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SHOULDER JOINT LOAD AND MUSCULAR ACTIVITY DURING LIFTING

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ABSTRACT. The aim of this study was to investigate load on shoulder joint and muscles in healthy adults during four different ways of lifting a moderate burden (12.8 kg) from floor to table level. The methods used were surface electrode EMG with linear envelope, and calculation of loading moments of force using a static biomechanical model. The shoulder load at the beginning of the lift was lowest for the straight knee lift. The peak load occurred in the late phase of all lifts. On completion of the lift, load and activation were considerably lower when the subject was allowed to step forward before putting the box down. The most activated shoulder muscles were the anterior and lateral parts of the deltoid, and the serratus anterior. Around 80% of the shoulder load moment was caused by the weight of the burden. Some 75% of maximum shoulder strength was required for lifting the burden.

Key words: Lifting, biomechanics, electromyography, ergonomics, shoulder joint

Some Swedish investigations, reviewed by Hagberg (12), show that 18% of a normal population reported pain from the neck and/or shoulder, and, in another investigation of normal Swedes, 23% said that they had suffered from neck pain during the year and 19% had experienced shoulder pain. In some occupations these figures are even higher. Some reported figures for prevalence of neck and cervical spine impairment are: shop-assistants 28%, packers in the food industry 37%, liquor shop assistants 35%. These examples represent work where repetitive lifting is an essential part. A report on sick leave at an engineering company (20), showed that 16% of sick leave for periods longer than four weeks was associated with neck-shoulder disorders. This proportion increased to 70% for people absent from work for a year or more.

Mechanical factors have been discussed as possible agents for various disorders of the locomotor system in the shoulder-neck region, as well as in other regions. The symptoms have been described by Hagberg (11). Connections have been reported between work with elevated arms, manual labour in contrast to office work, number of objects lifted, static work related to shoulder-neck disorders (13, 19).

Work situations with shoulder load may roughly be divided into two types. The first is when the arms are kept elevated for a considerable period, causing the problem of static muscular contraction common among for example typists, workers assembling electronic component cards, hairdressers and dentists. The second type of work entailing shoulder load is associated with movements similar to lifting, for instance as part of more complex movements during handling of material or tools. This latter type is studied in this report.

Members of our project have reported earlier on the lumbar spine (7), hip, knee and ankle joints. Some studies on various aspects of lifting deal with the problem of maximum burden weight using direct recordings of the maximum weight a person could lift once, or the burden the subject believed he could lift repeatedly for a certain period (8, 21). Another approach is that of Chaffin and collaborators (3, 5, 9). They developed models for calculating loading moments of force on the major joints, and compared the values obtained to muscular strengh values, in order to determine maximum lifting capacity and limiting muscle groups. Further, the physiological costs of lifting have been studied and used to study effects of different burdens and ways of performing lifts (18).

The function of the shoulder joint has been described by e.g. Inman et al. (15), Dempster (6) and Poppen & Walker (25). Much of the functional EMG investigation undertaken has been reviewed by Basmajian (1). Many of these studies concern unresisted movements, most often limited to the anatomical planes. Sigholm et al. (26) attempted a more systematic study of shoulder muscle activation in relation to load.

Table I

Muscle	Electrode placing
Deltoid, ant.	Distal to origin of clavicular portion
Deltoid, lat.	Inferior to acromion
Deltoid, post.	Distal to origin of spinous portion
Pectoralis major	
Clavicular	Distal to clavicular origin
Sternocostal	Lateral to sternocostal origin
Infraspinatus	Inferior to deltoid, lateral to trapezius and superior to latissimus dorsi
Latissimus dorsi	Caudal to inferior angle of scapula
Trapezius	
Upper	On descending part, close to antero-lateral margin, midway between occiput and acromion
Lower	On ascending part, at about level of vertebra T7–8
Serratus anterior	Just caudal to medial wall of axillary fossa
Coracobrachialis	Medial aspect of uper arm, just distal to pect. maj.
Biceps brachii	On distal upper arm over caput breve
Triceps brachii	On midpoint of posterior upper arm over caput longum

EMG has been recorded during occupational activities such as car driving (17) and manipulative work at different heights (16). One method of evaluating effects on various muscles of longer periods of work is to use EMG together with signal analysis, giving information concerning the development of fatigue in the muscles. This method has been used for shoulder muscles (10, 14).

This project is concerned with the assessment of load on, and muscular activity in, different parts of the locomotor system during lifting. The aim was to obtain basic values for mechanical load and levels of muscular activation while lifting moderate loads, and to study how these values were affected by different ways of lifting. An important topic is the distribution of load over different parts of the locomotor system. An instruction for lifting, intended to give minimum load on the low back, may cause increased load on some other region, e.g. knees or shoulders. In the effort to minimize and optimize load on the entire locomotor system, it is hence important to be able to analyse the causes and the effects of loads on the various parts of the body.

The following specific questions were put:

- What is the magnitude of the loading moment of force on the shoulder joint caused by lifting moderate burdens, and how does it vary during the lift?
- What proportion of the loading moment of force is caused by the weight of the burden?
- To what level are shoulder muscles activated during lifting?
- How are the levels and time distribution of the loading moment and muscular activity influenced by different lifting techniques?

MATERIALS AND METHODS

Fifteen healthy volunteers took part in the study. In an early phase of the project, only direct EMG was used, and consequently the results could not be quantified. During the course of this investigation a new technique (described below) was developed. From the later phase of the study the results from five subjects are presented and evaluated. These EMGs folow the same trend as those from the first part of the study.

Average height was 1.85 m and weight 75.8 kg, for the subjects whose EMGs are presented. Eight subjects were similarly chosen for biomechanical calculations for three types of lifts, and five subjects for the fourth type. EMGs are presented for five of these eight subjects. Average height was 1.83 m and weight 77 kg for the group of eight subjects. No subject suffered from shoulder problems, or had undergone surgery in the shoulder region.

The task was to lift a specially designed two-handled box from the floor to a table in front of the subject. The height of the table was adjusted to the height of the subject's umbilicus. The box measured 20 cm (height), by 40 cm by 25 cm, and weighed 12.8 kg including added weights. The handles were placed 24 cm above the bottom. In each handle, a strain gauge measured the forces applied vertically. The forces from the handles were added and recorded. The subject stood with the arms hanging before each lift, and returned to the same position after completing the lift. All the lifts were performed at moderate speed and took about 2 sec. Subjects were allowed at least 2 min rest between successive lifts.

For electromyographic (EMG) recording of shoulder and arm muscular activity, pairs of flexible, disposable surface electrodes were attached to the skin over the muscle bellies, in the direction of the muscle fibres, with an interelectrode distance of 0.03 m. The muscles and the placing of the electrodes are presented in Table I. EMGs from four muscles were recorded simultaneously. Full-wave rectified, time-averaged electromyograms (linear envelope) were recorded (Devices AC8 amplifiers and pen-recorder). For control purposes the amplified, unfiltered EMG signals were recorded in parallel on a UV recorder (Honeywell Visicorder 1508). All signals were simultaneously displayed on an oscilloscope (Tektronix RM565) connected in parallel. This was used particularly for testing electrode function.

To allow intra-individual and inter-individual comparisons, a normalization was performed. Muscle activity is presented as the ratio (TAMP-R; Time Average Myoelectric Potential Ratio) of the time-averaged myoelectrical potential recorded during the experiment, divided by the activity recorded during an isometric voluntary maximum test contraction. To ensure that the test contractions were isometric, the subject was fixed to a specially designed chair. For five of the subjects, isometric maximum shoulder flexion moment of force was recorded at 90 degrees of flexion while fixed to the special chair. The resistive force was measured perpendicular to the arm with a strain gauge (Bofors KRG-4), and recorded on one recorder. To obtain maximum muscle moment, the effective moment arm was measured and multiplied by the force. Subjects were instructed to increase their contraction level for approximately two seconds, and to maintain it for three seconds.

The lifts were photographed using a motor-driven camera (Olympus OM-1, 24×36 mm film) at 4 frames/sec and a focal distance of 5 m. A vertical reference bar with length indications was placed near the subject in the focal plane of the subject. For time synchronization of the EMG recordings and the photographs, an optical time indication panel was designed. This paced the time marks of the EMG recorders and showed time on a light emitting diode display, visible on the photographs.

The films obtained were used for the biomechanical calculations. The pictures were projectd directly from the negatives on a "graphics tablet" (Tektronix digitizer, 4953). The digitizer was connected to a graphics terminal (Tektronix 4012) connected to a Nord-10 computer (16-bit mini) at the Karolinska Institute computer centre. Using the digitizer, the coordinates of the reference points and the positions of thebilateral axes of motion for the major joints were entered. Time, recorded from the light emitting diode timer, was also entered. The recorded body position was displayed on the graphics terminal, which enabled the reseacher to check his action. This mechanical model uses static mechanics. For the calculations, it was necessary to know the masses, and the locations of the centres of mass, of the relevant body segments: hand, forearm and arm. These parameters were tkane from Dempster. The loading moment of force about the bilateral axes of motion for the shoulder joints (and other joints), was calculated for each picture taken during the lift.

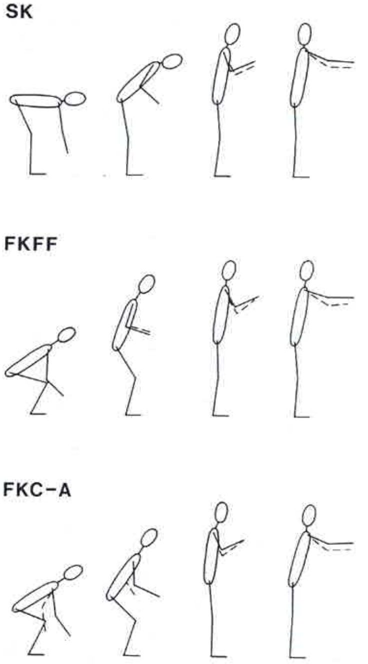

Fig. 1. Modes of lifting studied. *SK,* with straight knees; *FKFF,* with flexed knees and burden lifted in front of knees; *FKCA,* with flexed knees and burden close to body and between knees.

The loading moment of force is the sum of the moments caused by the weights of the segments and the moments caused by external forces, here the weight of the box lifted. The calculated moment of force for a certain joint axis is a useful parameter for comparisons between various ways of performing a task, and it also describes the load during the course of a task. The loading moment is counteracted by a net muscular moment of equal magnitude. The program also allowed us to change the weight of the burden for a given series of pictures. This function was used to calculate the contribution to the loading moment of the body segments alone (i.e. the posture itself), by using a burden factor of zero.

From individual strength values for five subjects and their calculated loading moments, a muscular strength utilization ratio (MUR) was calculated.

In order to display the moments calculated, a set of subroutines for graphic display (UPAG-6) was made. This enabled us first to see that analysed posture displayed on the terminal and later over a HASP communication line to an IBM 370/165 computer displayed on a cathode ray plotter (Calcomp 835), which produced a plot on film. The digitizer system was also used for rescaling the EMGs from the original recordings to uniform time and activity axes.

Lifting techniques

The following four types of lift were studied (Fig. 1):

1. Lift with straight knees (SK).
2. Lift with flexed knees and burden lifted in front of knees (i.e. far from pelvis) (FKFF).
3. Lift with flexed knees and burden between knees (i.e. close to pelvis) (FKCA).
4. Lift performed with flexed knees but subject allowed to step forward to bench before completion of lift (FKCB).

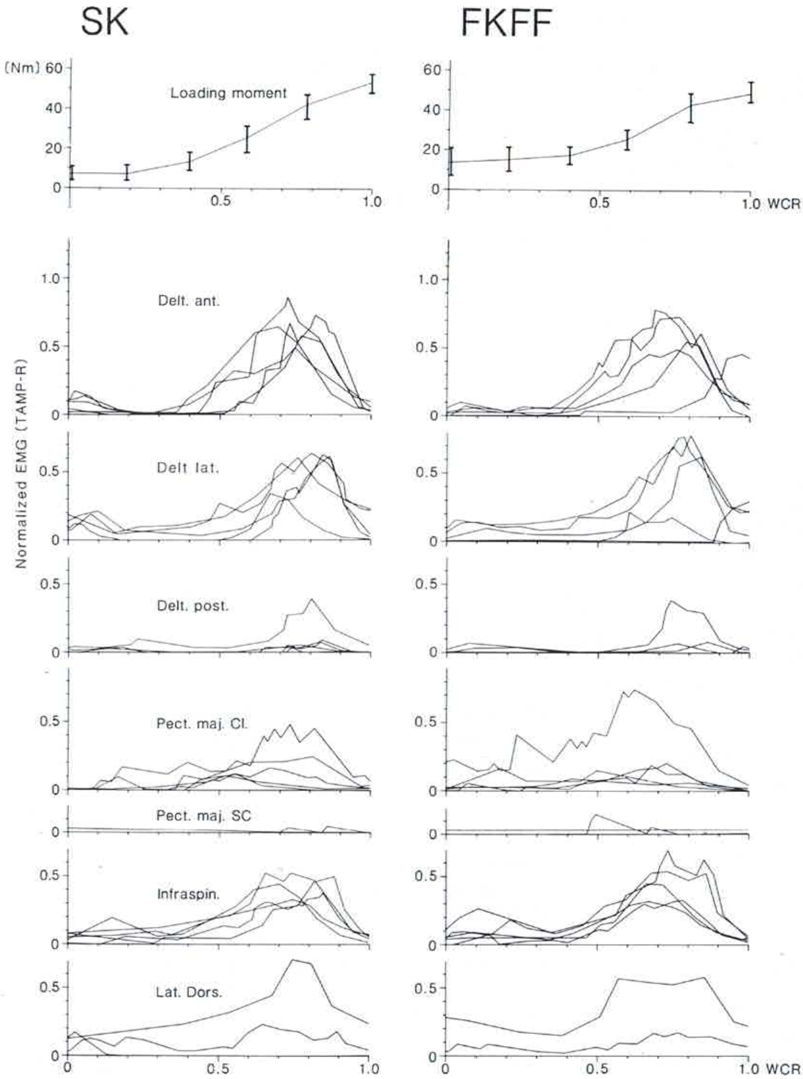

Fig. 2. The upper left and right graphs show the loading moment of force (mean with 95% confidence intervals) for the SK and FKFF lifts. Time is expressed as a working cycle ratio (WCR). The other graphs show individual muscular activity curves from five individuals from seven muscles: anterior, lateral (middle) and posterior parts of deltoid; clavicular and sternocostal portions of pectoralis major; infraspinatus and latissimus dorsi. Activity normalized to TAMP-R. "Segmentation" of curves is a result of rescaling process. "Missing" curves indicate absence of activity.

During FKCA and FKCB lifts, the subjects chose which of the handles was to be placed most anteriorly in the starting position, and thus which shoulder joint was to receive the higher load. The FKCB lift was performed by five subjects only.

Time is expressed as a working cycle ratio (WCR), where 0 corresponds to the time when the box left the floor and 1 to the placing of the box on the table.

The loading moments at the beginning and at the end of the different lifts were compared using ANOVA. Some orthogonal comparisons were included. A statistical computer program, STATPAC-IS developed at the Karolinska Institute, was used for this.

RESULTS

Loading moments

At the beginning of the straight-knee lift (Fig. 2) there was little loading moment on the shoulder joints because the arms were kept almost vertical and thus aligned with the line of action of the accel-

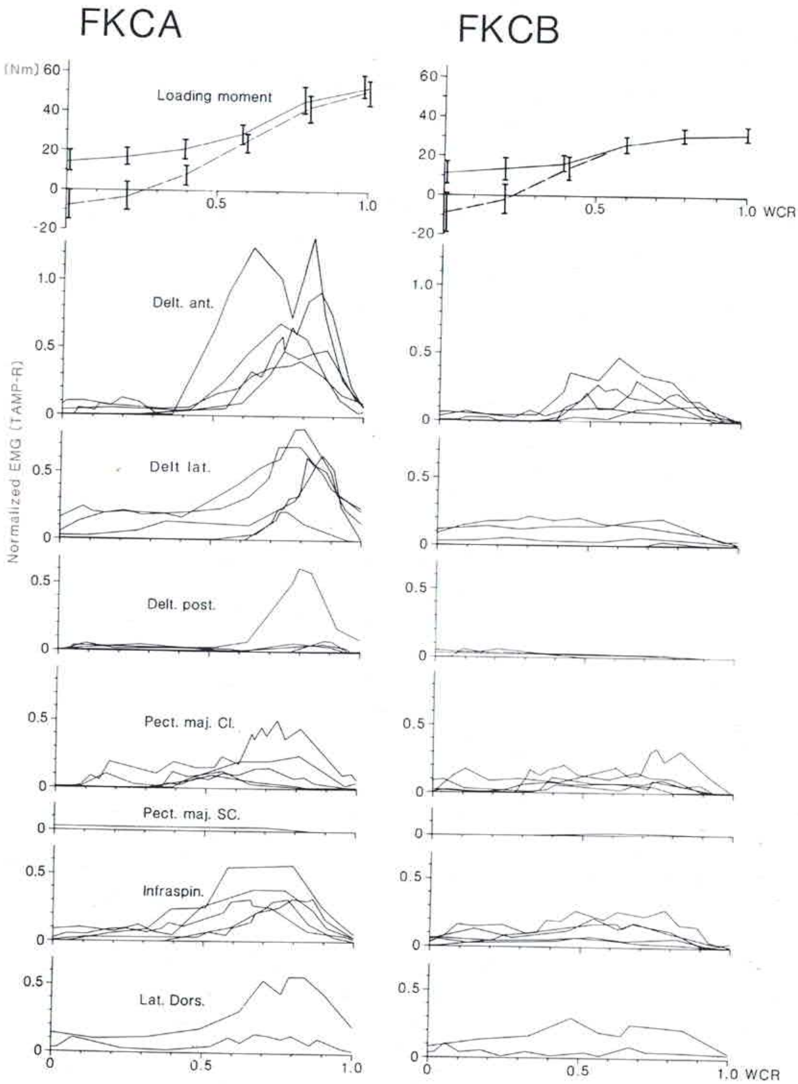

Fig. 3. As in Fig. 2, but for FKCA and FKCB-lifts. Dashed lines in uppermost graphs indicate loading moment for least loaded shoulder joint.

eration force of the box. This principal condition was not changed by the beginning flexion of the elbow joints. From about 0.2 WCR there was a steady increase in loading moment until completion of the lift. The symmetrical lift, gives both shoulder joints the same load.

Compared to the lift with straight knees (SK), the initial loading moment of the FKFF lift was higher; 14 Nm for FKFF against 8 Nm for SK lifts. The loading moment changed little until 0.4 WCR, when it increased more rapidly to the final maximum value.

The loading moment from the lift with flexed knees and the burden lifted between the knees and close to the trunk (FKCA) (Fig. 3) gave different levels of load on left and right shoulder joints, and hence these loading moments are displayed separately. Five out of eight (all right-handed) chose to take the higher load on the right shoulder joint, by holding the right hand more anteriorly. The curve representing the more loaded shoulder joint was very similar to the moment curve for the FKFF lift. The initial value was almost exactly the same.

The curve showing the loading moment of the least loaded joint was different from that of the previous curves. The loading moment at the beginning of the lift was −7 Nm, with a range between −20 and +4 Nm. All but one of the subjects were negative. A flexing loading moment was caused when the common centre of mass of the upper extremity and of half of the box was located behind a vertical line through the humeroscapular joint. At 0.4 WCR five subjects still gave negative values but the final value was close to that of the more loaded joint, namely 32 Nm. From 0.6 WCR the curve approached and ran parallel to that of the more loaded shoulder joint.

The results from the FKCB lift are shown to the right in Fig. 3. The loading moment for the less loaded shoulder is plotted from the start of the lift until it approximates the curve for the more loaded shoulder. This lift was analysed biomechanically for five subjects all of whom had chosen to take more load on the right shoulder joint. Because of the steps forward, this lift had a longer duration than the others, giving a more compressed time scale. The start of the lift was carried out in the same way as for the FKCA lift and gave a similar result. The mean for the right shoulder joints on completion of the lift was 32 Nm, which is lower than in the FKCA lift.

Statistical analysis supported the observation that the means for loading moments at the start of the lifts were different ($p<0.05$). The mean for SK lifts was significantly ($p<0.05$) less than the others. Among the others no difference was discernible. Also, the means from the ends of the lifts were significantly ($p<0.001$) different. Further analysis revealed that the difference between FKCB and the other lifts was significant, FKCB being lower.

Muscular activity

Levels of muscular activity during the four types of lift are presented in Figs. 2–5. In contrast to the loading moments, muscular activity shows much variation. Individual activity patterns for five subjects are presented.

The anterior part of the deltoid showed a tendency to bimodality of activation during the SK lift (second uppermost diagram in left column of Fig. 2). At the start of the lift there was a small peak before 0.2 WCR, after which all subjects returned to very low levels. The late maxima occurred between 0.6 and 0.8 WCR and reached levels between 0.5 and 0.8 TAMP-R. During the FKFF lift (Fig. 2, right column), the maxima resembled those of the SK lift, except for one subject. The first peak was vaguer during FKFF. In the FKCA lift two subjects showed activity near or over 1.0 TAMP-R. The others remained at about the same levels as for the SK and FKFF lifts. In the FKCB lift, with advance to table, all maxima lay below 0.6 TAMP-R, and all but one below 0.4. The maxima were more scattered in time during this lift, as seen also for other muscles.

The middle (lateral) part of the deltoideus acted similarly to the anterior part, in the latter part of the SK, FKFF and FKCA types of lift. For most subjects, the peaks reached values around 0.6 TAMP-R. During the SK lift there was a small initial peak. In the FKCB lift the levels of activity were generally lower, the highest individual maximum being only around 0.2 TAMP-R. Two individuals showed no activity.

Except for one subject, the posterior part of the deltoideus showed very little activity in all lifts. The activity in the clavicular part of pectoralis major (pect. maj. cl.) was rather unspecific compared to the anterior deltoid. Only one subject exceeded 0.3 TAMP-R. The sternocostal part of pectoralis major showed little or no activity. This is consistent with its main mechanical potential as an adductor. Only two of the subjects showed activity.

The infraspinatus is a lateral-rotator and forms part of the rotator cuff. In the first two types of lift the pattern was similar to that of the anterior part of the deltoid. During the SK lift the levels of activity had maxima up to 0.5 TAMP-R between 0.7 and 0.8 WCR. FKCB results in lower maxima than the other lifts; all maxima were below 0.3 TAMP-R.

In three out of five subjects the latissimus dorsi showed no activity. Only one subject exceeded 0.2 TAMP-R.

The levels of activity in some thoracoscapular muscles are shown in Fig. 4. During the SK lift, the upper part of the trapezius muscle showed only low activity until 0.4 WCR. In the FKFF lift the individual maxima were lower, and all levels were lower than 0.4 TAMP-R. The activity patterns were rather similar during FKCA and FKCB lifts; during FKCB all but one subject remained below 0.3 TAMP-R.

The activity from the lower part of the trapezius during the SK lift reached values around 0.5 TAMP-R. Most subjects increased their activity rapidly initially. In the FKFF lift the levels of activity reached about the same maximum values as in the SK lift. When the subjects lifted with flexed knees and were allowed to step forward (FKCB), activity in the later part of the lift was reduced below 0.4 TAMP-R.

The serratus anterior muscle is important for upward rotation of the glenoid and for protraction of the scapula. During the SK, FKFF and FKCA lifts, all subjects exhibited marked activity around 0.7 WCR, with levels all higher than 0.45 TAMP-R. Dur-

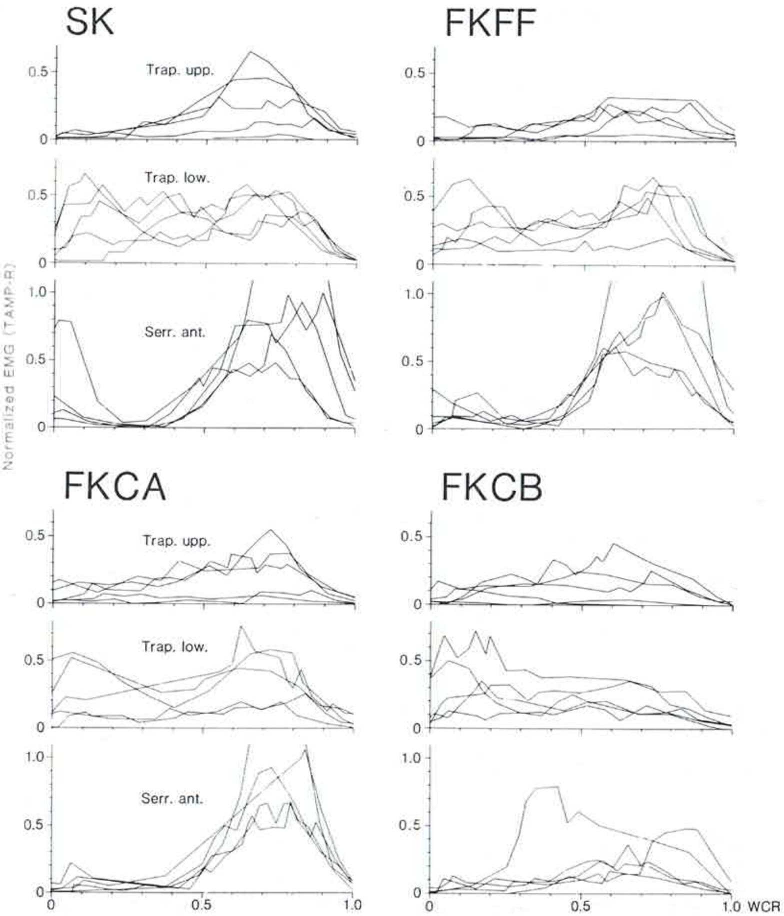

Fig. 4. Muscular activity during all four types of lifts in upper and lower part of trapezius and in serratus anterior.

ing the FKCB lift (steps forward) the levels are lower. The normalized activity from one subject exceeded 1.0 so markedly that the test contraction may have been incorrectly performed.

The levels of activity of some scapulobrachial muscles are shown in Fig. 5. Activity in the coracobrachialis is mostly below 0.4 TAMP-R in all lifts. Activity in the biceps brachii showed great variation between subjects in all types of lifts. In the first three types, individual maxima ranged between 0.2 and 1.1 TAMP-R, and individual peaks and maxima were spread along the time-scale. Activity levels were lower than 0.8 TAMP-R in the FKFF lifts and below 0.6 in the FKCB lifts. During all lifts studied, the triceps brachii gave activity levels below 0.2 TAMP-R.

In Table II data compiled from the load moment at the completion of the lifts are presented.

DISCUSSION

Assumptions

By combining mechanical calculations of joint load with electromyographic recordings of muscular activity levels, we intended to study the load on the shoulder joint and shoulder girdle during various types of lifting, and the reaction of the muscles involved when counteracting the external load. Since the lifts were performed in the sagittal plane, the loading moment of force about a bilateral axis of motion through the shoulder joint gives a measure of mechanical load on the shoulder joint and, to some extent, on the shoulder girdle.

Including measurements of the level of muscular activity gives a fuller picture of the distribution of load in the region. The EMG data both give information concerning the physiological response to an imposed mechanical load and may also, with proper

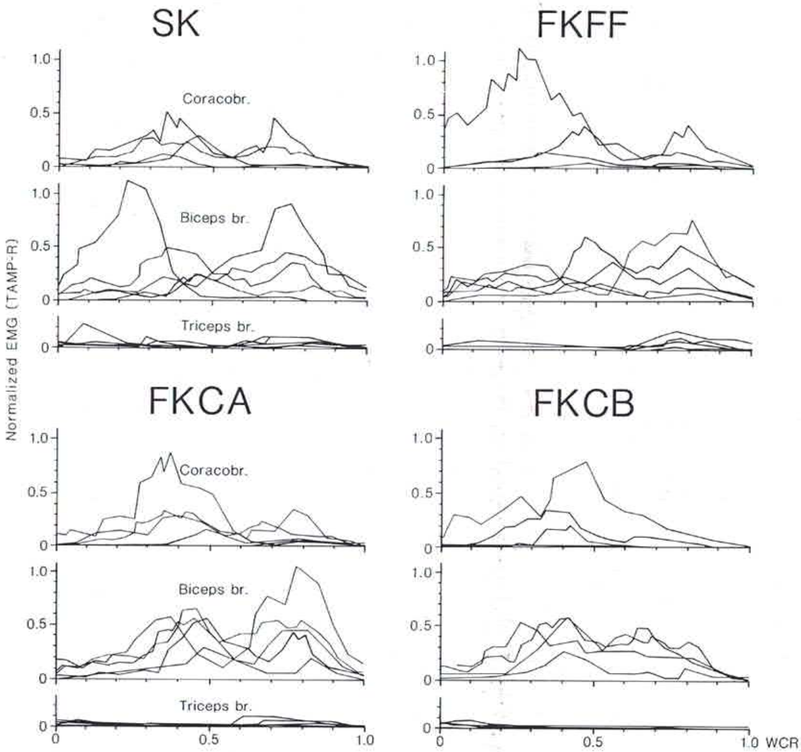

Fig. 5. Muscular activity during all four types of lifts in coracobrachialis, biceps brachii and triceps brachii.

caution, be used as a measurement of load on various muscles (1, 2, 24).

In any situation including movement, one has to consider dynamic effects on the force system (4). The real mass forces may thus differ from those calculated on the basis of statics. When dealing with slow movements, such as the lifts studied, we consider it justified to use statics because of the reduction achieved in the complexity of data sampling and calculations. Some authors advocate the use of dynamic mechanics (22), but they discuss mainly the effects on the low back, which are not the same as the effects on the shoulder joint.

An example of a curve of force vs. time, from the box handles, is presented in Fig. 6. The horizontal line shows the weight of the box. During the acceleration phase, the force in the handles is greater than the weight. When the curve passes below the weight, retardation starts. The distance from the weight line represents the linear acceleration or the retardation of the box. In this diagram, area = force × time, which equals impulse. Since the impulse received by

Table II. *Mean and half-width of 95% confidence intervals for loading moment, proportion of loading moment caused by burden, and quotient (MUR) of loading moment and individual strength*

All values are from end of lifts

	Loading moment, Nm		Fract. burden, %		MUR, %	
Lift type	Mean	I/2	Mean	I/2	Mean	I/2
SK	50	5	80	2	74	14
FKFF	49	4	81	1	72	18
FKCA	51	5	80	2	77	12
FKCB	32	4	82	5	46	10

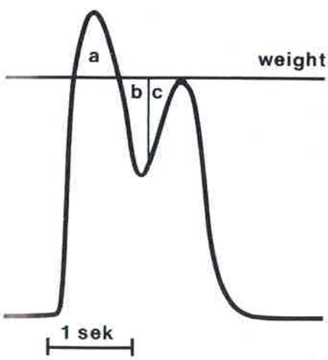

Fig. 6. Recording of force registered in handles of box. FKCA lift. Horizontal line indicates resting weight (126 N).

an object equals the change in its linear momentum, area in the diagram is proportional to change of speed. The highest peak represents an acceleration of 2.78 m/s^2 and the downward peak a retardation of 4.11 m/s^2. The acceleration corresponding to area *a* results in a speed of 0.94 m/s upwards. Area *b* equals *a*. This means that at the point in time, marked by the vertical line, the vertical speed is zero. The retardation during *b* will change to a downward acceleration during *c*, giving a "touch down" speed of 0.58 m/s. These values can be compared with the highest vertical burden speeds calculated by Leskinen et al. (22); 1.25–1.29 m/s.

In the initial phase of the lift the effective weight was higher than the weight used in the calculations of the loading moments, which will thus be somewhat too low. This effect is limited by the fact that, here, the burden was closest to the vertical line through the shoulder joints. In the middle part of the lift the calculated moment was somewhat too high, but in the final phase, where the moment is highest in all types of lift, the dynamic effect was small.

Calculation of the load moment caused by the weights of the body segments alone was performed using coordinate data sampled from lifts where the subjects really lifted a burden, but entering a burden weight of zero. The subject would probably have moved slightly differently if he had actually performed the lift without the burden, due to compensation for equilibrium in joints other than shoulder joints. Thus the calculated moment cannot be interpreted as being a proper simulation of lift with no burden, but can, however, be used for calculating the fraction of load moment caused by the burden in the lifts recorded.

It is desirable to be able to use the linear envelope signal for comparisons between muscles and between subjects. For this, some sort of normalization is needed (24). Here we related the activity levels recorded during the lifts to a reference level recorded from the same muscle, on the same occasion, during a standardized isometric voluntary maximum contraction (IMVC). Similar methods have been used by other groups (23). We believe the advantages of this technique outweigh the disadvantages. No prediction had to be made about the nature of the relationship between force and EMG level.

The loading moment in the sagittal plane for the shoulder joint is counteracted by muscular moments, and is thus transmitted to the shoulder blade. The joint and the muscles also transmit the segment cut (segment-to-segment) force. This force, which in this case represents the weights of the arm and and burden, should not be confused with the joint cut (bone-to-bone) force acting inside the joint. The downward force on the scapula must be counteracted by elevating muscles such as the upper part of trapezius, the levator scapulae and the rhomboids. Thus, the forces are transmitted to the cervical spine and to the skull. If the burden was held with the arm hanging at the side, the forces transmitted to the cervical spine and the neck joints would approach the weight of the arm and burden. If a loading moment is imposed on the scapula together with the downward force, the muscles fixing the scapula to the thorax must increase their pull. In this way the loads on the shoulder joint, on the cervical spine and on the neck joints are closely related.

Loading moments and EMGs

Of the muscles stabilizing the shoulder girdle, the upper and lower parts of the trapezius and the serratus anterior muscles can all act as upward rotators of the scapula (glenoid upwards) (6). During the later part of the lifts, an upward rotating muscular moment is needed, and at the same time there is also protraction of the scapula. The serratus anterior was highly and consistently activated during this phase of the lift (except for FKCB). Also the upper part of trapezius showed considerable activity.

In all the lifts where the subjects was not allowed to step forward, the loading moments at the end of the lift reached values around 50 Nm. The height of the table and the distance to it made it possible to put down the box in one way only. The loading moment in the initial phase was more influenced by the lifting technique than in the final phase. The straight-knee lift (SK) gave, initially, the lowest value (for both joints) because the arms were nearly vertical and the action line of the weights of the box and body segments passed near the shoulder joints.

If the object cannot be lifted in between the knees, and the lift is performed with flexed knees, then the burden must be lifted in front of the knees (FKFF). This will result in a higher load on both shoulder joints than in the straight-knee lift. This conclusion, for the shoulder joint, tallies with other investigations. Chaffin (3) deprecated strict rules for how all lifts should be carried out, allowing subjects to adapt their lifting technique to circumstances. Leskinen et al. (22) have reported that lifting with flexed knees may give higher load on the lumbar back than lifting with straight knees, when the burden is too big to be lifted close to the body. Kumar (18) found that lifting with straight knees led to the least energy expenditure.

When lifting with flexed knees and one handle held more forward than the other, the initial load on the more loaded joint was about the same as during the FKFF lift.

Initially the moment on the less loaded shoulder joint was negative for the FKCA and FKCB lifts, and had to be counteracted by extensor muscles. Since loading moments of different direction are counteracted by different groups of muscles, with different mechanical prerequisites, the levels cannot be compared directly.

We also wished to study the effect of a reduced trunk-table distance on the late phase shoulder load, when the box was put on the table. This was the FKCB type. Reducing the distance led to a reduction of the mean maximum value by 23 Nm (42%). Thus it is important to maintain as short a horizontal distance as possible between body and burden in all parts of the movement.

The main sources of muscular flexing moment at the shoulder joint are the anterior part of the deltoideus and the clavicular part of the pectoralis major (6). The coracobrachialis also contributes but this is a small muscle. The peak of deltoideus activity occurred at 0.6–0.7 WCR, although the mechanical load attained its highest value at 1.0 WCR. This decrease in muscular activity before reduction of mechanical load could be explained by the excentric contraction performed by the muscles during the last part of the lift. Much of the acceleration of the burden is achieved by the knee and hip extension.

Stepping forward gave considerable reduction of the activity of the anterior part of deltoideus. The coracobrachialis and the clavicular part of pectoralis major participated with low to moderate activity. Since the coracobrachialis is lying close to biceps, and since the activity was recorded using surface electrodes, part of the activity recorded may have originated from biceps.

Two muscle "capsules" stabilize the shoulder joint. The outermost layer is the deltoideus. The innermost consists of subscapularis, supraspinatus, infraspinatus and teres minor, together constituting the rotator cuff. In this study, the rotator cuff is represented by the infraspinatus muscle. This muscle was quite active, in a pattern resembling that of the middle and anterior parts of deltoideus. The activity in infraspinatus, and possibly other rotator cuff muscles, could be explained by the need to fix the humeral head of the glenoid cavitas, but an externally rotating moment may also be needed to counteract the internally rotating moments set up by the anterior part of deltoideus and by the clavicular part of pectoralis major (if this tendency is not counteracted by the grip).

The middle (lateral) part of deltoideus was activated similarly to the anterior part. There was little abduction, in relation to the sagittal plane, in the glenohumeral joint during the lift. However, as a lift proceeds and the scapula is protracted and rotated anteriorly, the middle part could contribute to the elevating (flexing) moment of force. The activity in the middle part probably reflects both this mechanism and a stabilizing function.

The posterior part of the same muscle gave very little activation. This is an example of how different parts of a muscle can be activated almost independently (7). During the types of lift studied the sternocostal part of pectoralis major also showed very little activity, in contrast to the clavicular portion.

The maximum for the part of the loading moments caused by the weight of the burden was 82%. Such values depend on the weight of the burden, here 12.8 kg. These values give information concerning what could be gained by reduction in burden weight. In this case the loading moment on the shoulder joint could be reduced by up to 82% by reducing the weight of the burden. These figures are lower (30%) for e.g. the lumbosacral joint (7), since this is influenced by greater segment masses.

The loading moments calculated present mechanical load as absolute values. These values are useful when comparing different ways of performing; but they are not so easy to conceive. Is 50 Nm "much"? One answer is to relate load moment to strength, which has the same dimension, as done by e.g. Westerling & Kilbom (27) to describe load, and by Chaffin and co-workers (5, 9) to predict lifting capacity. In our study the MUR-value (Muscular strength Utilization Ratio) for both the FKCA and SK lifts was 77%. This means that when lifting a moderate burden our healthy subjects had to use some 75% of their maximum shoulder flexor strength. The actual shoulder flexion angles were close to 80 degrees during the completion of the lifts. Isometric maximum strength varies through the sector of movement, but the change from 90 degrees to 80 degrees should be small.

Conclusions

For all lifts the highest load moment was reached late in the lift.

The loading moment at the beginning of the lifts was lower for the straight-knee lift.

In the late phase, the least load was found in the lift where the burden was deposited nearer the trunk (FKCB). It is thus important that the distance from the trunk to the destination of the burden is kept short. This is supported by the electromyograms recorded in the study.

The activity in the anterior part of deltoideus was higher than in the clavicular part of pectoralis major, both considered to be important flexors of the glenohumeral joint.

There was considerable activation of the muscles between the trunk and the shoulder girdle (trapezius and serratus anterior).

To lift the moderate weight used in the experiments (12.8 kg) the healthy subjects had to use around 75% of their maximum strength.

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