

Neuromuscular Coordination of Squat Lifting, I: Effect of Load Magnitude

Background and Purpose. In this study, we examined changes in kinematic and electromyographic (EMG) measurements of the coordination (ie, the relative timing of joint movements and muscle activity) of a squat-lifting task in response to lifting increasing loads. **Subjects.** Fifteen male industrial workers served as a sample of convenience. **Methods.** Subjects lifted a weighted crate containing 15% to 75% of their maximum lifting capacity using a symmetrical squat-lift technique. Movement kinematics were obtained with videography. The relative phase between joint motions was derived. The EMG activity of the vastus lateralis muscle (VL) and the erector spinae muscle (ES) was recorded, and the relative timing of their onsets and peaks was estimated. **Results.** The relative phase of movement between joints such as the knee and lumbar spine changed in a quasi-linear fashion with increasing load during lifting but not during lowering. The relative time of onset of ES EMG activity and its peak activity changed in a manner consistent with the interjoint relative phase results. The timing of VL events were not affected by increasing the load. **Conclusion and Discussion.** Relatively continuous changes in interlimb coordination occur when increasing the load lifted from an initial squatting posture. Changes in EMG relative timing partially corroborate the kinematic evidence for changes in coordination with load scaling. The results indicate the need for further study to determine whether the observed changes in coordination are beneficial or detrimental to the musculoskeletal system. Clinicians should evaluate performance of this task under a range of task conditions. [Scholz JP, Millford JP, McMillan AG. Neuromuscular coordination of squat lifting, I: effect of load magnitude. *Phys Ther.* 1995;75:119-132.]

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Numerous studies analyzing manual lifting have appeared over the past 30

years, motivated in part by the fact that load handling, including lifting, is

a major etiological factor in low back injury.¹⁻⁴ Much of the published work on manual lifting has examined the effect of the lifting technique, load magnitude, rate of lifting, or some combination thereof on the safety^{3,5-8} or efficiency^{6,9} of the task. "Safety" is usually defined with respect to musculoskeletal injury to the back and more often than not viewed in terms of lumbosacral compression. Despite tremendous advances in our understanding of lifting performance, the incidence of lifting-related back inju-

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ries does not appear to have decreased appreciably.¹⁰⁻¹² Moreover, controversy exists about which lifting technique is "safest" under any given set of lifting conditions.^{13,14}

Lifting moderate to heavy objects requires coordination of participating neuromuscular components such that control of the load's trajectory to a desired endpoint is achieved while maintaining balance and minimizing stress on the joints of the body. The difficulty faced in meeting these different requirements is reflected in the high incidence of injury, especially to the back, when lifting moderate to heavy loads. Grieve¹⁵ noted some time ago that our lack of knowledge about patterns of movement during lifting could prove to be a stumbling block to understanding the mechanisms of back injury. Therefore, it is surprising that so few studies have explored the coordination (ie, the relative timing of joint movements and muscle activity) of this act.^{12,16-20} The coordination of the limbs and trunk during lifting has also been ignored in published guidelines of lifting performance.²¹

Typically, studies of lifting have defined the technique used by subjects in terms of the initial lifting posture alone,^{5,6,8,9,11,22-26} without considering changes in neuromuscular coordination that might occur as task variables are varied. One may ask, then, whether the coordination of a squat lift, which begins from a bent-knee, relatively straight-back posture, remains the same regardless of factors such as the magnitude of the load or the speed of the lift, or is even the same among different individuals. If changes in neuromuscular coordination occur as task variables are varied, can these changes be attributed to differences in the capacity of muscles controlling different movement components (eg, knee and back) to overcome the inertia and change in the momentum of the load (ie, a purely mechanical effect)? Or are they the result of nonreflex changes in the timing of muscle activation? Do such coordination changes increase the stress on the musculoskeletal system or actually reduce the magnitude of

stress that would occur otherwise? Are there individual differences in how individuals coordinate the body even when lifting from the same initial posture?

As noted, the task of lifting objects requires the resolution of several task demands. An understanding of how the nervous system solves those demands could prove useful for setting realistic guidelines for worker performance. Such information could also be essential to develop effective programs for training workers how to lift. The experiments described in this article represent a continuing effort in our laboratory^{17,18} directed toward enhancing our understanding of such issues.

Studies of the Coordination of Lifting

Changes in the coordination of lifting with increases in load magnitude have been reported. Such analyses generally have been restricted to relations between the motion of one segment (eg, the trunk) and the load.^{8,19,20} Davis and colleagues,²⁰ for example, reported that as the load was increased from 0 to 40 kg, there was an increasing delay of the onset of back extension with respect to vertical motion of the pelvis during squat lifting. More recently, Schipplein et al¹⁹ reported that subjects who were allowed to lift in a self-selected manner changed the coordination of their body from a quasi-squat-lift pattern to a pattern more similar to a back lift (actually a trunk-kinetic lift²²) as the load was increased from 50 to 250 N. Neither of these studies used specific measures of interjoint coordination.

Direct measures of interjoint coordination during lifting have been used in several recent lifting experiments.^{17,18,27,28} In studies similar to our own, Burgess-Limerick and colleagues^{27,28} calculated the relative phase of motion between two joints based on each joint's instantaneous position and velocity (see Scholz¹⁸ and the "Method" section). They reported an increased phase delay between knee and hip extension during lifting

with relatively small changes in load.²⁸ As was true for the study of Schipplein et al,¹⁹ subjects were allowed to lift the load freestyle, making interpretation of the reported changes in coordination difficult. Moreover, all of these studies used loads that were chosen independent of each subject's lifting capacity.

A comprehensive study of the coordination of squat lifting was recently conducted by Scholz.^{17,18} Specific measures of coordination were used. Six male college students lifted loads that were 0% to 75% of their estimated maximum lifting capacity (MLC), beginning from a bent-knee, relatively straight-back posture. The effect of changing the load magnitude on the task's coordination was studied. Another goal of that work was to identify collective variables²⁹ for this task.

Collective variables are measures of the coordination of essential movement components that provide a quantitative description of a behavior's coordination. There are fewer collective variables than the many possible coordination measures that could be defined for a system composed of very many components. Collective variables often take the form of measures of the relative timing or relative phasing between key movement components (see "Method" section).²⁹⁻³⁴ The results of the reported experiments suggested that as few as two or three relative timing measures may be adequate to characterize the coordination of the squat-lifting task.¹⁷ The putative collective variables (assuming a bilaterally symmetrical lift) were a measure of coordination between (1) the lower extremity and the back, (2) the lower and upper extremities, and (3) joints within the lower extremity. These measures, particularly the relative timing of knee-lumbar spine motion, were shown to change in a relatively continuous fashion as the load to be lifted was made heavier. The influence of increasing the load to be lifted on the coordination between pairs of lower-extremity joints, however, was relatively small. Thus, changing a task parameter (ie, load) led to changes in the coordination of lifting, even when begun from ap-

proximately the same initial squatting position.

In a more recent extension of these experiments, we examined the behavior of continuous estimates of coordination, the relative phase between the motion of different joints (see "Method" section), and we examined coordination during lowering the load as well.¹⁸ Decreases in the stability of coordination between different lower-extremity joints and a trend toward greater instability of knee-lumbar spine coordination were found as the load was increased.¹⁸ The results of the relative-phase analyses were generally consistent with the earlier relative-timing results for the lifting phase of the task. Phase lags between joint motions were much smaller during lowering than during lifting the load; tended to be of opposite sign (eg, knee motion lagging back motion during lowering and back motion lagging knee motion during lifting); and were less influenced by the magnitude of the load, if at all.

The purposes of the experiments reported in this article were (1) to determine the extent to which changes in the timing between knee and back muscle activation contribute to previously observed changes in coordination of the knee and lumbar spine joint movement and (2) to extend our previous work on the coordination of this task to experienced workers whose job required lifting moderate to heavy loads. Therefore, electromyographic (EMG) activity from the vastus lateralis muscle (VL) and the erector spinae muscle (ES) was recorded in addition to obtaining kinematic data. Although many studies of lifting have measured EMG activity,^{2,36-43} only a few have explored issues related to neuromuscular timing^{36,37} and these studies were limited in scope. We hypothesized that measurements of

Table 1. Subject Height (in Meters), Weight (in Newtons), and Estimated 100% Maximum Lifting Capacity (MLC) (in Newtons)

	Height (m)	Weight (N)	MLC (N)
\bar{X}	1.79	842.4	578.2
SD	0.06	123.7	95.7
Range	1.68-1.89	645.0-1,147.6	444.8-741.3

EMG timing would be consistent with coordination measurements derived from movement kinematics. In addition, we hypothesized that the effect of increasing the weight lifted on the coordination of this task in this group of industrial workers would be the same as was found for college students studied previously.^{17,18}

Method

Subjects

Fifteen male subjects, ranging in age from 26 to 52 years ($\bar{X} \pm \text{SD}$) = 35.1 ± 7.6), were recruited as the sample of convenience from salaried staff of two local companies and the University of Delaware. Physical characteristics of the sample are presented in Table 1. The subjects chosen engaged in manual lifting as part of their jobs. All subjects signed an approved informed consent form after passing an initial musculoskeletal screening examination administered by a licensed physical therapist. Musculoskeletal screening was performed to ensure that subjects had no apparent asymmetries of posture or of movement of the pelvis or spine and that the length of their legs were approximately equal.⁴⁴ Subjects had no history of recurrent back pain nor any recent episodes of back injury.

Preparation

Two shuttered ($1/500$ -s) video cameras were synchronized, and their signals input to two industrial video recorders* to record the motion of the body. Reflective markers were attached to major landmarks of the right extremities and to the trunk using adhesive Velcro®.[†] The markers were placed at the base of the fifth metatarsal, lateral malleolus, lateral femoral condyle, lateral humeral condyle, greater trochanter, posterior superior iliac spine, and immediately inferior to the tip of the acromial process laterally. A marker was positioned on the crate to be aligned with the styloid process of the ulna when the subject grasped the crate's handles. For the purpose of these experiments, the subject's hand was considered to be rigidly connected with the forearm. Relevant joint angles were defined between coordinates of three adjacent marker locations (eg, knee: lateral malleolus-lateral femoral condyle-greater trochanter; Fig. 1).

Markers mounted on balsa wood fins were placed at the midline of the spinous process of the 1st sacral (S-1), 3rd lumbar (L-3), 12th thoracic (T-12), and 7th cervical (C-7) vertebrae. The first three markers were used to define the lumbar spine angle. The C-7, acromion, and lateral humeral condyle markers defined the angle of shoulder flexion-extension.

A Peak Performance Technologies Inc (PPTI) calibration frame[‡] (approximately 3 m³) was filmed prior to each experiment and later used to calibrate the measurement volume. Each video field (60 frames per second) of both videotapes was digitized off-line using the PPTI motion analysis system. This system has been shown to generate reliable and accurate measurements when tested in a similar measurement volume in our laboratory.⁴⁵

Electromyographic records were taken from the left ES and from the left VL using the Therapeutics Unlimited Inc Model-544 EMG system.[§] Preamplifier electrode assemblies were used to detect the EMG signal at the skin

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†Velcro USA Inc, 406 Brown Ave, Manchester, NH 03108.

‡Peak Performance Technologies Inc, 7388 Revere Pkwy, Suite 601, Englewood, CO 80112-9765.

§Therapeutics Unlimited Inc, 2835 Friendship St, Iowa City, IA 52245.

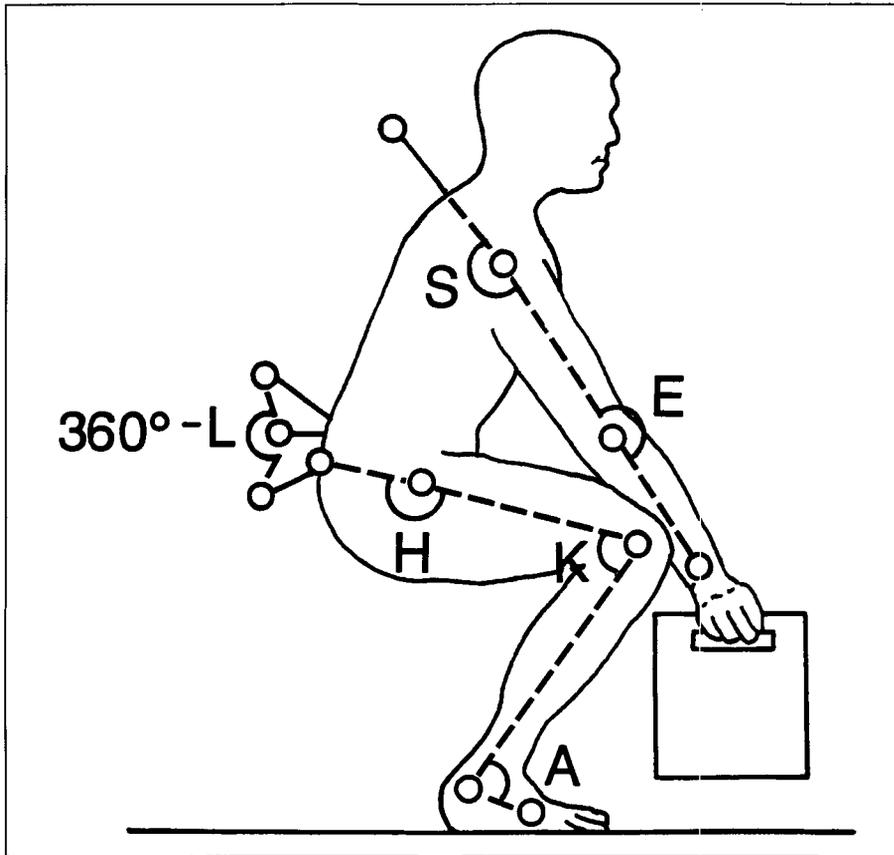


Figure 1. Schematic of body, marker locations, and angle definitions during squat-lifting task (A=ankle, K=knee, H=hip, L=lumbar spine, S=shoulder, E=elbow).

surface. The ES electrode was placed over the belly of the muscle, approximately 5 cm lateral to the spinous process of the third lumbar vertebra. The VL electrode was placed midway between the greater trochanter and the superior ridge of the patella, approximately 5 cm lateral to the midline of the thigh. Both electrode assemblies were applied in a vertical orientation after cleaning the skin with alcohol and lightly abrading it to reduce skin resistance. The EMG signals were amplified (10K gain factor), high-pass filtered (75-Hz cutoff), and digitized on-line at a rate of 360 Hz. The EMG sampling rate was chosen to be a multiple of the 60-Hz video sampling rate to facilitate alignment of the kinematic data and EMG signals. This rate, although somewhat low compared with that used in many studies, was considered adequate for the purpose

of identifying events (ie, onsets and peaks) related to the EMG envelope and is consistent with sampling rates used in many published studies that investigated similar EMG events.⁴⁶⁻⁵²

On a different day, each subject's MLC was determined. A maximum-effort lifting motion against a load cell¹¹ was performed from a squatting position, with the subject's knees flexed and the back relatively straight. The load cell was mounted on a platform on which the subject stood. The subject pulled upward on a "T" handle attached by a chain to the load cell. The average peak force (F_{avg}) produced for three trials was used to estimate MLC (Tab. 1) according to the equation:

$$(1) \text{ MLC(lb)} = 5.522 + 0.372F_{avg}(\text{lb})$$

established in an earlier (unpublished) study.

After applying the reflective markers and EMG electrodes, each subject was shown a short videotape illustrating the lifting technique to be used in the experiment. Subjects were instructed to lift the weighted crate to the height of their waist from a starting posture with the knees bent, back relatively straight (squat lift), and feet positioned symmetrically. Subjects were carefully monitored to ensure that they began each trial in the generally defined position. The exact amount of initial knee flexion and back extension at lift onset, however, was not controlled because of a desire to maintain some degree of external validity.

Several test trials with 30% MLC in the crate were then performed to ensure that the subjects were following instructions. After resting, an additional lift was performed with 60% MLC to determine the most comfortable distance of each subject's feet from the crate. In all trials, the near edge of the crate was positioned immediately in front of or slightly between the subject's big toes. Once determined, the positions of the crate and the subject's feet were marked on the floor with chalk and maintained throughout all trials.

Experimental Procedure

Prior to each experimental trial, subjects were reminded of their task by saying: "Bend your knees, reach for and grasp the handles of the crate. When ready, lift the crate four consecutive times with your legs, not with your back, and do so at your preferred speed." They were told to make certain that the crate was fully down between lifts.

A trial consisted of a subject performing four consecutive task cycles (ie, lifting and lowering) with one of the prescribed loads (ie, 15%, 30%, 45%, 60%, and 75% MLC). Only the last three task cycles of each trial were analyzed. One trial at each load (N=5) constituted a "block" of trials. Approximately 3 to 5 minutes of rest was

¹¹Back, Leg, and Chest Dynamometer, Therapeutic Equipment Corp, Clifton, NJ 07012.

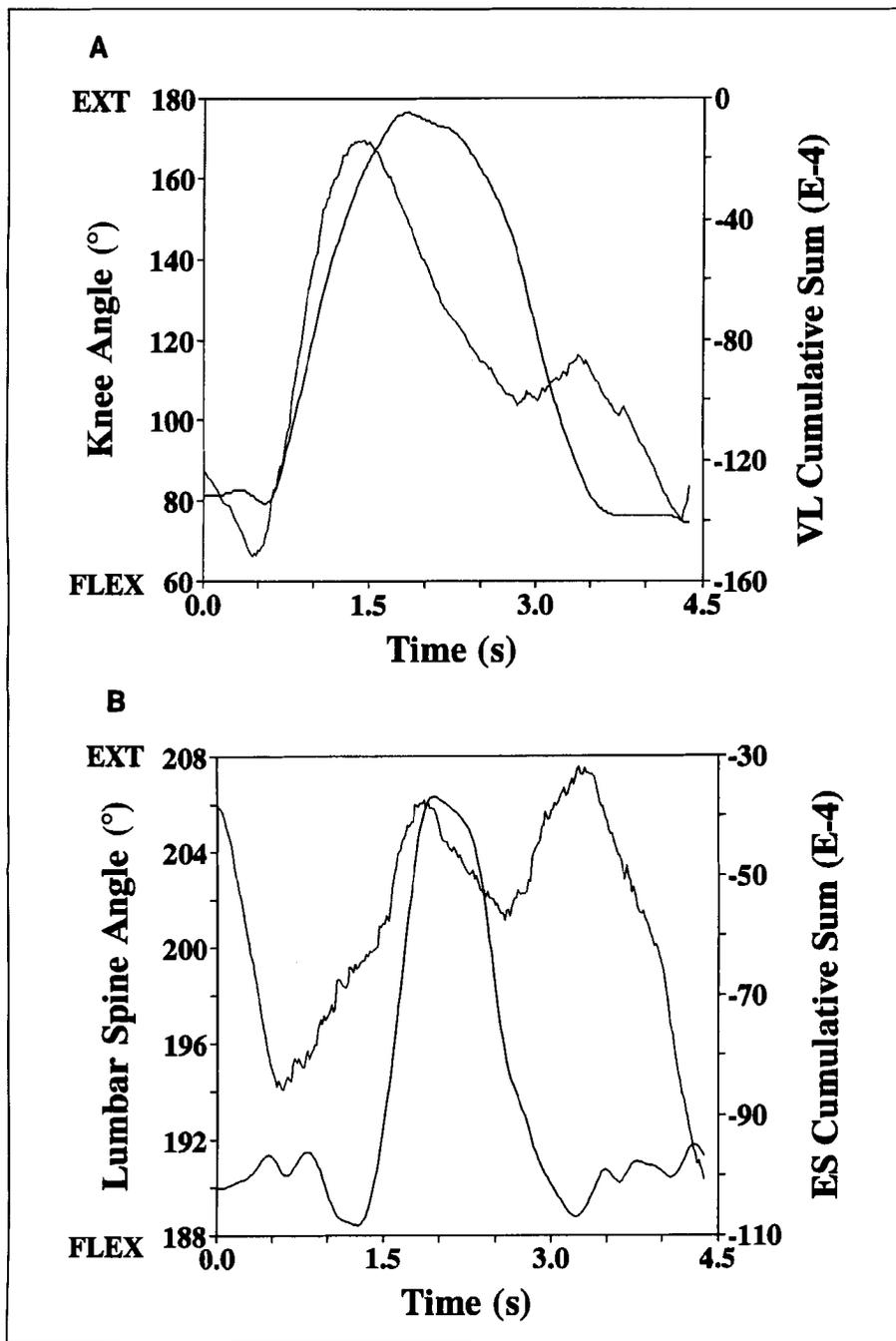


Figure 2. Cumulative sum histogram of (A) vastus lateralis muscle (VL) and (B) erector spinae muscle (ES) electromyographic signals (light lines) and the corresponding knee (A) and lumbar spine (B) angles (dark lines) for the one task cycle. (EXT=extension, FLEX=flexion.)

provided between trials within a block. Four blocks of trials were performed by each subject, each followed by a 15-minute rest period. The loads were presented in ascending order during the first two blocks and in descending order during the remaining two blocks. Additional rest was allowed whenever requested. Only

one subject requested additional rest (prior to the last block of trials).

Data Reduction

After calibrating the measurement volume, each trial from the two videotapes was digitized using the PPTI semiautomatic digitizing module,

yielding 12 lifts for each load condition (3 lifts per load \times four blocks of trials). The resulting two-dimensional raw data were filtered (3 Hz) using a fourth-order, zero-lag Butterworth digital filter to minimize errors introduced during digitization.⁵³ This cutoff frequency was chosen after spectral analysis of the digitized data revealed that the lifting movements occurred at frequencies of less than 1 Hz.

A direct linear transform was used to obtain the three-dimensional coordinates of each marker.⁵⁴ A link-segment model was used to calculate sagittal-plane (ie, flexion-extension) angles for the ankle, knee, hip, lumbar spine, shoulder, and elbow joints from the digitized, transformed coordinates of the reflective markers. We obtained three-dimensional data to allow the possibility for examining movement in the coronal plane. The analyses presented in this article are confined to those of the two-dimensional data.

Electromyographic signals were rectified in software. Cumulative sum histograms of each EMG signal then were obtained using the procedure of Ellaway⁵⁵:

$$(2) \quad S_i = \sum(x_i - X)$$

where S_i is the cumulative sum up to and including the current sample i , X is the mean rectified EMG signal over the entire trial, and x_i is the rectified EMG magnitude at each sample (Fig. 2). The summation is taken over all samples up to and including the current sample. This method was originally used by Ellaway to analyze peristimulus time histograms of neural events. We used this procedure to make determination of EMG onsets and peaks more reliable than determining these events from the rectified EMG signal (see "Results" section).

Independent Variables

The independent variables were the percentage of MLC lifted, the order of load presentation (ie, ascending or descending), and the phase of the task (ie, lifting or lowering). Note that the term "phase" as used here has a differ-

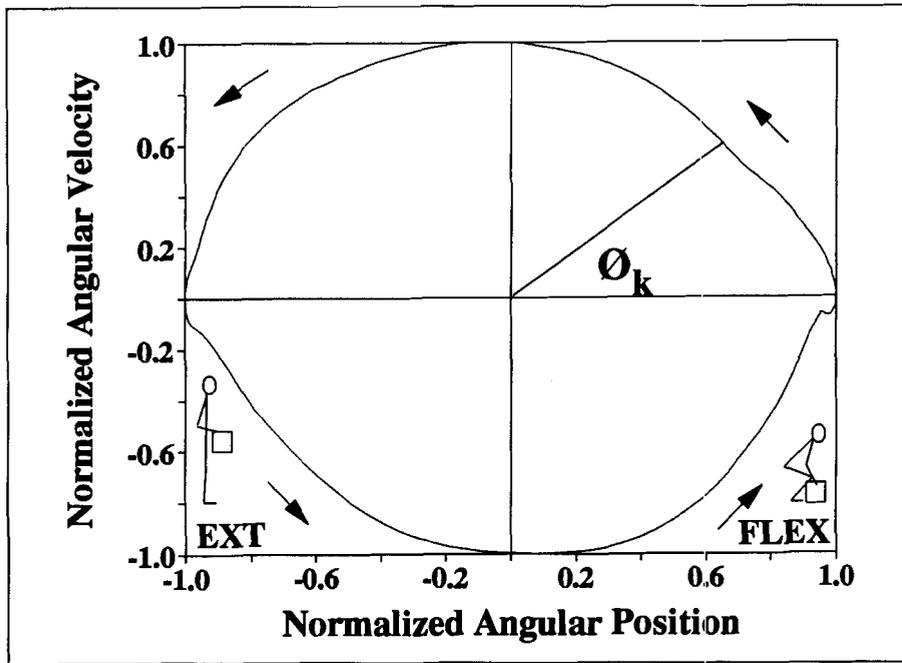


Figure 3. Normalized phase plot of knee motion (angular position versus angular velocity) for one complete task cycle (ie, lifting and lowering), and the phase angle (ϕ_k) for one sample in the knee trajectory. (EXT=extension, FLEX=flexion.)

ent meaning than that of the dependent variable. The lifting phase began when the crate left the floor and ended when the crate reached peak vertical displacement. The beginning of continuous downward motion of the crate defined the beginning of the lowering phase, which ended when the crate reached the ground.

Dependent Measures

Initial posture. To test the degree to which subjects assumed an initial flexed-knee, straight-back posture, we measured the initial knee and lumbar spine angles at the onset of the lift.

Task period. The period (in milliseconds) of the lifting phase and the lowering phase of the task was obtained for each task cycle.

Relative phase (ϕ). To assess the coordination of this task, relative-phase variables were calculated between each of the following joints: (1) knee and lumbar spine (ϕ_{KL}), (2) knee and shoulder (ϕ_{KS}), and (3) knee and hip (ϕ_{KH}).¹⁷

Relative phase is defined as the difference between the phase of movement (ϕ_i) of any two joint or coordinate pairs (eg, $\phi_{KL} = \phi_K - \phi_L$) at each data sample.⁵⁶ The phase of a particular joint's motion is

$$(3) \quad \phi_i = \tan^{-1}(V_{\text{norm}}/P_{\text{norm}})$$

where i is the joint of interest (ie, hip, knee, lumbar spine, shoulder), V_{norm} is the angular velocity normalized on each half-cycle of movement, and P_{norm} is the normalized joint angle.

The normalized joint angle and the angular velocity normalized on each half-cycle of movement are calculated as

$$(4) \quad P_{\text{norm}} = [2P/(P_{\text{max}} - P_{\text{min}})] - [(P_{\text{max}} + P_{\text{min}})/(P_{\text{max}} - P_{\text{min}})]$$

and

$$(5) \quad V_{\text{norm}} = V/|V_{\text{max}}|$$

where P is the actual joint angle at the time or sample of interest, P_{max} and P_{min} are the maximum and minimum

angles over an up-down movement, and $|V_{\text{max}}|$ is the absolute maximum angular velocity. This procedure has the effect of normalizing the angular position and velocity signals to the interval $[-1,1]$, which is necessary to apply the trigonometric function.

Note that although relative phase is expressed in units of degrees, its meaning is different from that of angular range of motion. This difference can be illustrated by considering the angular position and velocity of the knee at a given instance in time. These two variables can be considered to represent the "state" of knee joint motion at that instant. The values can be plotted on an X,Y (angular position versus angular velocity) graph. Similarly, position-velocity pairs at all other collected samples for a given trial can be plotted on the same graph. Connecting all such position-velocity pairs in sequence produces a quasi-circular plot if the joint motion is quasi-sinusoidal. An example of such a plot of normalized angular position-angular velocity pairs for one lifting-lowering cycle of knee movement is presented in Figure 3.

Each point on such a plot can be represented by either the X,Y-coordinate pair or by the polar (phase) angle formed between the positive X axis and a line connecting the origin (0,0) with the X,Y point. This angle is ϕ_i in equation 3. A full task cycle (ie, lifting and lowering) from the start (in this task, knee flexion) to the final (knee flexion) joint position encompasses a phase angle of 360 degrees. The phase angle ϕ_i indicates where in its cycle of motion the joint or segment is at sample i , a given instance in time. If two joints, "j" and "k," are at the same point in their movement cycle at sample i (ie, $\phi_{j,i} = \phi_{k,i}$), for example, if both joints are reaching full extension simultaneously, then their relative phase of motion (ϕ_{jk}) is 0 degrees. If joint "j" reaches full extension at the same time that joint "k" reaches full flexion, then $\phi_{j,i}$ is 180 degrees and $\phi_{k,i}$ is 0 degrees, so that their phase difference ($\phi_{jk} = \phi_{j,i} - \phi_{k,i}$) is 180 degrees. Thus, the relative phase variable provides a

Table 2. Period (in Seconds) (\pm Standard Deviation) of Lifting and Lowering Phases of Task as a Function of Load Magnitude

Phase of Task	Percentage of MLC ^a				
	15%	30%	45%	60%	75%
Lifting	1.224 \pm 0.082	1.225 \pm 0.069	1.225 \pm 0.077	1.282 \pm 0.080	1.335 \pm 0.081
Lowering	1.489 \pm 0.102	1.525 \pm 0.115	1.573 \pm 0.112	1.699 \pm 0.120	1.871 \pm 0.170

^aMLC=maximum lifting capacity.

quantitative estimate of the coordination between two joints or segments at each data sample.

Relative phase variability. Across all data samples, the standard deviation of each ϕ_i measurement was calculated separately for the lifting and lowering phases of each task cycle as measures of the stability of coordination. The mean of this temporal standard deviation was then obtained for trials of the same load condition.

Electromyographic relative timing.

The relative time of occurrence of ES and VL EMG activity onsets and of ES and VL EMG activity peaks within a lift cycle was determined from the cumulative sum histograms as, for example,

$$(6) \quad RT_{VL} = (T_{VLon} - T_0) / \tau$$

where RT_{VL} is the relative time of onset of VL EMG activity in a lifting cycle, T_{VLon} is the time of onset of VL EMG activity, T_0 is the time of initiation of vertical crate motion, and τ is the period of the lifting phase of the task (ie, the time from initiation of vertical crate motion to completion of the lifting phase). All other EMG timing measurements were determined similarly.

Statistical Analyses

The effects of the independent variables and their interaction on the dependent measures were tested by repeated-measures analyses of vari-

ance (ANOVAs) using the SYSTAT statistical package.* Tests for simple main effects and planned comparisons were performed where appropriate. Because of the large mass of data, the dependent measures for each subject were averaged across lifts of each condition before performing the ANOVAs. Because of a technical problem, EMG data were unavailable for 1 subject. Therefore, many statistical analyses were performed using data from 14 subjects.

Results

Initial Posture

The initial knee angle just prior to the crate leaving the floor averaged (N=15) between 77.0 \pm 4.0 degrees (15% MLC) and 81.0 \pm 4.1 degrees (75% MLC) (180 $^\circ$ =full extension) across all loads. Knee angle was not affected by either the load or its order of presentation. Lumbar spine angle at lift onset decreased on average from 190.5 \pm 1.0 degrees at 15% MLC to 187.7 \pm 1.0 degrees at 75% MLC ($F_{4,56}$ =15.3, P <.001). Given our marker locations, a neutral lumbar spine angle was approximately 195 degrees. Thus, the spine was flexed from neutral by 4 to 7 degrees on average across all load conditions at the start of the lift. Total lumbar spine excursion during the lift consistently averaged about 20 degrees across all subjects and loads.

Task Period

The period of each task phase changed with increasing load magnitude, and more so for lowering than for lifting ($F_{4,56}$ =15.3, P <.001; Tab. 2). The period of lowering was always longer than that of lifting by about 300 milliseconds with lighter loads and 500 milliseconds with heavier loads. Planned comparisons revealed that the lifting period increased only after 45% MLC (P <.05), whereas the period of lowering the load increased for all loads greater than 30% MLC (P <.05). Individual differences were also apparent. For example, six subjects had no substantial increase in the period of lifting across all load changes. Only one subject exhibited an increase in task period across all loads, and only for the lowering phase of the task.

The effect of load on task period also depended on the order (ie, ascending or descending) of load presentation ($F_{4,56}$ =4.24, P <.01). For loads less than 45% MLC, subjects moved more slowly during both lifting and lowering when the loads were presented in ascending order compared with descending order. For loads greater than 45% MLC, the opposite was true. The differences, however, were small in magnitude (ie, 18–24 milliseconds).

Relative Phase

The order of load presentation had no significant effect on any of the relative phase measures. Increasing the load affected the mean relative phase measures differently during the lifting and lowering phases of the task. Reported F values, therefore, are for the interaction between load (%MLC) and phase of lifting.

Knee-lumbar spine. Figure 4 illustrates qualitatively for subject DS's data how knee-lumbar spine coordination changed with increasing load during the lifting phase of the task. Each curve in the figure was obtained by (1) normalizing the angular position of both joints to the interval [-1,1], as is done when calculating ϕ_{KL} (equation 4); (2) normalizing the period of the lifting phase of each trial to 100%⁵³;

*SYSTAT Inc, 1800 Sherman Ave, Evanston, IL 60201.

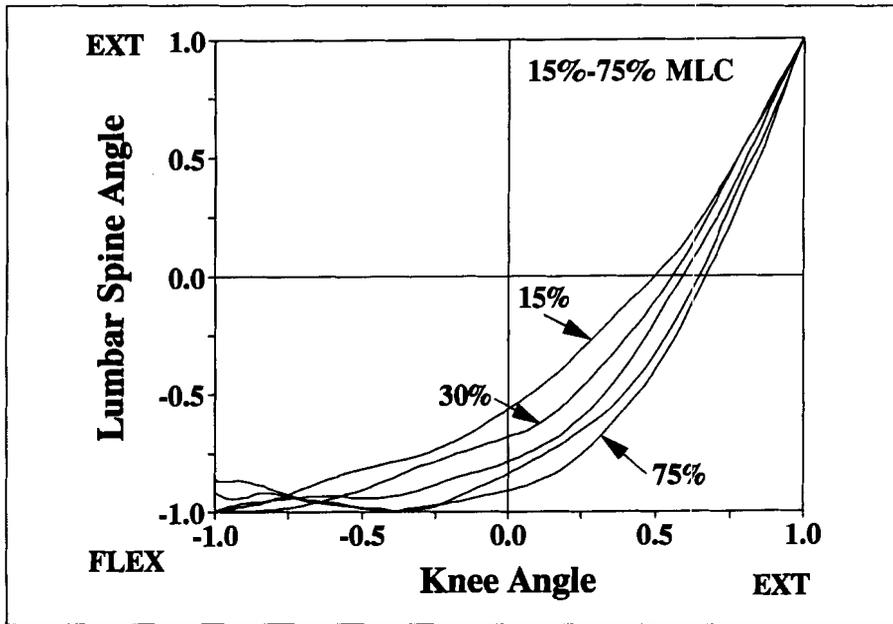


Figure 4. Normalized knee-lumbar spine relative motion plots of subject DS during the lifting phase for each load condition (ie, 15%-75% MLC). (EXT=extension, FLEX=flexion, MLC=maximum lifting capacity.)

(3) averaging, for each load, all 12 normalized movement trajectories for each joint; and (4) plotting the normalized knee trajectory against the normalized lumbar spine trajectory.

Lumbar spine extension clearly lagged further behind knee extension during lifting as the load was made heavier, whereas the load had no effect on ϕ_{KL} during the lowering phase ($F_{4,56}=22.8$, $P<.001$). Planned comparisons of ϕ_{KL}

for adjacent load conditions revealed significant differences between all adjacent loads when lifting (all $P<.01$), but no such differences when lowering the load (Fig. 5). Although knee extension exhibited a phase lead on lumbar spine extension during lifting, lumbar spine flexion led knee flexion during lowering.

Knee-shoulder. Technical difficulties made it possible to calculate ϕ_{KS} on only 10 subjects. Shoulder motion always lagged behind knee motion during lifting and led knee motion during lowering of the load ($F_{4,36}=3.7$, $P<.05$). The lag during lifting increased by about 10 degrees between 15% and 75% MLC (Fig. 6). The increase in phase lag of shoulder extension behind knee extension, however, was significant only between 30% and 45% MLC and between 60% and 75% MLC ($P<.05$). There were no significant differences in ϕ_{KS} between adjacent load conditions for the lowering phase.

Knee-hip. Hip extension lagged slightly behind knee extension when lifting, and hip flexion led knee flexion during lowering of the load ($F_{4,52}=15.5$, $P<.001$; Fig. 7). The lag of hip extension during lifting increased significantly between adjacent loads as the load increased between 30% and 75% MLC ($P<.05$). The increase across all loads, however, was small, amounting to only 6 degrees, on average. During lowering, ϕ_{KH} was unaffected by the load.

Relative Phase Variability

The within-trial SD of ϕ_{KL} ($F_{4,56}=10.97$, $P<.001$), ϕ_{KS} ($F_{4,36}=7.98$, $P<.001$), and ϕ_{KH} ($F_{4,52}=28.63$, $P<.001$) increased by 5 to 10 degrees with increasing loads. Except for ϕ_{KH} , the increase was similar for both phases of the task and was significant only between 45% and 60% MLC and between 60% and 75% MLC ($P<.05$). The greatest effect was found for ϕ_{KL} . Load and phase interacted to influence ϕ_{KH} ($F_{4,52}=15.32$, $P<.01$). Variability increased significantly between all adjacent load conditions ($P<.05$), and was slightly

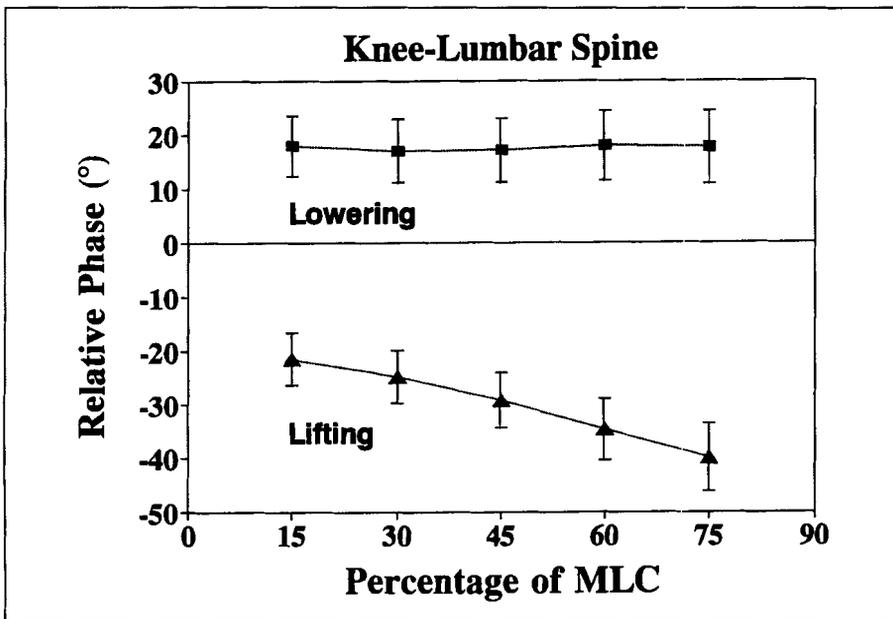


Figure 5. Average (all subjects) mean relative phase between knee and lumbar spine motion for each load for lifting and lowering phases of the task (\pm standard error of the mean). (MLC=maximum lifting capacity.)

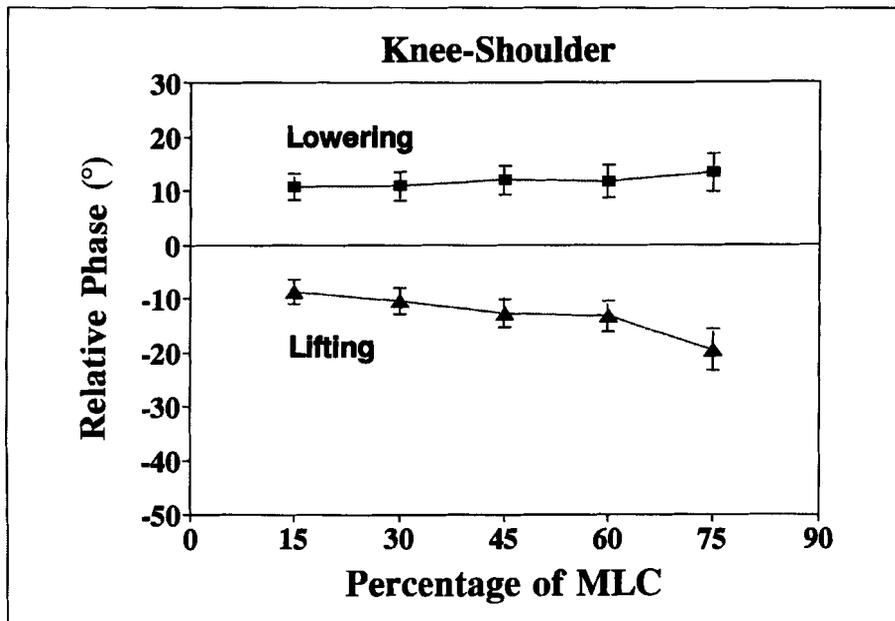


Figure 6. Average (all subjects) mean relative phase between knee and shoulder motion for each load for lifting and lowering phases of the task (\pm standard error of the mean). (MLC= maximum lifting capacity.)

higher during the lowering phase of the task.

Electromyographic Relative Timing

We calculated intraclass correlation coefficients (ICCs) to evaluate test-retest reliability of the estimates of EMG relative timing obtained from the cumulative sum histograms.⁵⁵ Measurements were repeated 6 months apart by the same individual (JPM). The ICCs (2,1) for the relative time of occurrence within the lift cycle of VL onset and peak of EMG activity and ES onset and peak of EMG activity were .9903, .9606, .8875, and .9991, respectively.

The VL minima that were clearly associated with the onset of knee extension were not apparent in most instances. The minimum in VL activity prior to lift onset for the majority of subjects actually occurred just prior to lowering the load on the previous task cycle. That is, after a pause in activity following the lifting phase, VL activity increased as the load was lowered and reached a peak during the interlift interval. The VL activity then de-

creased gradually to a new minimum at the completion of the succeeding lifting phase (Fig. 8A).

When considering these minima as the onset of VL activity and comparing them with minima for the onset of ES activity, a significant increase in the lag between VL and ES onsets of EMG activity was found to occur with increasing load ($F_{4,52}=13.2$, $P<.001$), that is, onset of VL EMG activity occurred earlier and onset of ES EMG activity occurred later as the load was made heavier.⁵⁷ The relative time of this VL minimum, however, is contaminated by the length of the interlift interval (ie, the pause between successive task cycles), which tended to be slightly longer with heavier loads.

In some instances, a minimum of VL activity did occur that was associated more closely with vertical crate motion and knee extension (Fig. 8B). Only six subjects, however, exhibited a sufficient number of such lifts to allow statistical analysis. The result of that analysis indicates that the relative time of this onset of VL EMG activity, expressed as a percentage of the lifting

cycle, did not change when lifting heavier loads ($P<.50$).

Distinct peaks of VL EMG activity and both onsets and peaks of ES EMG activity, however, could be readily identified for the majority of lifts for all subjects. The relative time of occurrence of peaks of VL activity did not change statistically across increases in load. In contrast, the relative timing of both the onsets of ES activity ($F_{4,52}=7.24$, $P<.001$) and the peaks of ES activity ($F_{4,52}=26.9$, $P<.001$) changed linearly with increasing load (Figs. 9 and 10). Although the onset of ES activity occurred prior to vertical crate motion when lifting light loads (hence, the negative lags in Fig. 9), it occurred nearly simultaneously with liftoff when lifting heavier loads. This effect, however, was rather small. Peak ES activity was affected more strongly by increases in load. It occurred later after the onset of lifting when the load was heavy.

Distinct onsets and peaks of ES activity could be identified during the lowering phase of the movement cycle in only 12 subjects. Neither the relative time of occurrence of onsets or peaks of ES activity were found to be significantly affected by increasing the load. Onsets and peaks of VL activity could not be identified consistently in any subject during this movement phase; VL activity remained relatively constant during lowering.

Discussion

The results of this investigation confirm and extend those of previous studies on the coordination of squat lifting.^{17-19,27,28} These results, obtained from experienced workers for whom lifting was part of their job, are generally consistent with the results of similar experiments on college students reported by Scholz.^{17,18} In particular, both lumbar spine and shoulder joint extension were found to lag further behind knee extension during the lifting phase of the task when lifting heavier loads (Figs. 4-6). A consistent effect of load on ϕ_{KL} during lifting occurred in 14 of the 15 subjects. In contrast, while lowering the load,

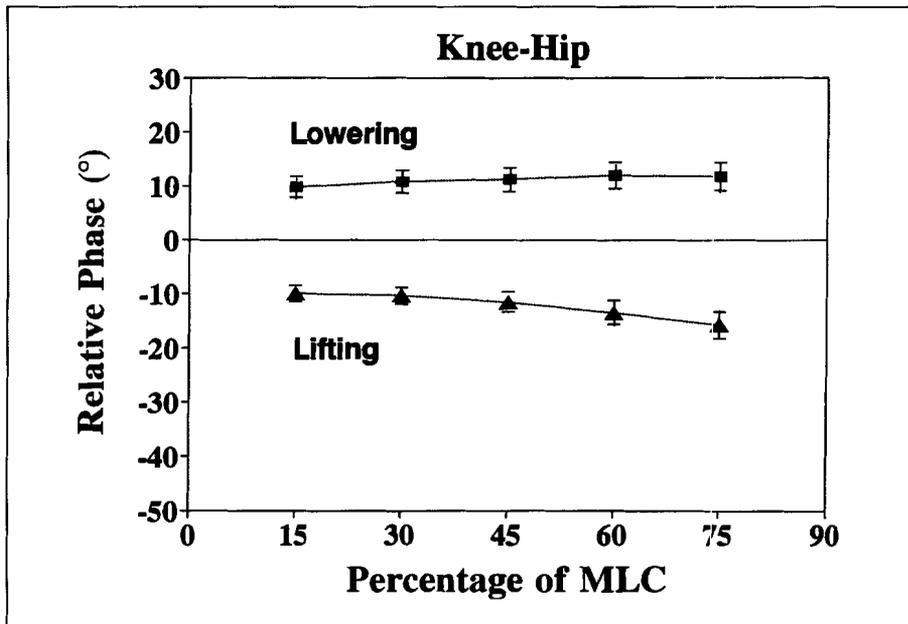


Figure 7. Average (all subjects) mean relative phase between knee and hip motion for each load for lifting and lowering phases of the task (\pm standard error of the mean). (MLC=maximum lifting capacity.)

lumbar spine and shoulder flexion always led knee flexion in time, and ϕ_{KL} and ϕ_{KS} were unaffected by increases in load.

The measured changes in relative phase were relatively small in magnitude, being about 20 degrees on average for ϕ_{KL} (Fig. 5). This value amounts to about 10% of the lifting cycle. The extent of the effect of increasing the load magnitude on knee-lumbar spine coordination can be seen qualitatively, however, in the representative mean relative motion plots depicted in Figure 4.

Whether this statistically significant and apparently linear change in coordination has clinical significance is an open question. We have developed a dynamic, biomechanical model to estimate joint torques and forces with the hope of gaining additional insights. Analysis of those data is currently under way. Ultimately, it may be necessary to obtain similar information on individuals with a history of lifting-related back injury to help resolve this question. Such studies will be difficult, however, because of a greater risk of injury.

In contrast to the results for ϕ_{KL} and ϕ_{KS} , the effect of load on knee-hip coordination differed from those reported previously.¹⁷ Scholz¹⁷ found that coordination of the knee and hip, revealed by discrete relative timing measures, did not change significantly across similar changes in load. Subjects in the current study showed a small but significant increase in the phase lag of hip motion behind knee motion after 45% MLC (Fig. 7), revealed by ϕ_{KH} . This finding supports the assumption that continuous relative phase estimates, which take into account both position and velocity, provide more sensitive indicators of joint coordination than do discrete relative timing measures.

Thus, when lifting the heaviest loads from an initial bent-knee, relatively straight-back posture, movement tended to occur in a clear distal-to-proximal sequence, similar to the sequence that occurs during raising the trunk after bending forward at the waist.⁵⁸

All measured relative phase variables were shown to become significantly more variable with increasing load.

The increases in within-trial variability, however, were relatively small. Nonetheless, the results suggest that it becomes increasingly difficult to maintain a stable coordination pattern when lifting heavy loads. The finding that the standard deviations increased most with loads above 45% MLC may indicate an upper limit on load for safe lifting using a squat lift.

That subjects began each lift from the prescribed bent-knee, relatively straight-back (or neutral spine) posture characteristic of squat lifting was evidenced by the analysis of starting knee and lumbar spine angles. The knee angle was always less than 90 degrees of flexion at the start of the lift and did not change significantly with increasing load. Although significantly more flexion was found for the lumbar spine angle at the beginning of lifts with heavier loads, the difference in this initial angle from neutral and across loads was quite small.

We did not attempt to experimentally control for task period because of concern about increasing the risk of injury when doing so while subjects lifted heavy loads. Significant changes in the period of lifting occurred only after 45% MLC. However, with the exception of ϕ_{KL} , the means and standard deviations of relative phase measures also changed significantly only beyond 45% MLC. Thus, the possibility must be considered that the measured changes in joint coordination resulted from changes in the task period alone, or an interaction of task period with load. Certainly, the coordination of numerous other tasks has been shown to be affected by changes in the period or frequency of movement.²⁹⁻³⁵

Nevertheless, several factors argue against changes in task period, or lifting speed (because amplitude was fixed), producing the observed coordination changes. First, previous studies²⁹⁻³⁵ have shown that deviations from in-phase coordination (ie, $\phi_{jk}=0^\circ$) occur as the movement period is decreased. Greater deviation from in-phase coordination when lifting heavier loads in our experi-

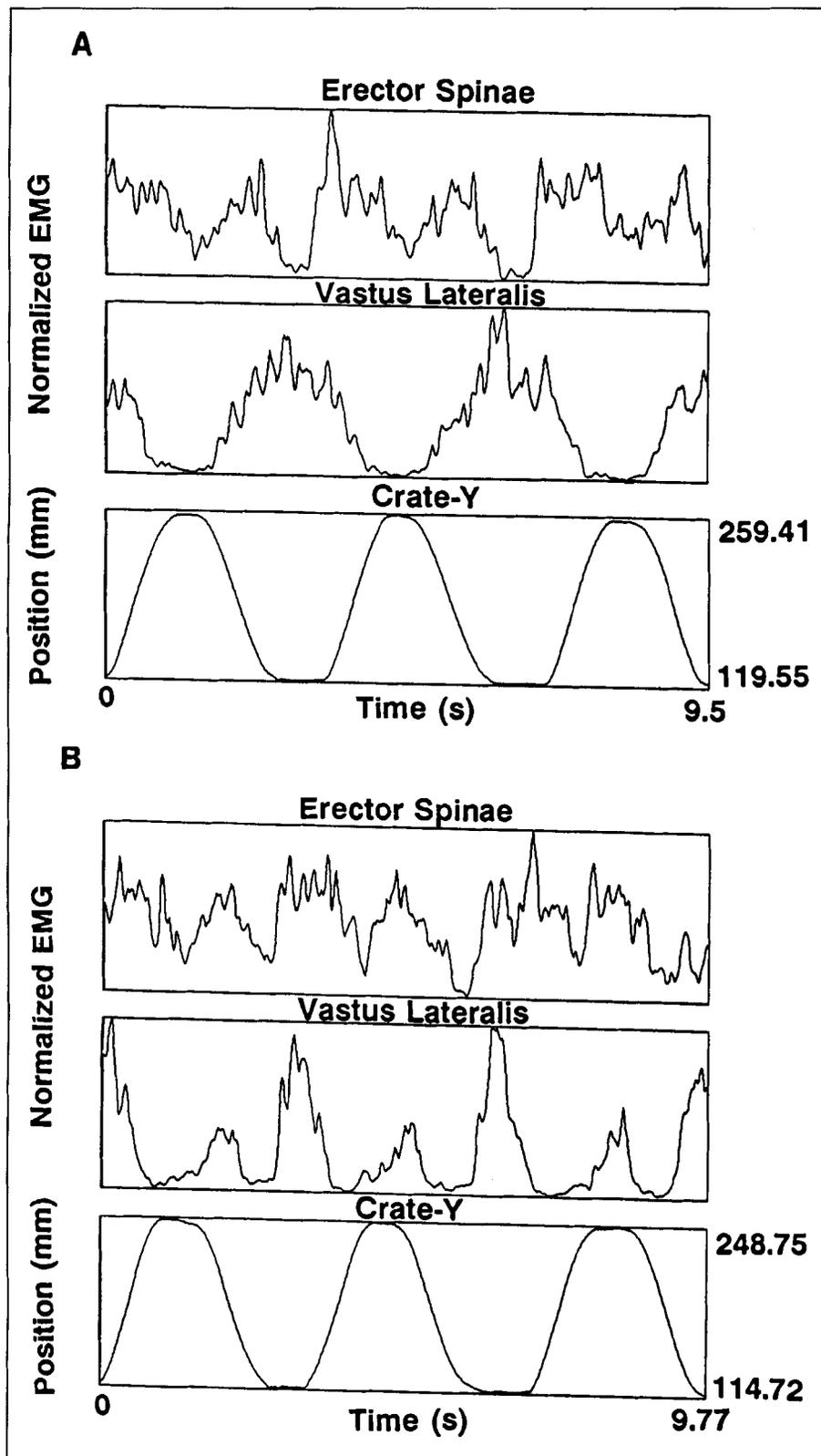


Figure 8. Electromyographic (EMG) activity of erector spinae and vastus lateralis muscles, normalized to maximum contraction, and vertical crate motion during one representative trial for which vastus lateralis muscle onset of activity was (A) poorly and (B) more clearly associated with onset of knee extension.

ments was associated instead with an increase in lifting period. Thus, it is possible that the increase in lift period that occurred when lifting heavier loads actually suppressed the magnitude of coordination changes that would have occurred otherwise, if both task period and load magnitude are control parameters for this behavior.²⁹ Second, the actual change in the period of lifting was quite small, amounting to less than 60 milliseconds on average between 45% and 60% MLC and between 60% and 75% MLC (Tab. 2). Third, the relative phase variable reflecting the greatest change in coordination was ϕ_{KL} . This variable changed significantly between all adjacent load increments, not only after 45% MLC when the lift period changed. Finally, the largest increase in task phase period with increasing load occurred during the lowering phase of the task. No significant changes in the relative phase of joint motions were found to occur across loads during this phase of the task.

Might the observed changes in coordination have resulted from purely mechanical factors alone? As the inertia of the load increased, the back extensor muscles may have been less capable than the knee extensors of overcoming the added inertia and accelerating the load. If so, this would lead to a phase lag between knee and back movement without a corresponding change in the lag between muscle onsets (ie, changes in neural activation). This possibility seems unlikely because slight increases in the lag between knee-lumbar spine motion occurred even with load changes spanning light loads (Fig. 5). Moreover, the effect of each load on the coordination of these joints differed between lifting and lowering phases of the task. Therefore, although mechanical factors certainly contribute, it seems unlikely that the change in ϕ_{KL} observed across relatively light loads was due to inertial effects alone.

We examined the EMG activity of back and knee extensor muscles to determine whether and how the changes in coordination observed kinematically were related to neuro-

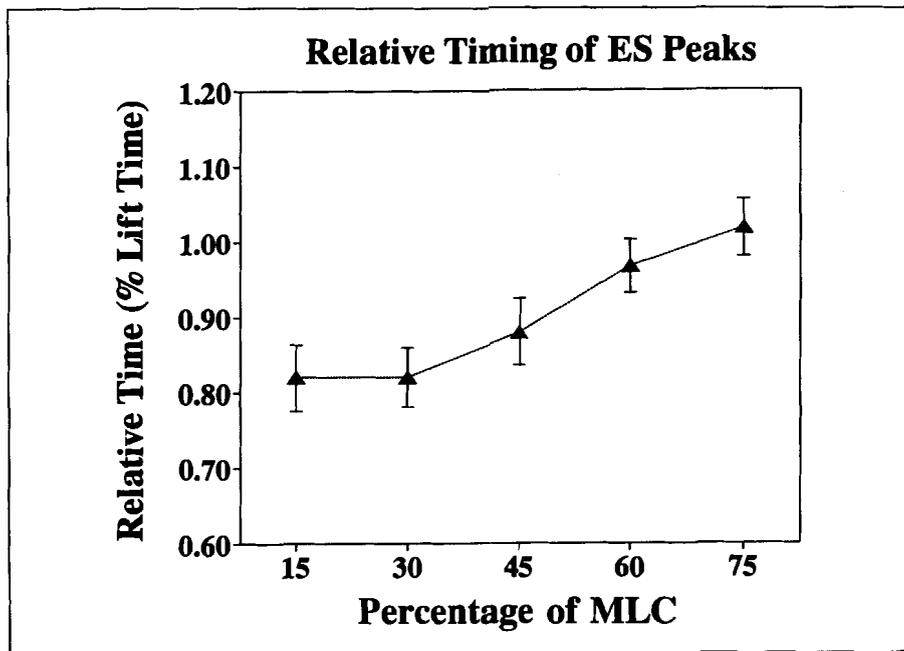


Figure 9. Relative time of occurrence of erector spinae muscle (ES) onset of activity during the lifting phase of the task as a function of load magnitude (percentage of maximum lifting capacity [MLC]). Averages across 14 subjects (\pm standard error of the mean).

muscular changes. Analysis of the ES and VL relative timing measurements

revealed mixed evidence for changes in neuromuscular timing underlying

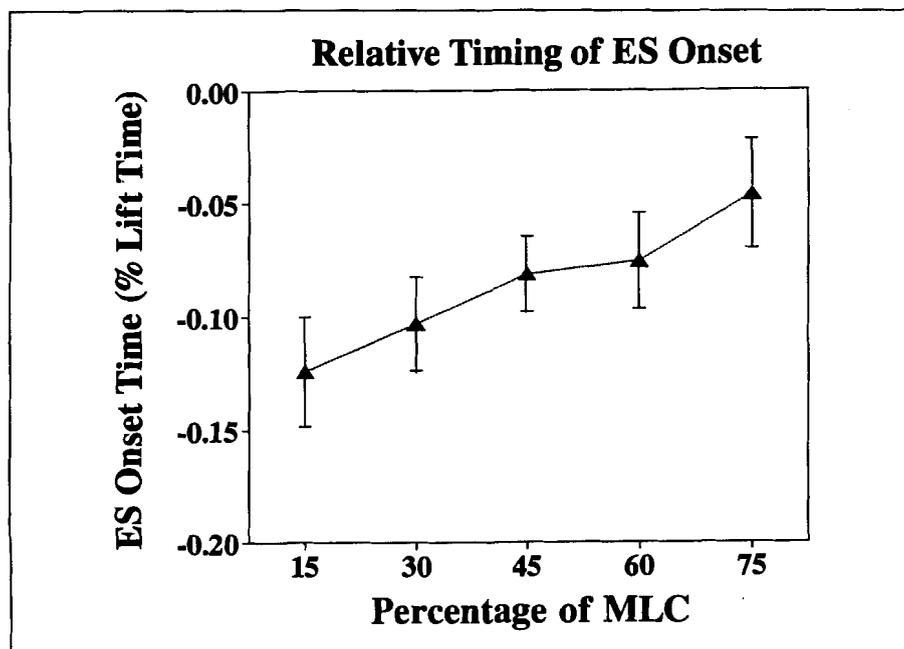


Figure 10. Relative time of occurrence of erector spinae muscle (ES) peak activity during the lifting phase of the task as a function of load magnitude (percentage of maximum lifting capacity [MLC]). Averages across 14 subjects (\pm standard error of the mean).

the kinematic results. A small but significant delay in the relative time of onset of ES EMG activity occurred as the load was increased in magnitude (Fig. 9). Increasing delays in ES peak activity with increased load were of greater magnitude (Fig. 10). Although we have no direct evidence for the origin of changes in EMG timing with increasing load, the change in the timing of the onset of ES activity is unlikely to be of reflex origin because the onset of ES activity occurred prior to the crate leaving the ground. The modification in onset of ES activity with increasing load is more likely to be due to task-specific adjustments in the central timing of muscle activation. Changes in the timing of ES peak activity may have originated from either centrally mediated or reflexive changes because ES peak activity occurred after lift onset.

It has recently been suggested that the timing of ES activity is more related to the posture of the lumbar spine during a lift than to the amount of load the subject lifts.⁵⁹ In that study, which manipulated spinal posture, peaks of ES activity were found to occur later in the lift cycle, when subjects maintained a kyphotic posture compared with a lordotic posture. In our experiments, lumbar spine posture was not specifically prescribed, although subjects were told to "keep their backs relatively straight." We found that the spine was slightly more flexed or kyphotic for all subjects prior to lifting the heaviest loads. However, the deviation from neutral with increasing load was minimal, and the spine extended approximately 20 degrees from the initial posture during the lift. Thus, the increased delay in ES peak activity found to occur with increasing load in our experiments is more likely due to the effect of lifting a heavier load than to the posture of the spine.

When combined with the fact that both EMG timing and mean ϕ_{KL} were unaffected by load changes when lowering the load, the significant change in ES timing found during lifting supports the conclusion that the change in ϕ_{KL} that also occurred during lifting was not a purely mechanical

effect. It is important to note, however, that there were individual differences in the degree to which changes in EMG timing and kinematic relative phase measurements covaried. For example, some subjects who exhibited a significant change in ϕ_{KL} across loads showed nonsignificant changes in ES event timing. Moreover, the EMG timing changes were small in magnitude. The available EMG equipment limited the number of muscles from which we could record, and our primary interest was in knee-lumbar spine coordination. Obviously, there are many other muscles that help to control motion of these segments, including antagonistic muscles and muscles acting across multiple joints.⁶⁰ Individual differences in the covariation of kinematic and EMG measurements of knee-lumbar spine coordination may be the result of differences in the use of two-joint muscles (eg, to simultaneously control knee and hip torques).⁶⁰

In contrast to ES timing, changes in neuromuscular timing of VL activity did not occur with increases in the load. Although VL minima (used to determine EMG onset of activity) could be identified for which the relative time of occurrence changed significantly with increasing load,⁵⁷ these minima were not clearly associated with knee extension at the onset of vertical crate motion (Fig. 8A). The relative time of onset of VL activity was uninfluenced by the load in six subjects for whom VL minima were consistently associated with lift onset (Fig. 8B). This was also the case for the relative time of occurrence of VL peak activity for all subjects. It is possible, however, that the activation of other quadriceps femoris muscles, including the biarticular rectus femoris muscle, is more closely related to the kinematic phase lags. Future experiments, therefore, will need to record from multiple muscles if the neuromuscular control of this task is to be better understood. For example, it may be that the vastus muscles are less important in controlling the knee during this task than is the combined action of biarticular muscles acting at this joint.^{60,61}

It should be noted that despite the general consistency of reported effects across subjects in this study, interesting individual differences were apparent. This finding was especially true for the relative phase of knee-lumbar spine motion and its relationship to measured changes in EMG timing. Moreover, the relative movement of the knee and lumbar spine during load acceleration appeared to fall into two dominant patterns based on the effects of the load. Individual differences are explored in detail in our companion article in this issue.

Conclusions

The observed changes in coordination and the individual differences found in this task occurred despite specific instructions about the lifting technique to be used. Thus, it can be concluded that beginning a lift from a posture generally associated with squat lifting does not ensure that an individual will lift the load in exactly the same manner regardless of the weight lifted, or perhaps in the face of changing other task variables such as movement speed. These facts need to be considered both when planning prevention programs for healthy workers and when instructing patients following back injury. The results of this investigation provide some insights about the nature of changes in coordination that occur as the result of lifting heavier loads. More work is clearly needed, however, to determine (1) the neuromuscular basis of the observed kinematic changes, (2) implications of these changes for the risk of injury, and (3) whether subjects can be taught to constrain the magnitude of such changes if these changes are shown to compromise joint integrity.

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