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Self-Selected Manual Lifting Technique: Functional Consequences of the Interjoint Coordination

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The pattern of movement self-selected by 39 subjects to lift light loads from 9 cm above the ground is described in kinematic and electromyographic terms. Hamstring length changes were estimated from hip and knee angular kinematics. Subjects adopted a posture at the start of the lift intermediate between stoop and full-squat postures. A consistent coordination between knee, hip, and lumbar vertebral joints during lifting was described through calculation of the relative phase between adjacent joints and found to be exaggerated with increases in load mass. During the early phase of lifting, knee extension leads hip extension, which in turn leads extension of the lumbar vertebral joints. Early in the lifting movement, when load acceleration is greatest, the erectores spinae are thus relatively long and shortening slowly. Both of these factors produce greater back extensor strength. Rapid hamstring shortening is also delayed, which enhances their strength, and coactivation of the monoarticular knee extensors and biarticular hamstrings observed early in the lifting movement suggested that the knee extensors contribute to hip extension through a tendinous action of the hamstrings.

INTRODUCTION

Manual lifting is consistently linked with a high incidence of occupational injuries (e.g., Jensen, 1988). It seems reasonable to suggest that the damaging physical effects of lifting might be partly a function of the postures adopted throughout the lifting movement and the patterns of joint movement involved.

Lifting techniques have been defined in terms of the posture adopted just before a load is lifted. It has been proposed (e.g., Davis and Troup, 1965; Trafimow, Schipplein, Novak, and Andersson, 1993) that, although intermediate postures are possible, the postures adopted to lift loads from a low level may be characterized in terms of two extremes. One is described as a stoop in which the knee joints are almost fully extended and the hip joints and vertebral column are flexed to reach the load. The second, described as a squat, is one in which the knee joints are fully flexed and the trunk is held in as vertical a posture as possible. The latter posture has been traditionally recommended as least likely to lead to injury (e.g., Bendix and Eid, 1983; Davis, 1959, 1967) and is still taught as such (e.g., McCauley, 1990).

Research results do not support such a uniform recommendation, however. Several authors (e.g., Garg, Sharma, Chaffin, and Schmidler, 1983; Noone and Mazumdar, 1992; Park and

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Chaffin, 1974) have cautioned that a technique involving a vertical trunk and fully flexed knees may not always minimize joint forces because the full squat may involve keeping the load farther away from the body than with other techniques. Other authors (e.g., Ayoub and Mital, 1989, p. 268; Parnianpour, Bejjani, and Pavlidis, 1987) have suggested that a single best method may not exist for any particular task. It has also been suggested (Ayoub and Mital, 1989, p. 51; Garg and Saxena, 1979) that the technique spontaneously adopted by the subject may actually be the least likely to lead to injury.

This last proposition—that a freely chosen lifting technique might be preferable in terms of injury prevention—raises the question of what postures and patterns of joint movement are adopted to lift low-lying loads when no prescriptive instructions are given. Self-selected lifting techniques have not been systematically described, although a number of opinions have been expressed.

Some authors (e.g., Troup, 1979) postulate that no natural lifting technique is identifiable. Others (e.g., Gagnon and Smyth, 1991, 1992; Genaidy and Asfour, 1989; Park, 1973; Park and Chaffin, 1974; Trafimow et al., 1993) have concluded that when no prescriptive instruction is provided, the posture adopted to lift a load involves a moderate range of knee flexion and an inclined trunk—that is, a posture intermediate between the extremes of a stoop and a full squat with a vertical trunk. A third opinion, either expressed explicitly or implied, is that the posture naturally adopted at the start of lifting is closer to the stoop and involves little knee flexion (e.g., Bendix and Eid, 1983; Davis, 1967; Garg and Saxena, 1985; Kumar, 1974; Mittal and Malik, 1991; Nag, 1991).

The differences in the foregoing statements are likely to be a consequence of differences in the characteristics of the lifting tasks observed. However, questions remain unanswered: What movement patterns are observed when people lift normally, and how do changes in the task characteristics, such as load mass, starting height, finish height, load size, and the like, influence these patterns?

Lifting techniques are not entirely described by the posture adopted at the start of the lift. A description of the interjoint coordination and how this coordination is altered by changes in the task may well be equally—or even more—important in describing the lifting techniques spontaneously adopted by subjects to lift low-lying loads.

One aspect of interjoint coordination in manual lifting that has attracted comment is a distal-to-proximal pattern of extension of the knee, hip, and lumbar vertebral joints. A lag between hip and lumbar vertebral extension has been noted during unloaded extension from a stooped position (Floyd and Silver, 1951; Tanii and Masuda, 1985) and when lifting (Burgess-Limerick, Abernethy, and Neal, 1992, 1993; Scholz, 1992, 1993a, 1993b). A similar relationship exists between knee and hip extension (Burgess-Limerick et al., 1993; Schipplein, Trafimow, Andersson, and Andriacchi, 1990). When a load is lifted, this coordination results in a delay after the start of the upward movement of the load before significant extension of the lumbar vertebral joints occurs (Davis, Troup, and Burnard, 1965; Grieve, 1974; Kumar, 1974; Troup, 1977). The magnitude of the lag between knee and lumbar vertebral extension, and consequently the delay after the start of the lift before significant lumbar vertebral extension, increases with increased load mass (Davis et al., 1965; Grieve, 1974; Kumar, 1974; Scholz, 1992, 1993a, 1993b; Troup, 1977).

The purposes of this research are to describe the kinematics, and particularly the coordination of the knee, hip, and lumbar vertebral joints, of subjects lifting light loads from floor height using a self-selected technique, and to investigate the effect of load mass on these kinematic patterns. Interpretation of the kinematic data is informed by surface electromyographic data collected from lower limb and back muscles and estimation of the time course of biarticular hamstring muscle length changes. The manual lifting task chosen for analysis is one in which both hands are involved symmetrically and movement occurs almost exclusively in the sagittal plane.

METHOD

Subjects

A total of 39 volunteers (20 women, 19 men) aged 18 to 26 years participated in the study. None had any history of serious back injury or recent minor back injury, nor had they ever worked full time in occupations involving manual handling.

Kinematic data were collected while each subject performed 100 lifts. Surface electromyographic data from six lower-limb muscles and erector spinae were also simultaneously collected from 20 of the subjects (10 female, 10 male). Subjects were given a brief, standardized explanation of the purpose of the experiment. No deception as to the purpose of the study was involved. Subjects were fully informed about the requirements of the experiment, which had previously been approved by the Human and Animal Experimentation Ethical Review Committee of the University of Queensland.

Procedure

The task involved symmetric bimanual lifting of loads from the floor to a shelf. The standardized instructions given to the subjects emphasized that they were to lift the load in "the way you would normally do the task, that is, the most comfortable way for you." Subjects were instructed to adopt a normal standing posture facing the load, with feet approximately parallel and an equal distance from the load. The distance of the feet from the load was self-selected, and subjects were allowed to vary this distance between trials as desired with the proviso that the feet remain stationary during each lift.

Five different loads (2.5 to 10.5 kg in 2-kg increments) were lifted. The dimensions of the weighted wooden blocks were 345 mm wide, 155 mm deep, and 110 to 250 mm in height in 35-mm increments. Cutout hand holds in each load ensured that the hands were 90 mm above the floor when the load was first grasped. The load was lifted to a shelf adjusted for each subject to be about shoulder height (range 1100 to 1200 mm) and comfortable arms' length.

Each subject performed two practice trials (one at each of the extremes of mass), and then 100 trials were performed in blocks of five lifts of each mass in ascending series. This 25-trial series was repeated four times. Time elapsed between trials was 5 s and between blocks, 30 s.

Two female subjects found that placing a 10.5-kg load on a shelf at arms' length and shoulder height placed an uncomfortable strain on their shoulder flexors during the practice trial. These two subjects did not lift the 10.5-kg load and thus performed only 80 trials each.

Kinematic Data Collection

Sagittal motion of ankle, knee, hip, and lumbar vertebral joint complexes and the load were estimated via automated digitizing at 100 Hz of 10 spherical reflective markers placed on the right side of each subject. Details of equipment, marker placements, and definition of joint angles are described by Burgess-Limerick et al. (1993). All angles were defined as included angles that increased when the joint extended or, in the case of the ankle, plantar-flexed. The angular position and hand displacement data were Butterworth low pass filtered with a 6-Hz cutoff before differentiating.

Kinematic Dependent Variables

Temporal measures. Each lifting trial consisted of a flexion phase when the hands were lowered to grasp the load, followed immediately by an extension phase. Flexion was defined as starting when the hand velocity exceeded 0.05 m/s downward and as ending when the hands reached their lowest position during the movement—that is, the global minimum in the hand vertical displacement time series. Extension was defined as commencing at the end of flexion and concluding when the angular velocity of lumbar vertebral extension returned to less than 10 deg/s.

Load kinematics. The maximum vertical velocity and maximum vertical acceleration of the hand during extension was determined for each trial. Constructing an automatic algorithm to reliably determine the appropriate local maximum in the vertical acceleration time series proved difficult because of a proliferation of turning points as a consequence of the double differentiation. However, with additional smoothing, the algorithm produced satisfactory results for at least 80% of the trials for each subject. Data concerning maximum vertical load acceleration from these trials only are included in the analysis.

Joint kinematics. Ankle, knee, hip, and lumbar vertebral angular position at the start of extension were calculated and expressed in terms of flexion from the position of each joint in normal standing. The maximum angular velocity during extension was calculated for each joint. The joint position (relative to normal standing) and angular velocity at the time of maximum vertical load acceleration were also calculated.

Relative phase. Interjoint coordination may be described in terms of the phase relationships between adjacent joint pairs. Relative phase between hip and knee joints and between lumbar vertebral and hip joints was calculated as previously described (Burgess-Limerick et al., 1993) by subtracting the phase angle (inverse tangent of normalized angular velocity/normalized angular position) of the distal joint from the phase angle of the proximal joint at each digitized point in time. A relative phase of zero indicates perfectly-in-phase (synchronous) coordination of adjacent joints, whereas deviations from zero quantify the extent to which coordination deviates from being perfectly in phase. A negative relative phase indicates the distal joint was leading the proximal in its cycle of states; a positive relative phase indicates that the proximal joint was leading the distal.

The pattern of interjoint coordination most commonly observed during the extension phase was quantified by calculating the minimum value of the hip-knee and lumbar vertebral-hip relative phase. The times at which these minimum values occurred were determined and expressed in absolute terms and as a percentage of the extension duration for that trial. The relative phase between each adjacent pair of joints at the time of maximum vertical load acceleration was also calculated.

Hamstring muscle length changes. An estimation of the length changes of the biarticular hip-extensor and knee-flexor muscle-tendon complexes (long head of biceps femoris, semitendinosus, and semimembranosus) was made using expressions for the effect of knee and hip joint rotation derived from direct measurement of 10 female and male cadavers (Kippers, 1990). Length changes were expressed relative to the length of the muscle-tendon complex in normal standing by expressing the angular position of each joint in terms of flexion from its position in normal standing and using the following formulas:

1. Biceps femoris length change (mm) = $1.155 \times$ hip angular flexion (deg) $-0.394 \times$ knee angular flexion (deg).
2. Semitendinosus length change (mm) = $1.220 \times$ hip angular flexion (deg) $-0.798 \times$ knee angular flexion (deg).
3. Semimembranosus length change (mm) = $1.055 \times$ hip angular flexion (deg) $-0.459 \times$ knee angular flexion (deg).

The estimated length of the hamstrings at the start of extension and the estimated maximum concentric velocity during extension were calculated from the angular position data for each trial. The length and velocity of the hamstrings at the time of maximum vertical load acceleration were also estimated.

Electromyographic Data Collection

Electromyographic (EMG) data were collected from 20 subjects by placing surface electrode units (QANTEC, Queensland, Australia) over seven muscles: the medial head of gastrocnemius, vastus medialis, rectus femoris, long head of biceps femoris, medial hamstring group, gluteus maximus, and erector spinae at the level of L3. Prior to placement of the electrode units, the skin was shaved, wiped with alcohol, and lightly abraded. Each electrode unit consisted of two active electrodes (spacing 20 mm), one reference electrode, common mode rejection circuitry, and preamplifier (100×).

The signal was passed via shielded cable to an oscilloscope for monitoring of signal quality and then to the main amplifier, where the signal was band pass filtered (10 to 1000 Hz) and amplified. A 50-Hz notch filter was utilized when necessary to remove interference from electrical sources. The amplified and filtered signal was then passed directly to a personal computer, where the analog signal was digitized at 1 kHz via a QANTEC A-D board and a waveform analysis package (WASP, QANTEC, Queensland, Australia). The EMG data were synchronized with the kinematic data by a common audio tone, which triggered both the collection of EMG data and the subsequent digitizing of the videotape.

Electromyographic Data Analysis

Raw EMG data from each subject were plotted and qualitatively analyzed in conjunction with kinematic data. In addition, the amplitude of the peak burst of EMG activity related to the initial phase of extension and, when this occurred, was determined for the knee-, hip-, and lumbar vertebral-extensor muscles examined for six trials in each mass condition for each subject. This process involved removal of any DC offset, full wave rectification, and linear envelope calculation via a moving average with 50-ms window. An algorithm was then used to identify the maximum value and the time (relative to the start of extension) at which it occurred. In those cases in which a later (and larger) burst of activity occurred (primarily this occurred in the hamstring muscle data), it was necessary to utilize the graphical capabilities of the EMG analysis software to manually estimate the time at which the peak of the first burst of activity occurred.

Statistical Analysis

Some subjects exhibited considerable kinematic variability during the first blocks of trials. For this reason the descriptive and inferential statistics were calculated from only the last 75 trials performed by each of the 39 subjects—that is, 15 trials in each of 5 mass conditions (2.5 to 10.5 kg). The average and standard deviation of the subject mean values for each dependent variable are reported. Evidence of reliable effects of increasing load mass across subjects was sought by calculating mean values of each dependent variable for each mass condition for each subject and then submitting these data to separate one-way repeated-measures analyses of variance (ANOVA) for each dependent variable. Probability values were Greenhouse-Geisser corrected for deviations from sphericity (Schutz and Gessaroli, 1993), and a Bonferroni correction was employed to maintain alpha below 0.05. Only those effects of load mass yielding Greenhouse-Geisser adjusted p values less than 0.001 were considered reliable as a consequence. For the two subjects who did not complete the heaviest weight condition, the mean values for each variable in the 8.5-kg condition were duplicated and submitted as the mean values for the 10.5-kg condition.

Tests of significance should be interpreted in concert with a measure of effect size. The effect size index for ANOVA (f) described by Cohen (1969) is a dimensionless number used to describe the degree of departure from no effect. It was calculated as the standard deviation of the sample means—that is, the standard deviation of the means for each of the five load mass conditions 2.5 to 10.5 kg—divided by an estimate of the within-conditions standard deviation (in this case, the pooled standard deviation described by Thomas and Nelson, 1990, was used). The calculation of f allows comparison of the relative magnitude of the effect of increases in load mass on different dependent variables. An f of 0.1 is considered by convention to be a small effect size, an f of 0.25 is considered a medium effect, and an *f* of 0.4 is considered a large effect size (Cohen, 1969).

RESULTS

Temporal Measures and Load Kinematics

Summary statistics for the temporal measures and load kinematics during lifting are presented in Table 1. The average duration of both flexion and extension phases increased reliably as the load mass was increased from 2.5 to 10.5 kg. Flexion duration (the time taken to bend from normal standing to grasp the load) increased from 0.89 s to 0.97 s, and extension duration (the time taken to return to normal standing) increased from 1.15 s to 1.29 s. For the same change in load mass, the average maximum vertical load velocity and acceleration during extension decreased reliably, from 1.9 m/s to 1.7 m/s and from 6.4 m/s^2 to 4.6 m/s^2. When the load lifted was 2.5 kg, the maximum vertical load acceleration occurred an average of 0.16 s after the start of extension (14.3% of extension duration). Increases in the load mass reliably delayed maximum vertical load acceleration in absolute and relative terms. When the load was 10.5 kg, maximum vertical load acceleration occurred 0.27 s after the start of extension (20.7% of extension duration).

Joint Kinematics

Ankle, knee, hip, and lumbar vertebral angular movement during a single trial is illustrated in Figure 1 in terms of angular position versus time and in terms of relative phase between adjacent joints as a function of time. The length changes of the hamstring muscle-tendon complexes estimated for this trial are also illustrated.

The average (and standard deviation) of subject mean absolute position of the joints in normal standing was 107 deg (6 deg) at the ankle, 168 deg (5 deg) at the knee, 78 deg (4 deg) at the hip, and 100 deg (4 deg) at the lumbar vertebral joint. At the position where the load is grasped (i.e., at the start of extension), the ankle was dorsiflexed an average of 25 deg from normal standing, the knee was flexed by 92 deg, the hip by 87 deg, and the lumbar vertebral joints were flexed by an average of 45 deg from normal standing position.

The variability within each subject of the position adopted at the start of extension was assessed by calculating the standard deviation of joint positions for the last 15 trials within each load condition for each subject. The average values across load and subject were 3 deg at the ankle, 6 deg at the knee, and 2 deg at both the hip and lumbar vertebral joints.

For 37 subjects the position of the knee joint at the start of extension was on average flexed between 40 and 130 deg from normal standing (absolute included knee angles between 120 and 40 deg). One subject adopted a posture at the start of extension that involved only 20 deg of knee flexion from normal standing (a stoop), whereas

TABLE 1

Summary Statistics for Flexion and Extension and Load Kinematic Dependent Variables

Dependent Variable	*Mean*	*sd*	$X^{10.5} - X^{2.5}$	*F*	*p*	*f*
Flexion duration (s)	0.93	0.14	0.08	44.114	<0.001	0.19
Extension duration (s)	1.21	0.17	0.14	31.734	<0.001	0.30
Maximum vertical load velocity (m/s)	1.79	0.26	−0.24	56.044	<0.001	0.31
Maximum vertical load acceleration (m/s/s)	5.43	1.45	−1.78	75.829	<0.001	0.45
Time of maximum vertical load acceleration (s)	0.21	0.07	0.10	63.840	<0.001	0.59
Time of maximum vertical load acceleration (%)	17.65	4.77	6.38	40.080	<0.001	0.51
Horizontal distance hand to ankle at start of extension (mm)	319	38	−8.7	5.365	=0.011	0.08

Note: $X^{10.5} - X^{2.5}$ refers to the average difference between 10.5- and 2.5-kg load conditions. Degrees of freedom = (4,152). The probability values reported in all tables are Greenhouse-Geisser adjusted probabilities (Schutz and Gessaroli, 1993).

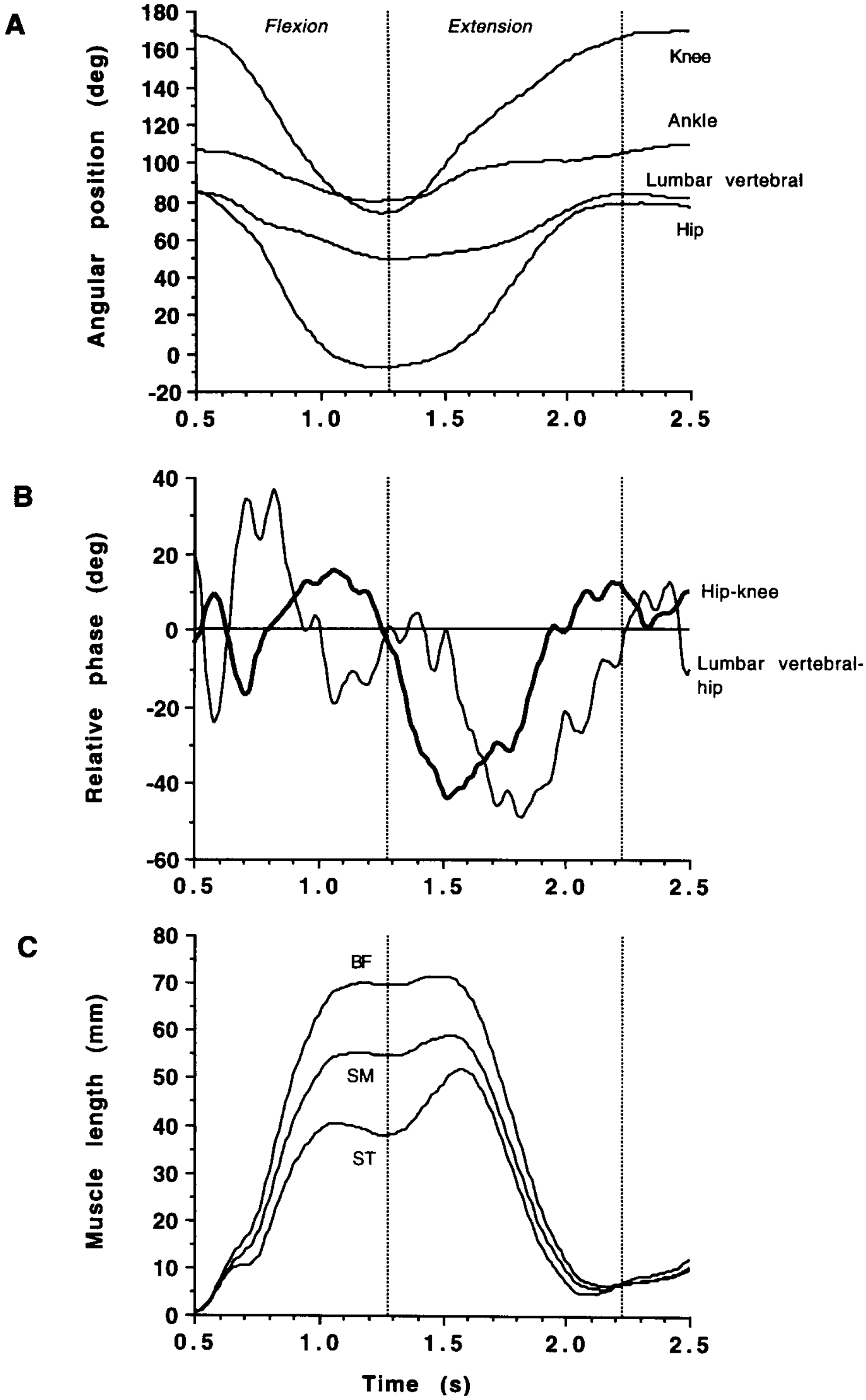

Figure 1. *Kinematic data for a subject performing a single lift of a load from 9 cm above the ground using a self-selected technique. Trial 100, subject EK, load mass = 10.5 kg. A. Ankle, knee, hip, and lumbar vertebral angular position as a function of time. B. Lumbar vertebral-hip and hip-knee relative phase as a function of time. Negative relative phase indicates the distal joint is leading the proximal in its cycle of states. C. Estimated length changes of biceps femoris (BF), semimembranosus (SM), and semitendinosus (ST) muscle-tendon complexes as a function of time. Estimated muscle lengths are expressed relative to the length in normal standing.*

another subject adopted a knee position 160 deg flexed from her normal standing position (a full squat).

Summary statistics for the joint kinematic data are presented in Table 2. The average overall effect of increases in load mass from 2.5 kg to 10.5 kg was to slightly increase the flexion of the knee, hip, and lumbar vertebral joints at the start of extension, although substantial individual differences existed in the response to changes in load, and only the changes at the hip and lumbar vertebral joints were statistically significant.

Interjoint Coordination

The most common pattern of relative phase relationships is illustrated in Figure 1B. In most cases knee extension initially led hip extension. This coordination is indicated by a negative local minimum in the hip-knee relative phase during extension, which is not preceded by a positive local maximum during extension. Similarly, a local minimum in the lumbar vertebral-hip relative phase during extension indicates that hip extension initially led lumbar vertebral extension in its cycle of states.

These patterns were quantified by calculating the minimum relative phase between each pair of joints during the extension phase for each trial. Subjects were deemed to consistently exhibit these patterns at each joint pair if the minimum relative phase values during extension in more than 60% of trials were negative and were not preceded by a positive local maximum during extension. This criterion was satisfied for the hip-knee relative phase data for all subjects and by data from 31 subjects for the lumbar vertebral-hip relative phase. Some of the remaining subjects exhibited highly variable lumbar vertebral-hip relative phase patterns, whereas in others, lumbar vertebral extension led hip extension. Minimum lumbar vertebral-hip relative phase data from these 8 subjects was omitted from the statistical analysis.

Summary statistics are presented in Table 3. The effect of increasing load mass on the interjoint coordination was to reliably increase the deviation from perfectly-in-phase coordination of both pairs of joints considered (Figure 2). The time of minimum relative phase in extension was also delayed in absolute and relative terms for both pairs of joints as the load mass increased from 2.5 to 10.5 kg (Table 3).

Hamstring Kinematics

Figure 1C illustrates the estimated length changes of the biarticular hamstrings relative to normal standing in a single lifting trial. During bending to pick up the load, the lengthening effect of hip flexion is greater than the shortening

TABLE 2

Summary Statistics for Joint Kinematic Variables

Dependent Variable	*Mean*	*sd*	$X^{10.5} - X^{2.5}$	*F*	*p*	*f*
Ankle position at the start of extension (deg)	25.12	8.85	0.71	1.7	=0.204	0.05
Knee position at the start of extension (deg)	91.89	25.00	4.99	4.9	=0.015	0.08
Hip position at the start of extension (deg)	86.81	9.30	2.28	12.2	<0.001	0.09
Lumbar vertebral position at the start of extension (deg)	44.97	9.77	2.31	18.0	<0.001	0.09
Maximum ankle angular velocity during extension (deg/s)	55.64	13.64	5.43	4.4	=0.252	0.14
Maximum knee angular velocity during extension (deg/s)	130.04	34.35	3.73	1.3	=0.268	0.05
Maximum hip angular velocity during extension (deg/s)	137.8	22.8	−0.49	0.32	=0.727	0.02
Maximum lumbar vertebral angular velocity during extension (deg/s)	69.12	18.23	3.078	3.9	=0.036	0.07

Note: The position of each joint at the start of extension is expressed in terms of flexion from average normal standing posture. Positive angular velocity values indicate ankle plantar-flexion and knee, hip, and lumbar vertebral extension. Degrees of freedom = (4,152).

TABLE 3

Summary Statistics for Relative Phase Variables

Dependent Variable	*Mean*	*sd*	$X^{10.5} - X^{2.5}$	*F*	*p*	*f*
Minimum hip-knee relative phase (MHK) during extension (deg)	−19.7	6.24	−8.1	55.3	<0.001	0.49
Time of MHK after the start of extension (s)	0.28	0.11	0.10	37.1	<0.001	0.36
Time of MHK after the start of extension (%)	23.6	8.7	5.6	24.323	<0.001	0.23
Minimum lumbar vertebral-hip relative phase (MVH) during extension (deg)	−43.2	15.1	−6.5	9.415	<0.001	0.15
Time of MVH after the start of extension (s)	0.57	0.11	0.12	24.606	<0.001	0.40
Time of MVH after the start of extension (%)	47.37	8.04	6.38	9.8	<0.001	0.28

Note: A negative value indicates that the distal joint of the pair was leading the proximal in its cycle of states. See text for details. Degrees of freedom for MHK = (4,152); for MVH = (4,120).

effect of knee flexion, and consequently the muscles lengthen. During lifting of the load, the shortening effect of hip extension is greater than the lengthening effect of knee extension, and the muscles shorten overall. Although the patterns of estimated length changes of the three muscles are similar in form, the effect of knee movement is greater on the semitendinosus than on the other hamstrings.

The differential effect of knee and hip flexion on the length changes of the three muscles is reflected in the average estimated lengthening of the hamstrings at the start of extension (25 mm longer than normal standing length for the semitendinosus, 49 mm for semimembranosus, and 64 mm for the biceps femoris). Similarly, the average maximum concentric velocity of the hamstrings during extension was 89 mm/s for semitendinosus, 96 mm/s for semimembranosus, and 115 mm/s for biceps femoris.

Kinematic Consequences of Hip-Knee Coordination

For all subjects, a negative local minimum in the relative phase of hip-knee coordination consistently occurred during extension, on average 280 ms after the start of extension or at about 24% of the extension duration. This indicates that knee extension led hip extension in its cycle of states early in the extension phase of lifting.

A closer examination of the estimated length changes of the biarticular hamstrings illustrated in Figure 1C reveals an interesting consequence of this pattern of coordination. Overall, during the extension phase of the lifting movement, the shortening effect of hip extension on the hamstrings is greater than the lengthening effect of knee extension, and thus the muscles shorten during the extension phase. However, the onset of rapid shortening is delayed by knee extension initially leading hip extension. The estimated shortening velocity of the biarticular hamstrings was relatively small early in extension. The relatively greater dependence of semitendinosus length on knee position frequently resulted in an estimation of a short period of lengthening in early extension.

This observation was explored quantitatively by describing the instantaneous kinematics at the time (in early extension) at which the vertical acceleration of the load was greatest.

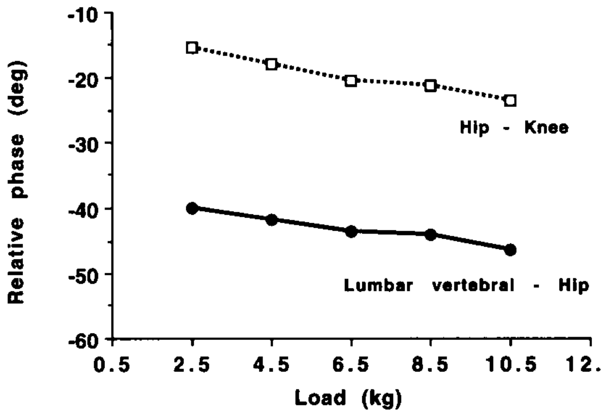

Figure 2. *Minimum lumbar vertebral-hip and hip-knee relative phase during extension.*

Summary statistics for these data are reported in Table 4.

Although the velocity of hamstring length changes during extension was estimated to reach, on average, minimum values of the order of −100 mm/s (i.e., shortening at 100 mm/s), the estimated velocity at the time of maximum vertical load acceleration (i.e., approximately 200 ms after the start of extension) was −21.4 mm/s for biceps femoris, −11.8 mm/s for semimembranosus, and +7.5 mm/s for semitendinosus.

The magnitude of the maximum deviation from synchronous coordination of the hip and knee (as measured by minimum hip-knee relative phase during extension) was noted earlier to increase with load. A similar observation was made when the hip-knee relative phase was calculated at the time of maximum vertical load acceleration (Figure 3A). Consequently, the estimated concentric velocity of the hamstring muscles at this time decreased as the mass of the load lifted was increased (Figure 3B). Even though the time of maximum vertical load acceleration occurred later in extension (in both absolute and relative terms) as load mass increased, the hamstrings were estimated to be shortening more slowly at this time as load was increased.

Electromyography

Qualitative examination of the EMG data revealed that the patterns of activity observed for each subject appeared consistent, although there was some variability among subjects. Figure 4 presents raw electromyographical data for the same trial as described in Figure 1.

Gastrocnemius activity was greatest late in the extension phase of the lift, when the load was being moved away from the body to be placed on the shelf. In about half the subjects a smaller burst of activity also occurred at the start of the extension movement.

Activity in the rectus femoris in most cases began in late flexion, peaking at or just after the start of extension and decreasing rapidly thereafter. A similar pattern generally occurred in vastus medialis, although on average the peak activity was later and activity more prolonged. In most cases vastus medialis activity remained elevated above baseline levels for 500 ms or more after the start of extension. The pattern of activity observed in the biceps femoris and

TABLE 4

Summary Statistics for the Value of Selected Kinematic Variables at the Time of Maximum Vertical Load Acceleration

Dependent Variable at Time of Maximum Vertical Load Acceleration	*Mean*	*sd*	$X^{10.5} - X^{2.5}$	*F*	*p*	*f*
Ankle position (deg)	22.68	8.59	−0.52	0.856	=0.406	0.04
Knee position (deg)	83.04	23.63	2.78	1.16	=0.314	0.05
Hip position (deg)	80.92	10.69	3.32	4.997	=0.019	0.10
Lumbar vertebral position (deg)	42.55	10.20	2.75	10.840	<0.001	0.09
Ankle angular velocity (deg/s)	23.78	12.30	9.03	13.37	<0.001	0.26
Knee angular velocity (deg/s)	79.61	31.71	16.44	8.415	=0.002	0.19
Hip angular velocity (deg/s)	45.71	17.19	−3.85	1.467	=0.236	0.07
Lumbar vertebral angular velocity (deg/s)	21.39	11.23	−3.80	3.361	=0.055	0.11
Biceps femoris length (mm)	60.71	10.88	2.81	3.037	=0.073	0.09
Semitendinosus length (mm)	24.70	16.06	2.01	1.323	=0.269	0.05
Semimembranosus length (mm)	47.15	10.73	2.50	2.808	=0.086	0.08
Biceps femoris velocity (mm/s)	−21.38	20.06	10.61	15.510	<0.001	0.19
Semitendinosus velocity (mm/s)	7.51	27.4	17.43	20.893	<0.001	0.23
Semimembranosus velocity (mm/s)	−11.83	19.37	11.41	19.498	<0.001	0.21
Hip-knee relative phase (deg)	−12.27	9.64	−9.39	30.288	<0.001	0.36
Lumbar vertebral-hip relative phase (deg)	−2.9	15.86	−0.70	0.432	=0.658	0.03

Note: Degrees of freedom = (4, 152).

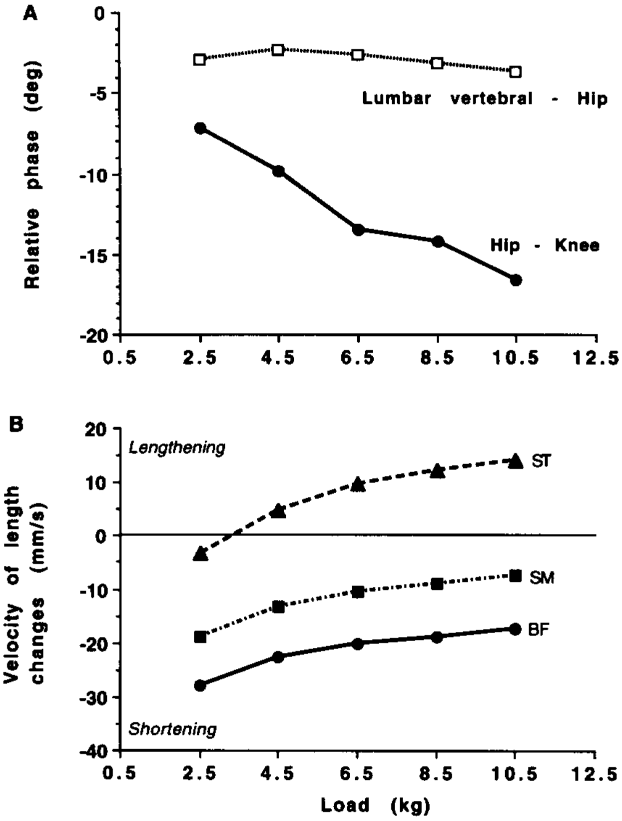

Figure 3. *Instantaneous kinematics at the time of maximum vertical load acceleration. A. Lumbar vertebral-hip and hip-knee relative phase. B. Estimated velocity of biceps femoris (BF), semimembranosus (SM), and semitendinosus (ST). A negative average velocity indicates that the muscle was estimated to be shortening.*

medial hamstrings was similar in most cases. Two bursts of hamstring activity were noticeable in about half the subjects. The peak of the first burst occurred just after the start of load movement, with a second burst later in the movement when the muscles were shortening rapidly. The activity of the hamstrings in the remaining subjects was more continuous throughout lifting.

Gluteus maximus was continuously active throughout the movement of the load in most subjects, although in about half the subjects the activity was noticeably larger during the first half of the extension movement. Little or no evidence was seen of a period of electromyographical silence in the erector spinae. In most subjects the erector spinae were active throughout the lifting of the load, and in about half the subjects, the activity was noticeably larger just after the start of load movement.

The peak amplitude (in arbitrary units) of the EMG linear envelope of the medial and lateral hamstrings, gluteus maximus, and erector spinae was increased with increases in load mass (Table 5). No reliable effect was observed for either rectus femoris or vastus medialis. The timing of peak EMG activity is illustrated in Figure 5.

Peak activity in the knee-extensor muscles occurred on average before the peak of activity in the hip-extensor muscle groups. Peak rectus femoris activity occurred on average before the peak activity of vastus medialis, and the peak activity in the hamstrings occurred on average before the peak activity in the gluteus maximus. Increases in load mass delayed the peak activity of vastus medialis, gluteus maximus, and erector spinae (Table 5).

DISCUSSION

The posture adopted at the start of lifting has been frequently used to characterize different techniques used in symmetric bimanual lifting. In the present research the average posture adopted at the start of the lift involved a moderate range of knee flexion (average 92 deg of flexion from normal standing), a similar amount of hip flexion (87 deg) and 45 deg of lumbar vertebral flexion from normal standing. The lifting technique self-selected by the majority of subjects involved a posture at the start of extension that might be described as intermediate between the extremes of stoop and full squat. This conclusion is consistent with data previously reported by Park (1973) and Gagnon and Smyth (1992).

The average range of lumbar vertebral flexion from normal standing observed in the current study (45 deg) is similar to the ranges previously reported during lifting from a starting posture involving moderate knee flexion (compared with 45 deg, Davis and Troup, 1965; 46 deg, Davis et al., 1965; 40 deg, Potvin, McGill, and Norman,

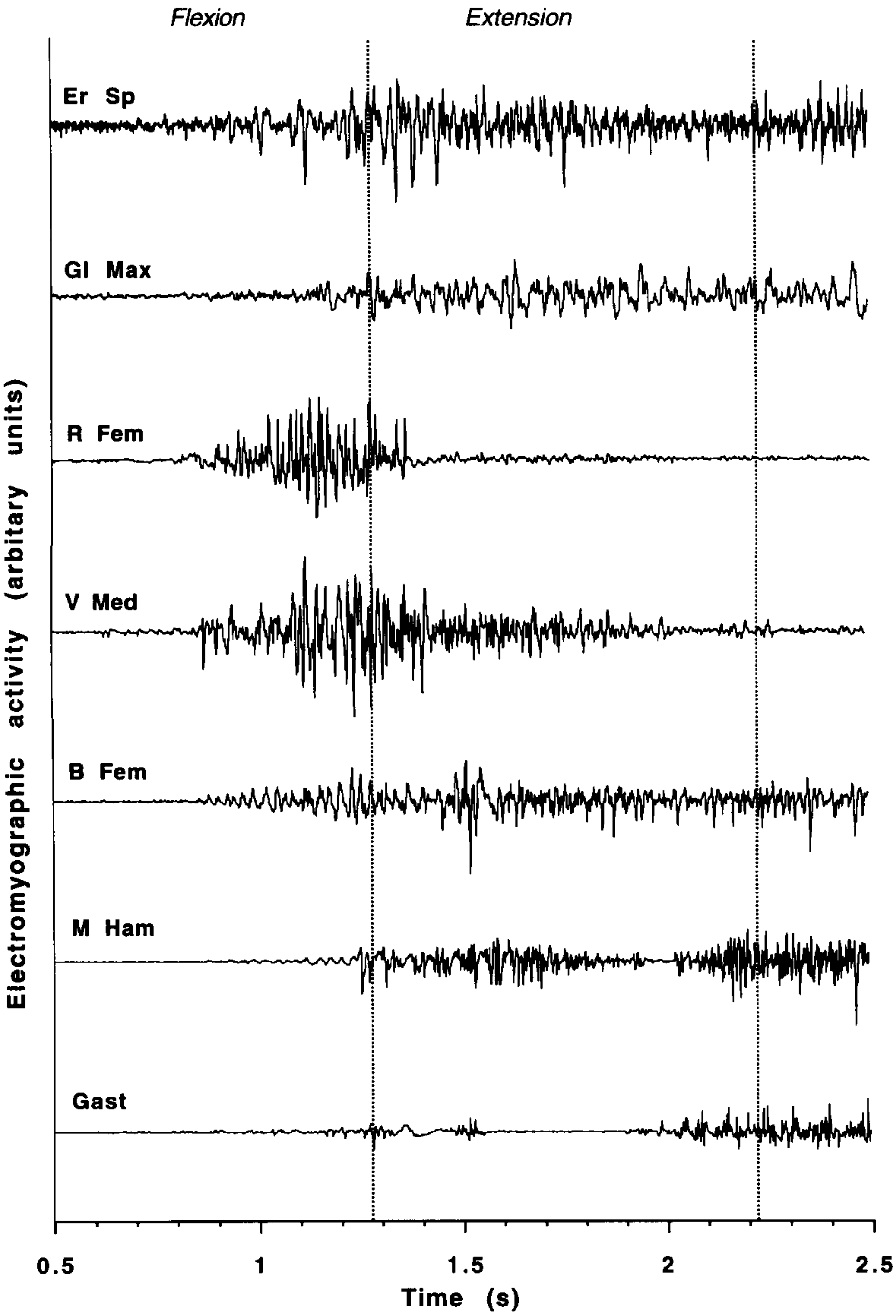

Figure 4. *Raw electromyographical data for medial gastrocnemius (Gast), vastus medialis (V Med), rectus femoris (R Fem), long head of biceps femoris (B Fem), medial hamstrings (M Ham), gluteus maximus (G Max), and erector spinae at the level of L3 (Er Sp) for one subject performing a single lift of a load from 9 cm above the ground using a self-selected technique. Trial 100, subject EK, load mass = 10.5 kg. Note: This is the same trial as that for which kinematic data are presented in Figure 1.*

TABLE 5

Summary Statistics for Amplitude and Time of Peak Linear Envelope Electromyographical Activity in Six Lower Limb and Trunk-Extensor muscles

Dependent Variable	*Mean*	*sd*	$X^{10.5} - X^{2.5}$	*F*	*p*	*f*
Peak vastus medialis (VM) amplitude (au)	1.83	0.68	0.12	1.549	=0.225	0.08
Peak rectus femoris (RF) amplitude (au)	1.84	0.70	0.21	1.576	=0.217	0.10
Peak biceps femoris (BF) amplitude (au)	1.50	0.57	0.33	8.689	<0.001	0.25
Peak medial hamstrings (MH) amplitude (au)	1.36	0.72	0.49	14.311	<0.001	0.24
Peak gluteus maximus (GM) amplitude (au)	1.67	0.71	0.33	6.286	<0.001	0.18
Peak erector spinae (ES) amplitude (au)	1.84	0.76	0.52	17.275	<0.001	0.25
Time of peak VM after the start of extension (s)	0.15	0.15	0.15	8.642	<0.001	0.34
Time of peak RF after the start of extension (s)	0.05	0.20	0.09	3.937	=0.015	0.24
Time of peak BF after the start of extension (s)	0.26	0.16	0.19	5.818	=0.003	0.44
Time of peak MH after the start of extension (s)	0.27	0.15	0.10	3.493	=0.162	0.23
Time of peak GM after the start of extension (s)	0.32	0.15	0.13	6.639	<0.001	0.28
Time of peak ES after the start of extension (s)	0.19	0.23	0.12	8.642	<0.001	0.20

Note: Degrees of freedom = (4,76); au = arbitrary units.

1991) and less than the 60 deg of lumbar vertebral flexion that is reported to occur when the spine is fully flexed (Potvin et al., 1991). In concert with the absence of an EMG silent period in erector spinae, these data suggest (consistent with Potvin et al., 1991) that in most cases, the paravertebral ligaments were not substantially stretched.

An adequate description of lifting technique requires consideration of the pattern of interjoint coordination, as well as the posture adopted at the start of the lift. A consistent

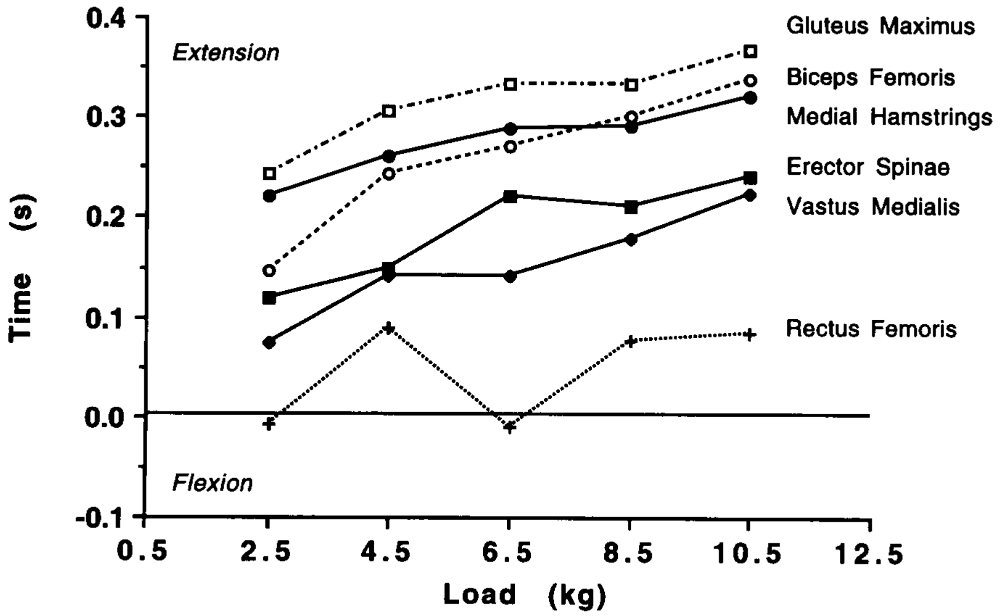

Figure 5. *The average time of peak linear envelope EMG activity for six lower limb and back muscles expressed relative to the start of extension (t = 0).*

distal-to-proximal pattern of coordination between knee, hip, and lumbar vertebral joints was seen in the majority of subjects, although some variability existed in the patterns of interjoint coordination observed.

A consequence of the observed interjoint coordination is a delay between the start of the lift and the attainment of significant lumbar vertebral extension. The interjoint coordination is exaggerated when load mass increases and this increases the lumbar vertebral extension delay. Grieve (1974) provided a figure of 200-ms delay before lumbar vertebral extension for subjects lifting 5-kg loads. Davis et al. (1965) reported a 500-ms delay when the load was 20 kg, and Grieve (1974) put the delay at 600 to 700 ms when the load was greater than 25 kg. The direct relationship between load mass and interjoint coordination does not, therefore, appear to be restricted to the range of loads examined here (see also Kumar, 1974; Scholz, 1992).

Although a number of authors (e.g., Davis and Troup, 1965; Davis et al., 1965; Grieve, 1974, 1977; Kumar, 1974; Scholz, 1992, 1993a; Troup, 1977) have commented on the degree to which lumbar vertebral extension is delayed after the start of lifting, Potvin et al. (1991) chose to emphasize the "more or less simultaneous" nature of the coordination between lumbar vertebral and hip joints. Potvin et al. noted that hip and lumbar vertebral extension occurs at the same time during lifting and not sequentially, as some biomechanical models have assumed. These apparently conflicting views can be reconciled by consideration of the relative phase between lumbar vertebral and hip joints during lifting (Burgess-Limerick et al., 1992). Such a description illustrates that although hip and lumbar vertebral extension does occur contemporaneously, the coordination systematically deviates from perfectly in phase and the magnitude of the deviation is load dependent.

In this light, it is instructive to note the data presented by Enoka (1988) as typical of an Olympic weight lifter lifting a load of the order of 100 kg from a height of 23 cm. During the first 400 ms after the start of the lift, 60 deg of knee extension occurred, accompanied by a small amount of trunk *flexion*. During the next 500 ms, 45 deg of trunk extension occurred and the knee extended a further 30 deg. It appears that trained weight lifters exhibit an interjoint coordination that is, at least qualitatively, similar to the pattern exhibited by the majority of (untrained) subjects observed in the present research.

The increased deviation from synchronous interjoint coordination that accompanied increased load in the present research is probably related to changes in load kinematics that also occurred. The effect of mass on interjoint coordination was accompanied by an increased delay before maximum vertical load acceleration occurred. The common dependence of load kinematics and the interjoint coordination on load mass supports the proposition that the delay observed in lumbar vertebral extension has the functional consequence of avoiding rapid lumbar extension early in the movement, when load acceleration is greatest (Davis and Troup, 1965; Davis et al., 1965). It is assumed that joint moments are also highest during this time, an assumption supported by de Looze, Toussaint, van Dieen, and Kemper (1993), who reported peak joint moments occurring at 25% of the extension duration when lifting 15.3 kg using two different techniques.

The observed pattern of interjoint coordination reduces trunk extensor muscular effort. The length-tension and velocity-tension characteristics of muscle complexes are such that strength is decreased when muscles are shortened and also decreases with increased concentric velocity (Hof, 1984). A partially flexed lumbar vertebral spine places the trunk extensor muscles in a relatively lengthened position. In this way, trunk extensor strength is increased with lumbar vertebral flexion and, if the velocity of lumbar vertebral extension is reduced (that is, the erector spinae shorten less rapidly), then strength is again increased (Marras, Joynt, and King, 1985). The observed delay before the attainment of rapid lumbar vertebral extension has the consequence of maintaining trunk extension strength early in the extension phase, when load acceleration is greatest, through avoiding rapid shortening of the trunk-extensor muscles at this time.

In the current study the average angular velocity of lumbar vertebral extension at the time of maximum vertical load acceleration was about 20 deg/s, which is considerably less than the average maximum lumbar vertebral extension velocity (70 deg/s).

The coordination observed between hip and knee during lifting from a low height, and the exaggeration of this pattern with increases in load mass, has been described previously (Burgess-Limerick et al., 1991, 1993; Scholz, 1993a, 1993b). However, here the interpretation of the consequences of knee extension leading hip extension is assisted by consideration of the estimated length changes of the biarticular hamstrings and the EMG data from knee- and hip-extensor muscles.

It has been proposed (van Ingen Schenau, 1989, 1990; Zajac, 1993) that biarticular muscles function in general to transport the mechanical output of monoarticular muscles to required joints—that is, they allow the monoarticular muscles to contribute to the movement regardless of the direction of the forces required to meet the task objectives. Coactivation of monoarticular and biarticular antagonists is considered to be highly functional in many skills, particularly those involving powerful lower-limb extension movements.

The coordination between knee and hip joints observed here during manual lifting has the consequence of delaying rapid shortening of the biarticular hamstrings. This has two functional consequences. Early in the extension phase the hamstring muscles are relatively lengthened and not shortening rapidly. The hamstrings are thus better able to contribute to an extension moment around the hip when load acceleration is greatest. A related consequence is that relatively isometric contraction of the hamstrings allows them to act to some degree as a tendon. In this way, tension simultaneously developed in the monoarticular knee-extensor muscles may contribute to the extension moment at the hip, while at the same time maintaining the external force vector in the direction required to meet the task requirement of maintaining balance.

Consistent with this interpretation, the EMG data revealed that coactivation occurred between the monoarticular knee-extensor muscle examined (vastus medialis) and the biarticular hamstrings during the first half of the extension phase. Similar coactivation has been observed during lifting by Ekholm, Nisell, Arborelius, Hammerberg, and Nemeth (1984); Nemeth, Ekholm, and Arborelius (1984); and Toussaint, van Baar, van Langen, de Looze, and van Dieen (1992). During lifting and lowering, similar coactivation was observed by de Looze et al. (1993). Toussaint et al. (1992) and de Looze et al. (1993) similarly concluded that the coactivation allowed the vasti to contribute to hip extension, as well as functioning to maintain balance.

The exaggeration of the distal-to-proximal coordination of knee, hip, and lumbar vertebral extension as load mass increases has been previously described. Davis and Troup (1965) interpreted this observation as a tendency of the subjects "to change the flexed knee lift into a stooping lift" (p. 327) when the load was heavy, a conversion they considered to be dangerous. However, the results of the present investigation suggest an alternative interpretation.

The observed coordination has functional consequences, and the changes that occur with increased load have the effect of reducing the muscular effort required. To say that the technique becomes more of a stoop as the load increases misses the crucial consequence of increased knee extension velocity, which is to delay the onset of rapid hamstring shortening. The large range of hip flexion and small range of knee flexion involved in adopting a stoop position means that the hamstrings are relatively lengthened at the start of extension. However, the hamstrings must then immediately shorten rapidly and continue to do so throughout the lifting of the load. Thus the capability of the hamstrings to produce tension and the possible contribution of monoarticular knee-extensor muscles to hip extension are both reduced. Rather than executing a stoop lift when the load mass increased, the majority of subjects in this investigation made increasing use of rapid knee extension early in the lifting of the load, and hence reduced the shortening velocity of the hamstrings early in the movement, when vertical load acceleration was greatest.

The human neuromuscular system habitually discovers and uses functionally relevant (i.e., successful) solutions to an infinitely diverse set of motor problems. In this case it appears that the lifting technique naturally adopted by many subjects involves a pattern of movement that is likely to reduce muscular effort. Further, the human system is sufficiently sensitive that alterations in interjoint coordination occur in response to relatively small changes in load mass.

This effect of load mass on the kinematics of self-selected lifting technique illustrates the influence of task characteristics on the patterns of movement that arise. This influence may explain to some extent the reported failure of training programs to achieve persistent changes in lifting behavior (e.g., Chaffin, Gallay, Wolley, and Kuciemba, 1986; Pheasant, 1986, p. 207).

In any case, if it can be assumed that muscular fatigue contributes to injuries suffered as a consequence of lifting (Edwards, 1988; see also Potvin, 1992), then a technique that reduces muscular effort may be preferred. Thus if an attempt is made to alter employees' lifting technique, then care should be taken to ensure that the normal coordination between knee, hip, and lumbar vertebral joints is not disrupted. Education in general lifting principles or guidelines and the use of exploratory learning techniques (Newell, 1991) to assist employees to discover individually appropriate modes of coordination that reduce effort may be more effective than prescriptive instructions of the "best" lifting technique.

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