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LOAD ON KNEE JOINT STRUCTURES AND MUSCULAR ACTIVITY DURING LIFTING

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ABSTRACT. The load on the knee joints during lifting has been less studied than low back load. Healthy subjects lifted a 12.8-kg box from floor to table-level in three different ways; 1) with straight knees, 2) with bent knees and the box in front of the knees, and 3) with bent knees and the box between the knees. The loading moment of force about the bilateral knee axis was calculated by means of a computerized static sagittal plane model. Electromyography was recorded from quadriceps and ischiocrural muscles. The beginning of the flexed-knee lifts caused a flexing loading knee moment of about 50 Nm and a knee angle of 90°. Straight-knee lifts gave all through the lift an extending loading moment. During the final phase of all lifts there was an extending loading knee moment of about 55 Nm and a knee angle of 0°. The three lifts were compared and discussed from a biomechanical and ergonomical point of view.

Key words: Biomechanics, EMG, ergonomics, human factors, models biological, rehabilitation

Lifting is widely recognized as one of the activities apt to elicite low back pain. Less is known about its role as an ethiological or forthbringing factor of knee pain. It is generally agreed that even a moderate load easily raises articular pain in osteoarthritic joints. A great deal of the patients with mild gonarthrosis work and they would benefit from information about how to avoid unnecessarily high load on knee structures. Data on forces acting on the knee joint are of great value for those involved in designing prosthetic and orthopedic devices and to orthopedic surgeons in correcting various deformities. At the same time, studies of forces and muscular function can be of considerable value to physiotherapeuts in rehabilitating postoperative or osteoarthritic patients, and to physical education instructors in improving performance and avoiding harmful forms of exercise.

The knee is one of the joints most susceptible to osteoarthritis, a serious medical and social problem. The pathogenesis is not clear, but some reports suggest that high mechanical stress is one important etiological factor (5, 26). It appears possible that repeated non-extreme loads to the articular cartilages may increase the rate of degeneration and the old concept that primary osteoarthritis is caused by repetitive impulsive loading has been revived (25).

The load on the low back during lifting has been studied by several authors (1, 8, 10, 23, 31). The load on the knee joints during lifting, however, has not earlier been studied in detail, as far as we know. There are several studies to be found on knee joint biomechanics (4, 12, 13, 18, 20, 22) and many reports has also been made concerning the activity of the knee muscles during various activities, such as in standing postures and bicycling (3, 7, 14, 15, 30), in various standing work positions (6, 11), in walking (2, 22, 24), in vigorous swing pahse (kicking) (32) and in sit-to-stand movements (16, 27).

The aim of the present study is to quantify and discuss the effects of loading on the knees and upon muscular activity utilizing three different modes of lifting a box from floor to table level. The following specific questions were analyzed.

1. What is the magnitude of the loading moment of force about the bilateral knee joint axis during the three different lifts?
2. What proportion of the total loading moment of force is caused by the weight of the burden and to what extent is the body position in itself contributing?
3. How can one compare the magnitude of the loading momemt of force during lifting with those involved in other activities?
4. How is the activity of the investigated muscles distributed over time for the various lifts?

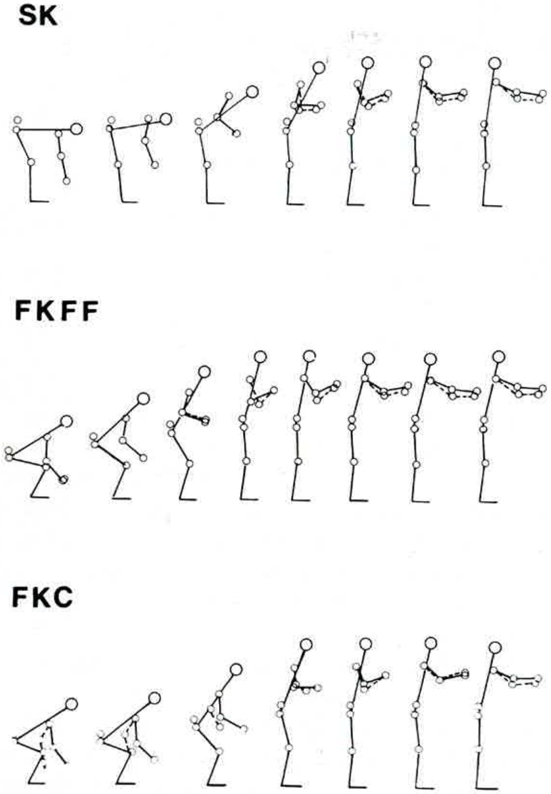

Fig. 1. Slightly modified computer displayed drawings of body segment positions during the three working-cycles studied when lifting a box of 12.8 kg from floor to table level. *SK*, lift with straight knees; *FKFF*, flexed knees, box lifted in front of knees; *FKC*, flexed knees, box lifted between the knees close to pelvis.

5. How much of the knee maximum muscular strength capacity is utilized during the lifts?

MATERIALS AND METHODS

15 subjects participated in the study (14 males, 1 female). Their ages ranged from 20 to 40 years and average height was 1.78 m. None of the subjects were suffering from knee pain, nor had previously undergone knee joint surgery, nor could attribute any longer periods of sick leave to knee joint disorders. During the course of the investigation, new techniques were developed, which explain why all cases are not included in the presented graphs. For illustrating the loading moments seven subjects (mean bw = 76.8 kg) were used and five of them (mean bw = 75.8 kg) for illustration of the muscular activity levels.

In all lifts the task was to lift a two-handled box of 12.8 kg from a position on the floor in front of the subject up to a bench whose height was adjusted to the level of the umbilicus of the subject. The subject started in an upright standing position with the arms at the sides, bent down and performed the lift, and ended by returning to the upright position. All lifts were performed at a moderate pace (the box was lifted from the floor to the bench in about two seconds). Between succeeding lifts the subjects was allowed at least two minutes rest.

Three types of lifts were studied (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 between the knees and close to the pelvis (FKC—flexed knees, close).

During the main part of the study muscular activity was recorded (Devices AC8) by means of full wave rectified, low pass filtered and time averaged electromyograms (linear envolope EMG) (33). In the beginning of the investigation only direct EMG was available. Two flexible disposable Ag–AgCl electrodes, with an interelectrode distance of approximately 30 mm, were placed over the muscle belly in the main direction of the muscle fibres. For control purpose, direct EMG was simultaneously registered on a UV-recorder (Honeywell, Visicorder).

To make possible comparisons of levels of muscle activity between different muscles and different individuals the EMG was normalized. A ratio, TAMP-R (time averaged myoelectrical potential ratio) for each muscle was used dividing the envelope EMG-level recorded at the particular lift phase by the activity level recorded during an isometric maximum voluntary test contraction for each muscle. In order to increase standardization of this maximum contraction the subject was firmly bound to a specially designed chair. These test contractions were performed for both knee extension and flexion in a knee angle of 90°. For simultaneous strength measuring three of the subjects carried out maximal isometric extension and flexion against a sling (around the distal calf) connected to a strain gauge (Bofors). The signal was amplified (Devices DC2) and recorded. The perpendicular distance from the sling to the bilateral axis of motion of the knee joint was measured. Thus the maximum muscular moment of the knee flexors and extensors could be calculated for these subjects.

A strain gauge was mounted in each of the handles of the box and the forces applied were thus measured. During the experiments the forces from two handles were summed up and recorded.

The lifts were photographed perpendicular to the sagittal plane by a motorized camera (Olympus OM-1, 24×36 mm film) at a picture frequency of 4 Hz. Some of the lifts were filmed with a 16-mm camera. To make time syncronization of the EMG-records and the film possible, a special time indication panel was designed. This "clock" also paced the time markers on the recorders and featured a light-emitting diod display with, a bar representation of time in milliseconds (visible on the photos).

The films from the experiments were used for the mechanical calculations. Photocopies were made or the pictures were projected directly from the negatives on a device for semi-automatic coordinate registration (Tektronix digitizer, 4953). This digitizer was connected to a graphic terminal (Tektronix 4012) which was connect-

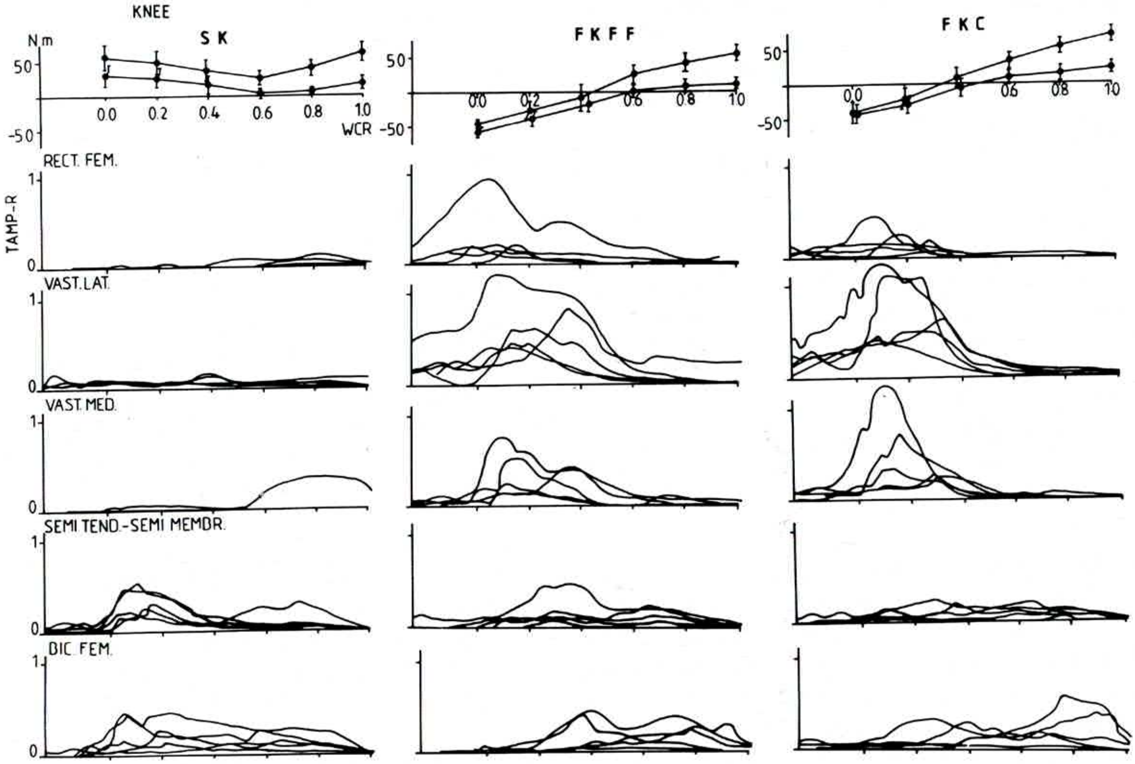

Fig. 2. Diagrams above: loading moment (y-axis) about the bilateral knee axis. Negative values show flexing loading moment, positive extending loading moment. Upper curve: total loading moment. Lower curve: moment due to burden subtracted. Means and 95% confidence, $n=7$. x-axis: time in WCR (working-cycle ratio units). Diagrams below: level of EMG-activity in TAMP-R (see text) in five knee muscles. $n=5$. Modes of lifting as in Fig. 1.

ed to a Nord-10 computer at the Computer central. The programs necessary for mechanical calculations and data administration were written by the group. The digitizer recorded the coordinates of the reference distance points and the positions of the bilateral motion axes of the major joints together with the time, recorded by the light emitting diod display. The entered body position was displayed on the graphic terminal which enabled the operator to control his actions. The coordinate data were stored on a magnetic disc. In a special program, mechanical calculations were performed. In this mechanical model for sagittal plane analysis static mechanics was used. In the model the lower extremities were handled symmetrically. The body segment parameters necessary for the calculations were taken from Dempster's anthropometric data (17). The knee angle and the moments of force (about the bilateral knee axis) caused by the weights of the body segments and by the burden were calculated for each picture during the course of the lift. The knee angle was calculated from the two lines connecting the points of localisation for the bilateral axis of the hip, knee, and ankle joints. The straight knee was defined as knee angle 0°.

The program also allowed us to change the weight of the burden used in the calculations for a given series of pictures. This function was utilized to isolate the role of body segments weight as a part of the total loading moment. A set of subroutines were used for graphic display (UPAG-6) of the calculated moments. The digitizer system was also used for rescaling the original linear envelope EMG-records to fit the presentation of the biomechanical parameters.

RESULTS

Fig. 2. shows the loading moment about the bilateral knee axis of the tibio-femoral joint and the EMG registration from five muscles passing the knee joint. From left to right the columns represent the three modes of lifting, as presented in Fig. 1, SK (straight knees), FKFF (flexed knees burden far away), and FKC (flexed knees burden close) respectively. At the top of each column the loading moment about the bilateral axis of one knee joint is illustrated, followed below by the normalized

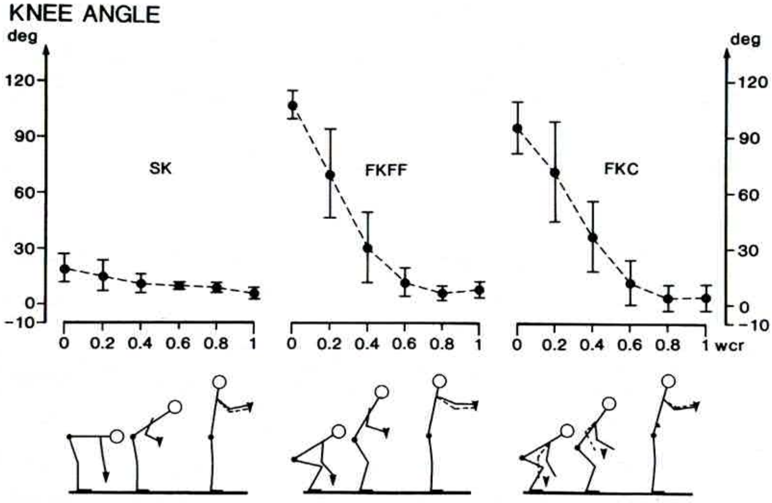

Fig. 3. Knee angle during the three lifts (same as in Fig. 1), pictured by stick figures. Negative values indicate hyperextension. *x*-axis: time, same as in Fig. 2.

EMGs of the muscles investigated. The loading moments are given as means at 95% confidence intervals for 7 subjects. The lower moment curves in each top diagram show the loading knee moment with the moment due to the weight of the burden subtracted. The horizontal axis shows time expressed relative to the total working cycle (=1.0). 0.0 indicates the time when the box has just left the floor and 1.0 is just as the box makes contact with the bench.

The loading moment curves show that there is an extending loading moment of force throughtout the straight-knee lift. In the two different flexed-knee lifts the moment changes from a flexing moment in the beginning (about 50 Nm) to an extending one in the middle of the working cycle. All lifts show an extending loading moment of about 50–60 Nm in the final phases. This difference, between the SK-lift on one hand and the FKFF and FKC on the other, is reflected in the muscle activities.

With straight knees the extensor muscles, the medial (vastus medialis) and lateral vastus (vastus lateralis) and the rectus femoris, show low activity throughout the lift. The biceps femoris (a flexor of the knee joint) is activated to a 0.50 TAMP-R level, and the semitendinosus/semimembranosus (also flexors in the knee) are activated to the same level during approximately the first third of the lift. These latter muscles are in also extensors of the hip joint, which complicates the interpretation of the results and will be further discussed below.

During lifting with flexed knees (Fig. 2. FKFF and FKC), there is medium to high muscle activity in the knee extensors in the first half of the working cycle. The knee flexor, biceps femoris, is mainly activated after WCR (working cycle ratio) 0.2–0.3, and the activity of the semitendinosus/semimembranosus is generally low. Vastus medialis and lateralis are more highly activated than rectus femoris.

During lifting with straight knees (SK), the moment of force caused by the weights of the body segments is higher in the beginning of the lift when the trunk is ventroflexed, than during the later phase of the lift when the trunk is more upright. At the end of the lift the elevated upper extremities are the only body segments contributing considerably to the extending loading moment of force. Consequently, the graph showing the moment of force caused by the body segments (lower line on moment diagram) is at its maximum at the beginning of the lift. The point of time in the lift where the total loading moment (upper graph) reaches its maximum depends upon the relative weight of the burden, since this contributes more to the loading moment at the end of the lift, when the moment arm for the burden is longest. At our chosen weight level of the burden the maximum occurs at the end of the lift. Note that, in spite of this, the biceps femoris and the semitendinosus/semimembranosus muscles are more activated in the beginning of the lift. This will be further discussed later on.

At the beginning of the lifts with flexed knees

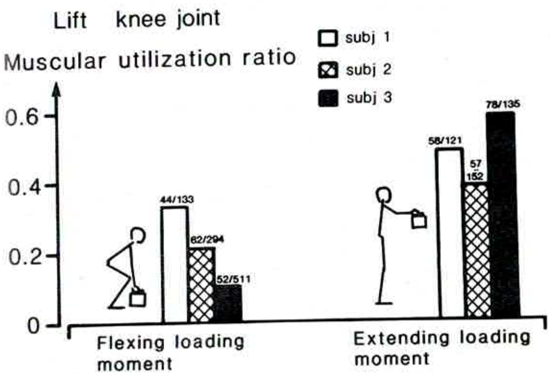

Fig. 4. The muscular utilization ratio is estimated as a quotient (*y*-axis) between the loading moment and the maximum muscular moment of the counteracting muscles recorded in a "max-test-contraction". The moments (in Nm) are also shown in absolute values for the three subjects as the figures on the top of each column. Left part of illustration: in the beginning of the lifts with bent knees (FKFF and FKC) there is a flexing loading knee moment present which will be counteracted by the quadriceps muscle. Right part: in the end of the three modes of lifting (FKFF, FKC and SK) there is an extending loading moment present which will be counteracted by the knee flexors.

(FKFF and FKC), most of the body weight will be behind the bilateral motion-axis of the knees and thus cause a flexing moment of force. This is particularly true when the burden is held in front of the knees (FKFF), since during this lift the burden will counterbalance most of the moments caused by body weight. With the burden between the knees (FKC) the trunk will be leaning a bit forward over the burden and the common centre of gravity (for the trunk and the burden) comes closer to the plane through the knees. The flexing moment of force will thus also be reversed into an extending moment somewhat earlier as what is the case with the FKFF lift. This is reflected in an earlier relaxation of the knee extensors during FKC.

After approximately WCR 0.6 the lifts should be alike from a mechanical point of view and this is also seen from the three diagrams in Fig. 2.

Knee angles during investigated lifts are presented in Fig. 3. When lifting with straight knees (SK) the mean knee angle ranges between 5 and 20°. The lifts with bent knees (FKFF and FKC) show knee angles of about 90–100° in the beginning of the lifting cycle and 0–10° at the end. No difference in knee angles is seen between these two modes of lifting. In the FKC lift (to the right in the diagram) some subjects had negative knee angles at the end of the lift. It is thus indicated that hyperextension of the knee joint is present in some subjects.

The quotient between the loading moment and the isometric maximum muscular moment have been calculated for three subjects (Fig. 4). In the beginning of the lifts with bent knees (FKFF and FKC), there is a flexing loading moment whose magnitude is divided by the maximum extending muscle moment. The ratio will show how much of the maximum capacity (i.e. the potential muscle power) that is utilized. This utilization ratio is higher in the end of the lift where the loading moment is extending. It should be considered that the maximum muscular moments were registered in 90° knee angle. The knee is straight (0°) at the end of the lifts and the knee-flexors are stonger in this position, i.e. they can produce a better muscular flexing moment (28). If a correction factor of 0.67 is introduced here according to Smidt's data (28), the muscular utilization ratio will be 0.32, 0.25 and 0.39 respectively. These figures are still higher han the corresponding values in the beginning of the flexed knee lift. This is partly due to the fact that the knee flexor muscles cannot exert the same level of maximum moment of force as the knee extensors can, and partly because the loading moment is higher in the end of the lift.

ANALYSIS

In the present study the patello-femoral compressive force for one subject (72 kg) has been estimated in the beginning of the FKFF-lift (Fig. 5). The moment arm for the extensor muscles' line of action as well as the maximum muscular strength (i.e. the maximum muscular moment) vary with the joint angle (12, 13, 19, 28). The patellar-tendon moment arm of 38 mm at a 90° knee angle (28) was used for an approximative calculation of the tension force in the patellar tendon.

The patello-femoral compressive force (R) was calculated using a biomechanical analysis (4, 20), which is presented in Fig. 6. The angle between the patellar and the quadriceps tendons was estimated to be about 100° (α in Fig. 6) when the knee angle is 90° and the flexing loading moment 55 Nm. The force in the quadriceps tendon will be nearly double that of the force in the patellar tendon. This is an interesting finding which might explain the fact that partial (or complete) ruptures are con-

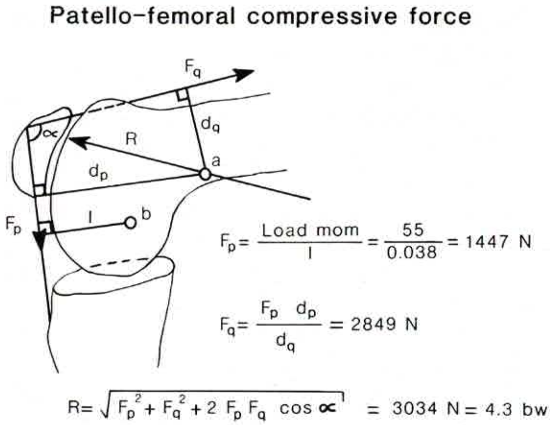

Fig. 5. The patello-femoral compressive force (R) calculated at a known flexing loading knee moment (55 Nm) and a knee angle of 90°. The laws of static equilibrium were used. f_q, force in quadriceps tendon; F_p, force in patellar tendon; ∝, angle between quadriceps tendon and patellar tendon, measured to 97°; a, the bilateral axis for the patello-femoral joint; b, bilateral axis for the tibio-femoral joint; d_q, moment arm for quadriceps tendon force to a; d_q, moment arm for patellar tendon force to a; l, moment arm (38 mm) for patellar tendon to b.

sidered to be more frequent in the suprapatellar parts than in the infrapatellar parts of the knee extensor apparatus (29). The value of the patello-femoral compressive force was calculated to about 3 kN which is of the same magnitude as the force in the quadriceps tendon. For this subject these forces are around four times body weight.

The tibio-femoral compressive force is the force acting between the tibial and femoral condyles. This compression may be approximated (with flexed knees) to the force in the patellar tendon (F=1 450 N) plus half of the gravitational force of the body segments above the knees (310 N). This compression will thus be around 1 760 N.

Other forces acting on this system, such as tension in the hamstrings, tension in gastrocnemius and actions in the ligaments of the knee, are ignored in the calculations above. The EMG registrations show some antagonistic activity in the hamstrings (Fig. 2) and muscle activity is also registered in the gastrocnemius (results will be published in a special report). Thus, the biomechanical situation is more complicated than what has been illustrated here. We know of no reports showing calculations of knee joint compressive forces when antagonistic muscle action is taken into consideration.

An extending loading moment of force about the knee joint as seen throughout the lift with straight knees and during the second half of the lifts with flexed knees has to be counteracted by structures posterior to the bilateral knee axis. These muscles are not highly activated (seen in Fig. 2) but the strain will probably add considerable tension.

An extending loading moment of 55 Nm (Fig. 2) which is present at the end of all three lifts studied, gives a calculated tensile force of about 2.2 kN in the ischiocrural tendons, if the moment arm to the bilateral knee axis for these tendons is approximated to 25 mm (28).

The gravity force of the body segments above one knee joint (about 310 N) should be added to these muscular forces to get the tibio-femoral compressive force which will thus be more than three times body weight.

Further studies on local knee joint biomechanics in the sagittal plane are under preparation.

DISCUSSION

The loading moments of force were calculated as if static conditions prevailed. The error thus introduced is considered to be small since the dynamic components during slow movements are comparably small and of the same magnitude in the three compared modes of lifting. The dynamic components will change the absolute values of the

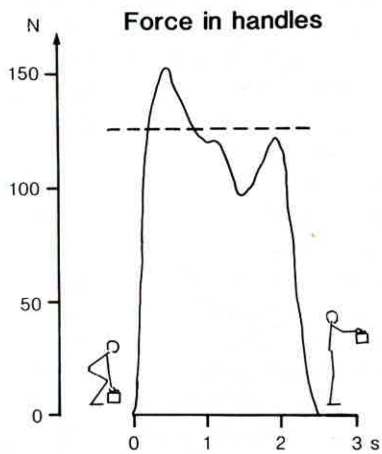

Fig. 6. Recording of force (from strain gauges) in handles during a typical lift of aobut 2.5 sec (χ-axis). The acceleration or deceleration of the box is illustrated as the force-curve changes in direction. Weight of burden: 126 N (dashed line).

Table I. *Summary of reported maximal muscular knee moments with respect to the bilateral knee axis during various activities. Knee angles within parenthesis*

Authors	Activity	Number of subjects	Mean weight (kg)	Max. moment knee extensors (Nm)	Max moment knee flecors (Nm)
Radcliffe (24)	Level walking	4	–	61 (20°)	20 (5°)
Morrison (22)	Level walking	9	82	35 –	28 –
Lindahl et al (19)	Isometric max. experiment	12	–	226 –	– –
Smidt (28)	Isometric max. experiment	26	82	120 (60°)	63 (5°)
Kelley et al. (16)	Rising from chair	6	–	110 (90°)	0
Wahrenberg et al. (32)	Kicking	6	76	180 (100°)	100 (70°)
Andriacchi et al. (2)	Ascending stairs	10	71	54 (65°)	48 (5°)
Andriacchi et al. (2)	Descending stairs	10	71	147 (60°)	43 (20°)
Ekholm et al. (11)	Machine milking	10	72	31 (108°)	56 (8°)
Ekholm et al. (11)	Isometric max. experiment	10	72	198 (90°)	113 (90°)
Ekholm et al. (present study)	Lifting	7	77	50 (90°)	55 (5°)

loading moments at the beginning of the lift. The dynamic components regarding the hands can be illustrated by the curve in Fig. 6 which shows the force registered in the handles of the box. The extra force affecting the hands caused by the acceleration of the box at the beginning of the lift is about 20% (25 N) of the gravity force of the box (125 N). This dynamic component increases the loading moment of the humero-scapular joint and does also influence the knee moment. In the beginning of the flexed-knees-lifts, where the loading knee moment is flexing, the force necessary to accelerate the box would decrease the calculated flexing loading moment about the knee. The linear and angular acceleration of trunk and thighs is low during slow lifting and would increase the flexing loading knee moment to some extent. On the other hand in straight-knee lifting the dynamic components of both the box and body segments increase the extending loading moment of the knee.

Quantifying the EMG always introduces some difficulties. Our method includes using the level of the linear envelope EMG during maximum voluntary isometric contraction as a reference. This level is dependent upon several factors which are difficult to standardize, such as mental state and motivation. It is easier to activate the motor units during dynamic conditions than during isometric. Hence, the EMG-level during the lifts occasionally exceeds the reference value (TAMP-R= 1.0). In our opinion this method of quantification provides definite advantages. The technique has been described and used earlier (9, 10).

Three of the muscles studied are biarticular which complicates the interpretation of their activity levels. The biceps femoris and the semitendinosus/semimembranosus are knee flexors as well as hip extensors. Their action on the hip explains why the activity during the SK-lift is highest in the beginning of the lift despite the fact that the extending loading knee moment is higher towards the end of the lift. This is most evident for the semitendinosus/semimembranosus. When lifting with flexed knees, the role of the ischiocrural muscles in the early hip joint extension, appears to be less important since the early peak of activation seen during the SK-lift is not present.

It has been shown by Molbech & Carlsöö (7, 21) that the ischiocrurals may cause an extending moment of force on the knee in a closed muscular chain (as in a standing position, where the feet are fixed to the ground) and when the knee angle is small (not flexed to more than 40°). So the activity levels seen in the ischiocrurals before the middle phase of the bent-knee lifts may be explained by the fact that these muscles are contributing to the knee extending muscular moment, and thus counteracting the flexing loading moment. Consequently the function of these knee flexor muscles may be a paradoxical contribution to knee extension and synergistic to the function of quadriceps. The rectus femoris is not activated to the same level as are the vastus medialis and lateralis. This is probably due to this muscle's function as a hip flexor as well. Since there is a flexing loading hip moment present, when bending the trunk forward, a working rectus femoris would increase rather than counteract this loading hip moment. These findings regarding the activity levels of the rectus femoris are in line with those found in earlier investigations (7, 11).

The flexing loading knee moment as such (around 50 Nm) is nearly of the same magnitude that was found in level walking (3, 24) and when ascending stairs (2) (see Table I). When rising from a seated position, unaided by the arms, a peak moment of 110 Nm was calculated (16) and when descending stairs a flexing loading peak moment as high as 147 Nm was found (2). Note however, that these maximal loading moments will arise at different knee angles which has great importance for the magnitude of the patello-femoral compression. The knee angle when climbing or descending stairs is, as an example, below 70° when the peak moments are present (2). Thus, it is not true that the patello-femoral compressive force when descending stairs (147 Nm) should be almost three times more than in lifting (50 Nm). When kicking a football (32) and rising from a chair (16) the knee angle is around 90° and the loading knee moments are high. In these activities the patello-femoral compression will be 2 and 3 1/2 times respectively more than the corresponding compression in lifting.

The extending loading knee moments during lifting are of the same magnitude as the peak moments found during walking on stairs (2) and as in machine milking (11). During level walking the peak extending loading moments are less, about the half, than what was found during lifting. In these studies, as in this one, the knee angles are close to zero (the knees are almost straight) when the most pronounced extending knee moments are present. An extending loading moment produces a tensile force in the ischiocrural tendons. The forces in the tendons will be most expressed during the end of all three lifts because we here find the highest extending loading moment. At the same time, in this straight-knee position, all tendon forces in the knee flexors compress the tibio-femoral joint as the angle between the joint surface and the muscle tendons is perpendicular. Consequently the tibio-femoral compressive force will be highest during the end of the lift.

The most important difference between the lifts with flexed and those with straight knees, with respect to the mechanical load upon the knee joint, is the existence of a flexing loading moment of force in the beginning of the lifts with flexed knees. In the lift with straight knees there is only an extending loading knee moment. The flexing loading knee moment has to be counteracted by a high tensile force in the knee extensor apparatus. As a consequence, the patella will be pressed against the femoral articular surface with a resulting high compressive force. This could lead to excess friction and osteoarthritis and may certainly initiate or increase pain in osteoarthritic knees. The patello-femoral compressive force increases with knee joint angle and will be most pronounced in the beginning of the flexed-knee lifts.

Regarding the biomechanical load on the knee joint there is no clear difference, between lifting with bent knees and the burden in front of the feet (FKFF) and lifting it from between the feet (FKC). The loading moments and the knee angles are about the same in both lifts. Consequently, the compressive forces in the knee joint are of the same magnitude. However, if one considers other regions such as the low back, there is a difference between these two modes of lifting with bent knees. The lift with the burden between the knees (close to the body, FKC) gives a lesser loading moment on the low back compared to the lift with the burden in front of the knees (FKFF and SK) (10).

From a biomechanical viewpoint regarding the knee joint, it is not obvious, that the flexed-knee lift is preferable to the lifts with straight knees. Bent knees cause a flexing loading moment which gives rise to high tension in the extensor muscles which produces high patello-femoral compressive forces. In these healthy subjects the loading moments were easy to counteract as the load was less than 40% of the potential muscle strength. When lifting with straight knees there is no flexing moment. The extending loading moment will produce high compressive joint forces in the tibio-femoral joint and the muscular utilization ratio is somewhat higher. Furthermore, in some subjects the knee joint is placed in an extreme position (hyperextended) and the overall patho-etiological effects of these factors are not clear. To optimize a floor-to-bench lift with respect to the knee joints, it is preferable to bend the knees to some extent in order to find the least loading moment. If the knee flexion is pronounced there will be a flexing loading moment, and if the knee bend is small there will be an extending loading moment. The "optimum" knee angle arises (focusing only on knee joint load) when the load on the knee joint is as small as possible, i.e. when the loading knee moment is around zero. This mainly depends upon the weight of the hand-held burden and the positions of the body segments above the knees. To get the "best" knee angle in a whole-body perspective the knee load must be compared to the load on other major joints, and one must also consider the individuals strength and other prerequisites.

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