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HIP LOAD MOMENTS AND MUSCULAR ACTIVITY DURING LIFTING

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ABSTRACT. The load on the hip joint during lifting was studied. Healthy subjects lifted a 12.8 kg box from floor to table level with straight and flexed knees in four different ways. The loading moment of force about the bilateral hip axis was calculated by means of a computerized static sagittal plane model. The highest load moment, 124 Nm, occurred initially in the straight knee lift and the compressive joint reaction force was 2.7 times body-weight. The lowest load moment, 82 Nm, occurred in the flexed knee lift with the burden moved close to the trunk, the compressive force was 3.2×bw. The load moment was also discussed in relation to the strength capability. The EMG levels of seven different hip muscles were normalized and expressed as a percentage of the recorded level of each muscle group during an isometric maximum voluntary contraction. The initial activity in the hamstrings was higher in the straight knee lift compared to flexed knees. The gluteus maximus was activated to a moderate level.

Key words: Biomechanics, EMG, ergonomics, joint compressive force, models biological

Slightly more than half of the reported cases of occupational diseases in Sweden are suspected to be caused by work related factors such as monotonous or strenuous movements and working positions (25). The lower limb accounts for 7.3% (25). Absence from work due to these diseases are frequently of long duration, 33% are absent 20 to 90 days (median value 24 days) (25). A careful study of the joint load enables us to give advice regarding load-reducing measures and contribute to the general "catalogue" of joint loads during different working positions. This is useful both in the occupational health service and in the work related rehabilitation medicine. The general aim of the present paper was to map the hip joint load, levels of muscular activity and movement-pattern during lifting. Our preliminary results have been reported earlier (7).

Most investigations concerning lifting deal with the low back (1, 10, 26) and one study (11) deals with the knee joint. Andriacchi et al. (2) reported the hip joint load during ascending and descending stairs. The mean maximum moment is 123.9 Newton-meters (Nm) walking up the stairs and 112.5 Nm walking down. The maximum strength of the hip extensors has been mapped (23) and the authors also have summarized the literature concerning hip muscle strength.

The EMG studies concerning hip motion show that the gluteus maximus is active during forward flexed positions (18, 19) and during extension (21, 32). Extension at the hip is usually initiated by the hamstrings (13, 28) and the gluteus maximus acts synergistically when a greater force is required (27, 28, 32). Fischer (13) noted minor activity in the gluteus maximus during lifting. The gluteus medius and minimus are activated before the gluteus maximus during extension (28). Carlsöö & Fohlin (4) observed no activity in the rectus femoris, the tensor fasciae latae and the sartorius muscles during extension of the lower limb. However, during flexion all three muscles are activated.

Johnston & Smidt mapped the hip motions in the sagittal plane during walking (15) and during different activities of daily living (16). The hip motion range during level-walking is 52°, during shoe tying (in the sitting position) with foot on the floor 121°, during stooping down to obtain object from floor 114° and during squatting 118°. The sagittal plane motions of the hip during stair-climbing has been reported to 42° (2) and 66° (16).

To our knowledge there is no study describing the load on the hip joint during lifting activities combined with a quantified EMG analysis. In the present study four different lifts were investigated and the purpose was to: (*a*) Quantify the flexing

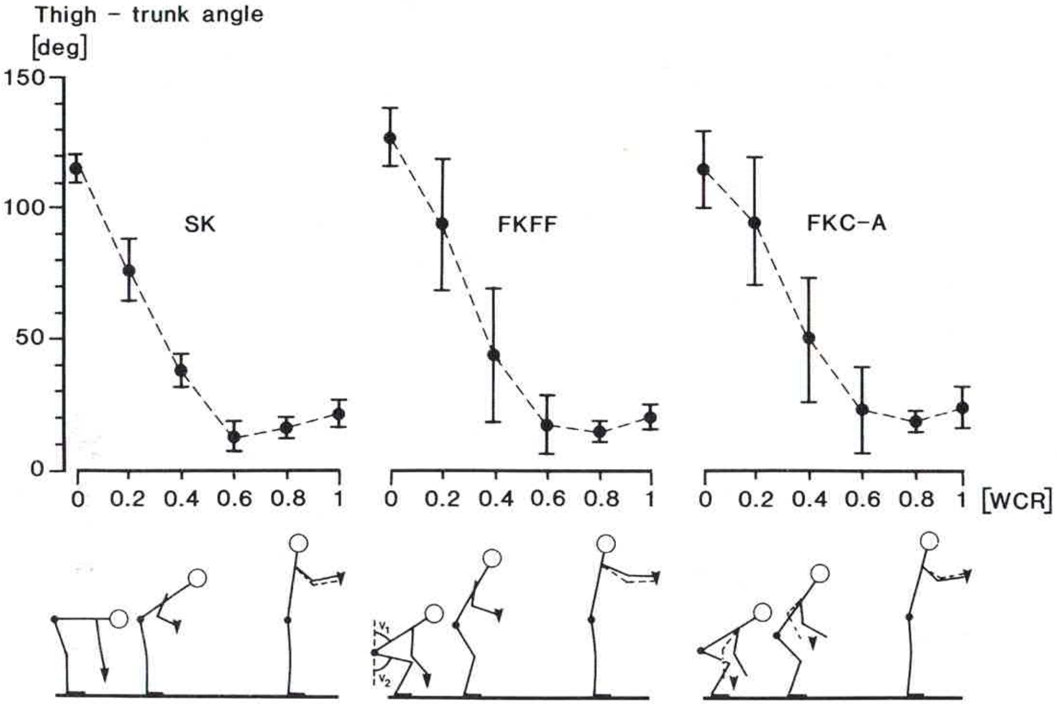

Fig. 1. Diagrams showing angle between thigh and trunk (vertical axis) during the different lifts. Time is expressed as working cycle ratio, WCR (horizontal axis). Confidence intervals are on the 95% level. SK is lifting with straight knees; FKFF is lifting with flexed knees and burden far from the body; and FKC is lifting with flexed knees and burden close to the body. Drawings below the diagrams illustrate different modes of lifting. v_1 is angle between trunk and vertical and v_2 is angle between thigh and vertical. Thigh–trunk angle is v_1+v_2. $n=7$.

loading moment acting about the bilateral hip joint axis, (*b*) map the levels of muscular activity by means of quantified EMG.

MATERIALS AND METHODS

Fifteen subjects participated in the study, mean age was 28 years (range 20–40). Their weights ranged from 70 kg to 83 kg, with a mean of 76 kg. Their heights ranged from 1.67 m to 1.89 m with a mean of 1.78 m. None of the subjects suffered from hip pain nor had previously been subjected to any joint surgery or had any periods of sick leave due to disorders of the musculoskeletal system. During the course of this investigation a new technique (described below) was developed. During the earliest phase of the project only direct EMG was available, and the biomechanical model and the calculations were not computerized. The results of the first subjects were evaluated with respect to EMG pattern and loading moment for the hip joint. These data were confirmed and extended in the latter part of the study, from which the presented graphs were taken. These graphs show results from male subjects, who are identically analyzed, with respect to the EMG and biomechanical methods. Seven subjects were used for biomechanical calculations in three of the four types of lifts and five for the fourth type. Five subjects were used to show the EMG results.

In all the lifts the task was to lift a two-handled box (12.8 kg) from floor level to a bench with the height adjusted to the level of the umbilicus. The subject stood upright with his arms hanging down, stooped down, performed the lift and returned to the same upright position. To avoid muscular fatigue the subjects were allowed at least two minutes rest between each lift. The handles of the box were 0.25 m above the bottom of the box. Each handle contained a strain gauge. The forces from the two handles were added together and recorded.

Four types of lifts were studied (their initial, middle and final phases are shown in the bottom of Fig. 1):

1. Lift with straight knees (SK);
2. Lift with flexed knees and the burden lifted in front of the knees (FKFF: flexed knees, far from);
3. Lift with flexed knees and the burden lifted between the knees and close to the pelvis (FKC-A: flexed knees, close);
4. This lift was performed with flexed knees as in FKC-A but the subject was instructed to advance (step forward) to the edge of the bench before putting the box down on the bench (FKC-B, not illustrated in Fig. 1).

All lifts were performed at moderate speed, i.e. approximately two seconds (±0.25 s). In order to make comparisons between the different modes of lifting, the time has been expressed as a working cycle ratio (WCR), which runs from 0 to 1.

The methods of the biomechanical calculations and EMG registrations have been described elsewhere (10). To summarize, the lifts where photographed (picture frequency 4 Hz) with a motor-driven camera perpendicular to the sagittal plane. From the pictures the co-ordinates of the bilateral motion axis for the major joints and reference points for length and the vertical axis were entered into a Nord-10 computer via a Tektronix digitizer 4593 and a Tektronix graphic terminal 4012. To synchronize EMG recordings and photographs an optical time-indicator panel paced time marks on all recorders and also had a light-emitting diod display with a bar representation of time. The time panel was visible on each photo. Time, recorded from the time display, was also entered into the computer. As the lifts were performed at moderate speed a mechanical model based on static mechanics acting in the sagittal plane was designed. It calculated the moments of force for the major joints from the sequence of body-positions entered for each lift. The hip loading moment was determined for one of the hips. The body segment parameters used were those reported by Dempster (5). The digitizer system was also used to rescale the EMGs to give diagrams equal in size to facilitate comparisons. From the given coordinates the computer also calculated the angles between the trunk and the vertical axis (v_1 in Fig. 1) and the thigh and the vertical axis (v_2 in Fig. 1). These two angles were then summed and defined as the thigh–trunk angle. The hip flexion angle (v_3 in Fig. 2) was also used. It was measured from the photographs taken during the experiments.

Seven hip muscles were investigated: the gluteus maximus, rectus femoris, long head of biceps femoris, semimembranosus/tendinosus, adductor magnus, gluteus medius and the tensor fasciae latae. The muscle activity was recorded by means of full-wave rectified low-pass-filtered and time-averaged EMGs (Devices M4, AC8) with a time constant of 0.25 s. Four channels of EMG could be used simultaneously. The amplified unfiltered direct EMG signals were recorded parallelly on a UV-recorder (Honeywell Visicorder 1508). This enabled a good control of possible disturbances hidden in the "integrated" EMG. On the right side of the body, two flexible disposable Ag-AgCl electrodes were attached to the skin with an inter-electrode centre distance of 0.03 m over each muscle belly in the main direction of the muscle fibers. For the thigh muscles the electrodes were placed half way between the knee and the hip. For the gluteus maximus they were placed at the middle of the muscular belly and approximately at the level of the tip of the trochanter major. For the gluteus medius and tensor fasciae latae they were placed over the centre of the muscle bellies between the trochanter major and the iliac crest, one third of the distance from the cranial point. In order to compare the EMG activity between different muscles and different individuals a normalization was performed. For each muscle we recorded the activity during an *i*sometric *m*aximum *v*oluntary *c*ontraction (IMVC). During the IMVC the subject was fixated in a specially designed chair and the hip and knee joints were both held in a midposition. However, testing the adductor magnus and the two abductor muscles the hip was held in the neutral position. Recording of the EMG activity during IMVC was performed before as well as after the experiment and the highest obtained activity was used as reference level. The normalized EMG is presented as the *t*ime-*a*veraged *m*yoelectrical *p*otential *r*atio (TAMP-R), i.e. the activity recorded during the experiment divided by the reference activity level. This method enables a comparison between intra- and inter-individual muscular activity levels (without having to find out whether the EMG is a linear function of the tension or not). The method has been described earlier (6, 8, 9, 14, 30, 33).

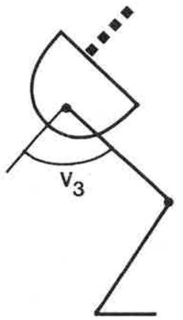

Fig. 2. Drawing illustrating hip flexion angle (v_3).

RESULTS

The magnitude and course of the thigh–trunk angle (Fig. 1) was mainly the same during the different lifts. However, during the initial and early (up to 0.4–0.6 of the working cycle ratio, WCR) phases of the flexed knee lifts FKFF and FKC the 95% confidence intervals were wider compared to the straight knee lift SK. Thus the subjects performed the straight knee lift more uniformly.

The flexing loading moments about one hip joint in relation to the working cycle are shown in the four diagrams at the top of Fig. 3. The greatest load, 124 Nm, occurred during the initial phase of the straight knee lift. During the flexed knee lifts the initial load was 105 Nm (FKFF) and 88 Nm (FKC). Except for the FKC-B lift the course of the loading moment was similar during the different lifts: High loads during initial and final phases due to long moment arms of the gravitational forces of the trunk and burden respectively. In between there was a minimum at about 0.6 WCR with the subject almost erect and the burden moved close to the trunk. Note that in the FKC-B lift the load remained low from 0.5 WCR to the end because the forward step made it possible to keep the burden closer to the trunk when the box was put down.

The level of muscular activity expressed as normalized EMG activity (TAMP-R) is shown below

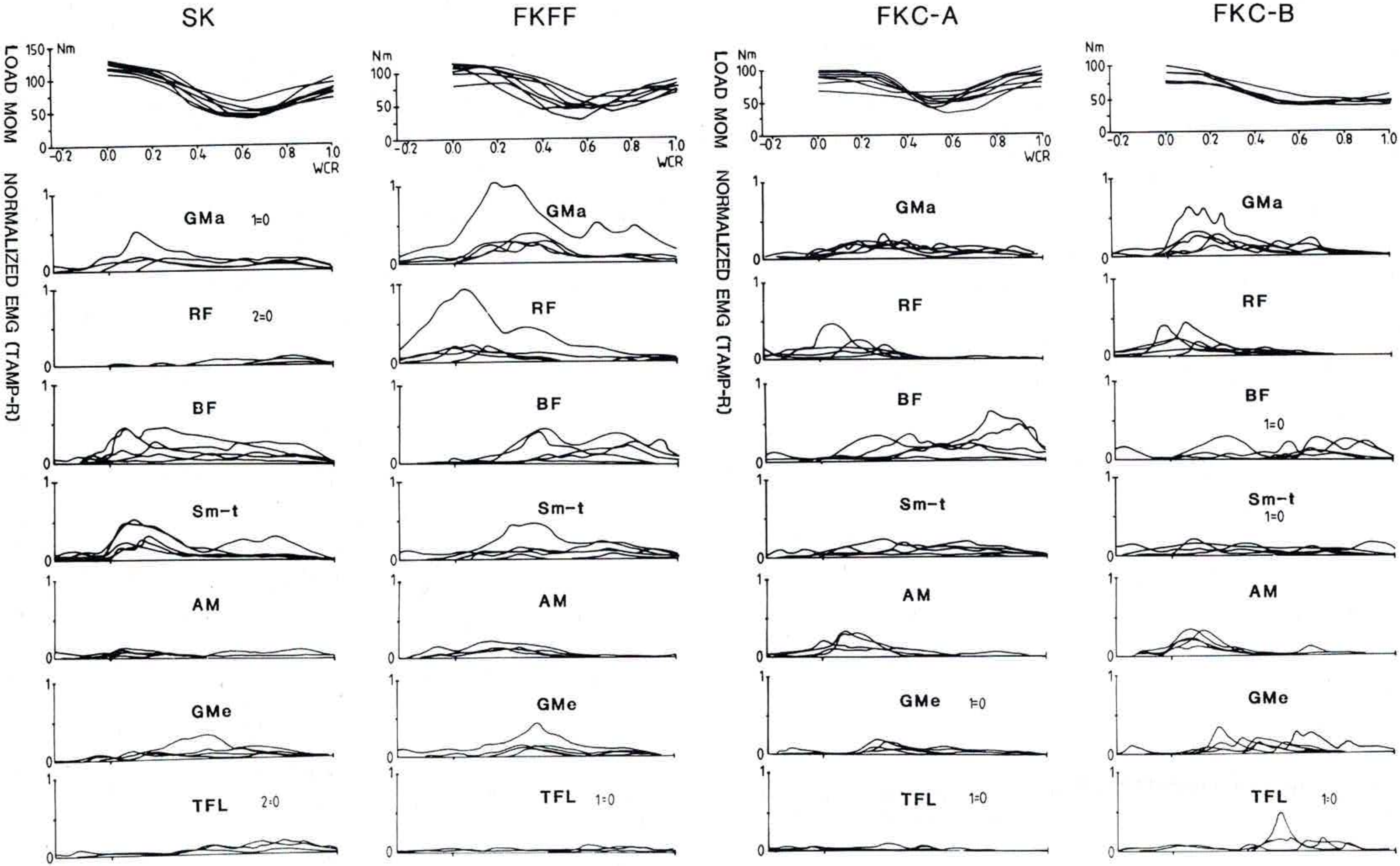

Fig. 3. Top: diagrams showing flexing loading moment of force about bilateral hip joint axis. Horizontal axis shows time expressed as working cycle ratio (WCR). Bottom: level of normalized EMG expressed as TAMP-R (vertical axis) in the gluteus maximus (GMa), rectus femoris (RF), long head of biceps femoris (BF), semimembranosus/semitendinosus (Sm-t), adductor magnus (AM), gluteus medius (GMe) and tensor fasciae latae (TFL) muscles. "1=0" means that one subject showed no activity. Values shown in the upper diagrams of loading moment SK, FKFF and FKC-A are of seven subjects and FKC-B of five subjects. In diagrams of muscular activity five subjects are shown. "Load mom" denotes the moment of force acting externally about one hip joint.

the load moment diagrams in Fig. 3. Comparing the different lifts, the gluteus maximus muscle (GMa) was activated to a low or moderate level during all lifts. In the straight knee lift one subject showed no activity (1=0) at all and another subject showed no activity at the beginning of the lift. During the flexed knee lifts the gluteus maximus of all subjects were active, and all were also active from the start of the lift. Most subjects showed higher activity in the first part of the lift compared to the latter. The rectus femoris (RF) showed mainly no activity during the straight knee lift. During the flexed knee lifts the rectus femoris showed low to moderate activity at the beginning of the lift. This is probably due to the flexing loading knee moment and is commented on elsewhere (11). The hip extensor muscles biceps femoris (BF) and semimembranosus/tendinosus (Sm/t) attained moderate levels of activity in the initial and early phases of the straight knee lift. During the flexed knee lifts, however, the activity occurred somewhat later. The adductor magnus muscle (AM), which acts as a hip extensor and adductor, was activated to a moderate level early in the FKC lifts but this activity was absent or weaker in the straight knee lift and FKFF lift. The hip abductor muscles gluteus medius (GMe) and tensor fasciae latae (TFL) were generally very little activated during the first part of the lifts but showed somewhat higher activity just before half the lift was performed. The gluteus medius was slightly more activated compared to the tensor fasciae latae. The activity late in FKC-B was partly due to the forward step.

Fig. 4 shows the mean loading moment as a function of the thigh–trunk angle. Time is indicated with symbols in the middle of the bidirectional 95% confidence intervals. In all lifts the load moment was decreasing with decreasing angle up to 0.6 WCR where the lowest loading moment occurred. Higher values were obtained in the final phase of the lifts. Both the load and the thigh–trunk angles showed greater range at 0.2, 0.4 and 0.6 WCR in the flexed knee lifts compared to the straight knee lift. Initially the range of the load was of the same magnitude in all lifts, but the thigh–trunk angle generally varied more during the flexed knee lifts.

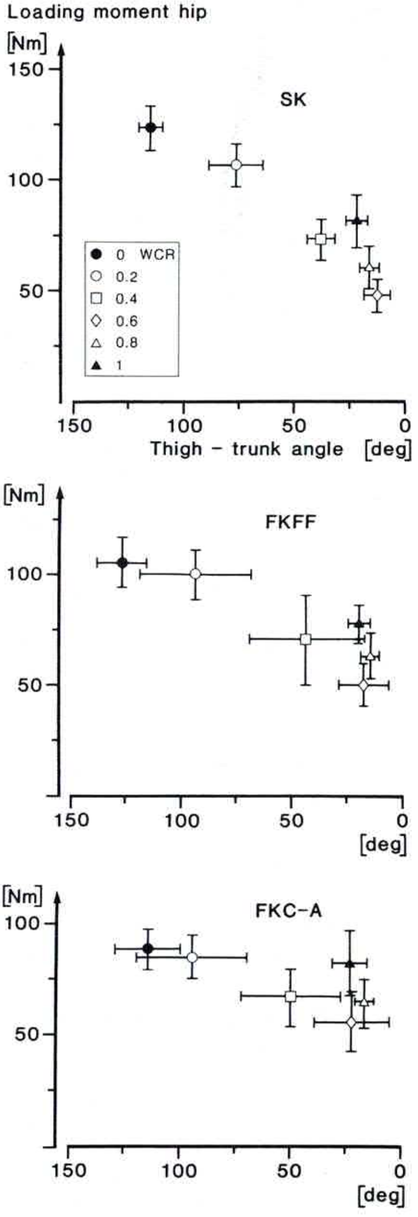

Fig. 4. Diagrams showing mean loading moment (vertical axis) as a function of thigh–trunk angle (horizontal axis) in seven subjects. Time expressed as working cycle ratio (WCR) is indicated with symbols in middle of bidirectional 95% confidence intervals. Filled symbols indicate beginning and end of lifts. For further code definitions of WCR, see insert of upper diagram. Thigh-trunk angle is defined in Fig. 1.

DISCUSSION

There are some difficulties associated with quantifying EMG activity. The reference activity level used in the normalizing procedure was recorded during isometric maximum voluntary contraction (IMVC) under standardized conditions in a given joint angle. However, the force during IMVC may

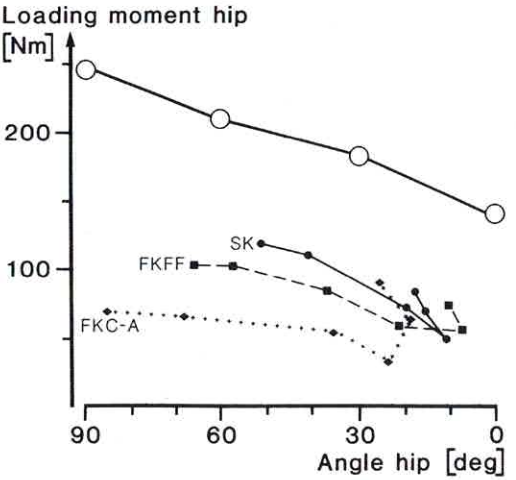

Fig. 5. Maximum hip extensor strength (open circles) (23) and the mean flexing loading moments about bilateral hip joint axis during different lifts. Hip flexion angle is shown on horizontal axis and defined in Fig. 2.

be changed 10% by changing the audio-visual feedback (29). The EMGs were recorded during lifting. This means that the contractions were performed with change of the muscular lengths, contrary to the isometric situation during recording of the reference level. The relationship of the amplitude of the EMG activity versus velocity of contraction and muscle length have been studied by Komi (20) and Vredenbregt & Rau (31) among others and a discussion of various EMG methods has been presented (17). During the initial phase of the straight knee lift there were no extreme positions (i.e. maximal lengthening or shortening of the muscles) except for the hamstrings. The lifts were also performed slowly. Consequently, the influence from these factors would be small on the EMG amplitude. The above described method of normalizing the EMG to provide comparable data on levels of muscular activation, offers several advantages. It allows comparisons of EMG levels between different individuals, different muscles and different experiments because the influence from skin resistance, subcutaneus fat layer, muscle volume and EMG-signal amplifying is neutralized.

The levels of muscular activity in the hip extensors gluteus maximus, hamstrings and adductor magnus showed a slight difference in their pattern. The hamstrings were more and earlier activated in the straight knee lift compared to the flexed knee lifts. In contrast to this, the gluteus maximus and adductor magnus seem to be more activated during the first third of the flexed knee lifts. In the straight knee lift the contraction of the hamstrings was the first occurring muscular activity, counteracting the flexing loading moment about the hip. These muscles tend to rotate the pelvis upward–backward on the hips, and as soon as the developed force has become large enough a movement will follow, i.e. extension of the hip. The contraction of the erector spinae comes somewhat later, meaning that the lumbar spine initially is hanging in its ligaments (10). Thus the extension of the trunk during straight knee lifting is initiated by hip extension, followed by extension in the lumbar spine. This has also been described earlier (12). In flexed knee lifts the vasti are the first muscles to contract, attempting of course to extend the knees (11). A little later the contraction of the gluteus maximus and adductor magnus follows. These are both big muscles and hence develop considerable force even at relatively low levels of activity. The muscles counteracting the late loading moment of the hip during the lifts were mainly the biceps femoris and gluteus maximus. In only a few cases the levels of activation exceeded 40–50% of the reference level. These 40–50% should not be considered as a low level of activity, since the burden was only about 13 kg and thus the reserve for heavier weights is small. The level of activity in the gluteus maximus agrees with what has been reported and so does the activity in the hamstrings, counteracting flexion at the hip (13, 27).

The range of motion in the sagittal plane utilized during the SK and FKFF lifts were of the same magnitude as those reported during stair-climbing (2, 16). During the FKC-A lift there was a larger range of motion utilized. The movement-pattern varied more in the flexed knee lifts compared to the straight knee lift. These differences in body-movements probably caused the within-lift differences in load moment (Fig. 4) due to changed moment arms of the gravitational forces. The hip angle is of great importance to the magnitude of the joint compressive force and will be discussed below.

The joint load was calculated by means of static biomechanics. The acceleration component would slightly change (increase) the load moments at the beginning of the lift. The dynamic force recorded in the handles is small, it exceeds the weight of the box by approximately 10–20% during the initial phase of all lifts (11), and the angular acceleration of the trunk and thigh is small during slow lifting.

When discussing the load, it may be useful to compare the loading moment with the maximum muscular strength that can be exerted at a given joint angle. The maximum hip extensor strength has been shown not to be influenced by the knee angle (23). Thus values for hip extensor strength can be used for comparisons with respect to the hip angle only. For comparisons, the mean maximum strength of the hip extensors (23) of ten healthy students is shown in Fig. 5 (open circles), also showing the load moments during the different lifts (filled symbols) in relation to hip flexion angle. The maximum strength decreases with decreasing joint angle. Contrary to this, the load during lifting decreased slightly with decreasing hip angle down to 5–20° where it increases abruptly. Consequently the subjects have to use more of their strength capability in this position. This may be expressed as the *m*uscular strength *u*tilization *r*atio (MUR), i.e. the actual loading moment at a certain joint angle divided with the maximum muscular moment at the same joint angle. Fig. 6 illustrates this ratio during the different modes of lifting. From this point of view the highest relative load during the flexed knee lifts occurred during the final phase, and for the FKC-A lift it was essentially higher compared to the initial phase. The FKC-B lift (not illustrated) differed from this: 31% of the strength capability was used in the final phase compared to 53% in the FKC-A lift. This is because the burden during the FKC-B lift was kept comparatively close to the trunk while concluding the lift. The use of maximum strength figures from other subjects (23) introduces an error. However, this comparison was considered relevant since the purpose was to discuss these relations in order to show the general principles of this kind of analysis. Furthermore, the age and training habits etc. of the strength analyzed group were fairly similar to the subjects of the lift study.

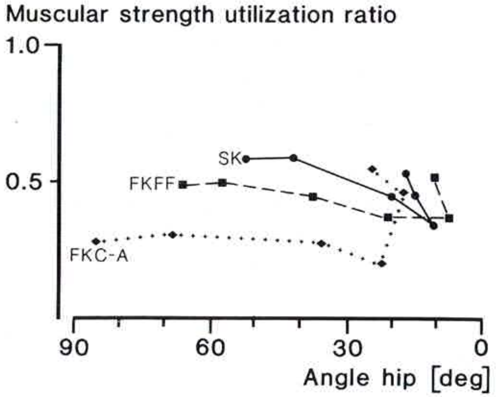

Fig. 6. Hip joint load expressed as mean muscular strength utilization ratio, MUR (vertical axis), i.e. actual loading moment at a certain joint angle divided with the maximum muscular moment at same joint angle. Hip flexion angle is shown on the horizontal axis and defined in Fig. 2.

The highest loading moments reported in the present study appeared initially in the straight knee lift, 36 Nm more than during the FKC lift. The load during the straight knee lift is comparable to the load during stair-climbing (2). The fact that the differences in load moment between the lifts mainly disappeared at 0.4 WCR and that the load moment for all lifts reached a minimum at 0.6 WCR is mainly an effect of the changes in length of the moment arms of the gravitational forces. The compressive force (or so-called bone to bone force) is one of the most valid parameters to describe the joint load. It is a very adequate measure when comparing load between joints. The compressive force is influenced by the loading moment, muscular moment arms and joint angle. The relations between these parameters and the resulting magnitude and direction of the compressive force during lifting have been reported (22, 24). With the use of the load moments and hip joint angles found in the present study the compressive forces were calculated to 3.2 times body-weight (loading moment 82 Nm) for the final phase of the lifts. This was the highest compressive force to occur during this kind of lifting. Contrary to the high load moment during the initial phase of the straight knee lift, the compressive force is somewhat lower initially than finally: 2.7 versus 3.2 times body-weight. The smallest compressive force occurs in the initial phase of the lift with flexed knees and the burden close (FKC), 2.6 times body-weight, due to a comparatively small loading moment (88 Nm) and a more favourable hip angle (85°) compared to the final phase.

The following conclusions may be drawn from this study:

1. The flexed knee lift with the burden initially kept close (FKC), causes the lowest load moment and compressive force. The load is greatest during the final phase of the lifts except when stepping forward before putting the burden down (FKC-B). This phase will therefore be limiting to people susceptible to pain due to load, for example patients

with osteo-arthrosis of the hip. If load reduction of the hip is desired, postures corresponding to the late phase of these lifts should be avoided. Note however, that ascending stairs causes the same level of load on the hip as these lifts.

2. Flexed knee lifts demand more mobility in the hip joints than the straight knee lift. This may be limiting for workers or patients with a reduced range of motion, for example caused by osteoarthrosis.

3. Since the level of muscular activity was low or up to moderate it seems that lifting of this kind is not likely to cause hyperactivity symptoms of the muscles or tendons, such as tendinitis. However, workers suffering from tendinitis of the hamstrings should particularly avoid straight knee lifts since the hamstrings are more activated in this lift. Discomfort from muscles and tendons shown to have very low or no activity during lifting, e.g. the gluteus medius and tensor fasciae latae muscles, is not necessarily an obstacle to lifting.

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REFERENCES

1. Andersson, G. B. J.: Back injuries during work. Report of the Swedish Work Environment Fund, no. 1, pp. 15–18, 1976 (in Swedish).
2. Andriacchi, T. P., Andersson, G. B. J., Ferlier, R. W., Stern, D. & Galante, J. O.: A study of lower-limb mechanics during stair-climbing. J Bone Joint Surg *62-A:* 749, 1980.
3. Carlsöö, S.: How man moves. Kinesiological methods and studies. Heineman, London 1972.
4. Carlsöö, S. & Fohlin, L.: The mechanics of the two joint muscles rectus femoris, sartorius and tensor fasciae latae in relation to their activity. Scand J Rehab Med *1:* 107, 1969.
5. Dempster, W. T.: Space requirements of the seated operator. WADC technical report. Wright Pattersson Air Force Base. Ohio, Wright Air Development Center, Ohio, 1955.
6. Dvir, Z. & Trousil, T.: Electromyographic study of several lower limb muscles during maintenance of dynamic balance in the frontal plane. Aggressologie *23:* 71, 1982.
7. Ekholm, J., Arborelius, U. P., Hammarberg, C. & Németh, G.: Loading moment on lumbosacral, hip, knee and shoulder joints and muscular activity during lifting. Report of the Swedish Work Environment Fund, pp. 215–227, Stockholm 1980 (in Swedish).
8. Ekholm, J., Arborelius, U. P., Hillered, L. & Örtqvist Å.: Shoulder muscle EMG and resisting moment during diagonal exercise movements resisted by weight-and-pulley-circuit. Scand J Rehab Med *10:* 179, 1978.
9. Ekholm, J., Arborelius, U. P., Fahlcrantz, A., Larsson, A-M. & Mattsson, G.: Activation of abdominal muscles during some physiotherapeutic exercises. Scand J Rehab Med *11:* 75, 1979.
10. Ekholm, J., Arborelius, U. P. & Németh, G.: The load on the lumbo-sacral joint and trunk muscle activity during lifting. Ergonomics *25:* 145, 1982.
11. Ekholm, J., Nisell, R., Arborelius, U. P., Hammarberg, C. & Németh, G.: Load on knee joint structures and muscular activity during lifting. Scand J Rehab Med, 16: 1, 1984.
12. Farfan, H. F.: Muscular mechanism of the lumbar spine and the position of power and efficiency. Orthop Clin North Am *6:* 135, 1975.
13. Fisher, F. J. & Houtz, S. J.: Evaluation of the function of the gluteus maximus muscle. Am J Phys Med *47:* 182, 1968.
14. Hanson, C. T., Joslow, B., Danhoff, J. V. & Alon, G.: Electromyographic response of the elbow flexors to a changing, dislocating force. Arch Phys Med Rehabil *62:* 631, 1981.
15. Johnston, R. C. & Smidt, G. L.: Measurements of hip joint motion during walking. J Bone Joint Surg *51-A:* 1083, 1969.
16. Johnston, R. C. & Smidt, G. L.: Hip motion measurements for selected activities of daily living. Clin Orthop *72:* 205, 1970.
17. Jonsson, B., Ericsson, B. E. & Hagberg, M.: Electromyographic methods in the analysis of muscular load and muscular fatigue effects during prolonged work. Report of the Swedish Work Environment Fund, no. 9, 1981 (in Swedish).
18. Joseph, J. & Williams, P. L.: Electromyography of certain hip muscles. J Anatomy *91:* 286, 1957.
19. Karlsson, E. & Jonsson, B.: Function of the gluteus maximus muscle. An electromyographic study. Acta Morph Neerl-Scand *6:* 161, 1965.
20. Komi, P. V.: Relationship between muscle tension, EMG and velocity of contraction under concentric and eccentric work. *In* New developments in electromyography and clinical neurophysiology (ed. J. E. Desmedt), vol. 1, pp. 596–606. Karger, Basel, 1973.
21. Merrifield, H. H.: An electromyographic study of the gluteus maximus, the vastus lateralis and the tensor fasciae latae. Diss Abstr *21:* 1833, 1961.
22. Németh, G. & Ekholm, J.: A biomechanical analysis of hip compression loading during lifting. Ergonomics, in press, 1984.
23. Németh, G., Ekholm, J., Arborelius, U. P., Harms-Ringdahl, K. & Schüldt, K.: Influence of knee flexion on isometric hip extensor strength. Scand J Rehab Med *15:* 97, 1983.
24. Németh, G. & Ohlsén, H.: *In vivo* moment arm lengths for hip extensor muscles at different angles of hip flexion. J Biomech, accepted for publication, 1984.
25. National Board of Occupational Safety and Health, National Central Bureau of Statistics: Occupational injuries 1979. Official statistics of Sweden. Liber för-

lag/Allmänna förlaget, Stockholm, 1982 (in Swedish with English summary).

26. Örtengren, R. & Andersson, G. B. J.: Electromyographic studies of trunk muscles, with special references to the functional anatomy of the lumbar spine. Spine *2:*44, 1977.
27. Pauly, J. E.: An electromyographic study of certain movements and exercises. I. Some deep muscles of the back. Anat Rec *155:*223, 1966.
28. Pauly, J. E. & Scheving, L. E.: An electromyographic study of some hip and thigh muscles in man. *In* Proceedings of the First International Congress of Electromyographic Kinesiology. Electromyography, Suppl. *1*, pp. 131–147, 1968.
29. Peacock, B., Westers, T., Walsh, S. & Nicholson, K.: Feedback and maximum voluntary contraction. Ergonomics *24:*223, 1981.
30. Perry, J. & Bekey, G. A.: EMG-force relationships in skeletal muscle. Crit Rev Biomed Engin *7:*1, 1981.
31. Vredenbregt, J. & Rau, G.: Surface electromyography in relation to force, muscle length and endurance. *In* New developments in electromyography and clinical neurophysiology (ed. J. E. Desmedt), vol. 1, pp. 607–622. Karger, Basel, 1973.
32. Weatley, M. D. & Jahnke, W. D.: Electromyographic study of the superficial thigh and hip muscles in normal individuals. Arch Phys Med *32:*508, 1951.
33. Zuniga, E. N., Truong, X. T. & Simons, D. G.: Effects of skin in electrode position on averaged electromyographic potentials. Arch Phys Med Rehabil *51:*264, 1970.

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