Strength profiles of the shoulder

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ABSTRACT - Strength profiles of the shoulder joint for three arm positions are measured experimentally to present quantitative data on the shoulder strength. The experimental profiles are compared with the corresponding theoretical profiles which are obtained by using a shoulder model. The calculations were made both with default muscle parameters and individually adapted parameters. The results show that the employed shoulder model gives reliable predictions and effects of training is demonstrated. Generally higher forces are predicted for adduction than for abduction. Appropriate adjustments of muscle parameters result in a better individual fit of the theoretical profile making it possible to use strength profile measurements diagnostically.

INTRODUCTION

To quantify and modify human shoulder performance or to take protective measures, there is a need to assess shoulder strength.

Maximum strength of the shoulder for a fixed position of the humerus can be defined as the ability to sustain the maximal external load applied. We use a force applied at the distal end of the humerus and orthogonal to the longitudinal axis of the humeral. The maximal strength defined in this way varies with the angle in the plane orthogonal to the humerus. Because of this we need to measure maximum strength for almost all directions in this plane. When we have enough data we may use the measured maximum forces around the circle to construct a closed curve - a so-called shoulder **strength profile.**

The shape of the strength profile depends on the position of the humerus. When a profile for a specific position of the humerus is needed, it is necessary to run the whole experiment for this position, which can be very expensive and time-consuming. What we need is a way to calculate strength profiles and this requires a reliable shoulder model.

Such a model is likewise of great value in interpreting results from strength measurements. It answers questions like: 'What muscles contribute to the strength in a particular direction in a given position?', 'What muscles have to be strengthened for a particular task?', 'What muscles may have been damaged?', 'How can a particular muscle group be trained?' etc.

Many studies have attempted to measure the strength of the shoulder. In some of these studies measurements of the shoulder strength were performed in a few directions only, often using external loads applied at the hand (e.g. Hinton, 1988; Greenfield et al. 1990), or at the lower arm. Experience has shown that such a load configuration, containing a structure with weak joints (the wrist joint and/or the elbow joint), cannot be expected to give a proper representation of the strength of the shoulder. Therefore, it is necessary to eliminate the wrist and the elbow function from the measurement by applying the external load directly to the humerus.

A detailed three-dimensional mechanical model of the shoulder is presented in Högfors et al. (1987) and this is developed further for use in this study. In the present study, we have measured the strength profiles of the shoulder experimentally to present quantitative data on shoulder strength and the contributions to it from the internal components.

Subsequently, we calculated theoretical strength profiles by using the shoulder model, which can predict the shoulder strength in a given direction. First this was done with default muscle parameters (DMP), which are derived from cadaver studies. Then the strength profiles were calculated using adjusted muscle parameters (AMP) for a specific subject who regularly participated in sports (kick-boxing).

MATERIAL AND METHODS

Experimental profiles

A person in a well-defined body position is requested to withstand an external load acting at some point of the upper arm. When performing such experiments on a subject with proper thoracic support it is quite possible to determine the maximal strength of the shoulder by applying the load at the distal end of the humerus. In order to have adequate support to maintain the same position during the experiment in different positions, an adjustable frame was developed. A device to measure the applied force (magnitude and direction), as well as a means of collecting data, were also needed. In the present study we are concerned with force profiles, i.e. the load is a pure force entirely in a plane perpendicular to the longitudinal axis of the humerus and applied at its distal end.

Positions of the upper extremity

The maximum external force on the arm in a plane perpendicular to the longitudinal axis of the hume-

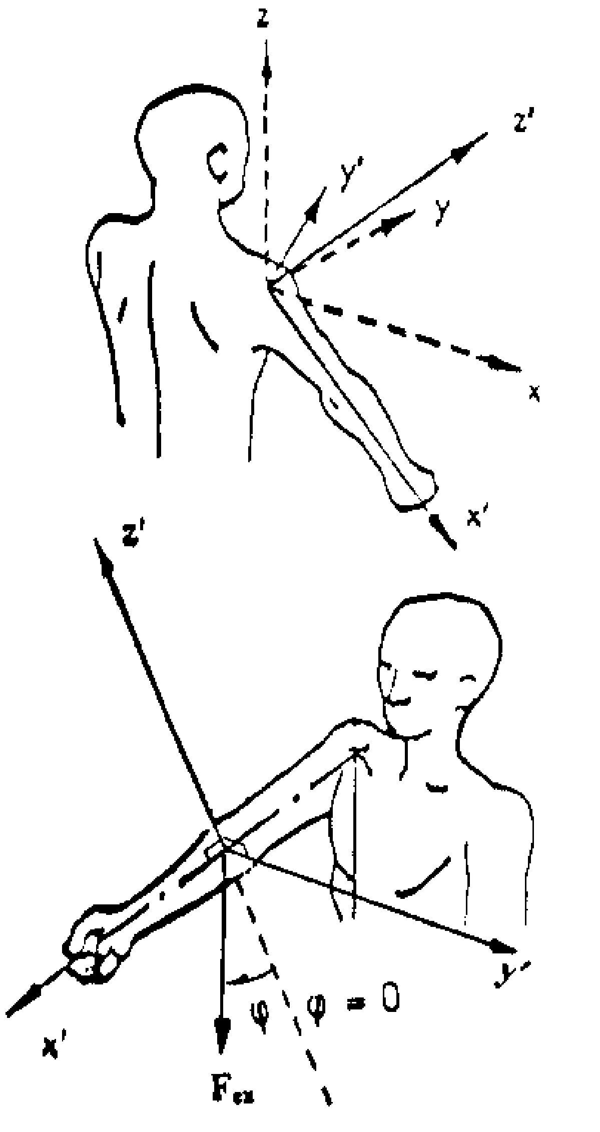

Figure 1 - *Description of the co-ordinate systems and the direction of the external force F_{ex}. The angle of force φ is measured between the negative humerus z′-axis and the force. The angle is positive for directions in the half space with negative y′ values. To show the angle φ clearly, the z′-axis and the y′-axis in the lower part of the figure are translated to the point of application of the external force. Because of the value zero of the humerus long axis rotation angle γ, the humerus z′-axis is always in the vertical plane spanned by the humerus x′-axis and the z-axis of the laboratory system.*

Table 1 - *Cardan angles for load cases (a, b and c).*

Load case	α	β	γ
a	0°	0°	0°
b	45°	0°	0°
c	45°	-36°	0°

rus was measured. This was repeated for many different directions around the arm and for three positions of the humerus. The direction φ of this maximum external force (F_{ex}) is shown in Figure 1.

For the laboratory co-ordinate system (Figure 1), the x-y plane is horizontal, with the y-axis in a sagittal plane and the x-axis in a frontal plane. The z-axis extends vertically upwards. The situations studied were those of a person standing in an upright position, i.e. 90° abduction as the starting position (α=0, β=0, γ=0), with the humerus rotated with an angle α forward from the frontal plane, elevated at different angles β and with the rotation angle γ around the longitudinal axis. These angles are Cardan angles and the rotations have to be taken in the given order and are around the body-fixed axes (about space-fixed axis in the opposite order). The attitude angles α, β and γ for the humerus are explained in Högfors et al. (1991) and Peterson (1994). The elbow angle is 175°, i.e. the arm is nearly straight. Three positions of the upper extremity are used in this study and the corresponding Cardan angles (taken in the order α, β and γ) are given in Table 1.

Experimental arrangement

The device for measuring the force applied by the arm consists of two short tubes of stainless steel which are connected to each other by four smaller tubes with longitudinal slits as shown in the Figure 2. These small tubes are weak enough to allow the measurement of the deformation caused by translating the inner tube relative to the outer one in radial direction. In order to measure the deformation in the small tubes, two strain gauges are bonded to the inner and outer surfaces of each tube. This mounting will indicate compression and elongation on both sides of the tube wall.

The entire device is rigidly mounted on a beam so that the outer tube remains fixed in space. Two Digital Strain Meters were used to measure the force components in the x- and y-directions. Thus we obtained both the magnitude and the direction of the force applied to the inner tube.

The experiments were performed on two male subjects who were thirty and thirty-nine years old, both right-handed. They did not have any history

of injury or symptoms related to the shoulder. Subject 2 participated in sports regularly.
The subjects were standing, with the right arm through the inner tube of the device. The subject was restrained in the frame during each experiment. Before testing, the subjects were trained to perform correctly, i.e. not changing their position during the experiment and also performing maximum force of a duration of at least one second.
A set of experiments for one position was divided into eight parts in a four day experiment. At the start of each part a warm-up session consisting of two maximal efforts was first performed to accustom the subject. Then five consecutive maximum test trials followed. These were separated by a rest of two minutes. After an additional twenty-minutes period of rest, this protocol was repeated once. At least two sets of experiments were carried out for each subject. This went on for eight days making a total of 80 measured directions. Subjects were not permitted to see the display of the force output.
Forces applied to the force device in terms of x-, and y-components were recorded into a file with a sampling rate of 20 Hz for 10 seconds. Each phase of an output signals from a specific direction was characterized by 200 points separated by equal time intervals. A program was made to transform the output signals to forces, magnitude and direction, using the calibration data.

Theoretical profiles
Of course such experiments as mentioned above could also be performed as theoretical computations using the shoulder model. It is then possible to compare not only absolute values of maximal strength in a particular direction but also, presumably more important, to compare computed and measured relative strength in all directions.
The anatomical description is presented in Högfors et al. (1987, 1991) and Peterson 1994.

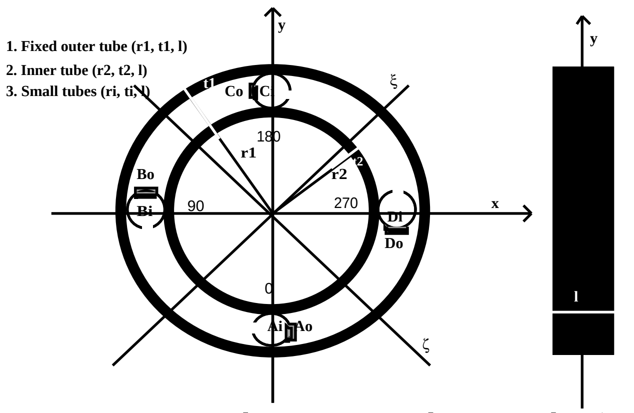

Figure 2 - *Force device.* ***1.*** *Fixed outer tube (r_1, t_1, l);* ***2.*** *Inner tube (r_2, t_2, l);* ***3.*** *Small tubes with longitudinal slits (r_i, t_i, l);* ***4.*** *A_i, A_o, B_i, B_o, C_i, C_o, D_i, D_o strain gauges (i and o are indexes for inside and outside of the small tube respectively).*

Table 2 -*Default Muscle parameters (DMP) in terms of normalized physiological cross-sections (NCSA) and upper limits of muscle forces (F_{max}).*

No.	Ref.	Muscle	NCSA (cm^2)	F_{max} (N)
1	1F1	Latissimus dorsi 1 (upper)	4,30	378,67
2	1F2	Latissimus dorsi 2 (lower)	5,18	456,16
3	2F1	Levator scapulae	3,14	276,64
4	3F1	Omohyoid	1,00	87,996
5	4F1	Pectoralis major (lower)	7,40	651,38
6	4F2	Pectoralis major(upper)	4,52	397,31
7	5F1	Pectoralis minor	3,42	301,17
8	6F1	Rhomboid major	3,80	334,52
9	7F1	Rhomboid minor	1,88	165,79
10	8F1	Serratus anterior (upper)	4,97	437,53
11	8F2	Serratus anterior (middle)	4,20	369,84
12	8F3	Serratus anterior (lower)	6,83	601,35
13	9F1	Sternocleidomastoid	1,00	87,996
14	10F1	Sternohyoid	1,00	87,996
15	11F1	Subclavius	1,00	87,996
16	12F1	Trapezius intermedial	4,54	399,27
17	12F2	Trapezius (lower downward)	3,97	349,24
18	12F3	Trapezius (upper)	6,60	580,75
19	12F4	Trapezius (clavicular part)	4,72	415,94
20	13F1	Deltoid medial	9,27	816,19
21	13F2	Deltoid posterior	8,58	755,37
22	13F3	Deltoid anterior	7,07	621,95
23	14F1	Coracobrachialis	3,15	277,62
24	15F1	Infraspinatus 1 (upper)	6,37	560,15
25	15F2	Infraspinatus 2 (lower)	6,67	586,64
26	16F1	Subscapularis 1 (upper)	5,15	453,22
27	16F2	Subscapularis 2 (middle)	4,65	409,08
28	16F3	Subscapularis 3 (lower)	6,37	560,15
29	17F1	Supraspinatus	6,30	554,26
30	18F1	Teres major	7,38	649,42
31	19F1	Teres minor	3,12	274,68
32	20Fc1	Biceps (long)	3,53	310,98
33	20Fc2	Biceps (short)	3,12	274,68
34	21F1	Triceps (long head)	9,98	877,99
35	21F2	Triceps (medial head)	8,98	789,71
36	21F3	Triceps (lateral head)	8,98	789,71
37	28F1	Brachialis	9,98	877,99

Forces acting on the rigid bodies are divided into muscle forces, joint contact forces and ligament forces. The muscles are modeled as stretched strings. Larger muscles are represented with more than one string. Some of these strings run over bony parts modeled by second degree polynomial surfaces, i.e. not in a straight line between the end points. The force directions are the tangents to the strings. The external forces acting on the arm are the weights of the upper arm, the lower arm and external loads which can be defined in terms of forces and moments.

The shoulder rhythm

It is necessary to know the relations between the orientations of the bones: the scapula, the clavicle and the humerus. This interplay between the motions of the constituent parts of the human shoulder is known as the 'shoulder rhythm'. The shoulder rhythm is based on the work by Högfors et al. (1991) and Karlsson and Peterson (1992).

Muscle parameters

With muscle parameters we mean such parameters as normalized physiological cross-sectional areas (NCSA) and upper and lower limits of muscle forces (F_{max} , F_{min}). These parameters and their relationships are defined in the shoulder model by Karlsson and Peterson (1992) as default muscle parameters, DMP (Table 2), which are derived from cadaver studies. It should be noted that the lower limit of the force (F_{min}) is equal to zero for all muscles.

Computation of the theoretical profiles

According to the description above and Figure 1 we obtained the following expressions for the exterior force in the laboratory system:

$$\mathbf{F}x = \mathbf{F}ex\left[\sin(\varphi)\sin(\alpha) + \cos(\varphi)\cos(\alpha)\sin(\beta)\right]$$
$$\mathbf{F}y = \mathbf{F}ex\left[\cos(\varphi)\sin(\alpha)\sin(\beta) - \sin(\varphi)\cos(\alpha)\right]$$
$$\mathbf{F}z = \mathbf{F}ex\left[\cos(\varphi)\cos(\beta)\right]$$

For a given force direction the program computes the maximum load by employing an iterative scheme in which the magnitude of the force (F_{ex}) is decreased or increased (based on whether the current force leads to infeasibilities or not) by a value equal to one half of the one used in the preceding step. The maximum force is considered to be found when this increase (decrease) becomes less than the assumed accuracy. The highest feasible value of the force used in the iteration is then taken as the maximum. The glenohumeral contact force constraint, the angle between the direction of the contact force vector in the glenohumeral joint and the normal to the cavitas glenoidalis was 32° for all computations.

Equilibrium equations

For our model with three internal bones (the clavicle, the scapula and the humerus) it is possible to obtain at most 18 equilibrium equations. As a means of obtaining a solution to the statically indeterminate force system (50 unknown forces) we have chosen to minimize the sum of the squared muscle tensions, i.e. minimize $\Sigma(F_i/A_i)^{-}$ (objective function).

RESULTS

The theoretical profiles were computed for all three load cases (Table 1) described earlier by using the shoulder model with the default anthropometric data (adult male) and DMP. The angle φ was increased by 2° in steps from 0° to 360°, which means that we computed shoulder strength in 180 directions for a fixed position of the upper extremity (see Figure 3).

To obtain a correct theoretical profile for a given subject we must adjust the muscle parameters. The problem is how to obtain the correct muscle parameters for an individual subject. One approach is based on the idea that training results in an increase in physiological cross-sectional areas (e.g. MacDougall et al. 1980). By using this fact we can adjust some muscle parameters for subject 2 who underwent the training program.

Experimental profiles were obtained for two healthy men for all three arm positions described earlier. For each subject, each arm position was examined, giving a total of six experiments each containing data from at least two trials.

The angle φ was randomized from 0° to 360° which means that we measured shoulder strength in at least 80 directions for a fixed position of the upper extremity. The whole range of the humerus-perpendicular plane was covered. The experimental strength profiles together with the corresponding theoretical profiles with DMP are shown in Figure 4. The force outputs in this diagram have been

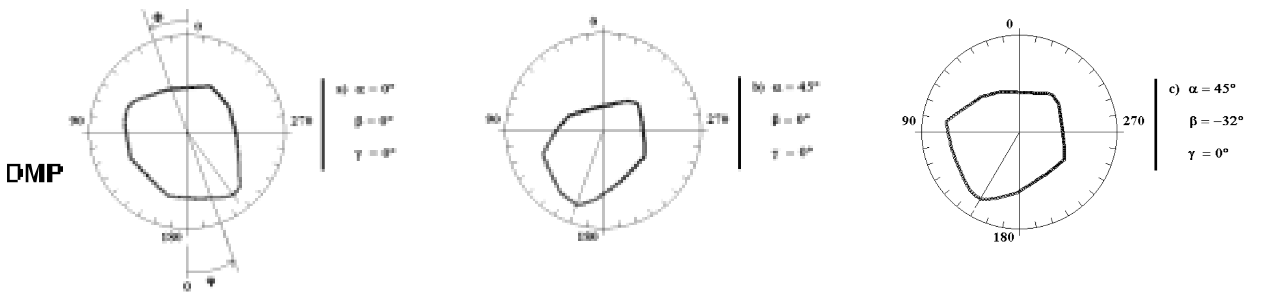

Figure 3 - *The theoretical strength profiles as produced by DMP (Table 2) for three load cases (a, b and c). Note that these profiles show the forces exerted by the shoulder in the direction Φ, not by the external force (Fex). The direction φ of the Fex is given in profile a).*

normalized so that the largest force has been set to 1.0. The same reference force has been used in all of the diagrams. The largest value is shown with one line through the center of the circle in these diagrams. The diagrams should be seen as looking from the shoulder towards the elbow. The humerus z´-axis is in the Φ=0° direction and the humerus y´-axis in the Φ=90° direction. An interpretation of the results of Figure 4 in terms of the commonly used terminology could be as follows: the adduction strength is considerably greater than the abduction strength. The human shoulder seems to be strongest when performing the action of pulling something towards the body and downwards.

Comparison between experimental and theoretical profiles with DMP

We compared the results by including both experimental and theoretical profiles for each subject and load case in one diagram (Figure 4). The computations of the theoretical profiles were performed with DMP. This is why we should not expect a perfect correspondence of these theoretical profiles to the experimental ones. However, even without adjusting the muscle parameters we obtained a profile which showed some common features and predicted precisely the direction of maximum strength in load cases b) and c). Appropriate adjustment of muscle parameters could result in a better individual fit of the theoretical profile.

Generally higher forces were predicted for adduction than for abduction. Furthermore, the theoretical strength profiles did not differ as much between upwards and downwards force ability as the experimentally measured profiles did. In load case a) when the humerus was in the frontal plane, it was noticed that the subjects did not manage to hold their arm straight, especially when pushing the support upwards (the weakest direction). The subjects actually flexed their elbow, presumably in order to get a more efficient muscle configuration. This may explain the differences between the experimental and the theoretical profile in load case a) for both subjects. Subject 2 appears to be relatively stronger in the upward direction, which perhaps can be attributed to the effects of boxing training.

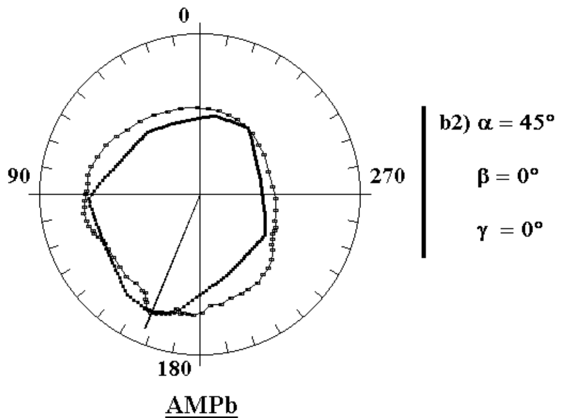

Figure 4 - *The theoretical profile with DMP for subject 2 and load case b) is compared with the corresponding experimental profile (Figure 3.b) in the same diagram*

Comparison between experimental and theoretical profiles with AMP

To fit the theoretical profiles with the corresponding experimental profiles we decided to study only one load case (case b) from subject 2. In this way we tried to identify the muscles which were affected by strength training.

We tried to adjust some muscle parameters for subject 2 and found that some muscles have larger physiological cross-sectional areas due to boxing training (see Table 3). The first muscle is serratus anterior (8F2) which is a major protractor of the pectoral girdle and as such is involved in all thrusting, pushing and punching movements where the scapula is driven forwards carrying the upper arm with it. Massive development of this muscle in boxers has been observed (Palastanga et al, 1989). The next muscle is trapezius which has an important function in stabilizing the scapula as a base for movements of the upper arm. A computation of the corresponding theoretical profile with the new parameters was performed. The new theoretical profile with AMP together with the corresponding experimental one is shown in Figure 5. We can see that an adjustment of muscle parameters (AMP) results in not only a better fit of the theoretical profile to the corresponding experimental one but also a profile which predicted precisely the direction of maximum strength when we compare the diagrams in Figure 3.b) (DMP) and in Figure 4 (AMP).

The shape of the last theoretical profile with AMP, which defines a convex diagram, can be divided into eleven sectors. At the beginning of each sector we can identify a group of muscles with different levels of activity. Some of these muscles have 100% activity for the whole range of the sector while the level of activity of the others changes within the sector - the activity of some muscles disappear and some new muscles become activated. Thus, the next sector starts with a new activated muscle group which changes the shape of the strength profile at the start of the new sector. In Figure 5 we can see the theoretical profile with AMP, these eleven sectors and the muscle group with 100% activity for each sector.

DISCUSSION

An interesting property of the theoretical strength profiles, which can be deduced from the way they are computed, is that a theoretical strength profile defines a convex diagram. This property follows from the assumption that the limit of the externally observed strength is the result of individual muscle force limits.

In the experimental profiles we cannot document the convexity for the whole range of the diagram. It was difficult to measure maximum forces in some sectors because of the shape of the shaft of the humerus at its lower end where we applied the external load. This resulted in an uneven angular density of measured maximum forces. When the subject tried to apply the maximum force in such a direction, a muscle group on this side slid away and changed the point of application. For this reason it could be better to use the external load higher up on the humerus where the shaft is almost cylindrical. However, this could cause problems with the larger muscle bellies.

Validation of the theoretical profiles is normally done by comparison with experimental profiles. The model prediction with AMP results in a better individual fit of the theoretical profile to the experimental one. To obtain a correct theoretical profile for a given subject, the muscle parameters need to be adjusted. This may be done systematically or, as here, by using an educated guess. However, even without adjusting the muscle parameters the model predictions show some common features. Thus, one may conclude that the default muscle parameters defined in the model are acceptable for use in modeling.

Results showed that the anterior Deltoid (13F3, Table 2) was not activated as much as expected.

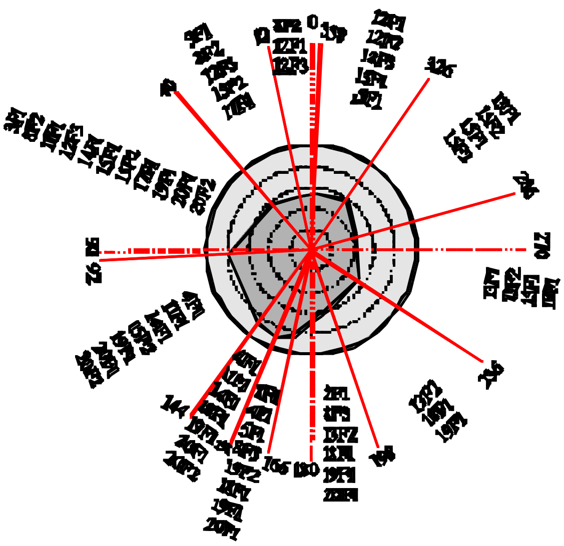

Figure 5 - *The theoretical strength profile with AMP for subject 2 and load case b) is divided into eleven sectors. Muscle group with 100% activity is shown for each sector.*

Table 3 - *List of the muscles which have the new 'adjusted' parameters (AMP). All other muscles have the same data as in Table 2.*

No.	Ref.	Muscle	NCSA (cm^2)	F_{max} (N)
3	2F1	Levator scapulae	4,52	397,74
4	3F1	Omohyoid	4,52	397,74
5	4F1	Pectoralis major (lower)	8,14	716,28
11	8F2	Serratus anterior (middle)	6,93	609,81
16	12F3	Trapezius (upper)	10,56	929,23
31	19F1	Teres minor	3,43	302,00

The reason for this is probably that no constraint is used on the contact force in the acromio-clavicular joint in the presently used theoretical model. Apart from this it is concluded from the explained results that the model gives a fair description of the considered experiment.

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REFERENCES

Greenfield, B.H., Donatelli, R., Wooden, M.J. and Wilkes, J. (1990) Isokinetic evaluation of shoulder rotational strength between the plane of the scapula and the frontal plane. *Am. J. Sports Med.*, **18,** 124-128.

Hinton, R.Y. (1988) Isokinetic evaluation of shoulder rotational strength in high school baseball pitchers. *Am. J. Sports Med.*, **16,** 274-279.

Högfors, C., Peterson, B., Sigholm, G. and Herberts, P. (1991) Biomechanical model of the human shoulder joint - II. The shoulder rhythm. *J. Biomechanics* **24,** 699-709.

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

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

MacDougall, J.D., Elder, G.C.B., Sale, D.G., Moroz, J.R. and Sutton, J.R. (1980) Effects of strength training and immobilization on human muscle fibres. *Eur. J. Appl. Physiol.* **43,** 25-34.

Palastanga, N., Field, D. and Soames, R. (1989) *Anatomy and Human Movement.* Heinemann, Oxford.

Peterson, B. (1994) On a model of the upper extremity. *'Advances in the biomechanics of hand and wrist'* p. 95-106. Schuind, F. (Ed), Plenum Press New York.