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SHOULDER EXTERNALLY ROTATING EXERCISES WITH PULLEY APPARATUS

Joint Load and EMG

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ABSTRACT. The mechanical load on the gleno-humeral joint and the muscular activity during shoulder external rotation resisted by a pulley apparatus were analysed using normalized, low-pass-filtered EMG recorded from the infraspinatus, deltoid, pectoralis major and trapezius muscles. The load moment about the longitudinal axis of the joint was compared with the distribution of maximum muscle moment over different angles throughout the range of motion. The effect of subject positioning on the joint load and the muscular activity was studied. The best adaptation between the curves for load moment and maximum muscle moment was obtained when the subject was positioned sitting with the pulley located 20 degrees posterior to a frontal plane through the shoulders at a distance of 1.3 m from the joint. Of the four muscles investigated, the infraspinatus was the most active. The method described might be used to find optimal designs of shoulder external rotation exercises with special regard to avoiding unintentional overloading of joint structures weakened by disease or trauma.

Key words: models biological, electromyography, biomechanics, physical therapy, rehabilitation

Shoulder disorders often induce different functional limitations, and thus different types of training must be considered in the rehabilitation phase. Depending on the individual purpose of the therapeutic exercise, different levels of mechanical resistance in terms of load moment are chosen (2, 3, 6, 16). The resistance may be applied manually or by a device; or by the help of body segment weights only. The induced load moment and also the muscular activity levels will influence the strain on the different structures involved during the movement. With this background, the present study was focused on mapping the magnitude of the joint load and levels of muscular activity during shoulder exercises

Pulley-resisted exercises specially designed for shoulder structures have been suggested earlier (1, 5, 7–9, 13, 14, 26). An ordinary pulley apparatus makes it possible to vary the load moment in a rather wide range with regard to the magnitude of the load and the shape of the load moment curve (1). The induced load moment (resistance from the device) is caused by the gravity force of the weights transmitted by the cord of the pulley (in the following denoted pulley cord force). The force in the cord multiplied by its moment arm to the longitudinal axis of the humero-scapular joint gives the loading moment of force acting externally about the joint. This load moment is counterbalanced by a moment, equal in size and opposite in direction, exerted by the muscles (i.e. the muscular moment) (20).

To be able to compare the load moment with the maximum voluntary muscle moment through the whole movement sector, as suggested by Williams (31) and Moritz (22), for example, the muscular moment at different joint angles must be known. The magnitude of the isometric muscular moment varies over the movement sector (e.g. 11, 22, 23, 31). While this has been studied for shoulder flexon, extension and abduction movements (31), we have not found in the literature any study concerning the isometric strength over the movement sector for shoulder external rotation. The possibilities of adapting the load moment caused by a device to that of the maximum muscular moment for the same joint angle during internal rotation exercise with a pulley have, however, recently been reported by our group (14).

The myoelectrical activity in shoulder muscles during various arm movements in anatomical planes has been studied by several authors (2, 4, 17, 18, 28, 33). Only a few EMG studies report on common shoulder therapeutic exercise (7, 13, 14, 30). Where shoulder external rotation has been included (7, 30), the rotation component has been part of a movement in a physiological pattern, e.g. flexion, external rotation and abduction. We have

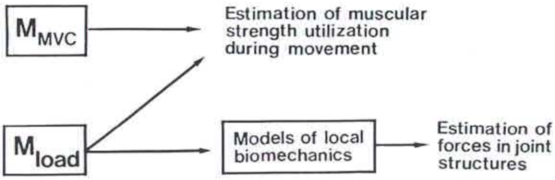

Fig. 1. Model illustrating how data about maximum muscular moment of force (M_{MVC}) and load moment caused by pulley (M_{load}) are used in present study.

not found any report concerning separate external rotation exercises.

The present study is part of a larger project concerning principles for optimizing therapeutic exercises for various types of patient. Some preliminary results concerning load moment and level of muscular activation during shoulder rotation exercises have been presented earlier (13). The general aim of the present study was to investigate the effect of different subject positionings in relation pulley apparatus with different weights applied to the pulley cord. The more specific questions were:

A. What is the level of activity in four different shoulder muscles during an exercise movement?

B. What is the magnitude of the gleno-humeral joint load moment acting about the longitudinal gleno-humeral joint axis and induced by the device?

C. How does the maximum isometrical muscular moment (strength) exerted by the external rotators vary through the sector of motion?

D. How can this kind of exercise be designed to allow optimal adaptation between the magnitude of the load moment and the maximum muscular moment over the movement sector?

E. What is the approximate magnitude of force in the infraspinatus (together with the teres minor) tendons during an exercise movement at an ordinary level of resistance?

MATERIALS, METHODS and ANALYSIS

The present study comprised four parts: recording of the muscular activity, calculation of the loading moment induced by the device, analysis of the forces required to counteract the applied load, and measurement of maximum muscular moment exerted by the shoulder external rotator muscles.

The model used in this study for measurement of the muscular moment during maximum voluntary contraction (M_{MVC}) and for calculations of the load moment (M_{load}) caused by the cord force from the pulley is illustrated in Fig. 1. A ratio between the load moment and the maximum muscular moment indicates an estimation of the proportion of the muscular strength utilized during the movement. Knowledge of the magnitude of the load moment (M_{load}) was also used for the analysis of local joint biomechanics, and allowed estimation of forces in joint structures.

After pilot studies, three different subject positionings in relation to the pulley used as the resistance device for shoulder external rotation were chosen for a more detailed analysis.

Subjects

Seven healthy subjects with informed consent volunteered for the muscle strength test. Five of them participated in the later EMG recordings of muscular activity. Their age, height and weight are given in Table I. All the measurements were performed on the right (dominant) arm. Before starting, the subjects were informed about the test and exercise procedures.

Exercise and training device

A pulley apparatus (LIC, Solna, Sweden) was used as the resistive device (Fig. 2). The force from the pulley cord (as well as that from the dynamometer during the strength tests) was applied perpendicular to the longitudinal axis of the gleno-humeral joint about which external and internal rotation occur. There were no flexing or extending moments acting about the elbow joint. The training exercise analysed was as follows: the subject was seated as shown in Fig. 2 on a bench with a support that held the right upper arm at a 30-degree abduction angle. The elbow was flexed 90 degrees and the forearm was unsupported. The subject was instructed to perform an external rotation in the gleno-humeral joint. The movement was performed in a motion sector from an 80-degree inward to a 60-degree outward rotated joint position as shown in Fig. 3 (in the following referred to as −80 degrees and +60 degrees respectively). The duration of the performance was approximately four seconds (a mean velocity of 35 degrees/s). The distance from the shoulder to the pulley,

Table I. *Sex, age, body weight and height of the 7 subjects participating in the study of maximal muscular strength for shoulder external rotators*

The 5 first subjects also participated in the EMG recordings during the pulley exercises.

No.	Sex	Age (yrs)	Body weight (kg)	Height (m)
1	M	28	88	2.00
2	M	27	72	1.89
3	M	43	78	1.87
4	F	37	65	1.69
5	F	34	60	1.71
6	M	24	67	1.74
7	F	22	55	1.62
$\bar{X}$		30.7	69.3	1.79

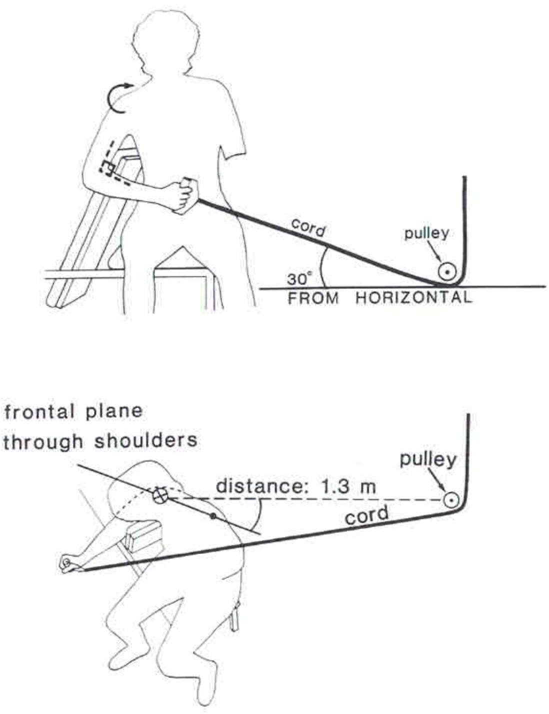

Fig. 2. Subject in seated position during measurements of maximum muscular moment and during pulley exercises. Right upper arm of subject supported in 30-degree abducted position. Elbow flexed 90 degrees, unsupported. Resisting pulley cord adjusted to maintain 30 degrees to horizontal plane through whole movement sector.

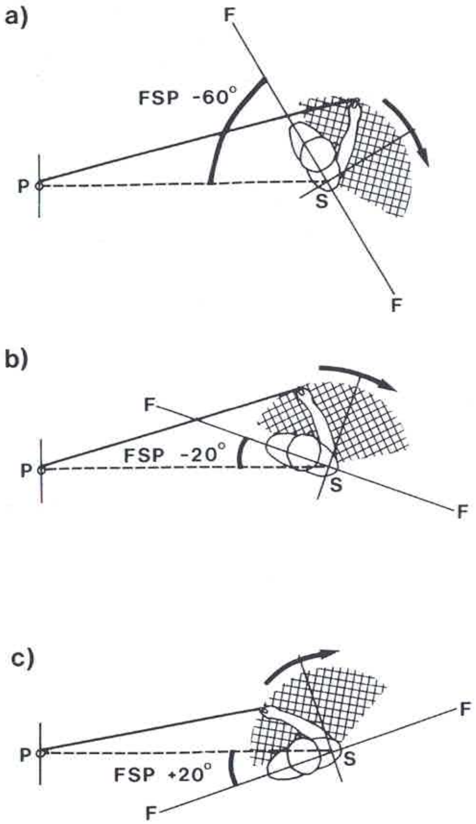

FIg. 3. Subject positioning in relation to pulley. *FSP:* angle between frontal plane and line from the shoulder joint to pulley at distance 1, 3 m; (*a*) −60 (*b*) −20 and (*c*) +20 degrees.

chosen as suitable from pilot study results, was 1.30 m in all the exercises.

To describe the different subject positionings during the exercise, a frontal plane through the shoulders (F-F in Fig. 3) and a line from the exercised shoulder to the pulley (S-P, broken line in Fig. 3) were defined. The angle between these two planes (frontal/shoulder-pulley, FSP in Fig. 3) will in the following be referred to as the subject positioning angle (FSP). A minus sign indicates that the subject was seated with the back somewhat turned towards the pulley. Subject positioning angles analysed were: (*a*) FSP=−60 degrees (Fig. 3 *a*). In this position the applied weights in the pulley gave cord forces of 13 Newtons (N) and 80 N. (*b*) FSP=−20 degrees (Fig. 3 *b*). Applied weights gave cord forces of 5 N, 13 N, 34 N and 80 N. (*c*) FSP=+20 degrees (Fig. 3 *c*). Applied weights gave cord forces of 13 N and 80 N. The applied weights were selected on the basis of the pilot studies.

Maximum muscular moment

Measurements of maximum strength were performed with the subejcts seated in the investigated exercise posture described above (Fig. 2). The maximum isometric voluntary muscular moment (isometric MVC) for the shoulder external rotators was measured at every 20th degree in the movement sector defined above. The sequence of joint angles was randomized. The duration of each contraction was 5 s and the subjects were allowed to rest for about 30 s between the contractions. A mechanical dynamometer (Salter 235, PIAB, Åkersberga, Sweden) was applied at the distal end of the forearm, and the perpendicular distance from the long axis of the arm to the line of the dynamometer force vector (the moment arm) was measured with a millimetre-gauge. The moment exerted by the muscles (i.e. muscular moment) was than calculated as the product of the force and the moment arm for the different joint angles.

Muscular activity

The muscles investigated were the infraspinatus, the posterior part of the deltoid muscle, the clavicular part of the pectoralis major and the upper portion of the trapezius. Muscular activity was recorded by means of surface (Ag–AgCl) electrodes applied in a standardized manner with a 0.03 m centre distance over the muscle belly parallel with the muscle fibres. The EMG signals were full-wave rectified, low-pass filtered and averaged over time using a time constant of 0.2 s (Devices amplifier AC 8). For control of possible disturbances hidden in the integrated EMG, direct EMG was recorded in parallel on a UV-recorder (Honeywell, Visicorder 1508). In order to make possible comparisons of levels of activity between different muscles or between different subjects, the EMG was normalized in the following way. A ratio (TAMP-R, Time Averaged Myoelectric Potential Ratio) was formed through division of the activity level recorded during the investigated exercise by the activity level during a maximum voluntary isometric contractions. The reference ('maximum') activity levels of the different muscles were obtained with static resistance in relevant directions for each muscle. Positions in the mid range of the full movement sector were chosen. The same type of normalization procedure has been used by Ekholm and co-workers (e.g. 7) and by Soderberg (29) among others. The muscle activity was correlated to the joint angle by use of an electrogoniometer (a potentiometer connected to a Devices amplifier) which measured the degree of joint rotation. Muscle activity was recorded continuously between 80 degrees inward and 60 degrees outward rotation, but determined only every 20 degrees.

With the greatest cord force, 80 N, some of the subjects were unable to complete the movement. The breaks (indicated by the length of curves of single subjects to the right in Figs. 6, 8, 9) occurred in different parts of the motion sector. Mean muscular activity levels in Fig. 7 was calculated for all values before each break-point.

Calculations of joint load moment induced by pulley apparatus

To calculate the loading moment of force about the shoulder, a simulation model was developed. The model was transformed into a computer program (FORTRAN 77) and run on a Nord-10 computer. The model used the investigated values for the subject-positioning angle (FSP) and the applied force in the pulley cord. The distance from shoulder to pulley was 1.30 m (as in the experimental part). The distance used from long axis of arm to handle in the hand was 0.35 m. The program calculated the load moment caused by the pulley apparatus about the longitudinal axis of the gleno-humeral joint for every angle in the range of motion. As the motion during the experiments was performed at low speed, dynamic forces were disregarded in the calculation model. Also, the load moment caused by the weight of the forearm and hand was excluded due to their negligible contribution (about 0.8 Nm) at this gleno-humeral abduction angle (30 degrees).

Calculations of forces in joint structures

To calculate the approximate magnitude of the load on different joint structures during the exercise, we chose the subject-positioning angle FSP=−20 degrees and the cord force 13 N with the joint angle 0 degrees (Fig. 3 *b*). Statical mechanical analysis was used (20).

The internally rotating load moment at this level of cord force was counterbalanced by the dorsal rotator cuff muscles (the infraspinatus and the teres minor) and the tendon force of these muscles was calculated. The geometry was reconstructed from anatomical sections (10, 19). Assumptions about distances and muscle vector angles were confirmed by dissection of one specimen. The angle between a sagittal plane and a cranio-caudal plane through the spine of the scapula, used in the calculations, was determined from goniometer measurements of the subjects participating in the study, with the arm in a 0-degree rotated position (i.e. the position analysed below).

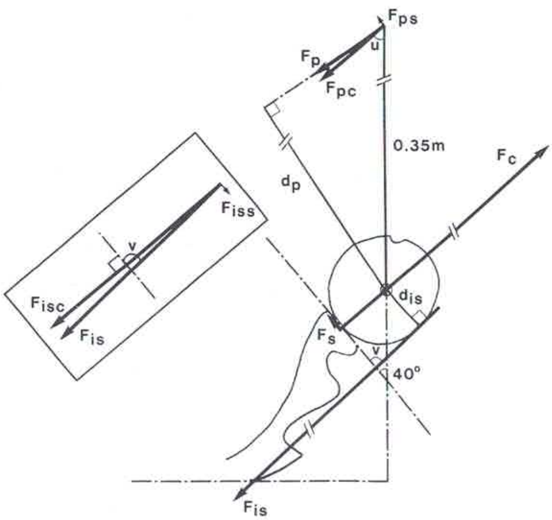

Fig. 4. Transverse plane through the gleno-humeral joint showing forces, distances and angles during pulley-resisted external rotation with joint in zero position. Subject positioning FSP = −20 degrees. F_p: pulley cord force (13 N) at 0.35 m distance from point of application of this force at hand to longitudinal axis of gleno-humeral joint (x). F_{pc}: joint compressive force vector and F_{ps}: shear force vector of F_p. *u*: angle of the force F_p to sagittal plane. F_{is}: muscular force of infraspinatus and teres minor (with its two components in compressive (F_{isc}) and shear (F_{iss}) directions indicated to left in Fig. Angle between glenoid and sagittal plane = 40 degrees. *v*: angle between plane of glenoid cavity and force F_{is}. F_c: compressive force of gleno-humeral joint. F_s: shear force. Vectors indicated are not proportional to calculated forces.

The geometrical model for calculations of shoulder muscular moment, tendon force and joint compression force is shown in Fig. 4.

Calculating the load moment:

$$M_{load} = F_p \times d_p \quad (1)$$

where

F_p = force in the pulley cord

d_p = perpendicular distance from gleno-humeral longitudinal joint axis to force vector of pulley cord F_p

Calculating d_p:

$$d_p = 0.35 \times \sin u \quad (2)$$

where

0.35 m = average distance from joint longitudinal axis to point of application of cord force to hand of subject

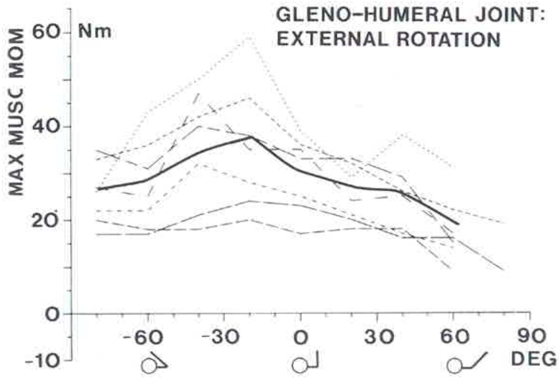

Fig. 5. Maximum isometric muscular moments of force in Nm (*y*-axis) for gleno-humeral external rotation at different joint angles (deg) (*x*-axis). Negative degree values indicate internally rotated joint positions. Broken lines: individual curves. Solid line: Computercalculated continuous mean curve of interpolated strength (n=7).

u = angle between force vector of F_p and sagittal plane

Since the load moment, M_{load}, is counterbalanced by a muscular moment, equal in size and opposite in direction, this muscular moment (M_{musc}) was calculated. The forces in the tendon structures were then calculated:

$$M_{musc} = F_{is} \times d_{is} \tag{3}$$

where

F_{is} = tne induced tendon force in infraspinatus and teres minor

d_{is} = the lever arm of tendon force F_{is}

The forces in the joint structures were separated into compressive and shear force components. Thus the compressive joint force was calculated:

$$F_c = F_{isc} + F_{pc} \tag{4}$$

where

F_c = compressive joint force

F_{isc} and F_{pc} = compressive force vector component from F_{is} and F_p respectively

The shear joint force was calculated:

$$F_s = F_{iss} + (-F_{ps}) \tag{5}$$

where

F_s = shear joint force

F_{iss} and F_{ps} = shear force vector component from F_{is} and F_p respectively

RESULTS

Maximum muscular moment

Fig. 5 shows the isometric maximum muscular moment exerted by the external rotators at different rotated angles in the gleno-humeral joint. The peaks were in the interval between 20-degree and 40-degree internally rotated joint positions. Peak average muscular moment occurred at about 20 degrees internally rotated gleno-humeral joint angle. As expected, there was great variation in the maximum moment (19–58 Nm) for the different subjects.

Load moment and muscular activity—effects of subject positioning

The calculated load moments at different joint angles during the external rotation are shown in Fig. 6 (left upper diagram, dashed lines). The filled circles indicate the average muscular strength of these five subjects also investigated for level of muscular activity. The particular subject positioning (frontal plane/shoulder/pulley line angle) is shown at the top to the right. The eight lower diagrams in Fig. 6 show the level of muscular activity for two of the cord forces investigated, 13 N (left) and 80 N (right) (presented as normalized EMGs throughout the sector of motion).

As can be seen in Fig. 6 (top left), the best possible co-variation between muscular (filled circles) and load moment of force (dashed lines) was achieved with the subject positioned to give a frontal plane/shoulder/pulley line angle (FSP) of −20 degrees, i.e. the direction to the pulley was 20 degrees from behind (Fig. 3). Of the muscles investigated, the infraspinatus reached the highest activity levels for both cord forces 80 N and 13 N. The levels of muscle activity were generally lower at the beginning of the movement sector (internally rotated positions) but increased more initially when greater load was used. The posterior part of the deltoid and the upper part of the trapezius showed low activity levels for cord force 13 N, but reached higher levels of acivity for 80 N. The clavicular portion of the pectoralis major, which is an internal rotator and thus an antagonist, showed very little activity at any cord force.

Fig. 7 shows that the mean muscular activation curves for the infraspinatus followed the same sequence as the corresponding cord forces. The mean activity levels from the infraspinatus varied from 1.2 to 0.4 TAMP-R depending on the cord force applied (80, 34, 13 and 5 N).

A subject positioning of FSP=+20 degrees (Fig. 8) gave a discrepancy between muscle strength and load moment resulting in no or low load at the beginning of the movement and an extensive load at

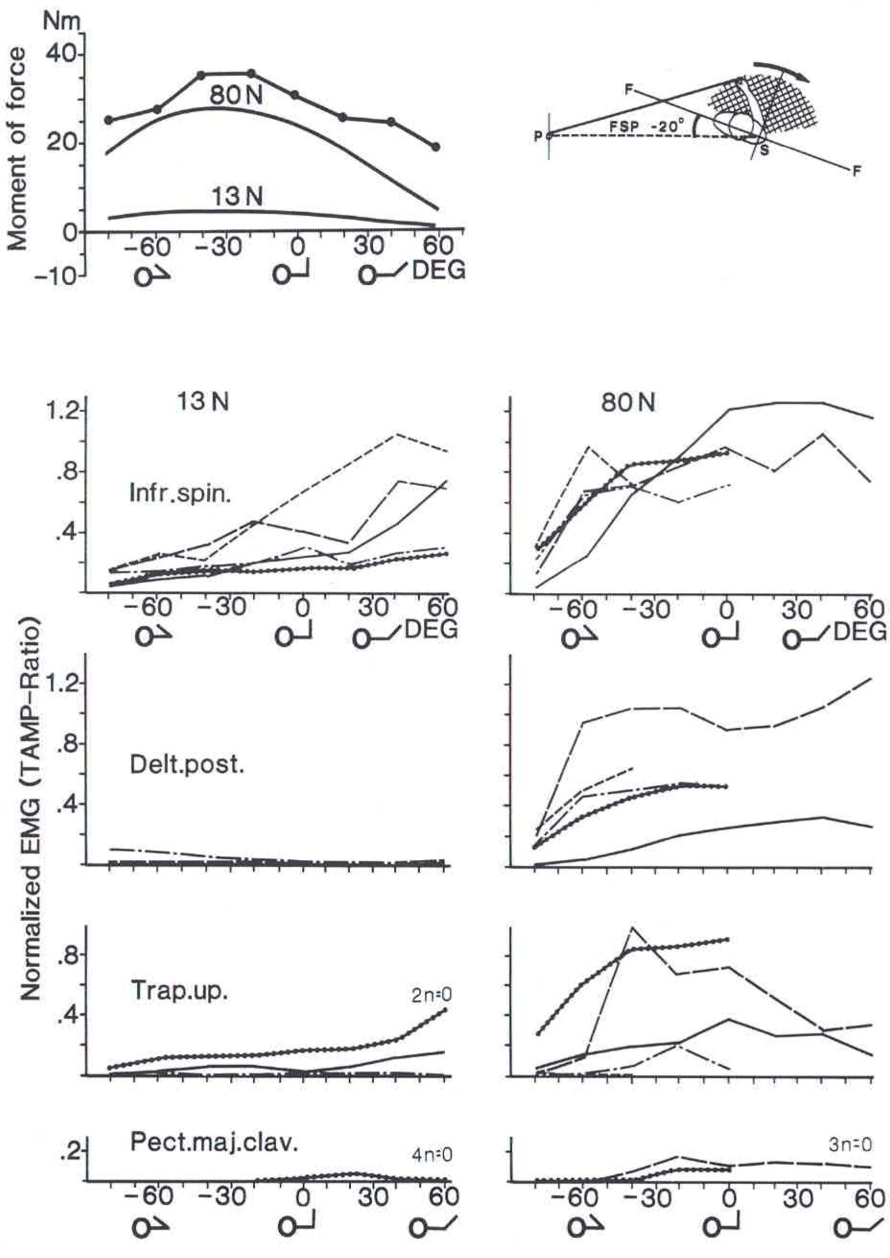

Fig. 6. Load moment, muscular strength and level of muscular activation. Top right drawing: subject positioning angle FSP = −20 degrees (see Fig. 2). Top left diagram: Moment of force (Nm) on *y*-axis and joint angle (deg) on *x*-axis. Negative degrees: internally rotated joint angles. Dashed lines: Calculated loading moment caused by pulley. Cord forces: 13 N and 80 N. Filled circles: Mean (n=5) of maximum isometric muscular externally rotating moment about gleno-humeral joint. The eight diagrams below show level of muscular activity as normalized EMG in TAMP-R (*y*-axis) in relation to joint angle (*x*-axis) for pulley cord forces 13 N (left) and 80 N (right) (n=5). Muscles: Infraspinatus (Infr. spin.), posterior deltoid (Delt. post.), upper part of trapezius (Trap. up.) and clavicular part of pectoralis major (Pect. maj. clav.) from top to bottom. Length of curve lines (80 N) indicates how far a subject was able to complete the movement.

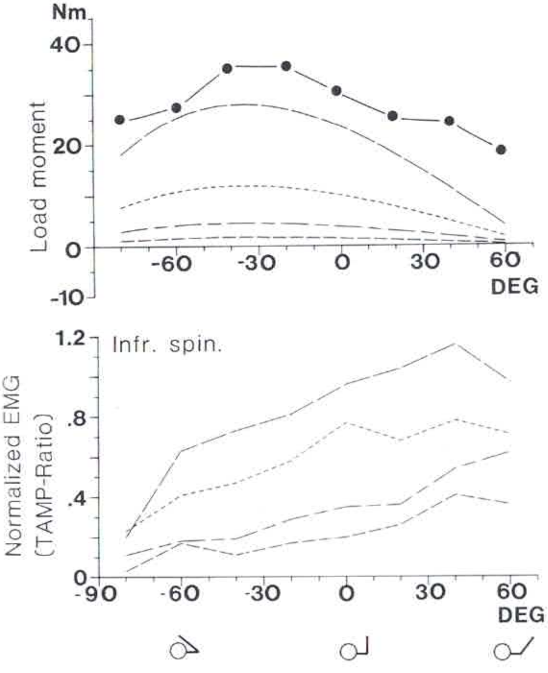

Fig. 7. Upper diagram: Load moment (*y*-axis) for cord forces 80, 34, 13 and 5 N (broken lines from top to bottom), joint angles (*x*-axis), subject positioning FSP = −20 degrees. Lower diagram: Mean level of muscular activity in infraspinatus (n=5) as normalized EMG in TAMP-R (*y*-axis) in relation to joint angle for corresponding cord force.

the end. The levels of muscle activity showed the same kind of basic pattern as for the −20-degree subject positioning angle. Using cord force 13 N, the infraspinatus was the most activated of the muscles investigated. The posterior deltoid showed very low activity levels except for one subject where the activity level rose at externally rotated joint angles. The upper trapezius and the clavicular part of pectoralis major showed very low activity levels for all subjects with this force. With cord force 80 N the activity level increased in infraspinatus more initially to high levels. The posterior deltoid showed low or moderate activity levels for four subjects and high levels for one subject. The upper trapezius showed low or moderate activity levels and the pectoralis major still showed very low activity.

With a subject positioning of FSP=−60 degrees (Fig. 9) there was an extensive load moment at the beginning of the movement (internally rotated joint angles) with cord force 80 N. In the last part of the motion sector the load moment was very low. But as the cord of the pulley apparatus here came in contact with the subject's trunc, the results from the calculation model (shown in Fig. 9) will be too low at the end of the movement. The levels of muscular activity at 13 N cord force showed the same pattern as the other two subject positionings. The force 80 N caused a high activity level over infraspinatus in the motion sector from 60 degrees internal rotation onward. The posterior deltoid reached moderate or high activity levels for two of the subjects. For one subject a high activity level was shown from the upper trapezius and the clavicular part of the pectoralis major, but the other showed low or very low levels form these muscles.

Joint and tendon forces

The approximate force in the infraspinatus and teres minor was calculated for the subject positioning represented by a frontal plane/shoulder/pulley line angle FSP=−20 degrees (Fig. 4). The pulley cord force 13 N applied to the hand at 0.35 m distance from the longitudinal axis of the shoulder joint induced a shoulder muscular moment of 3.8 Nm when the arm was kept in the sagittal plane.

The muscle vector force of infraspinatus (and teres minor) (F_{is}) was found to be directed 87 degrees (*V* in Fig. 4) to the plane of the glenoid cavity, and its lever arm to the motion axis (d_{is}) was 0.02 m. This induced a tendon force (F_{is}) of 190 N. The compressive (F_c) joint force component was calculated perpendicular to the plane of the glenoid fossa (directed 40 degrees to the sagittal plane) through the center of the humeral head. The compressive component of the pulley force (F_{pc} = 13 N) and the muscular force (F_{isc} = 190 N) gave a gleno-humeral joint compressive force of 203 N (approx. 0.3 times body weight). The shear joint force components in the plane of the glenoid were found to be very small in this pulley position and joint angle: The shear components of the pulley force (F_{ps} = 2 N, anteriorly directed) and of the muscle force (F_{iss} = 10 N, posteriorly directed) gave a total gleno-humeral joint shear force of 8 N, posteriorly directed. This means that the glenoid and the surrounding structures must keep the head of the humerus forward with a force of 8 N to compensate for this activity (= a joint reaction shear force of 8 N, directed anteriorly).

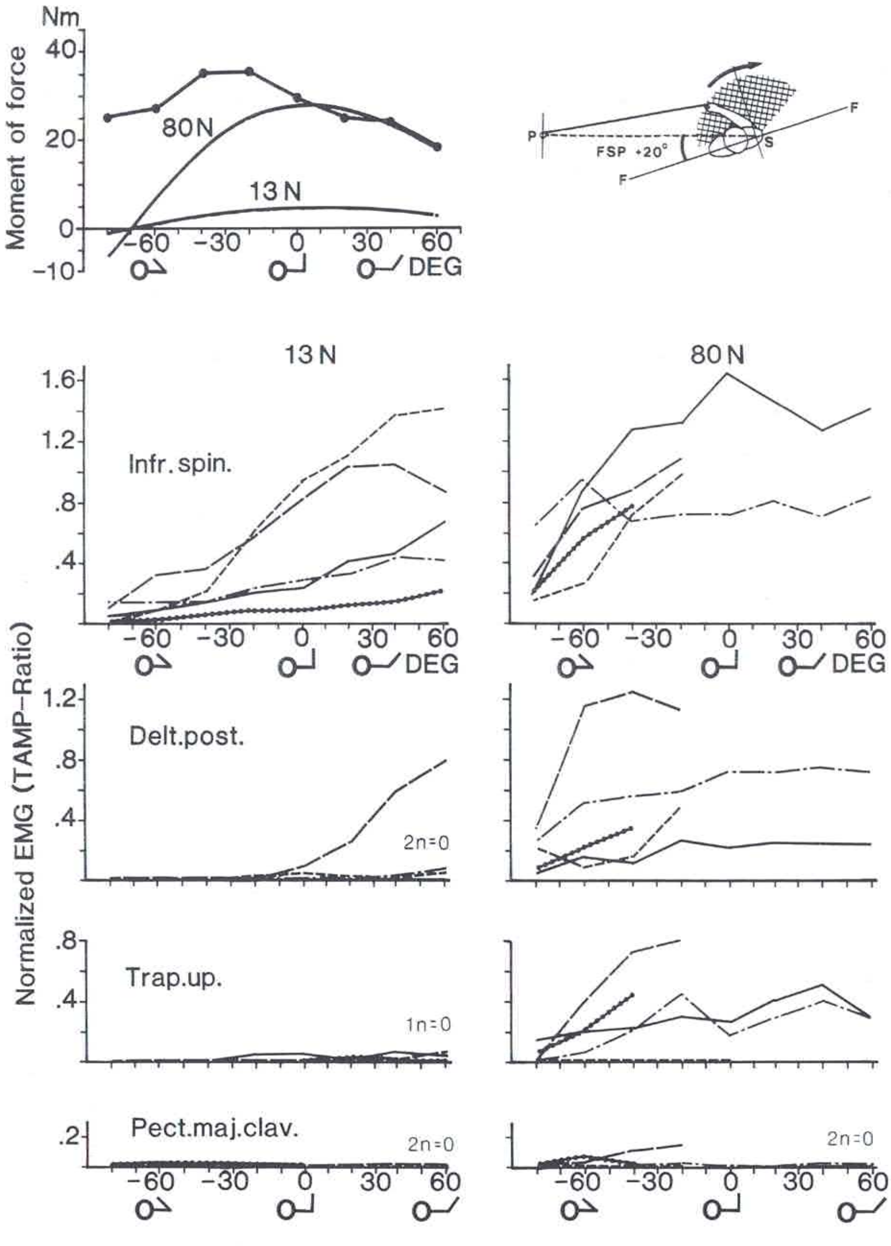

Fig. 8. As in Fig. 6 but diagrams for subject positioning angle FSP = +20 degrees. Pulley cord force: 13 N and 80 N.

DISCUSSION

Training devices require some attention to the effect of different adjustments which can be performed in order to allow a relevant resistance through the whole range of motion. In the present study the maximum muscular moment of force for isometric external rotation varied through the range of motion with regard to amplitude. The angular loci of the peak also varied to some extent. The results show that by placing the subject in different positionings in relation to the pulley, it is possible to obtain changes in the resistance in terms of the load moment curve, and this can be utilized to match the average curve of muscular strength dis-

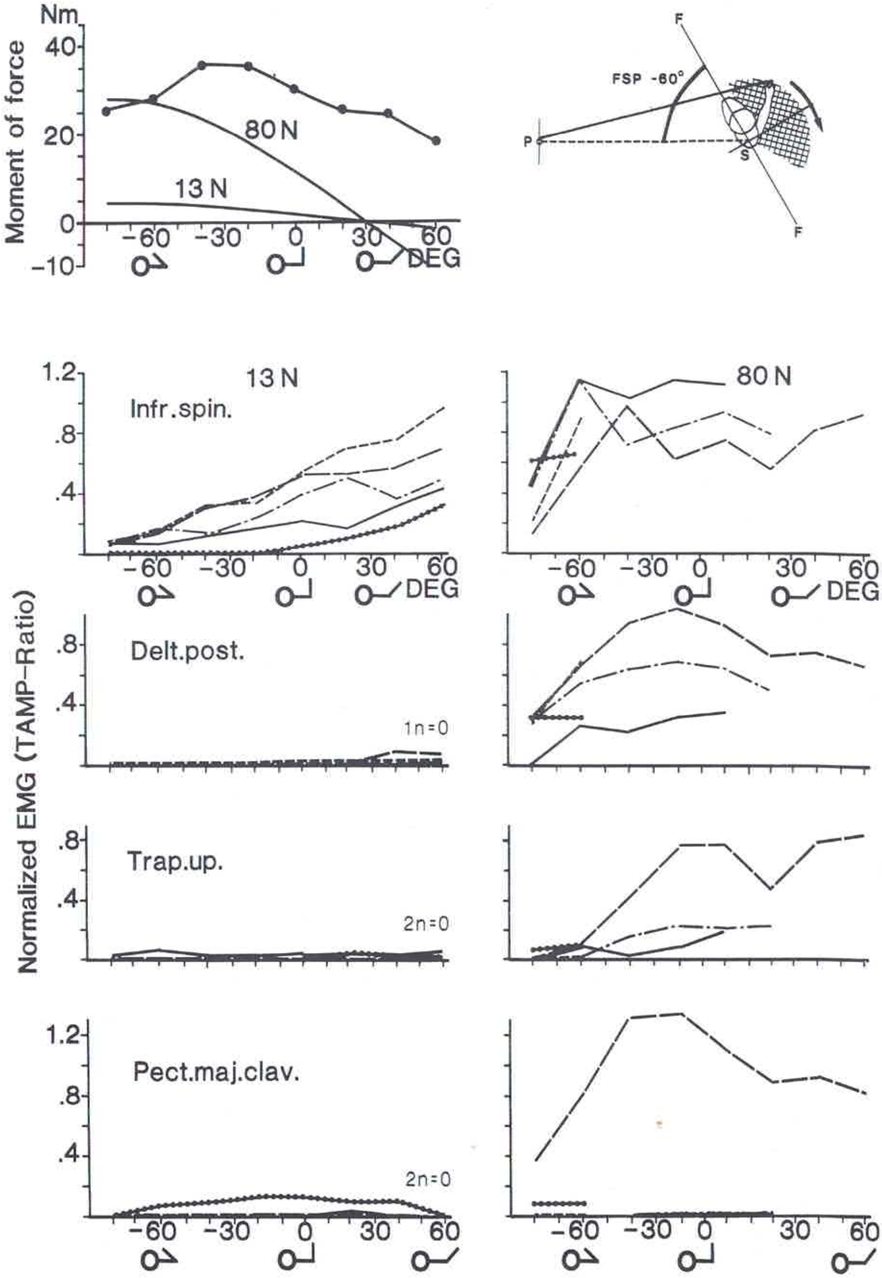

Fig. 9. As in Fig. 6 but diagrams for subject positioning FSP = −60 degrees. Pulley cord force: 13 N and 80 N.

tribution in the range of motion. For the healthy subjects studied, the optimal positioning was the one with a frontal plane/shoulder/pulley angle of −20 degrees, i.e. with the cord coming 20 degrees from behind. The subject positioning angle +20 degrees gave too low a load moment at the beginning of the motion sector, while the −60-degree angle gave too low a value at the end, for adequate strength training. An aspect that might be more important clinically is, however, that, for the same subject positionings, there appears a risk of too great a load moment at the end of the movement sector with positioning FSP = −20 degrees and at the beginning of positioning FSP = −60 degrees, when moderate or great cord forces are used.

Some patients might have an abnormal distribution of maximum muscular moment. In order to design individual exercises one might measure the maximum muscular moment at three or four different joint angles. These data can be plotted on a graph. If the load moment for different subject positionings with different cord forces (e.g. from this present study or (14) concerning shoulder internal rotation) is transferred to over head film, it is possible to compare and chose the most adequate training positioning. Thereafter, depending on the purpose of the training (strength- or endurance-increasing effects) the cord force can be chosen that gives the desired muscular utilization level through as much as possible of the movement sector. Recommendations exists for the optimal proportion of MVC for different training goals (12, 15), but further research is needed in this field also. The benefits of calculating a muscular strength utilization ratio (applied load/muscular moment) and arranging training situations where this ratio can be kept at a desired submaximal level have been discussed in other studies from our group (1, 8, 13, 14, 23). For instance, small weights such as 0.5 kg in positioning FSP = −20 degrees have been used in patients with RA for muscular training and pain reduction purposes (25).

The interruptions, when the greatest pulley cord force 80 N was used, did not always occur before or when the load moment had reached its maximum. One possible explanation could be that there might be a delay of fatigue in relation to the strength utilization peak. But load sharing might also occur between structures, causing extreme load on some structure that gives pain and strength inhibition, e.g. in some part of the tendo-osseous juntion of origo or insertion. For explanation, further research is needed.

The described method of estimating the required force in the rotator cuff tendons contains several assumptions, and should be considered as an approximate calculation, and the results should be considered as a contribution to the discussions about the magnitude of strain during exercises. The values were calculated for one specimen and will change with sex, age and individual factors. The load will change with joint position (ranging from 0.6 to 4.5 Nm for the situation we calculated) and applied load. In this study EMGs were recorded from some selected muscles. Co-contractions in the supraspinatus, the subscapularis and the anterior deltoid can add more compressive force to the joint. The compressive joint reaction force of 0.3 times body weight might be compared with that of 0.9 times body weight, which is the force developed when holding the arm in a 90 degree abduction positoin according to Poppen (1978) (27).

An increase of the muscle activity level when the muscles were shortened could be seen in our study. This is a well-known phenomenon (summarized by Winter (32)). The different EMG curves recorded over the infraspinatus had the same sequence as the corresponding load moment curves. It should be noted that EMG reflects muscle activity and not the force exerted by the muscle. The activity levels exceeding 1 TAMP-R from the infraspinatus and the posterior deltoid are probably explained by the fact that they contract concentrically during the exercise in contrast to the isometric test for the reference activity level.

The clavicular part of pectoralis major can mechanically be considered as an antagonist to infraspinatus. High levels of activity were recorded from pectoralis major for one subject at high cord forces with positioning FSP = −60 degrees, e.g. with the pulley much behind the subject. This might be a reaction to the extending loading moment caused by the pulley in that position.

There are discrepancies in the literature regarding activity in the posterior deltoid during external rotation movements (4, 7, 28, 30, 33). This diversity might be due to differences in magnitude of resistance. In the present study the deltoid was activated to a high extent when high resistance was used. With lower resistance it was only little activated. Sullivan et al. (30) as well as Ekholm et al. (7) found high levels of activity with different movement patterns regardless of external or internal rotation involved when combined with abduction motion.

From long term training studies of rheumatic patients there have been reports of positive effects of exercises (24). For rheumatic joints a biomechanical consideration of the loads during the exercise ought to be of special importance. Subject positioning with the frontal plane/shoulder/pulley line angle −20 degrees, as was suggested as optimal in this study, has been used as training position in a clinical study (25) where patients with rheumatic shoulder pain reported less pain after a training period compared with a control group. It is suggested that the method described in the present study might be used in the design of training exercises.

CONCLUSIONS

The present study shows how to avoid unintentional excessive load in parts of the movement sector duing therapeutic exercises using a pulley apparatus. The load moment can be adjusted to a magnitude appropriate to the intention of the training almost throughout the range of motion.

Adjustment of the subject's position in relation to the pulley will change the relation between the maximum muscular moment about the longitudinal axis of the gleno-humeral joint and the induced load moment at different joint angles. The load moment for the corresponding joint angle can thus be changed for the same pulley cord force. For healthy subjects, the best adaptation between these two parameters was found when the frontal plane/shoulder/pulley line angle was −20 degrees, with the pulley a little behind the subject (at a distance of 1.30 m from the joint). It is implied that this training positioning can be used by patients with a generally lowered muscular moment, e.g. patients with rheumatic or osteoarthrotic disorders. Their distribution of the maximum muscular moment throughout the movement sector can be assumed to be approximately the same as for the healthy subjects, but the magnitude will be at a lower level.

To find the optimal positioning where for example tendinitis or partial tendon ruptures, it may be in the supra- or infraspinatus muscles, are assumed to be interfering with the magnitude of muscular strength in parts of the movement sector, it is suggested that maximum muscular moment be measured at three (or more) different joint angles. Comparisons can then be made between these data and the different load moment curves presented in this study. When the patient has been positioned in the optimal frontal plane/shoulder/pulley line angle, the level of the patient's maximum muscular moment and the purpose of the therapeutic exercise will determine the choice of applied pulley cord force.

The average level of muscular activity during the exercise follows the same sequence as the level of load moment caused by the pulley apparatus. This exercise thus activates the infraspinatus more than the other muscles studied.

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