Shoulder muscle force predictions - comparison of two models

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Abstract - To assess shoulder muscle loads during manual work, different methods are available. In this study, two models are compared for ten shoulder muscles: one based on a mechanical description combined with the assumption of an optimization principle, and one based on EMG recordings. The comparison showed similarities in activation patterns for two of the shoulder muscles (m. deltoideus, pars posterior and m. biceps brachii), but significant differences for the other muscles regarding:

1) muscle force levels (m. deltoideus, pars anterior, m. latissimus dorsi, m. pectoralis major, and m. subscapularis);

2) force direction giving peak muscle force (m. deltoideus, pars anterior & medialis, m. latissimus dorsi, m. supraspinatus, and m. triceps brachii);

3) range of force directions giving muscle forces more than half of the peak muscle force (m. deltoideus, pars anterior & medialis, m. pectoralis major, m. subscapularis, m. infraspinatus, and m. triceps brachii).

In general the EMG-based model predicts higher muscle forces as well as muscle activity for a wider range of external force directions, compared to the mechanical/optimization model. This may be explained by co-contraction of the shoulder muscles.

INTRODUCTION

Knowledge on the relationship between physical job demands and internal shoulder muscle forces may improve the ability to prevent work-related disorders. Since internal shoulder forces cannot be measured directly, indirect methods and modeling are needed. In the last few years, detailed three-dimensional models for shoulder muscle force predictions have been developed based on a detailed mechanical description (Karlsson and Peterson, 1992; van der Helm, 1994). The problem of indeterminacy when solving the equations for muscle forces has in these models been approached by choosing an objective function (cost function) to be minimized. As an alternative, an EMG-based shoulder model for describing static work situations has recently been developed (Laursen, 1996). The purpose of the present study is to compare the shoulder muscle forces predicted by these two types of models for a static arm position, while performing hand forces in varying directions.

METHODS

The two models compared were an optimization-based shoulder model which is now available as PC software (Makhsous et al, 1994), and an EMG-based model (Laursen, 1996). Both models calculate shoulder muscle forces as a function of external load for a given arm position, but based on different principles. The optimization model is based on a study of shoulder muscle anatomy (Högfors et al, 1987). Here, the shoulder muscles are modeled as straight or curved strings. Broad muscles are modeled as several strings, representing a total of 38 independent muscle parts, of which 16 are used in the comparison between the models. The muscle forces are calculated as the set of forces which satisfies moment and force constraints, and minimizes the objective function:

$$\Sigma_{\mathrm{muscles}\ i}(F_i/A_i)^2 \qquad \text{(Eq. 1)}$$

where A_i is the physiological cross-sectional area of the muscle obtained from cadaver studies. The muscle forces F_i are restricted to range from zero to the maximum muscle force, F_{max} (Karlsson and Peterson, 1992). F_{max} is calculated from the A_i using a specific tension of 70 N/cm^2.

The EMG model (Laursen, 1996) is a model based on EMG/force calibrations for individual muscles. Recordings were obtained from ten muscles or muscle parts crossing the glenohumeral joint, therefore this model is restricted to this part of the shoulder. The EMG recordings were derived from six young female subjects. For each subject, the relationship between external, submaximal hand force in three dimensions and EMG for each of the shoulder muscles was analyzed by multiple linear regression using a spatial resolution of 15 degrees. The relationship between EMG and force was calibrated for submaximal forces for the force direction giving the steepest EMG increase. Thereby, muscle force relative to the maximum muscle force was expressed as a function of external hand force, for each of the subjects.

To assess the effect of cross-talk from other muscles, the same analysis was applied to the time-differentiated EMG signal. This signal is expected to contain considerably less crosstalk (Winter et al, 1994), because of the combined effect of "tissue filtering" (Basmajian and deLuca, 1985) damping the higher frequencies of the EMG spectrum from fibers more distant to the electrodes, and the time-differentiation of the signal increasing the weight of the higher frequencies.

By comparing the results from these two signals, the presence of crosstalk could be considered. Since the scales of the two signals were different, the differentiated signal was normalized to the same peak value as the corresponding unprocessed signal.

To produce comparable results from the two models, similar postures were chosen: the arm position was a slightly abducted upper arm (10 degrees), and a slightly flexed elbow (elbow angle approx. 150 degrees). The external hand force was chosen to have a constant magnitude corresponding to 15-20% MVC, dependent on the force direction. Since the optimization model was based on male data and the EMG-based model on female data, the maximum external forces were different for the two models. In order to have the same relative hand force compared to MVC, the absolute forces were chosen to be different: 19 N and 11-15 N for the optimization and EMG-model, respectively. The model calculations were performed for 24 different force directions in the horizontal plane. For muscles in the EMG-based model corresponding to several muscle parts in the optimization model, the weighted mean was calculated for these parts in the optimization model.

For both models and each of the muscles, three parameters were determined from the muscle force patterns: the peak of the relative muscle load among the 24 directions of external force, the direction of this peak, and the half-width (number of directions giving muscle load above the half of the peak relative muscle load). When testing differences between the two models, a sign test was used, testing whether the median of the EMG-model results for the six subjects differed from the results for the optimization model. Significance was assumed for $p<0.05$.

Table 1- *The results for the three parameters describing the muscle force variation. Results for the EMG-based model are mean of the results for the six subjects. Significant differences are marked with an asterisk (*).*
§) One subject excluded.

	Peak Muscle relative load %MVC		Peak load direction (degrees)		Half width (degrees)	
Muscle	Opt. model	EMG model	Opt. model	EMG model	Opt. model	EMG model
M. deltoideus, pars anterior	0.3	15.9*	15	70*	15	152*
M. deltoideus, pars medialis	16.9	19.6	345	23*	90	180*
M. deltoideus, pars posterior	19.3	17.5	315	338	120	132
M. latissimus dorsi	6.9	17.2*	315	233*	165	230
M. pectoralis major	10.6	27.1*	135	152	135	173*
M. supraspinatus	22.5	25.3	90	30*	165	163
M. subscapularis	4.6	20.7*	195	240§	90	148*
M. infraspinatus	17.2	22.6	30	8	240	185*
M. biceps brachii	24.9	18.7	105	105	105	145
M. triceps brachii	12.4	16.1	240	293*	90	165*

RESULTS

The results were compared for the ten muscles or muscle parts included in the EMG-based model. Figure 1 shows the model results for the ten muscles, and Table 1 shows the results for the three parameters. For the EMG-based model, there are curves corresponding to each of the six subjects included in the study. For four of the muscles, the EMG-based model predicted higher muscle loads, calculated as % of the maximum muscle force. This was the case for m. deltoideus

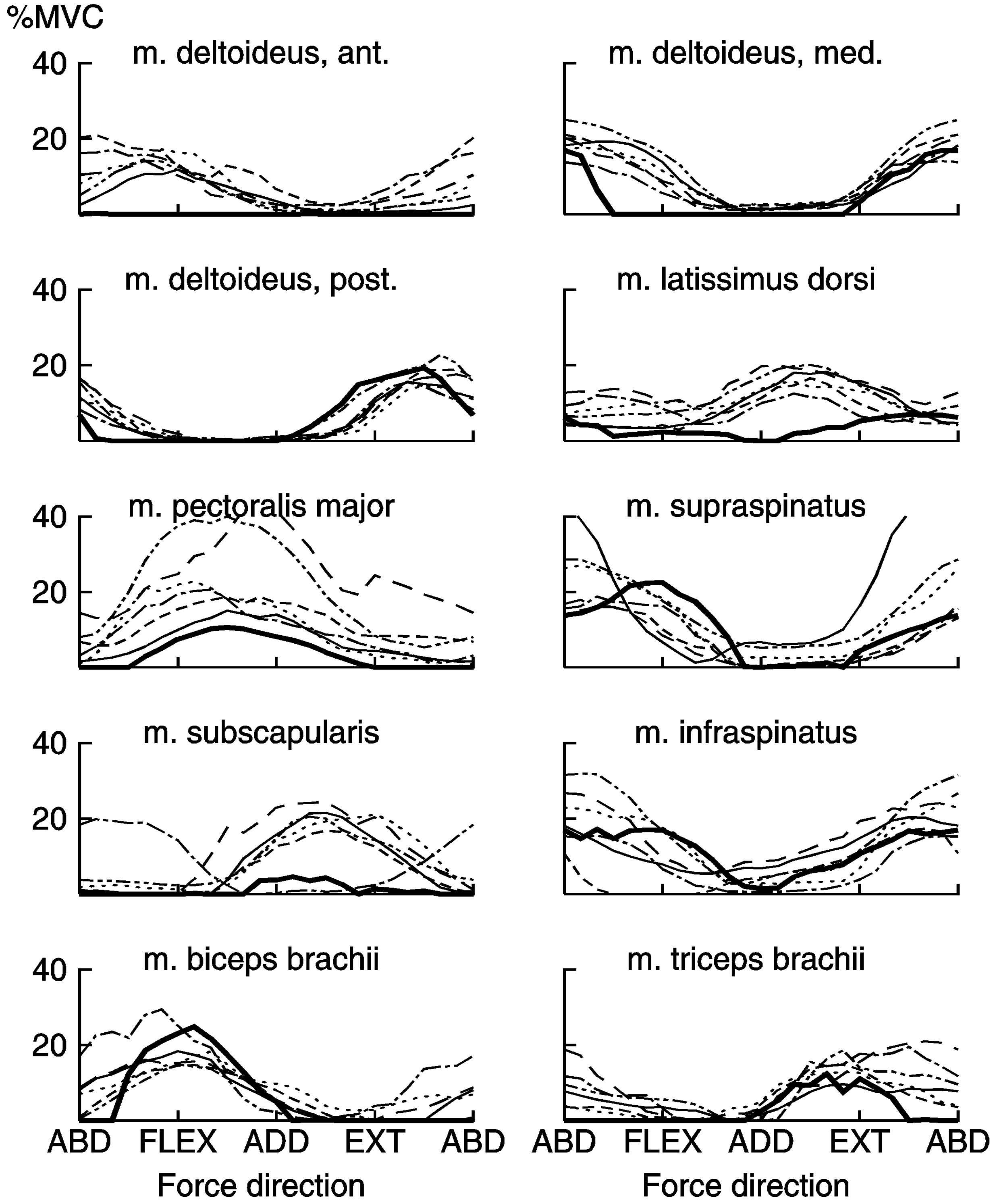

Figure 1. *Model predictions of muscle force for external hand force in 24 direction. Thick line: Optimization model. Thin lines: EMG-based model, each line corresponds to data from one subject.*

(pars anterior), m. latissimus dorsi, m. pectoralis major, and m. subscapularis. For the remaining muscles, the activity levels predicted by the two models were comparable. For five of the muscles, there were coincidences between the directions of maximum activation. However, for m. deltoideus (pars anterior & medialis), m. latissimus dorsi, m. supraspinatus and m. triceps brachii, these directions were significantly different. The EMG-based model predicted a broader half-width for six muscles (m. deltoideus (pars anterior & medialis), m. pectoralis major, m. subscapularis, m. infraspinatus, and m. triceps brachii). In general, the half-width had less variation for the EMG-based model (132°-230°, mean of six subjects), compared to the optimization model (15°-240°).

DISCUSSION

When comparing the two models, the individual variation should be considered. This variation can be assessed from the six curves for the EMG-based model. When this is taken into account, a good agreement between the two models was found for some muscles for all three parameters (m. deltoideus (pars posterior) and m. biceps brachii). For other muscles, there were minor deviations (m. supraspinatus and m. infraspinatus). For the remaining muscles (m. deltoideus (pars anterior & medialis), m. latissimus dorsi, m. pectoralis major, m. subscapularis, and m. triceps brachii), there were substantial deviations between the two models. Several of the muscles were in the optimization model represented by several muscle parts, and the corresponding EMG recording may not represent the whole muscle. This could explain some of the deviations for m. latissimus dorsi, m. pectoralis major, and m. subscapularis. Another explanation may be crosstalk between EMG signals. In order to assess this, the results from the original and differentiated EMG signals were compared.

In most cases, the differences between the original and differentiated signal were small. One of the largest differences was for m. deltoideus (pars medialis), and is shown in figure 2. Consequently, the difference between the models may only to a minor extent be explained by crosstalk from other muscles or muscle parts. This is also confirmed by a comparison between surface and intramuscular EMG from m. infraspinatus (Laursen, 1996). A more probable explanation to the differences in half-width between the two models could therefore be the choice of objective function in the optimization model.

For m. latissimus dorsi, m. pectoralis major, m. subscapularis, and m. deltoideus (pars anterior), the optimization model predicts much lower peak relative muscle loads, compared to the EMG-based model. For m. latissimus dorsi, the low maximum EMG (mean 290 μV), as well as recorded maximum extension force may indicate that this muscle was not fully activated. If this was the case, the EMG model would predict too high muscle loads due to the EMG/force calibration. For m. subscapularis and m. deltoideus (pars anterior), this is probably not the case, and these muscles had high maximum EMG (mean 1.4 mV for m. subscapularis and 1.0 mV for m. deltoideus (pars anterior)).

Although the two models give comparable results for some of the muscles, there was a general tendency of higher muscle loads according to the EMG-based model, compared to the optimization model. This may be caused by co-contraction of the shoulder muscles, in order to stabilize shoulder and arm in addition to overcome the external load. If this is the case, the optimization model will underestimate the muscle loads.

Both models offer advantages as well as disadvantages. The EMG-based model reflects the

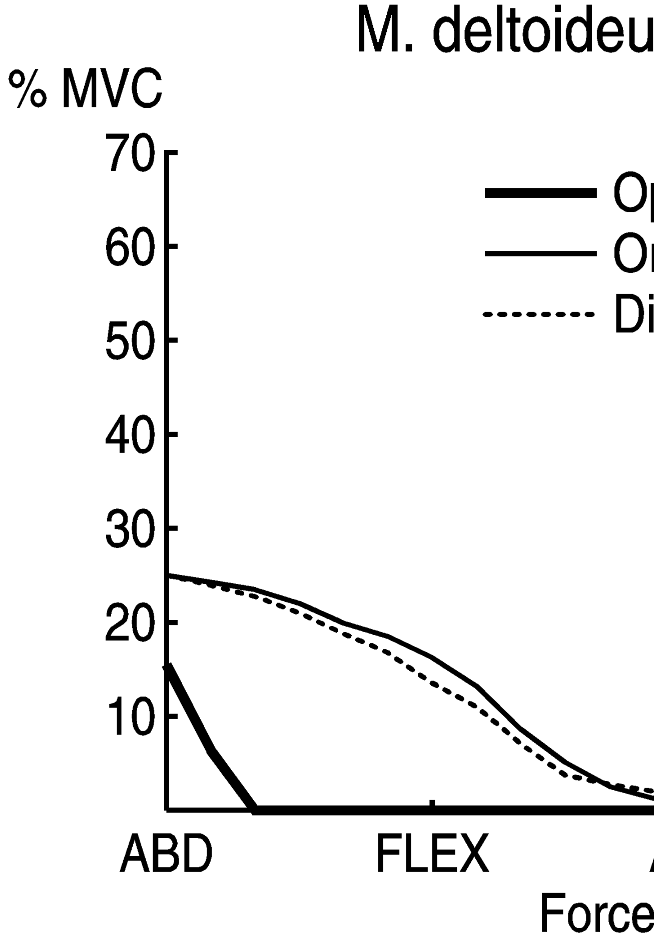

Figure 2. *Assessment of the influence of crosstalk. The differentiated EMG signal is assumed to contain less crosstalk than the unprocessed EMG.*

actual muscle activation pattern making this model useful studying especially muscle load changes caused by different working conditions, but the model may be biased by difficulties in the EMG/force calibration, and the resultant moment of the summed muscle forces do not necessarily equate the moment of the external force. The optimization-based model has the advantage being more general applicable, including all positions and load conditions, but the calculated muscle forces must be regarded as a "minimum force set", which may underestimate the true forces.

REFERENCES

Basmajian, J. V. and De Luca, C. J.(1985) *Muscles alive. Their functions revealed by electromyography*, 5th edn, pp. 1-561. Williams & Wilkins, Baltimore.

Högfors, C., Sigholm, G., Herberts, P. (1987) Biomechanical model of the human shoulder -I. Elements. *J Biomech* **20**, 157-166.

Karlsson, D. and Peterson, B. (1992) Towards a model for force predictions in the human shoulder. *J Biomech* **25**, 189-199.

Laursen, B. (1996) Shoulder muscle forces during work. EMG-based biomechanical models. PhD.-thesis, National Institute of Occupational Health, Copenhagen.

Makhsous, M., Peterson, B., Sieminski, A. (1994) Shoulder version 1.0. Centre for Biomechanics, Chalmars University of Technology, Gothenburg.

Van der Helm, F.C.T.(1994) A finite element musculoskeletal model of the shoulder mechanism. *J Biomech* **27**, 551-569.

Winter, D.A., Fuglevand, A.J., Archer, S.E. (1994) Crosstalk in surface electromyography: Theoretical and practical estimates. *J Electromyogr Kinesiol* **4**, 15-26.