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Changes in Trunk Dynamics and Spine Loading During Repeated Trunk Exertions

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Study Design. Trunk and hip kinematic and spinal loading changes were documented as experienced subjects performed a set of standard free-dynamic lifting tasks at various intervals throughout a 5-hour repetitive manual materials handling session.

Objectives. To document how spine loading changes as a function of repetitive lifting during an extended period of lifting cycles.

Summary of Background Data. Many studies have evaluated spine loading as a function of a specific lifting task, but no studies could be found in the literature that evaluated how spine loading may change with repeated exposure to a lifting task.

Methods. Ten experienced material handlers were recruited. Each was required to transfer 11 pallets of 23 kg boxes at a rate of 125 lifts per hour. Before and after unloading each pallet, subjects were asked to perform a set of five *standard* free-dynamic lifting tasks. Electromyographic activities of 10 trunk muscles were recorded along with kinematic and kinetic data. An electromyography-assisted model was used to evaluate spine loading during the standard lifting tasks.

Results. Subjects significantly changed their motion patterns throughout the lifting session. Trunk range of motion, velocity, and acceleration decreased in the sagittal plane, whereas these same measures increased for the hip. Trunk moment also decreased by 7% during the standard lifts. These changes were accompanied by a redistribution of muscle recruitment patterns, resulting in a decrease in spine compression and an increase in anterior/posterior shear.

Conclusions. This study has shown that spine loading patterns do indeed change with repetition, suggesting that one needs to monitor these changes throughout repetitive lifting tasks if low back disorder risk is to be minimized. [Key Words: electromyography, spine loads, trunk kinematics] **Spine 1997;22:2564–2570**

Most biomechanical investigations of spine loading have investigated the loads that occur on the lumbar spine in response to particular activities such as lifting or exerting force against a dynamometer in a specific posture.[4,5,7,11] These studies have described the spine loading patterns that occur as subjects perform a limited number of exertions during specific yet brief periods of lifting. These results then may be used to generalize the loading pattern that one might be exposed to under similar repeated exertions. Hence, spine loading patterns observed during a few lifts often are extrapolated throughout a work day to help understand the risk of low back disorder (LBD) that one may be exposed to during the course of an entire work day.

There is evidence in the literature to suggest that the trunk may behave differently as a function of time during repetitive exertions. Parnianpour et al[14] observed an increase in lateral moments in subjects attempting to perform repetitive and fatiguing flexion–extension tasks. Fathallah et al[3] have documented diurnal changes in trunk motion throughout the day. Casual observations by those tracking incident trends in industry have suspected that risk of back injury may change throughout the work day; however, these observations may be confounded with psychosocial or psychological factors. Thus, these observations suggest that repetitive lifting may change the nature of the loading pattern on the spine, which would in turn affect the risk of an occupationally related LBD.

The primary objective of this study was to document whether and how spine loading may change as a function of repetitive lifting during an extended period of lifting cycles. A secondary objective was to determine the biomechanical mechanism responsible for this suspected change in spine loading, if indeed it did occur.

■ Methods

Subjects. Ten men who worked as item selectors at a local warehouse volunteered as subjects in this study. None had a prior history of LBD. The participants' ages ranged from 19–49 years, with an average age of 27.2 years. Work experience ranged from 0.25–23 years in a warehouse setting. The average stature of the subject population was 180.3 ± 7.1 cm, and the average weight was 97.8 ± 8.4 kg.

Experimental Design. Trunk kinematics, hip kinematics, trunk moments, and spine loading were evaluated as the subjects performed a set of five standard test exertions at various points throughout a long repetitive manual materials handling session. The standard test exertions consisted of lifting a 23-kg

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load in a sagittally symmetric position from knee height to waist height. This standard test was performed while the individual stood on a force plate and repeated five times during each set. Trunk positioning relative to this force plate was documented using a goniometric position documentation system described by Fathallah (Fathallah F. Coupled spine motions, spine loading, and risk of occupationally related low back disorders. Unpublished doctoral dissertation, The Ohio State University, 1995). This system used a series of six potentiometers to track pelvic tilt and position relative to a set point on the force plate. This system, along with a trunk goniometer (Lumbar Motion Monitor [LMM]) located at L5–S1, made hip, pelvic, and spine kinematic measurements possible. This standard test set was performed before the participants were exposed to any repetitive trunk loading tasks. Subjects then were asked to perform a manual materials handling task that required them to depalletize/palletize (transfer) 11 pallets of boxes within a 5-hour period. This repetitive task required the subjects to handle boxes averaging 23 kg at a rate of 125 boxes per hour. This rate of lifting matched the rate that they experienced in their jobs. Before and after transferring *each* pallet load, subjects were asked to perform the standard test set. A brief (2-minute) period of time typically was required to position and instrument the worker on the force plate so that he could perform the standard lift. This provided a brief break period before and after the standard test was performed. Thus, this study documented a standard lift periodically throughout a 5-hour work day of intense materials handling.

The dependent measures in this study consisted of both trunk and hip kinematics and estimated trunk loading characteristics as a function of time into the lifting session. Trunk kinematics variables included trunk angle range of motion, velocity, and acceleration. Hip kinematics included hip angle range of motion, hip velocity, and hip acceleration. The moment or torque imposed about the trunk also was evaluated during the standard test exertion. Spine loading was evaluated in terms of the compression and two shear forces (anterior/posterior and lateral) acting on the spine during the standard exertions.

Spinal loading was evaluated using an electromyography-assisted biomechanical model developed in our laboratory during the past decade.[4,5,7–10,12,15] In general, the model measures electromyographic activity to predict muscle forces that are acting on the spine. Given the trunk geometry (derived from subject anthropometry), the model evaluates instantaneous spinal loading by summing the major muscle groups' force contributions in each direction (compression *vs.* shear). Muscle force is assessed by considering the relative amount of electromyographic activity (percentage of Max) in a muscle and multiplying this value by the cross-sectional area of the muscle and the muscle gain (force per unit area). Muscle force then is modulated by the muscle length–strength relation and the force–velocity relations. In addition, a fatigue modulation factor was used in this study. The electromyographic fatigue modulation factor was included to control for the effects of muscle fatigue on muscle force interpretation. Because modeling was performed under "closed loop" conditions (predicted spinal moments were compared with measured spinal moments), a fatigue adjustment factor was derived for each subject and used to "normalize" the electromyographic signal so the relation with muscle force was established and verified during *each* standard test exertion set. To establish this fatigue modulation factor, the electromyography per unit trunk moment relation was observed during the course of the lifting exertions. The model was iterated to accurately determine the decline in electromyography per unit trunk moment for each set of calibration exertions. Knowing this factor, it was possible to modulate the electromyography-to-force relation used in the model.

The spinal loads estimated in this study were the maximum values of compression force, anterior–posterior shear, and lateral shear forces on the lower back at the lumbosacral joint. The trunk moments included the maximum values of sagittal bending, lateral bending, and axial twisting moments.

Apparatus. An LMM was used to collect the trunk motion data during the standard test. The LMM is essentially an exoskeleton of the spine in the form of a triaxial electrogoniometer that measures instantaneous three-dimensional position, velocity, and acceleration of the trunk.[6] The LMM measures the rotational (not translational) changes in position of the thorax relative to the pelvis and processes position information to derive velocity and acceleration.The lightweight design of the LMM allowed the data to be collected with minimal obstruction to the subject's movements. Hip position, velocity, and acceleration were documented using a custom-made hip goniometer.

Electromyographic activity was monitored through the use of bipolar electrodes spaced approximately 3 cm apart at the 10 major trunk muscle sites. The 10 muscles of interest were the left/right pairs of the erector spinae, latissimus dorsi, internal obliques, external obliques, and rectus abdominis. Ten electrode pairs were placed in the standard locations described in Mirka and Marras.[13] The subject then was placed into a structure that allowed static maximum exertions to be performed in six directions. These maxima were performed to allow all subsequent electromyographic data to be normalized. The six exertions consisted of sagittal flexion, right lateral flexion, left lateral flexion, right twist, and left twist at an upright posture, as well as sagittal extension with the trunk at a 20° forward flexion angle. After each maximum exertion, 2 minutes of rest was provided, in accordance with Caldwell et al.[1]

A force plate (Bertec 4060A, Bertech Corpo, Worthington, OH) and a set of electrogoniometers measured the external loads and moments placed at L5–S1 during the various calibration exertions.

The electrogoniometers measured the relative position of L5–S1 with respect to the center of the force plate, along with the subject's pelvic angle. The forces and moments were translated and rotated from the center of the force plate to L5–S1 in this manner (Fathallah F. Coupled spine motions, spine loading, and risk of occupationally related low back disorders, Unpublished doctoral dissertation, The Ohio State University, 1995).

All signals from the above equipment were collected simultaneously through customized Windows-based software developed in the Biodynamics Laboratory. The signals were collected at 100 Hz and recorded on a 486 portable computer using an analog-to-digital board.

Procedure. During the *standard* exertions, the subject lifted a 23-kg box from a sagittally symmetric position at a slow, smooth pace (controlled by the subject). The lift started at the subject's knee height and ended in his upright position. The *standard* (calibration) lifts were run under "closed-loop" con-

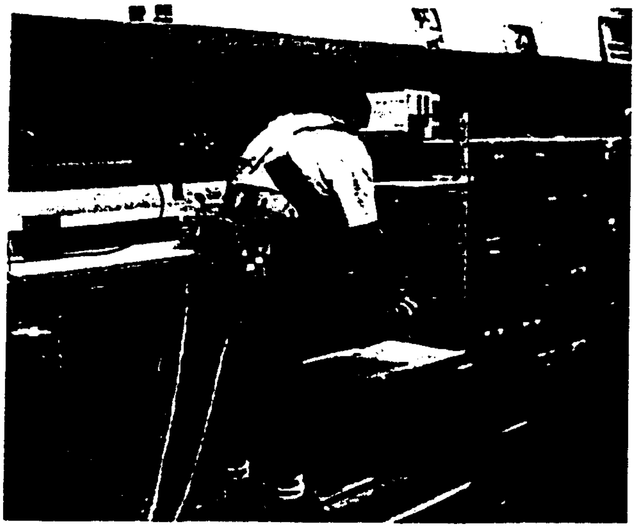

Figure 1. The standard test required the subjects to perform several sagitally symmetric lifting exertions while standing on a force plate. A box of 23 kg was lifted from a platform at knee height while trunk and hip motion and trunk muscle activity were monitored.

ditions; that is, internal moments were validated with measured external moments. Before and after each set of calibrations, neutral trunk position data were collected to document the LMM, force plate, and electrogoniometers values representing an upright, static posture for each subject. During each standard test set, the lift was repeated five times. Figure 1 shows a participant performing the standard task.

Data Analysis. Descriptive statistics (means, standard deviations, minimum and maximum values) were computed for all of the dependent variables. Graphic representations were used to help display various relations. Dependent variables were examined to identify outliers (due to equipment failure), which then were excluded. Analysis of variance statistical analyses then were performed on all the dependent variables. For all significant independent variables, *post hoc* analyses, in the form of Tukey multiple pairwise comparisons, were performed to determine the source of the significant effect(s).

■ Results

Kinematics

Trunk range of motion, velocity, and acceleration characteristics all changed in a statistically significant manner ($P < 0.01$) during the testing period (between standard test sets #1 and #12). The average range of motion exhibited by the subjects in the sagittal plane decreased by more than 6° throughout the test day (Figure 2). This corresponds to a reduction in trunk flexion angle of more than 15% during the course of the testing period. Changes in lateral and twisting range of motion also were found to be statistically significant during the testing period; however, the magnitude of these differences was very small. Average lateral range of motion increased gradually throughout the testing period by approximately 1° during the 12 standard tests, with a peak range of motion occurring during the 10th standard test set (3.9°). Average twisting range of motion decreased by approximately 1.5° among the standard test sets. The peak range of motion in the twisting plane also was observed on the fourth standard test set (3.4°). Even though these differences are statistically significant, they are most likely biomechanically irrelevant.

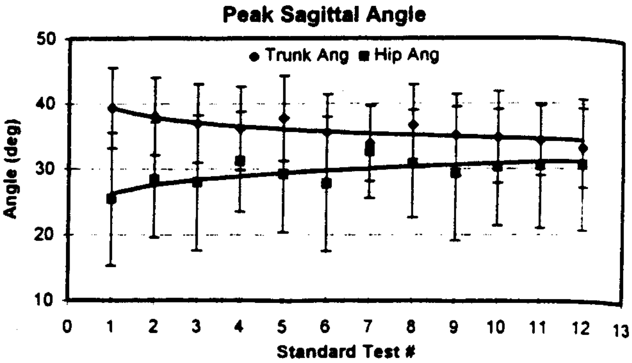

Figure 2. Sagittal range of motion decreased throughout the day. This 15% reduction was offset by an 18% increase in the hip flexion range of motion.

Changes in trunk velocity in the three planes of the body followed a similar trend to that of trunk range of motion. A gradual decline in sagittal plane trunk velocity occurred during the course of the 12 standard test sets. Sagittal plane trunk velocities were in the range of 44–47°/sec during the first five standard test sets and reached velocities of 39–40°/sec during the last three standard test sets. Thus, a maximum average velocity change of approximately 8°/sec was noted among the standard test sets (Figure 3). Velocities within the lateral and transverse planes were fairly small, averaging 5.2°/sec and 2.6°/sec, respectively. Velocity in the lateral plane increased but by less than 1°/sec during the standard test sets. Twisting velocity decreased, but the magnitude of this decrease was less than 1.5°/sec. Therefore, these motion changes in the lateral and twisting planes were judged to be biomechanically insignificant.

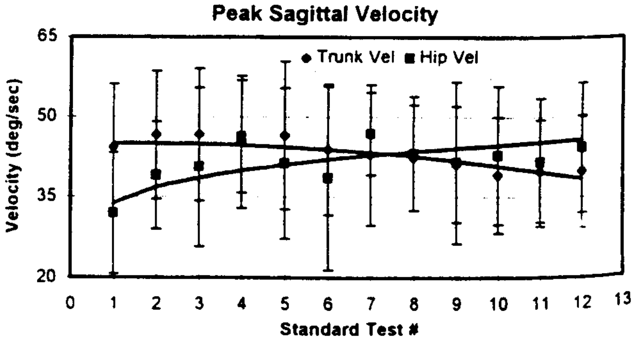

Figure 3. Peak trunk extension velocity decreased approximately 8°/sec. Hip extension velocity increased by a similar amount.

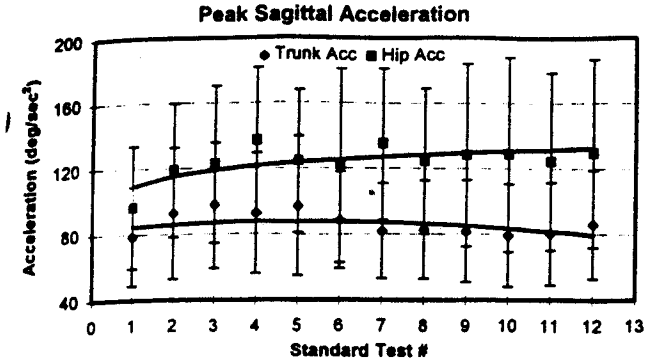

Figure 4. Trunk extension velocity decreased significantly from a peak of 97°/sec^2 in the second and fourth standard tests to a value of 80–86°/sec^2 in the last two standard tests. Hip extension acceleration increased from 96 to 129°/sec^2 throughout the day.

Peak trunk accelerations also were found to change significantly during the 12 standard test sets in each trunk plane. As with trunk range of motion and velocity, however, the only biomechanically interesting trend occurred in the sagittal plane. Sagittal plane accelerations were greatest in the second and fourth standard test sets at approximately 97°/sec^2, as shown in Figure 4. These accelerations then decreased to between 79–86°/sec^2 during the last six standard test sets. Average accelerations in the lateral plane differed only on the third standard test set, where acceleration increased to approximately [illegible]°/sec^2 from a range of 18–23°/sec^2 on the other standard test sets. Twisting accelerations were of even lower magnitude. In general, the twisting accelerations decreased from a high of 15°/sec^2 during the first three standard test sets to a minimum of 8–9°/sec^2 during the 10th and 11th standard test sets.

Hip position, velocity, and acceleration changes were statistically different ($P < 0.01$) during the course of the 12 standard test sets, but only in the sagittal plane of the body. The most significant biomechanical change occurred in hip flexion range of motion. During the course of the 12 standard test sets, range of motion increased from a minimum of 25.4° on the first standard test set to more than 30° on the last standard test set. This indicates an increase of nearly 18% during the course of the standard tests (Figure 2). Similar trends were noted for hip velocity (Figure 3) and acceleration (Figure 4) in the sagittal plane. Hip velocity increased from approximately 32°/sec on the first standard test set to more than 44°/sec on the last test set. Hip acceleration increased from 96°/sec^2 originally to approximately 129°/sec^2 on the last trial.

Trunk Moments

External moments supported by the trunk in the three planes of the body are shown in Figure 5. The moments supported by the trunk in the sagittal plane differed significantly ($P < 0.01$) among the standard test sets. The supported moment (as measured by the force plate system) was observed to decrease from approximately 199 Nm during the first two standard test sets to between 185 and 186.5 Nm on the 11th and 12th standard test sets, respectively. This trend represents an approximately 7% decrease in moment during the standard test sets. It is interesting to note that this decrease in imposed moment corresponded to the decrease in sagittal plane range of motion and the increase in hip sagittal plane range of motion.

Spine Loading

Spine loading was evaluated using the electromyography-assisted model described earlier. The model performed well, with the average R^2 between the predicted and measured moment in the sagittal plane of 0.87. The minimal average R^2 for a trial was 0.82, and there were no statistically significant differences in model performance between model runs as a function of the various standard test sets. Average absolute error in the predicted *versus* measured lifting moment was 21.9 Nm. The average predicted gain was physiologically reasonable (average 28.4 Ncm^{-2}). Both of these parameters did not change in a statistically significant fashion among standard test sets. These results indicate that the estimates of spinal load from the model were reasonable.

Significant changes in spine loading were observed in all three directions of loading as a function of the standard test sets. The trends in compression and anterior/posterior shear are shown as a function of the standard test sets in Figure 6. As indicated in these figures, peak compression significantly decreased by approximately 10% from the first standard test session to the final standard test session; however, both anterior/posterior and lateral shear forces increased during the same period by approximately 35%. This is particularly significant for anterior/posterior shear, where the shear force increased by more than 190 N, thereby increasing the load on the spine to a value close to the shear fatigue strength of the neural arch.[2] The magnitude of the lateral shear forces were biomechanically trivial, which is not unexpected given the sagittally symmetric nature of the standard task.

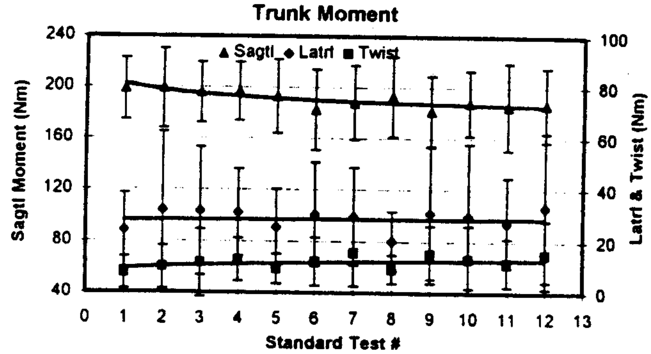

Figure 5. Sagittal lifting moment (approximately L5–S1) decreased 7% throughout the day. Variation in the lateral and twisting moments demonstrated no consistent trends.

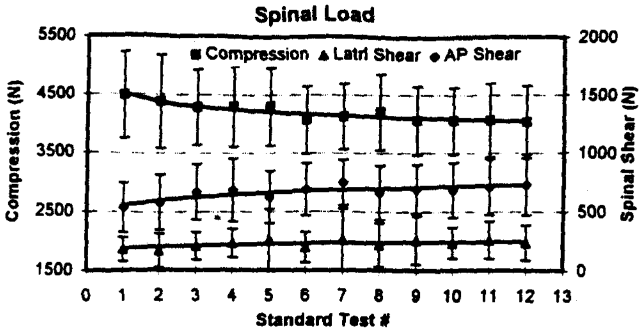

Figure 6. Spinal loads were computed using an electromyography-assisted biomechanical model. Predicted levels of compression decreased approximately 10% throughout the day. Conversely, anterior–posterior shear forces on the lumbosacral junction increased 35%, indicating increased risk of spinal injury.

■ Discussion

This study demonstrated that there are both kinematic and kinetic changes that occur in the body during a prolonged work period when experienced materials handlers are exposed to repetitive lifting exertions. These changes are interrelated in that the changes in trunk and hip kinematics imposed a different muscle recruitment pattern in the trunk muscles, which, in turn, resulted in a change in spine loading patterns.

The majority of the significant kinematic and kinetic changes occurred in the sagittal plane of the body. This was expected because the standard task involved a sagittally symmetric exertion. The most obvious kinematic change involved a trade-off between trunk and hip motion. Trunk range of motion, velocity, and acceleration all decreased during the testing periods, whereas hip range of motion, velocity, and acceleration increased during this same period. These changes also were accompanied by a change in the trunk moments that the subjects exposed themselves to while performing the standard test sets throughout the work session. During the course of the standard test sets, trunk moments decreased by 7%. These changes may be a function of worker experience. Naive subjects may have responded with very different motion patterns of the hips and trunk.

There are two potential explanations for how the trunk moment was reduced during the course of the lifting session. First, the trade-off between trunk and hip motion may have been a mechanism by which the subjects minimized the moment arm between L5–S1 and the load, thereby changing the load path and the subsequent inertial dynamics. Second, the trade-off between trunk and hip motion may have reduced the moment attributed to the trunk mass. The reduction in trunk angle may have reduced the trunk mass that was flexed forward of L5–S1 in the sagittal plane. Thus, instead of bending and increasing the moment imposed on the spine due to the increased distance of the trunk center of gravity relative to L5–S1, subjects elected to bend from the hips and lower the body while in a more upright trunk posture to lift the box.

It is not known why subjects changed their lifting strategy during the course of the standard test sets. One hypothesis may relate to the potential for fatigue occurring during the course of the study. If subjects were experiencing increased muscular fatigue as the lifting bout progressed, then they may have reduced their trunk angle in an attempt to restrict the extensor muscle's (erector spinae) length to a region that maximizes their potential force output (the strongest potion of the length–tension relation). Therefore, subjects may compromise the performance of the large hip extension muscles for an increase in performance of the trunk extensor muscles. Another consideration may be related to subject experience. Many back injuries are expected to occur at the end range of motion (*e.g.*, ligament tears, disc herniation, *etc.*). The experience of the subjects who participated in this study may have prompted the subjects to incorporate a larger margin of safety in their motion pattern as they fatigued. In addition, these subjects may be attempting to increase stability through these kinematic and kinetic changes in behavior.

In this study, no statistically significant increase in lateral moments during the course of the lifting exertions was observed, unlike that noted by Parnianpour et al.[14] This may be a function of differences in the experimental testing procedures. The previous study used a dynamometer that locked the pelvis in place, thereby measuring motion from trunk up. In this study, a sophisticated movement tracking system developed in the authors' laboratory was used that is expected to be far more accurate than a dynamometer system. Thereby, whole body lifting activities were monitored, and the moments imposed on the spine were able to be evaluated more precisely.

The trade-off in trunk and hip motion patterns, along with the change in sagittal plane moment loading, resulted in a change in the loading pattern experienced by the spine. Spine compression was reduced; however, this reduction in compression was achieved at the cost of increased anterior/posterior shear. Anterior/posterior shear (due to load and body mass as well muscle reactions)increased by more than 42% because of the change in trunk/hip motion and the corresponding change in trunk moment. This large increase in anterior/posterior shear may be especially alarming in light of the reduced tolerance to shear observed in the spine.[16] Some believe that that the tolerance to shear for repetitive lifting should be considered to be 1000 N.[12]

To understand how these changes in body kinematics and kinetics result in changes in the loading pattern of the spine, one must consider the role of the muscle recruitment patterns in loading the spine. Trunk muscle activities during the standard testing conditions are shown in Figure 7. These activities are represented in terms of normalized electromyography per unit moment supported by the trunk. A redistribution of muscle activ-

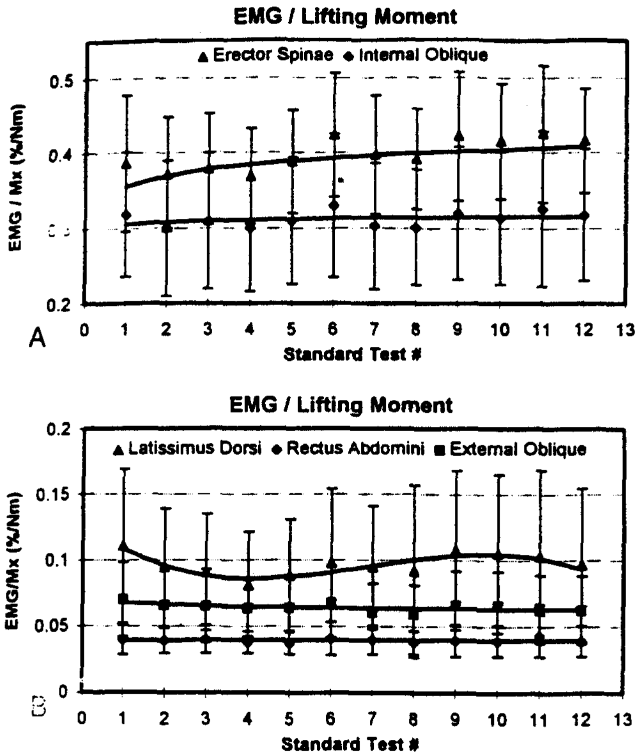

Figure 7. **A**, *Increased erector spinae muscle activity* (normalized electromyography per unit of trunk lifting moment) indicates the useful force generated by these muscles was reduced throughout the day. **B**, Consequently, the lifting moment was transferred from the erector spinae muscles to the more anterior–posterior oriented oblique muscles. As the internal obliques take on a larger share of the exertion, a greater posterior shear load and reduced compressive load were applied to the spine. There were no significant changes in the relative activities of the rectus abdominis, external obliques, or latissimus dorsi.

ity was observed during the course of the standard test sessions that was partially responsible for the change in spine loadings observed. Trunk compression was reduced during the standard test periods in part because the subjects managed to reduce the imposed lifting moment. When the electromyography per unit moment was evaluated during each standard test, however, an 8% increase was observed in the erector spinae muscles. Increased erector spinae behavior indicates the useful activity generated by these muscles was reduced throughout the day, possibly because of fatigue. The change in relative erector spinae activity also coincided with changes in hip and trunk motion throughout the work period. Increases in hip flexion might indicate that the these muscles were lengthened, thus requiring increased activity to account for the length–strength relation change in the muscle. This would also change the muscle angle and the loading vector, thereby contributing more to anterior/posterior shear and less to compression. In addition, because the erector spinae muscle angle increased, the lifting moment was partially transferred from the erector spinae muscles to the more anterior–posterior oriented internal oblique muscles. As the internal obliques took on a larger share of the exertion, a greater posterior shear load was applied to the spine. Therefore, as the test session progressed throughout the day, spinal load was influenced by changes in the external loads as well as changes in muscle coactivation patterns.

The notion that certain muscles would fatigue, thereby causing changes in trunk muscle coactivation patterns, is reasonable, given the evidence in the literature. Trafimow et al[17] showed that when the quadriceps muscle was fatigued, subjects would shift lifting style from a squat lift to a back lift, which requires more external moment support by the trunk and less by the legs. This change in moment support certainly would be expected to change the muscle recruitment pattern and, thereby, the coactivity pattern of the trunk muscles.

There are several limitations of this study. First, as indicated in all the figures, there was significant individual variability associated with all the trends noted. The results reported here simply describe the general trend associated with the study. Therefore, even though the trends observed were statistically significant, individuals might respond differently than indicated by the general trend. Second, as noted earlier, the changes observed in this study may be a function of muscular fatigue. This study was not designed to test for or determine the degree of muscle fatigue because the objective was to determine the effects of repetitive exertions on spine loading. Testing for individual muscle fatigue (*via* spectral analysis) would significantly increase the testing time and therefore interfere with the experimental objectives. It should be noted that the modeled biomechanical loads were assessed by examining the dynamic range of electromyography relative to the applied lifting moment, thereby conditioning the electromyographic-force relation that may be due to fatigue. Therefore, it can be inferred that the changes observed here were due to changes in the muscle recruitment pattern, which *may* be due to general physiologic fatigue. Finally, these results emphasize the significance of muscle coactivity in determining loading of the spine. Thus, models that do not have the fidelity to account for changes in muscle recruitment patterns during repeated exertions should be used with caution when designing workplaces.

These results have several implications for the design of repetitive lifting tasks. First, they indicate that extended lifting sessions may be problematic in that they increase the shear loads to which the spine is exposed. From an applied standpoint, this is particularly important when considering the overtime situations where manual materials handling tasks are involved. It also should be emphasized that these experimental standard tests were all sagittally symmetric. The spine loading patterns expected as a result of the performance of more realistic industrial lifting tasks would involve more asymmetric lifting tasks. This would be expected to further increase the shear loading on the spine. Second, given these findings, work schedules could be designed in such a manner that the detrimental effects of repetitive work (shear loading) can be minimized. Further research is needed, however, to determine how these increases in muscle coactivity and changes in the recruitment patterns might be minimized through the introduction of microbreaks throughout the work day.

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