were almost constant, assuring the absence of muscle activation and contraction dynamics. Although the task was isometric we could discriminate muscle activation on- and offset by changing force direction. We therefore believe that we applied a methodology in which muscle function can be analysed by recording EMG (read muscle activation) which is free of signal modification due to motion and activation non-linearitys, as was earlier recognized by Flanders and Soechting (1990).

Five parameters were required to describe the estimated muscle activation $A(\phi)$. These parameters were estimated from 20 aEMG samples resulting in a fourfold data reduction. Three relevant parameters, i.e. the principal action and the clockwise on- and offset of the
aEMG, were accurately estimated for all selected shoulder muscles with an average standard deviation of 4°, 10° and 5° respectively. The signal-over-noise ratio at maximum activation, i.e. the principal action, is highest and results in an estimate with the least variation. The twofold difference between the clockwise on- and offset of the aEMG is likely to originate from the fact that the falling (negative) slope of activation \( A(\phi) \) was generally steeper than the rising (positive) slope of activation (Fig. 3), resulting in a higher signal-over-noise ratio in the terminal steep part of the aEMG recordings relative to the initial part of muscle activation. Because the peak activity comprehends 180 or more degrees, the number of 20 samples of aEMG seems to be sufficient for a reliably estimation of the directional parameters.

4.2. Method in relation to muscle function

The external torque changes sinusoidal and symmetrically around a single joint axis (Fig. 1d) and muscles were expected to contribute for a section of only 180° of the externally exerted forces. In absence of interaction between the shoulder muscles, the peak activity of the individual shoulder muscles was expected to be symmetrical, being solely dependent on their moment arms and the required external moment. The external moment however, is the result of the weighted contribution of all muscles with positive (and negative) moment arms. This is e.g. illustrated by the co-activity of the descending and ascending parts of the \( m. \ trapezius \), having different lines of action but active in the same range of force angles, and the fact that most muscles are activated over a range of more than 180° (Fig. 3).

The musculo-skeletal system is a three-dimensional structure with a redundant number of (multi-articual) muscles. The action of the muscles is interactive, generating positive primary torques with respect to the external force but also generating (secondary) torques around other joint axes (degrees of freedom), which must be compensated for by antagonistic muscles (De Groot, 1998). These interactions and compensating forces resulted in the asymmetry of EMG activity (Figs. 2 and 3).

Flanders and Soechting (1990) treated the asymmetry, which seems to be a significant feature of the shoulder muscle activation, in a different way. They estimated one or more symmetrical activity peaks using a cosine function, assuming the symmetry of moment arms increase and decrease described above. The disadvantage of their method is the increase of the number of parameters with each added peak of activity. Theoretically, a muscle activity peak exists for each of the degrees of freedom affected by the muscle and for each co-activated muscle. For multi-articual muscles like the \( m. \ latissimus \ dorsi \), the number of theoretical activity peaks would be at least 7, i.e. the number of degrees of freedom in the shoulder mechanism (Van der Helm, 1994a; De Groot, 1998). Consequently, this would result in about 29 parameters to be estimated. In the method presented here, the lumped effect of these ‘secondary’ moments was represented by the asymmetry of muscle activation.

In the recorded EMG’s, the range of muscle activation did not correlate positively with the increase in the number of degrees of freedom. The glenohumeral muscles, i.e. \( m. \ infraspinatus, m. \ teres \) major and \( m. \ deltoideus \), scapulo-thoracic muscles, i.e. \( m. \ trapezius \) and \( m. \ serratus \), and thoracic-humeral muscles, i.e. \( m. \ latissimus \) and \( m. \ pectoralis \) major, cross one, two and three joints respectively. The glenohumeral muscles were active over a range of about 270°, while the \( m. \ pectoralis \) major was symmetrically active over just 180°. The large \( m. \ pectoralis \) seemed only to be involved in generating the externally desired torque, while the selected glenohumeral muscles are also active while generating antagonistic moments and seem also involved in co-ordination or ‘tuning’ of muscle torques.

The principal action of the different parts of the deltoid muscles rotated as expected according to their moment arms, i.e. clockwise from anterior to posterior (Fig. 3). This effect is not observed for the descending clavicular and transverse and ascending scapular parts of the trapezius muscles, Fig. 3. With the counter-clockwise rotation of the muscle fibres also the number of affected joints changes. Due to the relative changing moment arms around the sternoclavicular joint and the acromioclavicular joints the function of the muscle parts change more than observed for e.g. the deltoid parts.

The least curve fitting results were obtained for the \( latissimus \ dorsi \) muscle (Fig. 3). Two peaks could be discriminated in the recorded muscle activation, one in the lower right quadrant (lateral adduction) and one in the upper quadrants (abduction). In the curve fitting procedure the direction of principal action was estimated in between both peaks at 115° which does not correspond to the adduction function of the muscle as also indicated from model simulation (Fig. 4c). The recorded activation peak around 0°-force direction may be the result from cross-talk of the underlying serratus muscle (Fig. 3). From our application using surface electrodes, cross-talk could not be quantified. However, wire electrodes, applied for rotator cuff muscles e.g. subscapular muscle (Meskers et al., 2004), are difficult to insert in the thin \( latissimus \ dorsi \) muscle.

4.3. Method in relation to model validation

The comparison of the recorded EMG’s (muscle activation) with the estimated muscle forces, calculated by means of inverse dynamic model simulation was added to illustrate the equivalence between both muscle activation and muscle force. Because isometric and isotonic, we assumed both relative muscle activation and relative
muscle force to be equivalent variables. Judging from Fig. 4c, and with a significant Pearsons correlation coefficient of 0.778 (p < 0.01) this assumption seems to be valid.

The shoulder model is the formalized algorithm of the anatomical representation that we all use to predict muscle function. The muscle force estimation in the inverse dynamic model was based on the mechanical muscle contribution (shoulder geometry, muscle moment arms) and the algorithm through which the muscle redundancy problem was solved. The observed correlation between model prediction and recorded muscle activation supports the hypothesis that in our experimental set-up muscle activation reflects relative muscle force. It is also a relatively accurate method to analyse the function of muscles in the shoulder by recording muscle activation only (Flanders and Soechting, 1990). The surplus value of the model is the prediction of muscle and joint-reaction forces, which cannot be recorded by EMG. The shoulder model also allows a priori simulation of interventions in the shoulder (Magermans et al., 2004).

The discrepancy between measurement and model, e.g. for the lattisimus muscle and the infraspinate muscle expresses either a possible artefact of measurement, or an imperfection in the assumptions of the model. Both arguments are illustrated:

A possible measurement artefact is the potential cross-talk of the serratus muscle recorded in the latissimus muscle activation as described in the previous section. If indeed the activation of the serratus muscle is recorded as cross-talk of the latissimus muscle, the latissimus muscle is only activated at the lower adduction quadrants, which is in concordance with the prediction of the shoulder model.

In comparison with model results, the activation of the infraspinate muscle showed the largest divergence. The principal action of the muscle activation is directed towards 53° with an activity in the upper and right quadrants, while the model predicts a principal action of muscle forces towards 235°. Equivalent recordings with wire EMG (Meskers et al., 2004) indicate a principal action 0° in their co-ordinate system (pointing upward, with the humerus at in the scapular plane, i.e. 30° angle with the frontal plane, 60° elevation and forearm positioned at 45° with the horizontal plane), which is equal to 45° in the co-ordinate axes of the humerus (this paper). The recorded muscle activation is therefore concluded to be reliable. Though not visible in the figure (Fig. 3) the infraspinate muscle is constantly activated above rest level.

The mechanical contribution of the infraspinate muscle estimated from the model was opposite of the recorded (and assumed reliable) muscle activation. The infraspinate muscle forces may be sensitive for the exact position of muscle insertion in relation to the task definition. This is illustrated by the fact that the model calculates predominant inward (medial) muscle forces, while the subjects indicated predominant lateral (outward) directed muscle activity. Though the external moment around the longitudinal axis of the humerus was zero, all muscles inserting on the humerus generate (intrinsic) axial moments that need to be balanced (sum of moments is zero). These muscle forces may therefore be very sensitive for their position of insertion, due to their small moment arms relative to the longitudinal axis of the humerus.

The comparison of activation recording and model simulation reveals sensitive details of both the method of recording of muscle activity (EMG of the latissimus muscle) and the modelling of the shoulder (infraspinate muscle). Despite some discrepancy, the correlation between recorded and muscle activation is high and the overall model seems relatively insensitive for individual muscle discrepancies.

The method of isometric EMG data acquisition and modelling serves functional, biomechanical and clinical purposes. The sub-set of shoulder muscles, which were recorded by means of EMG, was conditionally chosen by the possibility of surface EMG recording and affected all degrees of freedom of the shoulder. The parameter match between force predicted by musculo-skeletal models and EMG will be an independent measure for the validity of the biomechanical models during static postures (De Groot et al., 1992; De Groot, 1998).

4.4. Application

In shoulder disorders, where co-ordination, pain or force reduction is involved, e.g. sub-acromial disorders, spasm, or habitual subluxation, the distribution of muscle forces will be affected. We expect this load sharing between the muscles to be reflected by the angular parameters, PA, I1 and I2 and the estimated parameters of the recorded EMG pattern will change in pathological situations. The application of isometrical force direction dependent EMG recording may turn out to be a new tool for clinical evaluation and follow-up of shoulder problems (Arwert et al., 1997; Meskers, 1998).

5. Conclusion

The method of isometric muscle activation recording of a rotating external force vector is a promising method for the validation of musculo-skeletal models. The method of rotating force addressed all muscles to become active for a specific section of force directions. Only five parameters were needed to accurately describe the muscle activation magnitude. The initial and terminal baseline intercepts and the principal action are relevant parameters that can be estimated accurately from 20 EMG recordings.
The methodology shows that shoulder muscle activation is not easily derived from text book anatomy, where only positive moment arms indicate the role of the muscle for specific tasks. Complex interactions through multiple degrees of freedom affected by the shoulder muscles result in muscle activity where the muscle is expected to have negative moment arms for the specific task. The overall match between recorded muscle activation and estimated muscle forces from musculo-skeletal models allows for the comparison these variables for model validation.

References


