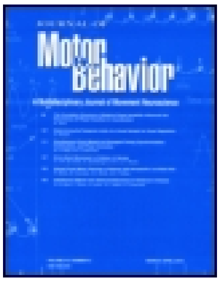

Journal of Motor Behavior
Publication details, including instructions for authors and subscription information:
http://www.tandfonline.com/loi/vjmb20

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Version of record first published: 02 Apr 2010.

To cite this article: Jaap H. van Dieën, Petra van der Burg, Tamara A. J. Raaijmakers & Huub M. Toussaint (1998): Effects of Repetitive Lifting on Kinematics: Inadequate Anticipatory Control or Adaptive Changes?, Journal of Motor Behavior, 30:1, 20-32

To link to this article: http://dx.doi.org/10.1080/00222899809601319

Journal of Motor Behavior, 1998, Vol. 30, No. 1, 20–32

Effects of Repetitive Lifting on Kinematics: Inadequate Anticipatory Control or Adaptive Changes?

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ABSTRACT. In the present study, the effects of repetition on the kinematics in discrete lifting were studied in 10 subjects who lifted a barbell weighing 10% of body mass at a determined speed and along a determined trajectory 630 times during about 40 min. Three-dimensional (3-D) kinematics of the feet, lower and upper legs, pelvis, and trunk were recorded in the first 3 and the final 3 lifting movements of each set of 70 lifts. Over time, trunk extension velocity in the initial 250 ms of the lifting movement decreased, reaching negative (increasing flexion) values in most subjects. In contrast, hip extension velocity increased. Those changes resulted in an increased phase lag between hip and trunk extension. Also, over time, subjects started the lifting movement with their legs more extended and their trunks further flexed. Finally, the motion of the trunk around its longitudinal axis (twisting) increased. The increase in phase lag between hip and trunk extension is interpreted as a consequence of fatigue—more specifically, as the result of a decreased rate of force development of the back muscles. The change in initial posture more likely is an adaptation that functions to retard further fatigue development.

Key words: anticipatory control, biomechanics, motor control, muscle fatigue

Muscle fatigue has been defined as any reversible decrease in the performance capacity of a muscle that results from its activity (Bigland-Ritchie & Woods, 1984). It is often assumed that, because of impaired coordination, muscle fatigue causes an increased risk of musculoskeletal injuries, (Bunkens, 1996; Roy, DeLuca, & Casavant, 1989; Seidel, Beyer, & Brauer, 1987). That hypothesis seems to rely on the view that movement coordination is based on preprogrammed control. It is implicitly assumed that in the unfatigued state, motor control is optimized or at least constrained so that the resulting kinetics and kinematics for the given motor act involve a relatively low risk of injury. If, when muscles are fatigued, the motor control is not adjusted to the changed characteristics of the effector organs, changes in kinetics will result. There will be overt changes of the joint kinematics (angular velocity and joint angle), which are, after all, integral measures of the joint moments produced by the muscles. Those changes in kinematics are expected to be adverse in nature with respect to the load on the musculoskeletal system. So far, in studies of humans, evidence supporting that hypothesis is scarce. In animal experiments, however, some evidence has been obtained. For instance, increases in bone strain have been demonstrated in foxhounds' tibias during intense uphill running on the hind legs; those increases were attributed to less optimal timing or scaling of muscle activity, or to both (Yoshikawa et al., 1994). But in that experiment, a rather specific motor act was studied, one in which the effect of impact loads was predominant.

The proposed insufficient adjustment of motor control during muscular fatigue might explain the epidemiological

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association of repetitive tasks and musculoskeletal disorders in humans. More specifically, repetitive lifting has been shown to be a risk factor for the development of low-back pain (e.g., Frymoyer et al., 1983; Garg & Moore, 1992). Experimental studies have shown that repetitive lifting causes a rapid development of back extensor muscle fatigue (Petrofsky & Lind, 1978; Potvin & Norman, 1993). Nevertheless, we have previously shown that in a task of continuous lifting and lowering, trunk kinematics were not affected (Dieën, Toussaint, Maurice, & Mientjes, 1996). In the line of reasoning outlined above, that finding would suggest that the nervous system adjusts the motor program so as to compensate for changes of the effector organ characteristics. Such flexibility is in apparent contrast with a form of preprogrammed control and seems easier to reconcile with the view that coordination is based on dynamic pattern generation. In the latter approach, coordination is seen as the emergence of stable kinematic patterns from the nonlinear interaction between all parts of the motor system. Such patterns are believed to be relatively robust to changes of the constituent parts. In the dynamic pattern-generation approach, muscles are viewed not as simple effectors operating under nervous command but as an integral part of a mutually influencing network with the nervous system and the environment. Changes in the mechanical capacities of the muscles are not isolated phenomena but coincide with changes of the network's dynamics. Rephrased in neurophysiological terminology, we hypothesized that information based on, for instance, type III and IV afference from the fatigued muscles might function to restructure the motor command so that the kinematic pattern is conserved. One might question, however, whether the kinematic pattern would be equally robust when unforeseen external perturbations are experienced, because the stability of the kinematic pattern under changing conditions seems dependent on a continuous flow of information within the network.

Results from studies on lifting in an unfatigued condition suggest that the instant of picking up the load can be considered a self-imposed perturbation of the global mechanics of a lifting movement (Toussaint, Commissaris, Hoozemans, Ober, & Beek, 1997). That perturbation is partly coped with in an anticipatory fashion and partly compensated for after the load is picked up. Commissaris and Toussaint (1997b) showed, however, that the compensations were not visible in the trunk kinematics. When compared with an unloaded upward movement, trunk kinematics were not affected by picking up the load. That finding implies that the joint moment around the lumbosacral junction, which controls the trunk kinematics, is properly scaled when a load is picked up. Because explicit feedback on load mass is available only after lift-off, that control is probably done in an anticipatory fashion (Commissaris & Toussaint, 1997b; Johansson & Cole, 1992). The latter assumption is supported by the fact that when subjects are unexpectedly confronted with a reduced load mass, the lumbosacral moment is initially not reduced, which causes an overshoot in the initial trunk kinematics (Commissaris & Toussaint, 1997b). Thus, in the unfatigued state, when loads of known mass are lifted, anticipatory control allows a trunk movement pattern that is robust with respect to the perturbation caused by picking up the load.

In our previous study on fatiguing repetitive lifting (Dieën et al., 1996), subjects kept the load in their hands continuously. Therefore, no potential perturbation of the trunk kinematics was present at the initiation of the lifting movement, and, consequently, no anticipation in the sense described above was required. Information pertaining to the changing balance between load mass and trunk muscle performance capacity that results from fatigue development may, in that case, be provided continuously by, for example, increased muscle spindle activity (e.g., Gantchev 1990; Nelson & Hutton, 1985). That increased muscle spindle activity will reduce the need to modify the central motor command. A much more natural task, however, would be to pick up the load at a low level, lift the load, release it at a higher level, bend over unloaded, and then again pick up a load at the low level. In such a discrete lifting task, the balance between trunk muscle condition and load mass has to be adjusted before feedback from the muscles is present. A memorized representation of that balance, derived from the previous lifting movement, could be used to accomplish that adjustment. In support of that possibility, Commissaris and Toussaint (1997a) have shown that information obtained in previous lifts is used in scaling anticipatory actions. The use of a memorized representation, therefore, involves central control. One may question whether such memory-based adjustments to muscular fatigue are as effective as adjustments based on instantaneous peripheral feedback. In general, instantaneously available sensory information seems to allow more accurate motor control than does memory-based information (Oudejans, Michaels, Bakker, & Dolné, 1996). Therefore, fatigue may very well affect trunk kinematics in a discrete lifting task much more than in a continuous lifting–lowering task. If the activation of the trunk extensor muscles is insufficiently adjusted to their fatigued state, trunk extension velocity would be expected to be reduced. That reduction would result in an increased phase lag between hip and lumbosacral extension, as has been shown to occur with increased load mass (Burgess-Limerick, Abernethy, Neal, & Kippers, 1995; Scholz, 1993a, 1993b; Scholz, Milford, & McMillan, 1995). Our primary aim in the present study was to test the hypothesis that, unlike its effects in continuous lifting, fatigue in discrete lifting would lead to an increase of the phase lag between hip and lumbosacral extension.

In our previous investigation (Dieën et al., 1996), we examined two-dimensional (2-D) kinematics to study the effect of fatigue in lifting. Parnianpour et al. (1988) have shown, however, that in repetitive iso-inertial trunk extension an increase of out-of-plane movements (lateral flexing and twisting) occurs. That finding was explained as follows: The excursion of the trunk was thought to be controlled in a feedback manner. If delays in a feedback loop increase, instabilities will occur in the to-be-controlled variable. A decreased rate of voluntary force development of the muscles and a decreased reflex motor time, which are well-documented effects of muscle fatigue (Bigland-Ritchie, Johansson, Lippold, & Woods, 1983; Häkkinen & Komi, 1983; Horita & Ishiko, 1987; Parnianpour, Nordin, Kahanovitz, & Frankel, 1988), would cause such an increased delay. Control over the trunk excursion in space will thus be diminished. If those kinds of changes occur in free lifting, our analysis in the earlier study would have missed them. In the present study, we therefore performed a 3-D rather than a 2-D analysis of the kinematics.

In summary, our aims in the present investigation were primarily to study the trunk kinematics in the sagittal plane during repeated discontinuous lifting in order to evaluate the effectivity of centrally mediated adjustments in anticipatory motor control to peripheral fatigue and, additionally, to study the trunk kinematics in the frontal and transversal plane in order to evaluate the accuracy of the control of the ongoing movement.

Method

Subjects

Healthy men (N = 10; age, 23.3 years [SD = 2.1 years]; height, 1.78 m [SD = 0.05 m]; body mass, 71 kg [SD = 8 kg]) participated in the experiment. All subjects signed an informed consent before participating. None had a history of low-back pain.

Procedure

The subjects were asked to lift a barbell weighing 10% of their body mass for approximately 40 min. No instructions were given about the starting posture or lifting technique. We used a motor-driven lifting device to standardize the time for a lifting cycle. That device consisted, in essence, of two movable horizontal arms upon which the barbell could be placed. The subjects only lifted the barbell; the lifting device lowered it. For details of the lifting device, see Toussaint et al. (1995). The subjects performed at least 80 practice lifts to become familiar with the lifting task. After those practice lifts, we allowed a sufficient resting period to ensure that the lifters were completely recovered.

The subject lifted the barbell to knuckle-height in upright stance. In the lowest position, the vertical distance between the barbell and the ground was 10% of his body height. The horizontal distance between the barbell and the most anterior part of the toes of the subject was 15% of his body height.

The durations of the lifting and unloaded lowering movements were 0.7 s each. In both the lowest and highest position, a pause of 1 s was made, during which the lifter kept his hands close to the barbell. The initiation of the movement of the lifting device was preceded by an auditory cue. Because the subject followed the movements of the lifting device, movement times were constant throughout the experiment. The experiment consisted of 9 bouts of 70 lifting movements. After each 35 lifts, the lifter rested in a standing posture during 5 cycles of the lifting device. The last three lifting movements of each bout were recorded, and the first three lifting movements of the first bout served as a reference condition. In total, 10 recordings were performed.

Data Collection

We used in this study a three-dimensional linked segment model (LSM; Kingma, Looze, Toussaint, Klijnsma, & Bruijnen, 1996) to examine the kinematics of the lifting movement. The model comprised eight segments, each of which was assumed to be rigid and connected to the others by frictionless pivot joints. The segments were two feet, two lower legs, two upper legs, a pelvis, and a trunk. To each segment, a brace with five markers was attached. The trunk brace was firmly strapped on the thorax. We recorded the position of the markers on the brace as well as the markers on anatomical landmarks to allow reconstruction of the position of the anatomical axis system at each instant of time from the position and orientation of the brace. A single marker attached to the barbell recorded the trajectory of the load. The anatomical axes were defined according to McConville, Churchill, Kaleps, Clauser, and Cuzzi (1980). During the lifting movements, we used a 3-D automatic video-based motion recording system (VICON; Oxford Metrics, four cameras) to record the positions of the brace markers at 60 Hz; we then used those positions as input for the LSM.

Data Analysis

We digitally filtered the coordinates of the segment center of mass position that had been calculated with the LSM, using a fourth-order Butterworth filter with zero phase lag at an effective cut-off frequency of 5 Hz.

To describe the joint angles, we determined Euler angles in the following sequence: frontal (flexion–extension), sagittal (lateral flexion), longitudinal (torsion) axis. Flexion and extension were defined by the angles between the anatomical axes of two successive segments. Flexion was defined as negative. Excursions out of the sagittal plane were analyzed only for the fifth lumbar to first sacral (L5–S1) joint. Lateral flexion and torsion were defined in the anatomical axis system of the trunk. We considered that the absolute value of the torsion and the lateral flexion angle would detect deviations from pure flexion and extension movements.

To describe the relative timing of the rotation in neighboring joints, we calculated the phase angles and relative phase angles of the lifting movement according to Kelso, Saltzman, and Tuller (1986). In short, the phase angle (inverse tangent of angular velocity scaled to a range of −1 to +1, divided by angular position scaled to a range of minus −1 to +1) of the distal joint was subtracted from the phase angle of the proximal joint at each point in time. We calculated time-series of joint angles, angular velocity, phase angles, and relative phase angles to describe the kinematics.

The data from three consecutive lifts were averaged and taken as representative of the lifting bout. A positive vertical barbell velocity indicated the start of the lifting phase. For all variables, the mean was determined over the first 250 ms of the lifting phase because kinematic changes resulting from inadequate anticipatory control were expected to be most pronounced in that part of the lifting movement. Compensatory reactions would be delayed because of premotor and motor times before a reaction to the sensed load mass occurred; those reactions would therefore hardly be noticeable in the first 250 ms of the movement (Kroll, 1973, 1974).

We used analysis of variance (ANOVA) with repeated measures to test for the effects of repetition (fatigue). Missing values (one complete trial for 1 subject) were estimated iteratively as described by Healy and Westmacott (1956). That procedure does not affect the estimated treatment effects, and the degrees of freedom were adjusted to the number of valid observations. The iterative estimation procedure allows one to use incomplete data sets of a subject, thereby avoiding the need to exclude those data. We used a stratified linear regression analysis, with the order of the recording as the independent variable, to indicate the direction of the significant change found in the ANOVA. In that procedure, one estimates an intercept per subject in order to accommodate for interindividual variation, whereas the slope estimate is based on the pooled data. In all tests, a 5% level of significance was used.

Results

The flexion–extension movements at the beginning of the first lifting bout are illustrated in Figure 1. The presence of considerable interindividual variation is clearly shown by the figure; that variation reflects, among other things, the fact that no lifting technique was prescribed. The quantitative analyses performed were all based on the period between the vertical lines marking the initiation of vertical load movement and 250 ms after that point. To give an impression of the stability of the kinematic pattern (i.e., its low within-subject variance), we plotted in Figure 2 some examples of the lumbosacral joint angle excursion in 4 randomly picked individual subjects. The results shown are the means of the three initial lifts, and the bars represent the standard deviations between the three consecutive lifts. The

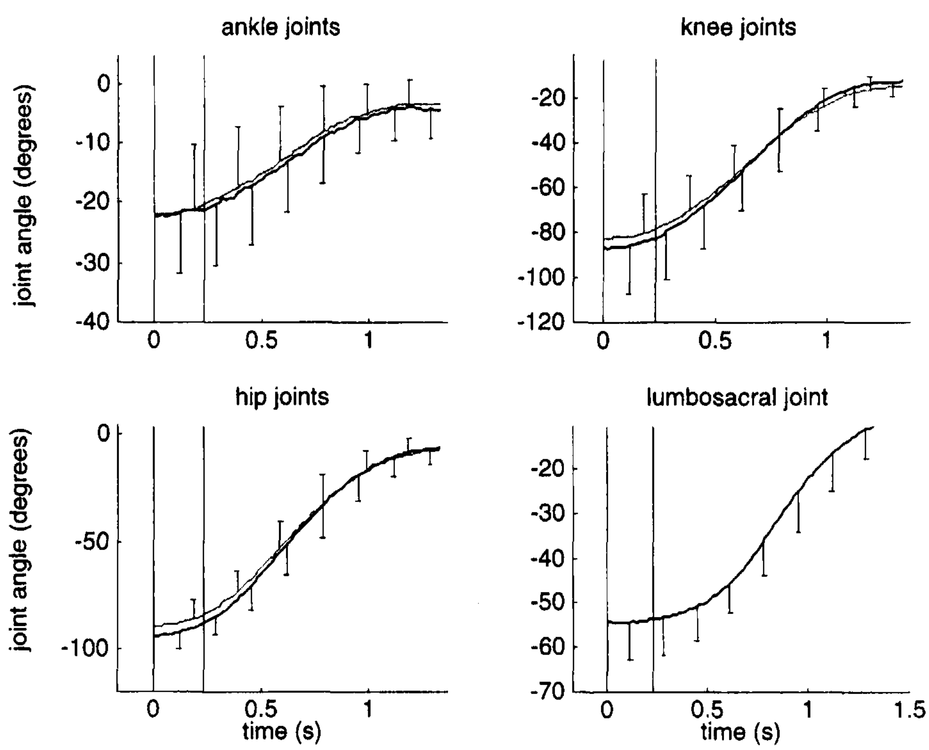

FIGURE 1. Joint angle excursions in the flexion–extension direction during the first three lifting cycles, averaged across subjects ($N = 10$). The vertical bars indicate 1 standard deviation. The period between the lines represents the first 250 ms after vertical load displacement was detected. All quantitative analyses refer to that period. Dark lines represent the left leg joint; light lines represent the right leg joint.

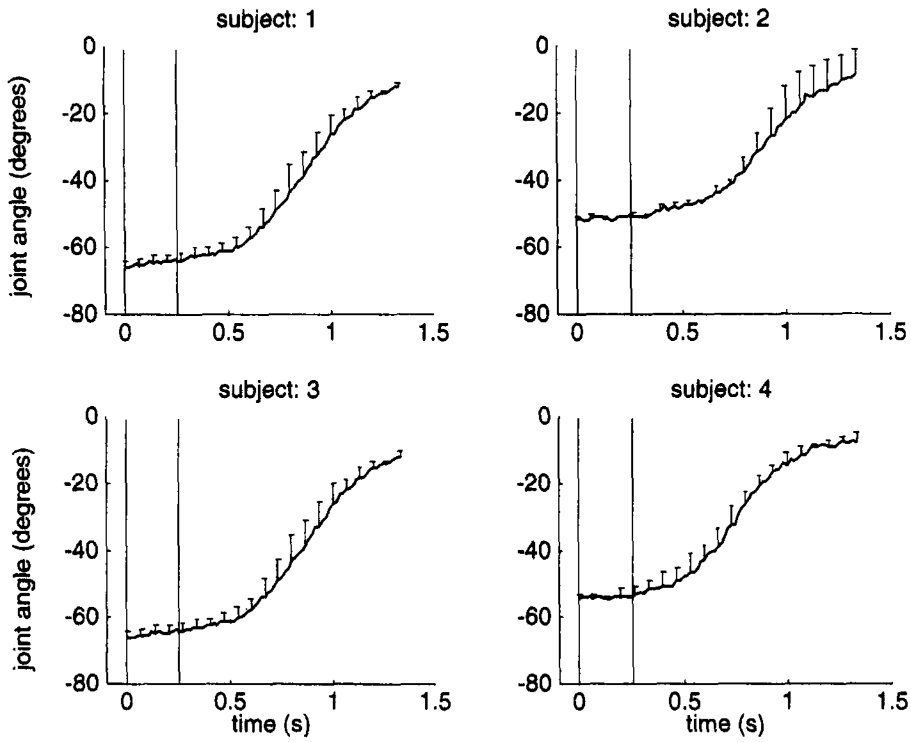

FIGURE 2. Lumbosacral joint angle excursions in the flexion–extension direction in the initial lifting cycles of 4 randomly picked subjects. The vertical bars indicate the standard deviation between three consecutive lifting cycles of the same subject.

median standard deviation during the 250 ms analyzed was always below 10% of the mean angle in that period.

From the curves in Figure 1, one can see that a time lag between trunk and leg joint excursions was present, as has been described previously (Burgess-Limerick et al., 1995; Dieën et al., 1996; Scholz, 1993a, 1993b). The time lag is even more clearly shown in Figure 3, where the angular velocities of the same joints are presented. Left–right differences in joint excursion and angular velocity were only minor.

Our hypothesis was that angular velocity at the lumbosacral joint would decrease with fatigue and, hence, with the number of repetitions. In Table 1, the results of the analysis of variance on mean angular velocities in the first 250 ms are presented. As can be seen, velocity was affected by repetition. The slopes of angular velocity against bout number, as estimated for those joints in which the ANOVA revealed a significant effect, are also provided in Table 1. The velocity in the hip joints was found to increase significantly (faster extension); whereas in the lumbosacral joint, velocity decreased significantly. The extension velocity of the lumbosacral and hip joints in the first (dark line) and last (light line) lifting bouts, averaged across subjects, are represented in the left-hand window of Figure 4. As can be seen, the velocity changes in the initial part of the lifting movement were corrected for in the latter part of the movement, resulting in significant inverse changes in velocity in the fourth 250-ms episode of the movement for all three joints: left hip, slope = –0.02 deg/s per bout, $t(88) = -1.81$, $p < .05$; right hip, slope = –0.02 deg/s per bout, $t(88) = -1.75$, $p < .05$; lumbosacral joint, slope = 0.04 deg/s per bout, $t(88) = 4.44$, $p < .001$. The trend in the angular velocities during the first 250 ms of the lifting movements and the concomitant individual data (displayed in the right-hand windows of Figure 4) showed that most subjects followed the general trend, especially with regard to lumbosacral angular velocity. Differences between subjects consisted mainly of differences in intercept. Regression analyses on individual data confirmed that observation for 8 of the 10 subjects.

As we expected, the angular velocity changes, depicted in Figure 4, led to an increase of the phase lag between hip and trunk extension (Table 1). That lag is illustrated in Figure 5, in which the relative phase angle of those joints is plotted for the first and last lifting bouts.

The posture adopted by the subjects at the initiation of the lifting movement changed over time (Table 2). An increase (decreased flexion) in the angles of the lower extremity was found, but it did not reach significance in the left knee. A decreased angle (increased flexion) was found in the lumbosacral joint. Overall, we can state that during the experiment the subjects showed a greater tendency to use the so-called back-lifting technique, in which leg joint excursions are minimized. Four subjects used more or less

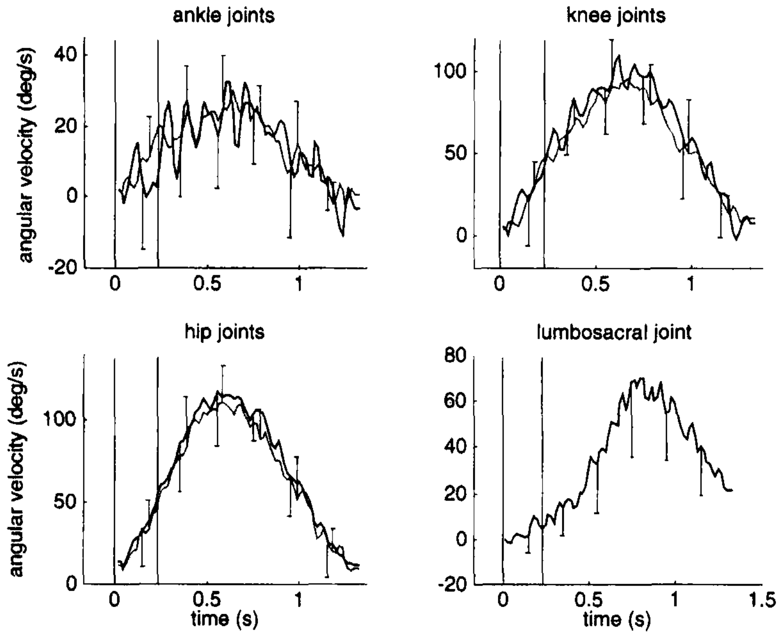

FIGURE 3. Angular velocities in the flexion–extension direction during the first three lifting cycles, averaged across subjects ($N = 10$). The vertical bars indicate 1 standard deviation. The period between the lines represents the first 250 ms after vertical load displacement was detected. All quantitative analyses refer to that period. Dark lines represent the left leg joint; light lines represent the right leg joint.

TABLE 1
Results of the Angular Velocity and the Phase Lag Analyses of Variance and the Regression Analyses

Variable Joint	*p* value, ANOVA	Slope regression (deg/s per bout)	*p* value
Angular velocity			
Left ankle	*ns*		
Right ankle	*ns*		
Left knee	*ns*		
Right knee	*ns*		
Left hip	.042	1.1	< .001
Right hip	.043	1.0	< .001
Lumbosacral	.003	–0.5	< .001
Phase lag			
Left hip-lumbosacral	.025	–0.01	< .001
Right hip-lumbosacral	.021	–0.01	< .001

Note. Analyses of variance (ANOVAs) were conducted on the angular velocities and on the phase lag between hip and lumbosacral extension (during the first 250 ms). In the regression analyses performed on those variables, the order of the lifting bout was the dependent variable.

a back technique from the onset of the experiment, 5 subjects started with a technique more like the leg technique, in which lumbosacral excursion is minimized, and 1 subject started by using an intermediate technique. The change toward a back technique is clearly illustrated in Figure 6. The angular excursions of a subject who initially used a technique resembling the leg technique are presented in the top windows. The technique used in the final bouts by that subject (light line, upper windows) closely resembled the technique used by the subject whose first and last lifting bouts are compared in the lower windows of the figure. As can be appreciated, the changes in lifting technique for the

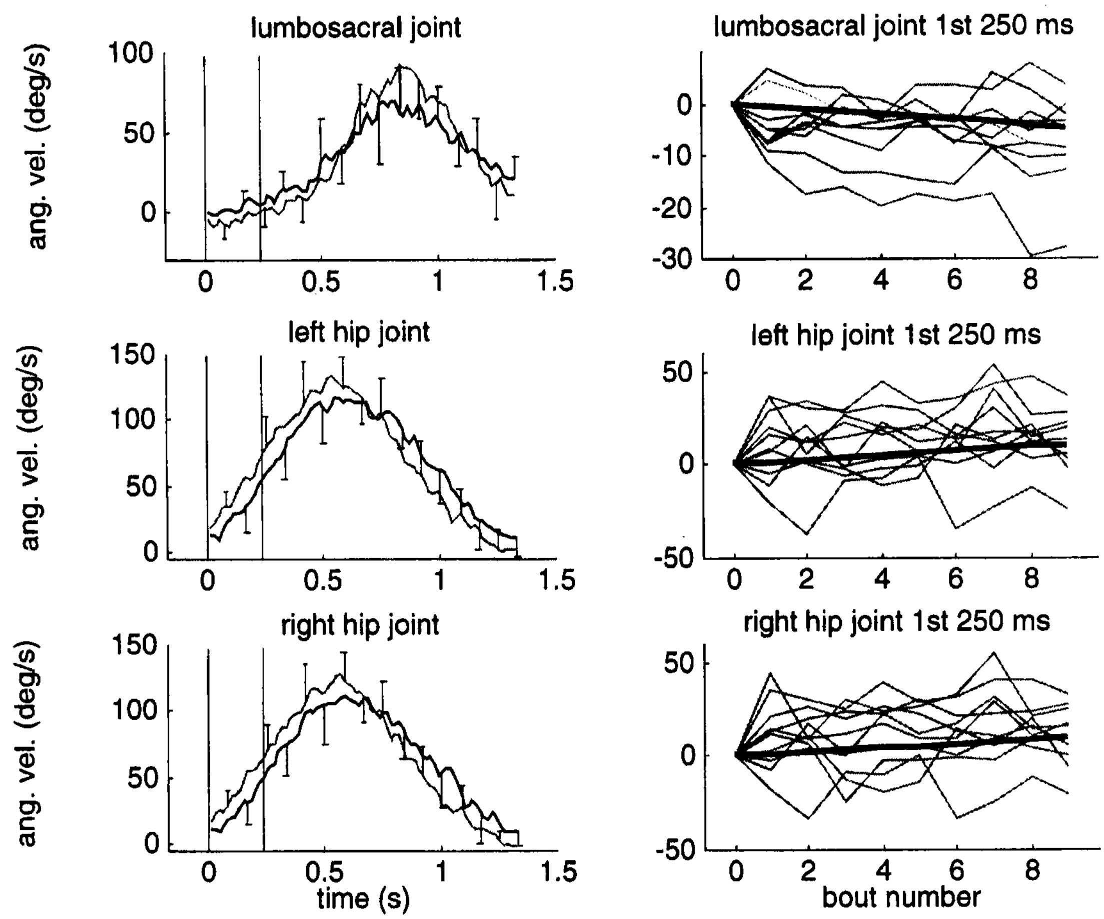

FIGURE 4. Left. Angular velocities in the initial lifting cycles (dark line) compared with those in the final lifting bout (light line), averaged across subjects ($N = 10$). The vertical bars indicate 1 standard deviation. **Right**. The change in mean angular velocities during the first 250 ms of the lifting movement per subject, plotted against the number of the lifting bout. The thick line indicates the regression line calculated on the basis of the pooled data from all subjects.

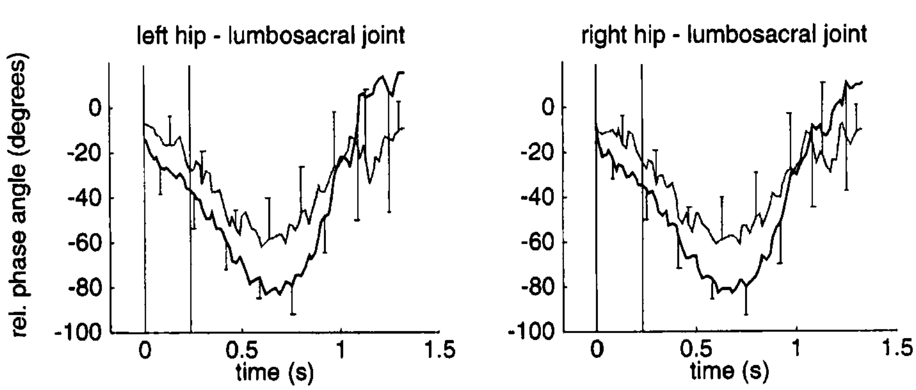

FIGURE 5. The relative phase angles between the hips and the lumbosacral joint in the initial lifting cycles (dark line) compared with those in the final lifting bout (light line), averaged across subjects ($N = 10$). A negative relative phase angle indicates that the distal joint was leading the proximal joint. The vertical bars indicate 1 standard deviation.

TABLE 2
Analyses of Variance and Regression Analyses on Initial Joint Angles and Out-of-Plane Ranges of Motion

Joint	*p* value, ANOVA	Slope regression (deg/ per bout)	*p* value
Left ankle	.016	0.6	< .001
Right ankle	.001	0.5	< .001
Left knee	*ns*		
Right knee	.010	1.3	< .001
Left hip	.039	0.6	< .001
Right hip	.032	0.6	< .001
Lumbosacral	< .001	–0.7	< .001
ROM lateral flexion	*ns*		
ROM twisting	< .001	0.5	< .001

Note. ANOVAs were conducted on initial joint angles and out-of-plane ranges of motion (ROM) at the lumbosacral joint, and the regressions of those variables were analyzed; the order of the lifting bout was the dependent variable in the regression analyses.

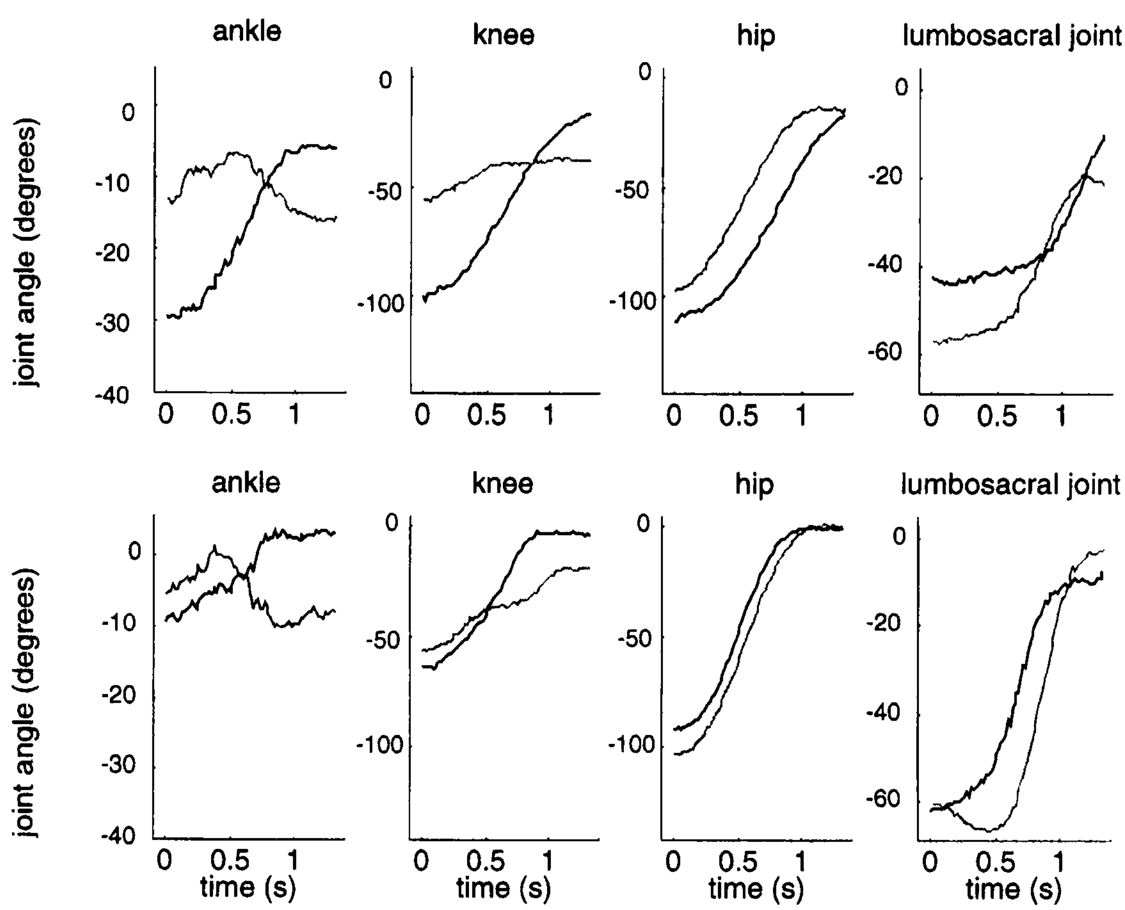

FIGURE 6. Angular excursions in the initial lifting cycles (dark line) compared with those in the final lifting bout (light line). **Top**. A subject who initially displayed a leg technique. **Bottom**. A subject displaying a back technique.

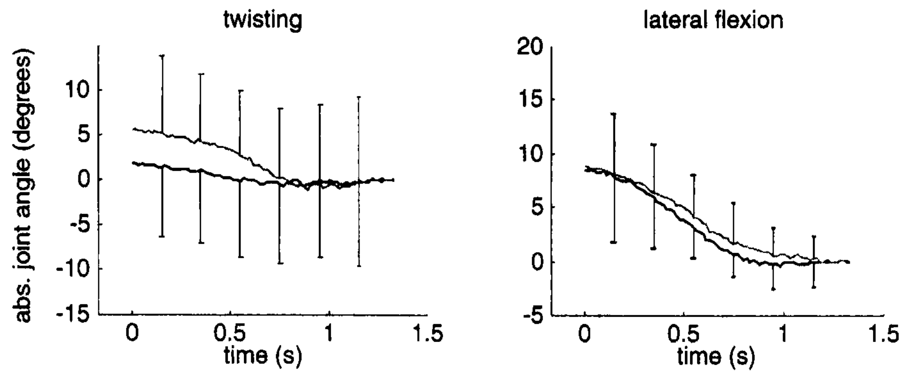

FIGURE 7. The angular excursions in the lumbosacral joint that were outside the sagittal plane in the initial lifting cycles (dark line) compared with those in the final lifting bout (light line), averaged across subjects ($N = 10$). The vertical bars indicate 1 standard deviation.

subject who initially used a back technique were qualitatively similar. In the final bout, leg joint excursions were further minimized and lumbosacral excursion increased. The same tendency was observed in all subjects. Also, the changes in lumbosacral angular velocity and, to a lesser extent, hip angular velocities were qualitatively similar in all subjects, as can be seen in the right-hand windows of Figure 4. Therefore, the data of subjects who started with different initial techniques were not analyzed separately.

Twisting motion increased significantly with repetition, whereas changes in the range of motion in lateral flexing did not reach significance (Table 2). That finding is illustrated by Figure 7, which compares the absolute angular excursions averaged across subjects between the first and last lifting bouts.

Discussion

Our main hypothesis in the present study was that repetitive lifting would cause a decrease of trunk extension velocity with the number of repetitions and, consequently, an increased phase lag between hip and trunk extension. The results presented clearly support our hypothesis. The hypothesis was based on the expectation that the back muscles would fatigue, but because we did not measure the force-producing capacity of the back muscles we can not ascertain that the effects found were indeed mediated by muscle fatigue. However, several authors (Jørgensen, Andersen, Horst, Jensen, & Nielsen, 1985; Petrofsky & Lind, 1978; Potvin & Norman, 1993) have shown that lifting with similar loads and durations is certainly fatiguing for those muscles. The subjective reports of our subjects confirmed that they felt fatigued. In view of the considerable number of practice trials performed, it seems unlikely that subjects changed their lifting strategy during the experiment on the basis of experience from the preceding lifts. Relating the changes found to back muscle fatigue, therefore, seems reasonable. However, alternative explanations for our findings will be discussed below.

Our expectation that a change in hip trunk coordination would occur was based on a generalization of previous findings on the coupling of perception and action. That generalization can be phrased as follows. Action requires information on the environment; that information has to be related to the capacity of the motor system. For example, reaching for an object requires information about ones distance from the object in relation to, for instance, arm length (Carello, Grosofsky, Reichel, Solomon, & Turvey, 1989). It has been asserted, vice versa, that action provides the actor with relevant information. For instance, freely moving perceivers have been shown to be more accurate than stationary perceivers in judging whether or not lofted balls can be caught (Oudejans, Michaels, Bakker, & Dolné, 1996).

In the task of lifting, information on load mass is needed in relation to the force-producing capacity of the to-be-active musculature. It is quite conceivable that the actual handling of the load can produce that information. Conscious estimation of the mass of a hand-held load appears to be quite adequate (Jones & Hunter, 1982). During fatigue, the consciously estimated load mass is higher than actual load mass is; that difference reflects the change of the balance between the load mass and the force-producing capacity of the muscle as the muscle's ability to produce force decreases (Jones, 1983). Thus the centrally available information on the balance between load mass and the force-producing capacity of the trunk muscles that is obtained during a lifting movement might be used to adjust in an anticipatory fashion the control of those muscles for the next lift.

It has been shown, however, that instantaneously available information is more useful for controlling action than memory-based information is (Oudejans, Michaels, Bakker, & Dolné, 1996; Oudejans, Michaels, Dort, & Frissen, 1996). An explanation for that finding might be that direct adjustments to the motor command are possible at a low level without any need for higher level memory-based information. In the case of lifting a load, the increased muscle spindle sensitivity that occurs with fatigue (Djupsjöbacka, Johansson, & Bergenheim, 1994; Gantchev, 1990; Ljubisavljevic, Jovanovic, & Anastasijevic, 1992; Nelson & Hutton, 1985) might be a very efficient means of achieving the goal of adjusting the motor command directly at a low level. That process, however, could function only in the presence of instantaneous information on load mass—thus, while an individual is handling the load. In the discrete lifting task studied here, that information was not present during the preparation of the lifting movement. Mechanically, picking up the load is a perturbation of the posture or movement prior to that instant. It has been shown that the disturbance is counteracted partially by anticipatory adjustments (Toussaint el al., 1997). Those adjustments thus have to be performed without the availability of instantaneous feedback on the balance between load mass and muscle capacity. That lack could conceivably lead to an underactivation of the fatigued muscles and a resulting decrease in the acceleration and, hence, velocity, or even to a negative (in the flexion direction) acceleration and, hence, velocity, in the joint controlled by those muscles.

Underactivation of the back muscles, which could explain the decrease in angular velocity at the lumbosacral joint, could occur if the onset of activation remains unaltered while the rate of force production of the muscles decreases (Bigland-Ritchie et al., 1983; Parnianpour et al., 1988). In experiments in which no external loads are handled, generally an earlier onset of agonist activity is seen with fatigue, and that activity appears to prevent changes in the kinematics (Arendt-Nielsen & Sinkjær, 1991; Bonnard, Sirin, Oddsson, & Thorstensson, 1994; Lucidi & Lehman, 1991). Unfortunately, the pattern of the back muscle activity in lifting does not allow EMG onset preceding the upward movement to be reliably determined. Therefore, we are not able to ascertain whether the changes in the timing of muscle activation in continuous lifting–lowering differ from the timing changes in discrete lifting.

The too low force production by the back muscles is not necessarily related to inadequate adjustments of the activation pattern. Alternatively, the muscles simply might no longer be able to produce the required force. That seems unlikely, however. Lifting a load of 10% of body weight is certainly a submaximal task. In addition, in our previous experiment (Dieën et al., 1996) subjects lifted similar loads at a comparable frequency and from a comparable initial position for 400 cycles, holding the load continuously in their hands. Nevertheless, no changes in trunk kinematics were found. In the present study, the changes in trunk kinematics were already discernible in the second lifting bout (Figure 4). It seems unlikely that subjects would be able to perform the full 630 lifting cycles when the required muscle force was already supramaximal after 70 cycles.

So far this discussion has focused on the change in trunk kinematics. However, hip extension velocity was shown to change as well. In contrast to trunk extension velocity, an increase was seen in hip extension velocity. Two reasons might be given for that increase. The first is that decreased torque production by the back muscles allows a faster backward rotatory acceleration of the pelvis and, hence, a faster hip extension. Thus, that finding may simply be secondary to the fatigue effect discussed above. We will address the second possible cause later in the discussion.

We have already explained that the changes in trunk kinematics and in hip trunk coordination were a consequence of the inadequate control caused by back muscle fatigue. As Latash and Anson (1996) have pointed out, however, changed kinematics in pathology should not be dismissed too readily as being a consequence of erroneous motor control. In contrast, the kinematic changes may reflect highly functional adjustments to changes in the system. Likewise, in the case of fatigue, the kinematics may be changed to compensate for the inability of the fatigued muscles to produce the forces originally produced. Earlier in the discussion, an inability of the muscles to produce the required force was dismissed as unlikely in the present experiment. Nevertheless, kinematics may also change because of functional adaptations aimed at avoiding or retarding discomfort and fatigue development. In other words, the lifting strategy could be modified to allow prolonged continuation of the task performance. In the present study, that explanation certainly deserves further scrutiny.

A functional role for a decreased angular velocity at the lumbosacral junction, in that sense, could be to shift power production from the back muscles to the hip extensor muscles. Joint power is the product of joint torque and angular velocity. Because joint torques are high initially in the lifting movement, the observed changes in velocity would certainly cause such an effect. Nevertheless, the functional relevance of that effect seems limited. First, as can be seen in Figure 3, angular velocity in the lumbosacral joint during the initial lifting phase was already very low in the first bout. Hence, little or no power was produced. Second, in 8 out of 10 subjects the velocity took on negative values during the later lifting bouts; thus, power was lost at the lumbosacral joint. It would be difficult to find a functional explanation for the negative change in velocity. In addition, if it is a functional adaptation, a similar change should have occurred in the continuous lifting task studied previously by Dieën et al. (1996), but they found no such change.

A second functional explanation could be that the decreased lumbosacral extension velocity allows the back muscles to remain at a longer muscle length for a larger fraction of the lifting phase. Staying at that longer length would keep them in a more optimal part of their length–force relationship and could thus limit fatigue and discomfort in these muscles. Again, that explanation is contradicted by the finding of predominantly negative velocities (lengthening of the muscles) in the later lifting bouts. Lengthening contractions are well known to cause extreme discomfort in muscles (Armstrong, Warren, & Warren, 1991; Fridén, Sjöström, & Ekblom, 1983).

Besides the changes in the angular velocities with repetition, a change in the posture from which the subjects lifted was found (Figure 6). In later bouts, the lifting technique more and more approached a pure back technique. In our view, that finding can be explained as a functional adaptation rather than as a consequence of fatigue. Because of the smaller excursion of the body center of mass in back lifting, as compared with leg lifting, the former technique is energetically more efficient (Welbergen, Kemper, Knibbe, Toussaint, & Clysen, 1991). In contrast to what is often assumed, the lumbosacral torque is not higher in back lifting than in leg lifting (Dieën, Creemers, Draisma, Toussaint, & Kingma, 1994). Therefore, exploiting the energetic efficiency of back lifting does not entail a higher load on the already fatigued back muscles. Furthermore, it may relieve the quadriceps muscles from fatiguing contributions to the lifting movement (Trafimow, Schipplein, Novak, & Andersson, 1993). The fact that no change toward back lifting was found in the previous experiment on continuous lifting might be explained by the strict instructions the subjects were given on lifting technique (Dieën, Toussaint, Maurice, & Mientjes, 1996). When subscribing to that position, one should consider further whether the changes in angular velocity at the hips and trunk are not secondary to that functional adaptation.

A comparison of the angular velocities in the initial lifting cycles found in the 4 subjects who used a clear back technique with those of the 5 subjects who primarily used a leg technique with respect to the angular velocity at the lumbosacral joint does not support that explanation (2 deg/s [*SD* = 4 deg/s] in leg lifting vs. 6 deg/s [*SD* = 5 deg/s]). The increase in hip angular velocity might in part be explained, however, by the change in lifting technique; the velocity was found to be 28 deg/s (*SD* = 3 deg/s) and 24 deg/s (*SD* = 7 deg/s) in the left and right hips in leg lifting and 43 deg/s (*SD* = 7 deg/s) and 40 deg/s (*SD* 8 = deg/s) in the left and right hips in back lifting. A Student *t* test revealed the difference to be significant at the .05 level for both the left and right hips. In conclusion, the change in hip angular velocity might (in part) be explained by a change in lifting technique used, a change that seems to be a functional adaptation of the movement pattern. That would then also explain in part the change in lumbosacral phase lag, because that change is codetermined by hip angular velocity. The data do not support the suggestion that the change in trunk angular velocity might be secondary to a change in lifting technique. If anything, the average values point in the opposite direction.

With repetition, the range of motion around the trunk's longitudinal axis (twisting) increased. Our expectation that trunk motions outside of the sagittal plane would be increased was based on dynamometer tests performed by Parnianpour et al. (1988). In contrast to our findings, those authors reported increases predominantly in lateral flexion. However, angular excursions in three dimensions can be directly compared only when the procedure for determining the angles is the same. We used an Euler decomposition in the order flexion–extension in the pelvis's axis system, lateral flexion, and twisting in the trunk's axis system. Unfortunately Parnianpour and his colleagues did not report the procedure they used. They hypothesized that the increase in motion out of the sagittal plane was caused by less precise feedback control that resulted from increased delays in both voluntary (Bigland-Ritchie et al., 1983; Parnianpour et al., 1988) and reflex (Häkkinen & Komi, 1983; Kroll, 1974) motor reactions. Alternatively, impaired feedback control of the movement might be explained by the decreased information transmission in muscle spindle afference that occurs with fatigue and which is expected to affect propriocepsis (Bergenheim, Johansson, Pedersen, & Djupsjöbacka, in press).

A final question to be addressed is what the consequences of the changes in kinematics found might be. From the perspective of whether the subjects remained capable of performing the prescribed task, those changes clearly have no meaning. In spite of changes in the pattern of joint rotations and the decreased precision of the trunk movements, as reflected in increased out-of-plane motion, the subjects were able to follow the prescribed trajectory of the load at the prescribed rate. That ability illustrates only the flexibility offered by the high number of degrees of freedom in the motor system. However, one may ask whether there is a higher order cost of maintaining task performance in the presence of fatigue while allowing for changes in the movement pattern. It has been suggested that fatigue would increase the probability of musculoskeletal injuries (e.g., Bunkens, 1996; Seidel et al., 1987). Some of the changes we found offer support for this assumption. First, the negative angular velocity implies eccentric action of the back muscles in the beginning of the lifting movement, while the lumbosacral torque and, thus, muscle forces are high. High-intensity eccentric actions are known to lead to reversible but extensive muscular damage (Fridén et al., 1983). In addition, the increased lumbosacral flexion might take the spine to its elastic limit. Extreme flexion, in combination with a high compression force, which also occurs at the initiation of a lifting movement, can produce herniation of the intervertebral disc (Adams & Hutton, 1982). That risk seems further enhanced by twisting of the spine (Gordon et al., 1991; Shirazi-Adl, Ahmed, & Shrivastava, 1986). It is questionable, however, whether the twisting angles found in the present study were large enough to cause that effect. Nevertheless, when performing an asymmetric lifting task, a similar increase in twisting, on top of already present asymmetry, may occur because of fatigue and thus put the spine at a greater risk. Those findings might explain the strong association between repeated asymmetric lifting and disc herniation (Kelsey, Gittens, & White, 1984).

REFERENCES

Adams, M. A., & Hutton, W. C. (1982). Prolapsed intervertebral disc. A hyperflexion injury. *Spine, 7,* 184–191.

Arendt-Nielsen, L., & Sinkjær, T. (1991). Assessment of dynamic muscle fatigue by EMG- and kinematic profiles. In P. A. Anderson, D. J. Hobart, & J. V. Danoff (Eds.), *Electromyographic kinesiology* (pp. 247–250). Amsterdam: Elsevier.

Armstrong, R. B., Warren, G. L., & Warren, J. A. (1991). Mechanisms of exercise-induced muscle injury. *Sports Medicine, 12,* 184–207.

Bergenheim, M., Johansson, H., Pedersen, J., & Djupsjöbacka, M. (in press). Alterations in information transmission in ensembles of primary muscle spindle afferents induced by increased intramuscular concentrations of lactic acid in heteronymous muscle. *Neuroscience Research.*

Bigland-Ritchie, B., Johansson, R., Lippold, O. C. J., & Woods, J. J. (1983). Contractile speed and EMG changes during fatigue of sustained maximal voluntary contractions. *Journal of Neurophysiology, 50,* 313–324.

Bigland-Ritchie, B., & Woods, J. J. (1984). Changes in muscle contractile properties and neural control during human muscular fatigue. *Muscle & Nerve, 7,* 691–699.

Bonnard, M., Sirin, A. V., Oddsson, L., & Thorstensson A. (1994). Different strategies to compensate for the effect of fatigue revealed by neuromuscular adaptation processes in humans. *Neuroscience Letters, 166,* 101–105.

Bunkens, H. (1996). *Back muscle fatigue and chronic low back pain.* Unpublished doctoral dissertation, University of Leuven.

Burgess-Limerick, R., Abernethy, B., Neal, R. J., & Kippers V. (1995). Self-selected manual lifting technique: Functional consequences of the interjoint coordination. *Human Factors, 37,* 395–411.

Carello, C., Grosofsky, A., Reichel, F. D., Solomon, H. Y., & Turvey, M. T. (1989). Visually perceiving what is reachable. *Ecological Psychology, 1,* 27–54.

Commissaris, D. A. C. M., & Toussaint, H. M. (1997a). Load knowledge affects low back loading and control of balance in lifting tasks. *Ergonomics, 40,* 559–575.

Commissaris, D. A. C. M., & Toussaint, H. M. (1997b). *The coordination of movement and equilibrium in a bimanual lifting task: I. A synergy between upper body and lower segments controls equilibrium.* Manuscript submitted for publication.

Dieën, J. H. van, Creemers, M., Draisma, I., Toussaint, H. M., & Kingma, I. (1994). Repetitive lifting and spinal shrinkage. Effects of age and lifting technique. *Clinical Biomechanics, 9,* 367–374

Dieën, J. H. van, Toussaint, H. M., Maurice, C., & Mientjes, M. (1996). Fatigue-related changes in the coordination of lifting and their effect on low back load. *Journal of Motor Behavior, 28,* 304–314.

Djupsjöbacka, M., Johansson, H., & Bergenheim, M. (1994). Influences on the γ-muscle-spindle system from muscle afferents stimulated by increased intramuscular concentrations of arachidonic acid. *Brain Research, 663,* 293–302.

Fridén, J., Sjöström, M., & Ekblom, B. (1983). A morphological study of delayed muscle soreness. *Experientia, 37,* 506–507.

Frymoyer, J. W., Pope, M. H., Clements, J. H., Wilder, D. G., MacPherson, B., & Ashikaga, T. (1983). Risk factors in low back pain: An epidemiological survey. *Journal of Bone & Joint Surgery, 65A,* 213–218.

Gantchev, G. N., (1990). Adaptive postural processes during changes in the functional capacity of the muscular system. In T. Brandt, W. Paulus, W. Bles, M. Dietrich, S. Krafczyk, & A. Straube (Eds.), *Disorders of posture and gait* (pp. 269–276). Stuttgart: Thieme.

Garg, A., & Moore, J. S. (1992). Epidemiology of low-back pain in industry. *Occupational Medicine, 7,* 593–608.

Gordon, S. J., Yang, K. H., Mayer, P. J., Mace, A. H., Kish, V. L., & Radin, E. L. (1991). Mechanism of disc rupture. A preliminary report. *Spine, 16,* 450–456.

Häkkinen, K., & Komi, P. V. (1983). Electromyographic and mechanical characteristics of human skeletal muscle during fatigue under voluntary and reflex conditions. *Electroencephalography and Clinical Neurophysiology, 55,* 436–444.

Healy, M. J. R., & Westmacott, M. J. H. (1956). Missing values in experiments analysed on automatic computers. *Applied Statistics, 5,* 203–206.

Johansson, R. S., & Cole, K. J. (1992). Sensory-motor coordination during grasping and manipulative actions. *Current Opinions in Neurobiology, 2,* 815–823.

Jones, L. A. (1983). Role of central and peripheral signals in force sensation during fatigue. *Experimental Neurology, 81,* 497–503.

Jones, L. A., & Hunter, I. W. (1982). Force sensation in isometric contractions: A relative force effect. *Brain Research, 244,* 186–189.

Jørgensen, K., Andersen, B., Horst, D., Jensen, S., & Nielsen, A. (1985). The load on the back in different handling operations. *Ergonomics, 28,* 183–196.

Kelsey, J. L., Gittens, P. B., & White, A. A. (1984). An epidemiologic study of lifting and twisting on the job and risk for acute prolapsed intervertebral disc. *Journal of Orthopaedic Research, 2,* 61–66.

Kelso, J. A. S., Saltzman, E. L., & Tuller, B. (1986). The dynamical perspective on speech production. *Journal of Phonetics, 14,* 29–59.

Kingma, I., Looze, M. P. de, Toussaint, H. M., Klijnsma, H. G., & Bruijnen, T. B. M. (1996). Validation of a full body 3-D dynamic linked segment model. *Human Movement Science, 15,* 833–860.

Kroll, W. (1973). Effects of local muscular fatigue due to isotonic and isometric exercise upon fractionated reaction time components. *Journal of Motor Behavior, 5,* 81–93.

Kroll, W. (1974). Fractionated reaction and reflex time before and after fatiguing isotonic exercise. *Medicine and Science in Sports and Exercise, 6,* 260–266.

Latash, M. L., & Anson, J. G. (1996). What are normal movements in atypical populations? *Behavioral and Brain Sciences, 19,* 55–106.

Ljubisavljevic, M., Jovanovic, K., & Anastasijevic, R. (1992). Changes in discharge rate of cat hamstring fusimotor neurones during fatiguing contraction of triceps surae muscles. *Brain Research, 579,* 246–252.

Lucidi, C. A., & Lehman, S. L. (1991). Adaptation to fatigue of long duration in human wrist movements. In P. A. Anderson, D. J. Hobart, & J. V. Danoff (Eds.), *Electromyographic kinesiology* (pp. 259–262) Amsterdam: Elsevier.

McConville, J. T., Churchill, T. D., Kaleps, I., Clauser, C. E., & Cuzzi, J. (1980). *Anthropometric relationships of body and body segment moments of inertia.* (Air Force Aerospace Medical Research Laboratory Tech Rep. no. AFAMRL-TR-280-119). Wright-Patterson Air Force Base, OH.

Nelson, D. L., & Hutton, R. S. (1985). Dynamic and static stretch responses in muscle spindle receptors in fatigued muscles. *Medicine and Science in Sports and Exercise, 17,* 445–450.

Oudejans, R. R. D., Michaels, C. F., Bakker, F. C., & Dolné, M. (1996). The relevance of action in perceiving affordances: Perception of catchableness of fly balls. *Journal of Experimental Psychology: Human Perception and Performance, 22,* 879–891.

Oudejans, R. R. D., Michaels, C. F., Dort, B. van, & Frissen E. J. P. (1996). To cross or not to cross: The effect of locomotion on perception of safety. *Ecological Psychology, 8,* 259–267.

Parnianpour, M., Nordin, M., Kahanovirtz, N., & Frankel, V. (1988). The triaxial coupling of torque generation of trunk muscles during isometric exertions and the effect of fatiguing isoinertial movements in the motor output and movement patterns. *Spine, 13,* 982–992.

Petrofsky, J. S., & Lind, A. R. (1978). Metabolic, cardiovascular and respiratory factors in the development of fatigue in lifting tasks. *Journal of Applied Physiology: Respiratory Environmental & Exercise Physiology, 45,* 64–68.

Potvin, J. R., & Norman, R. W. (1993). Quantification of erector spinae muscle fatigue during prolonged dynamic lifting tasks. *European Journal of Applied Physiology, 67,* 554–562.

Roy, S. H., De Luca, C. J., & Casavant, D. A. (1989). Lumbar muscle fatigue and chronic lower back pain. *Spine, 14,* 992–1001.

Scholz, J. (1993a). The effect of load scaling on the coordination manual squat lifting. *Human Movement Sciences, 12,* 427–459.

Scholz, J. (1993b). Organizational principles for the coordination of lifting. *Human Movement Sciences, 12,* 537–576.

Scholz, J. P., Milford, J. P., & McMillan, A. G. (1995). Neuromuscular coordination of squat lifting, I: Effect of load magnitude. *Physical Therapy, 75,* 119–132.

Seidel, H., Beyer, H., & Brauer, D. (1987). Electromyographic evaluation of back muscle fatigue with repeated sustained contractions of different strength. *European Journal of Applied Phsyiology, 56,* 592–602.

Shirazi-Adl, S. A., Ahmed, A. M., & Shrivastava, S. C. (1986). Mechanical responses of a lumbar motion segment in axial torque alone and combined with compression. *Spine, 11,* 914–927.

Toussaint, H. M., Commissaris, D. A. C. M., Hoozemans, M. J. M., Ober, M. J., & Beek, P. J. (1997). Anticipatory postural adjustments prior to load pick-up in a bi-manual whole body lifting task. *Medicine and Science in Sports and Exercise, 29,* 1208–1215.

Toussaint, H. M., Winter A. F. de, Haas, Y. de, Looze, M. P. de, Dieën, J. H. van, & Kingma, I. (1995). Flexion relaxation during lifting: Implications for torque production by muscle activity and tissue strain at the lumbo-sacral joint. *Journal of Biomechanics, 28,* 199–210.

Trafimow, J. H., Schipplein, O. D., Novak, G. J., & Andersson, G. B. J. (1993). The effects of quadriceps fatigue on the technique of lifting. *Spine, 18,* 364–367.

Welbergen, E., Kemper, H. C. G., Knibbe, J. J., Toussaint, H. M., & Clysen, L. (1991). Efficiency and effectiveness of stoop and squat lifting at different frequencies. *Ergonomics, 34,* 613–624.

Yoshikawa, T., Mori, S., Santiesteban, A. J., Sun, T. C., Hafstad, E., Chen, J., & Burr, D. (1994). The effects of muscle fatigue on bone strain. *Journal of Experimental Biology, 188,* 217–233

Submitted December 20, 1996
Revised March 7, 1997