SPINE Volume 23, Number 6, pp 706–716

An Assessment of Complex Spinal Loads During Dynamic Lifting Tasks

Fadi A. Fathallah, PhD,*† William S. Marras, PhD,* and
Mohamad Parnianpour, PhD*

Study Design. An electromyogram-assisted free-dynamic lifting model was used to quantify the patterns of complex spinal loads in subjects performing various lifting tasks.

Objectives. To assess *in vivo* the three-dimensional complex spinal loading patterns associated with high and low risk lifting conditions that matched those observed in industrial settings.

Summary of Background Data. Combined loading on the spine has been implicated as a major risk factor in occupational low back disorders. However, there is a void in the literature regarding the role of these simultaneously occurring complex spinal loads during manual lifting.

Methods. Eleven male subjects performed symmetric and asymmetric lifting tasks with varying speed and weight. Reactive forces and moments at L5–S1 were determined through the use of electrogoniometers and a force plate. An electromyogram-assisted model provided the continuous patterns of three-dimensional spinal loads under these complex lifting tasks.

Results. The results showed that complex dynamic motions similar to those observed in risky industrial tasks generated substantial levels of combined compressive and shear loads. In addition, higher loading rates were observed under these conditions. Unlike loading magnitudes, loading rate was a better indicator of dynamic loading because it incorporated both the duration and magnitude of net muscle forces contributing to total spinal loading during the lifting conditions.

Conclusions. Quantification of spinal combined motions and loading *in vivo* has not been undertaken. This study provided a unified assessment of the effects of combined or coupled motions and moments in the internal loading of the spine. Dynamic lifting conditions similar to those observed in risky industrial situations generated unique complex patterns of spinal loading, which have been implicated to pose a higher risk to the spinal structure. The higher predicted loading and loading rate during asymmetric lifting conditions can be avoided by appropriate ergonomic workplace modifications. [Key words: complex spinal loading, ergonomics, lifting, low back disorder risk] **Spine 1998;23:706–716**

From the *Biodynamics Laboratory, The Ohio State University, Columbus, Ohio; and †The Liberty Mutual Research Center for Safety and Health in Hopkinton, Massachusetts (currently).
Partial funding has been provided by the Ohio Bureau of Workers' Compensation, Division of Safety and Hygiene, Columbus, Ohio.
Acknowledgment date: January 23, 1997.
First revision date: May 14, 1997.
Acceptance date: July 31, 1997.
Device status category: 1.

Low back disorders (LBDs) in occupational settings have been considered the most significant musculoskeletal disorders in both cost and prevalence.[4,5,27,50,64] Manual materials handling in general and lifting activities in particular have been implicated most often in relation to the risk of occupation-related LBDs.[7,57,58,60,61]

Recent investigations of LBD risk factors in industrial settings have emphasized the role of three-dimensional trunk motion characteristics in conjunction with workplace factors in relation to the risk of LBD.[41,42] In addition, results of a follow-up study of these efforts have shown that elevated combined (complex) dynamic motions of the back were patterns specific to groups with increased LBD risk.[17] However, the effects of these complex motions on the spinal structure *in vivo* are not well understood. These spinal motions are expected to generate complex loading patterns on the spinal elements (*e.g.*, combined lateral shear and compressive loading). Pope et al[49] have expressed the importance of quantifying such complex combined loads; however, they indicate that such a task is difficult to investigate under both field and laboratory conditions. Biomechanical changes in spine tolerance are also expected to occur when these risk factors occur simultaneously. Shirazi-Adl[54–56] has demonstrated how strains in spinal disc anulus fibers are dramatically increased under combined lateral and twisting conditions, reaching levels that may exceed the tissue tolerance limits. The findings in these latter studies as well as those in other studies *in vitro* have implicated combined loading of the spinal structure as a mechanism for back injury.[2,8,22,26] Therefore, for prevention of LBD, it is essential to quantify the types and magnitudes of the three-dimensional mechanical loading experienced by the spinal structure when subjected to dynamic combined or complex motions. Knowledge of this type of information allows identification in detail of the situations that compromise the integrity of the structure and thereby helps reduce injuries resulting from these conditions. However, there is a void in the literature about the role of these simultaneously occurring complex spinal loads during manual lifting. In most biomechanical studies in which spinal loads have been investigated, these loads have been quantified in terms of univariate statistics (*e.g.*, average or maximum loading) without assessing their temporal occurrences in conjunction with their magnitudes.

Therefore, the main objective of this study was to describe quantitatively the magnitudes and patterns of the simultaneous three-dimensional spinal loading occurring under complex lifting conditions that are similar to those expected in industrial lifting situations.

Methods

Study Design. Eleven healthy male subjects volunteered to participate in this experiment. Mean ± SD age was 28.4 ± 4.4 years, mean stature and weight were 180.7 ± 3.7 cm and 78.6 ± 10.8 kg, respectively. A questionnaire was administered to each subject to ensure that there was no history of serious back disorders, and to screen subjects with current back-related discomforts.

The experiment consisted of a three-way within-subject design. The independent variables included lift type, speed of lift (duration of movement), and weight handled. These variables were chosen to replicate complex motion conditions under varying workplace parameters similar to those observed in the industrial settings.[41,42] The lift types investigated were symmetric lifts and asymmetric lifts. The sagittally symmetric lifts were chosen to replicate closely the kinematic conditions observed in the low-risk groups of a companion industrial study.[17,42] Whereas, the asymmetric lifts were designed to generate multiplanar complex trunk motions that were more unique to the medium and high risk groups (risk groups were defined on the basis of historical LBD-related incidence rates).[42] Lift duration (speed) was set at three levels: low (2 seconds per lift), medium (1.5 seconds per lift) and high (1 seconds per lift). These lifting rates were chosen to replicate dynamic motion patterns that corresponded to motion patterns of the three occupational LBD risk groups: low, medium, and high risk.[17,42] Three weight levels were considered: low (22 N), medium (67 N), and high (156 N). The weight levels were determined on the basis of the distribution of weights observed in industrial tasks.[41,42] The low weight level corresponded to a value between the 25th and the 50th percentile, the medium level between the 50th and 75th percentile, and the high level between the 75th and 100th percentile of the weight distribution. These weights were also chosen to match the moment conditions of the risk database so that the patterns of three-dimensional spinal loading during complex trunk motions could be investigated.

The dependent variables consisted of the three-dimensional spinal loading patterns (magnitudes and rates) at L5–S1 in terms of the compressive, anteroposterior shear, and mediolateral shear forces.

Apparatus and Modeling. Three-dimensional continuous position, velocity, and acceleration of the trunk were determined using the lumbar motion monitor (LMM).[41,42] Three-dimensional external forces and moments around L5–S1 were monitored by the combination of a Bertec 4060A force plate (Bertec, Worthington, OH), and two electrogoniometers were used to determine the continuous location and orientation of the joint at L5–S1 in three-dimensional space. One monitor, the L5–S1 locator, provided the three-dimensional distance vector between the force plate center and the center of the joint at L5–S1 (three-dimensional locator), and the second monitor, the pelvic orientation monitor, provided the orientation of the joint at L5–S1 with respect to the pelvis.[16,18] This information was necessary to correct for three-dimensional motions of the trunk relative to the force plate. This system provided a mechanism to monitor torque or moment around L5–S1 during a lift. The full description of the method and its validity and reliability was detailed elsewhere.[16]

An electromyogram system collected signals from 10 pairs of bipolar silver–silver chloride surface electrodes affixed over the specific locations of 10 muscles of interest. The electrode locations for the corresponding muscles were: 1) right and left latissimus dorsi: most lateral portion of the muscle at T9; 2) right and left erector spinae: at L3, 4 cm from the spinal midline; 3) right and left rectus abdominis: 2 cm from the umbilicus level, 3 cm from the abdomen midline; 4) right and left external oblique: 10 cm from the abdomen midline at 45°, 4 cm above the iliac crest; and 5) right and left internal oblique: 3 cm above the posterior superior iliac spine, 10–12 cm from the spinal midline at a 45° angle (lumbar triangle).[37,40]

The electromyogram signals were first amplified and low-pass filtered (1 kHz). The filtered signals were hardware rectified and processed by a 20-msec moving average window (integration constant). An asymmetric reference frame[37,40] was used to solicit static maximum voluntary contractions (MVCs) in six directions (flexion, extension, left and right twist, and left and right lateral bending). These exertions were used to normalize the electromyogram signals.

An electromyogram-assisted model provided estimates of the internal moments required to achieve the balanced equilibrium conditions.[23–25,35,38,39,43] The model assumes that to achieve dynamic equilibrium during a lifting task, the external moments generated around L5–S1 must be balanced by moments generated internally by the musculature of the body. Previous efforts were only capable of assessing spinal loads while the pelvis was restrained in a fixture (*e.g.*, Granata and Marras[24]). The current study extended the modeling so that it could predict and validate spine loading during whole-body, free-dynamic, complex lifting conditions, thereby eliminating the need to restrain the pelvis to a fixture.[16] It should be noted that this electromyogram-assisted modeling approach provides estimates of the total loads generated at the center of the joint at L5–S1.

During the experiment, the subject lifted a wooden box that was filled with the weight associated with the prescribed condition. The box was 30.5 cm × 30.5 × 23 cm with two handles (3.8 cm in diameter and 11.4 cm in length) centered at its sides. For each condition, the subject was provided with an auditory signal (loud tone) indicating the lift pace by identifying the start and end of each lift phase. This tone was necessary to control the lift duration (speed). All the analog signals gathered from the devices described earlier were collected at 100 Hz through a 12-bit, 32-channel analog-to-digital (A–D) converter connected to a 386-based microcomputer. Figure 1 shows a subject performing a symmetric lift.

Experimental Protocol. Pairs of surface electrodes were applied to their respective sites using standard preparation procedures.[36] The subject was provided with written instructions detailing various parts of the experiment. The subject first performed the six static MVCs in random order. A 2-minute rest period was allowed between two exertions.[9] After all six MVCs were collected, the subject was ready to perform the dynamic lifting tasks. The subject was first fitted with the Lumbar Motion Monitor along with the proper attachments of the three-dimensional locator and the pelvic orientation monitor. The subject was permitted time to familiarize himself with the tasks, and to ask pertinent questions. Each task consisted of two lifts and one lowering, with tones indicating the start and end of each part.

In the symmetric condition, the box (weight) was placed on a platform in front of the subject, just above knee height and at a horizontal distance equal to his arm length. At the onset of the tone, the subject was asked to lift the box from its location to a position as close as possible to the body while maintaining straight arms (this puts the vertical distance of the center of load at approximately 20 cm below the iliac crest level). For the complex (asymmetric) conditions, the box was placed in front of the subject in the same manner as in the symmetric condition; however, in this case the subject was asked to set the box down on another platform placed to his right at an angle perpendicular to the midsagittal plane. The platform height was set at the level of the subject's iliac crest and was placed at approximately an arm's length distance horizontally. For all trials, the subject was instructed to maintain the arms and legs as straight as possible (*i.e.*, to perform a stoop lift). Within a given type of lift (symmetric or asymmetric), the three weights and speeds were presented in random order to control for carryover effects. To avoid the potential of fatigue, the subject was given at least a 60-second rest between exertions. The subjects were instructed to take an additional rest period whenever they so desired.

Statistical Analysis. The performance of an electromyogram-assisted lifting model was assessed by three measures: 1) R^2 statistic, 2) average absolute error, and 3) muscle gain. R^2 represents the percentage of variance in the actual (measured) moments explained by the predicted moments; therefore, providing a measure of trend agreement between the two moments. The average absolute error reflects the magnitude difference between the measured and predicted moments. The muscle gain parameter relates to how the muscles force production capabilities per unit area (in Newtons per square centimeter) match existing anatomic data. For each trial, gain is determined by comparing muscle-generated (predicted) moments with measured, applied moments at L5–S1. Physiologically valid values for gain range between 30 and 100 N/cm^2.[45,51,65] The measured extension moment generated around L5–S1 compared with the extension moment predicted by the electromyogram-assisted model during a symmetric lift is shown in Figure 2. The association between these moments is indicated by R^2, which in this instance was 0.98 with an average absolute error of 14.6 Nm and gain of 75 N/cm^2.

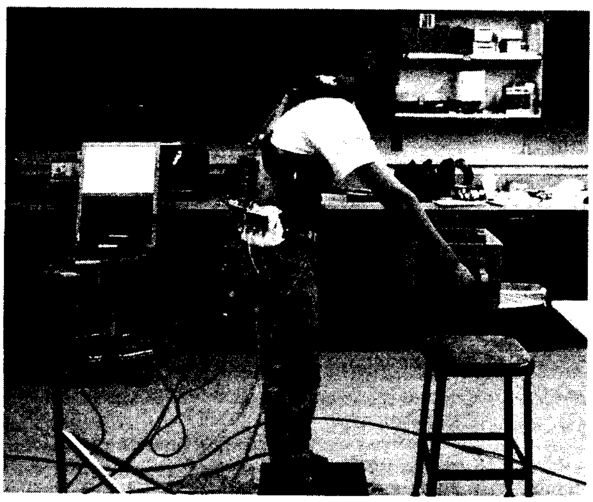


Figure 1. A subject shown performing a symmetric lift.

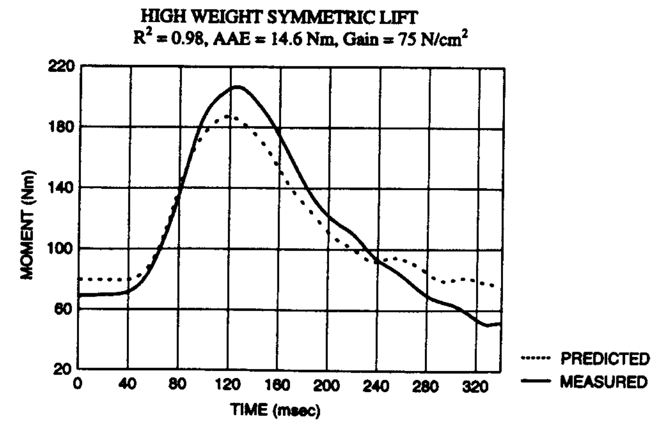


Figure 2. Measured extension moment *versus* the predicted moment of a typical symmetric lift.

To investigate the effects of weight, speed and lift type (symmetric *vs.* asymmetric) on the internal loading of the spine, multivariate analysis of variance (MANOVA) was conducted. The internal spinal loading was characterized by average maximum compression, anterior shear forces, and lateral shear forces acting on the lumbar spine. The simultaneous occurrence of combined loading was quantified by bivariate (two-dimensional) distributions of compression and anterior shear forces and compression and lateral shear forces. Note that only the lifting portions of the tasks were considered in all analyses in this study. The bivariate distributions were statistically compared among conditions, using the two-dimensional Kolmogorov–Smirnov test.[15] This test facilitated the comparison among various conditions with respect to the extent of simultaneously occurring forces on the lumbar spine. Because the experimental exertions were truly dynamic, rate of loading was also quantified. This parameter has been shown to be a sensitive measure for reflecting the effect of dynamic loads on spinal tissue tolerance *in vitro*.[68] Loading rate was estimated by differentiating the compression and shear forces with respect to time (in Newtons per second). For each condition, the average maximum loading rate for all subjects was determined within each of the three spinal loading directions (compression, anterior, and lateral shear). Multivariate analysis of variance (MANOVA) was performed to determine the effect of load, lifting rate, and asymmetry on the magnitude and rate of net joint reaction forces (compression, lateral, and anteroposterior shear forces).

Table 1. Multivariate Analysis of Variance (MANOVA) and Analysis of Variance (ANOVA) for Maximum Spinal Loading: Compression (COMP), Lateral Shear (LATSHR), and Anterior Shear (ANTSHR)—Type I Error Probabilities Shown Only for Significant Effects

		ANOVA		
Condition	MANOVA	COMP	LATSHR	ANTSHR
Lift type (L)	< 0.001	< 0.001	< 0.001	< 0.001
Weight (W)	< 0.001	< 0.001	< 0.001	NS
Speed (S)	NS	NS	NS	NS
L × W	< 0.001	< 0.001	< 0.001	NS
L × S	NS	NS	NS	NS
W × S	NS	NS	NS	NS
L × W × S	NS	NS	NS	NS

NS = not significant.

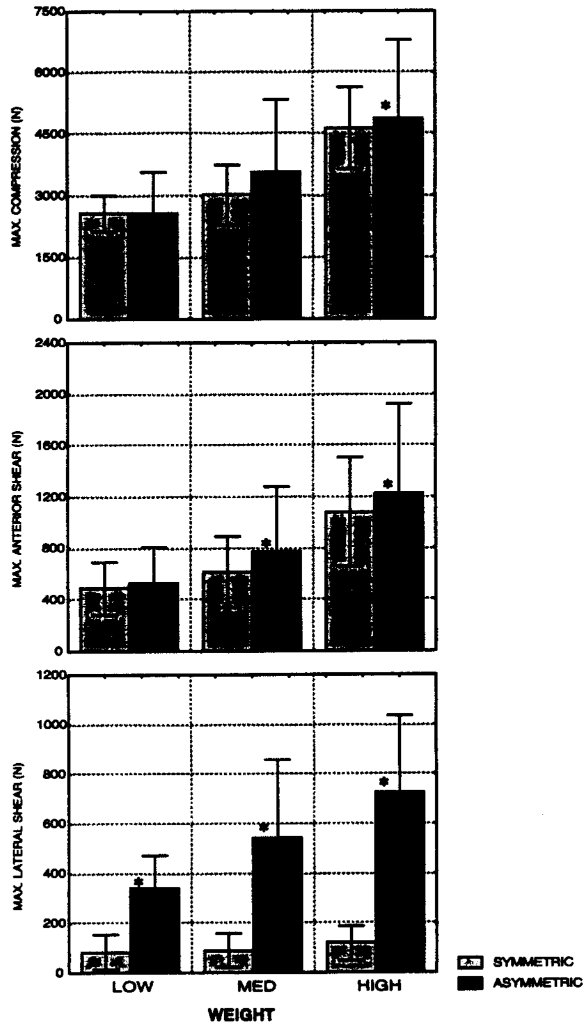
■ Results

Extension moments for all conditions and all subjects (198 total trials) were predicted with an average R^2 of 0.82 ± 0.23, with 86% of those trials above 0.5 (0.7 correlation) and 41% above 0.9 (0.95 correlation). The average absolute error of the predicted moments were 23 ± 19 Nm and 21 ± 12 Nm, for the sagittal and lateral directions, respectively. Subject gain values averaged to include all exertions were 74.14 ± 22.8 N/cm^2, which falls within the physiologically acceptable range of 30–100 N/cm^2. Thus, the model was able to predict external trunk moments reasonably and was therefore assumed to predict spinal loads reasonably.

The results of the MANOVA for the magnitudes of maximum three-dimensional loading showed a significant interactive effect between weight and lift type ($P <$ 0.0001; Table 1). This interaction was significant for both compression and lateral shear ($P < 0.0001$), but not for anterior shear. In Figure 3, the nature of this interaction is depicted. With the exception of lateral shear under symmetric lifts, maximum spinal loading significantly increased with increases in the box weight ($P < 0.001$). For the low weight condition, only lateral shear produced a significant difference between the two lifting types. With increase in the lift weight, the anterior shear was significantly different between the lifting types; and compression was significantly different only for the medium weight condition (Newman–Keuls; $P < 0.05$; Figure 3).

To explore further the nature of combined (complex) loading conditions on the spine, the simultaneously combined compression and anterior shear, and combined compression and lateral shear were investigated under each weight lift type combination. The combined compression and anterior shear observed for all subjects for each of the weight levels for both the symmetric and asymmetric lifts are shown in Figure 4. For each weight level, the bivariate (compression–anterior shear) distribution under the asymmetric condition was statistically different from the symmetric condition (two-dimensional Kolmogorov–Smirnov; $P < 0.01$). When compared with their respective symmetric lifts, the asymmetric lifts had a pronounced shift in their distributions toward the regions of increased combined compression and anterior shear. For example, the high weight asymmetric lift condition had approximately 22% of the total data observed under combined levels of more than 4000 N of compression and 400 N of anterior shear. In contrast, the respective high weight symmetric distribution had approximately 13.5% of the total data observed under these combined loading levels.

The bivariate distributions of compression and lateral shear observed in all subjects for each of the weight levels for both the symmetric and asymmetric lifts are depicted in Figure 5. As shown earlier, the magnitude of spinal



* Symmetric versus asymmetric, significant at p < 0.05

Figure 3. Average maximum loading ± standard deviation in all spinal loading directions under each weight level for the symmetric and asymmetric lifts. Note that the high weight asymmetric condition is considered a high risk situation, whereas the low weight symmetric condition would be considered a low risk situation.

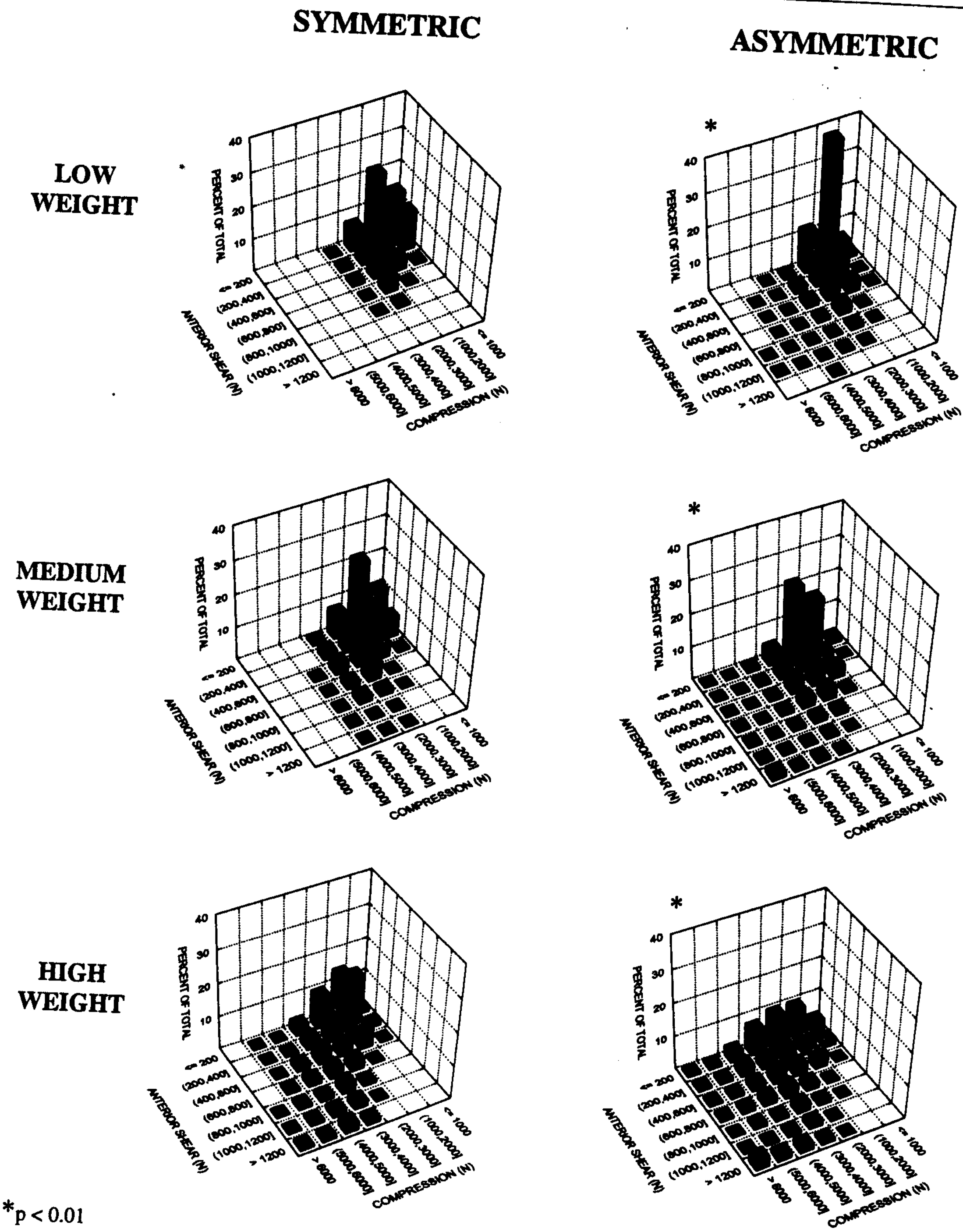


Figure 4. Bivariate compression and anterior shear distributions for each weight and lift type combination.

loading significantly increased with the increase in weight; however, for each weight level, the patterns of combined loading differed between the symmetric and asymmetric lifts. Similar to the combined compression and anterior shear test, the asymmetric conditions had substantial shift in their distributions toward areas of simultaneously occurring high magnitudes of compression and lateral shear. However, under symmetric lifts, the bivariate distributions showed that lateral shear was maintained at rather low levels throughout the range of compressive loading. For all three weights, more than 90% of the time the lateral shear was maintained at less than 200 N during symmetric lifts (Figure 5). In contrast, during asymmetric lifts, lateral shear was observed at increased levels for a considerable portion of the time. For instance, for the high weight asymmetric lifts, approximately 40% of the total data were observed at lateral shear levels exceeding 200 N with some instances

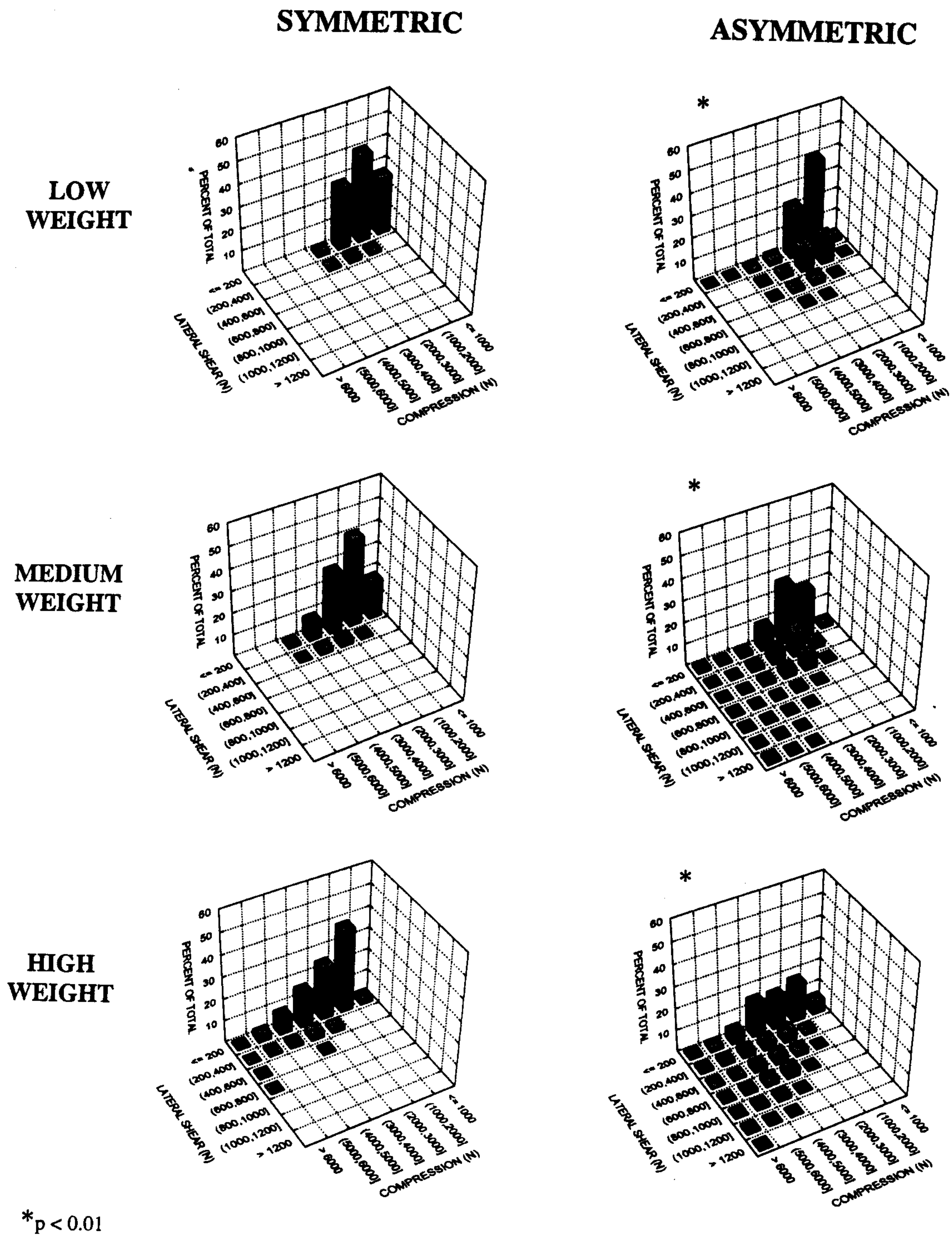


Figure 5. Bivariate compression and lateral shear distributions for each weight and lift combinations.

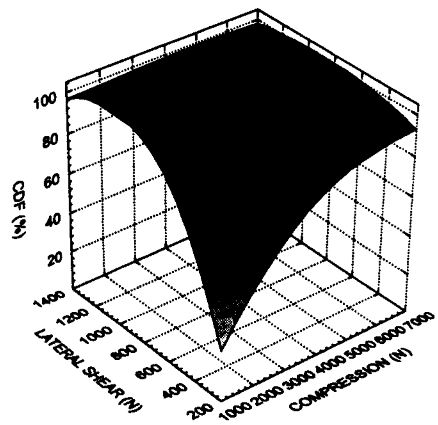
reaching higher than 1200 N, combined with substantial magnitudes of compressive loading. This observation is further explored in Figure 6, in which the bivariate cumulative distribution functions of the high weight symmetric and asymmetric lifts are depicted. In this figure, the presence of combined loading is indicated at elevated magnitudes during asymmetric lifts. It should be noted, as in the case of combined compression and anterior shear, with each weight level the difference between the symmetric and asymmetric lifts was statistically significant (two-dimensional Kolmogorov–Smirnov; $P < 0.01$).

The MANOVA of spinal load magnitudes indicated that there were no statistically significant effects related to the speed of lift (Table 1). However, the results of the MANOVA for the loading rates showed a significant main effect caused by speed and an interactive effect between speed and lift type (complex loading; $P < 0.001$;

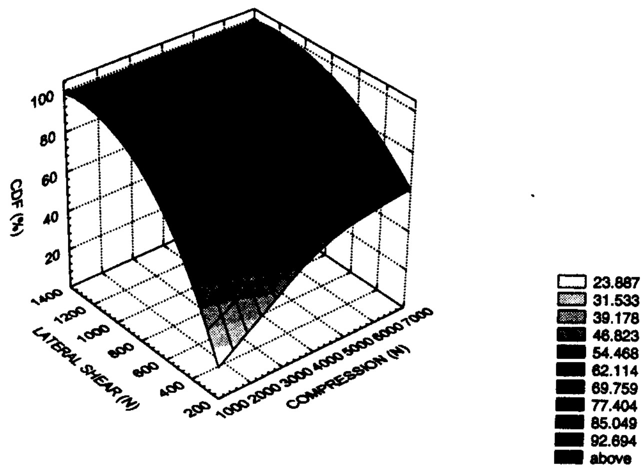
Table 2). In addition, there was a significant interaction between lift type and the weight handled ($P < 0.001$) as a function of lift speed (Figure 7). With the exception of lateral shear under symmetric conditions, within each weight level the loading rates significantly increased with speed ($P < 0.05$). In many instances, the loading rates under the high speed conditions were almost double those observed under the low speed conditions. The loading rates increased with weight of the load, with distinct patterns observed between the two lifting types. Within each weight level, loading rates under the asymmetric (complex) lifts were almost consistently higher than those of the symmetric lifts. This was especially evident under the high weight condition for compression and lateral shear.

■ Discussion

Results of previous epidemiologic studies have implicated trunk combined or simultaneously occurring motions and loading as potential risk factors for occupation-related low back disorders and have emphasized the importance of quantifying three-dimensional factors to understand better the loads on the spine during work.[4,5,29,33,48] However, before this research, the levels of simultaneously occurring loads that may become problematic were not well understood. The results of this study have provided estimates of how the combined three-dimensional loads act on the lumbar spine during various lifting conditions. Although the compressive forces reported in these results were within the range reported in results of several studies that shared similar experimental conditions,[1,10,12,14,19–21,28,32] the three-dimensional loading experienced in the spinal structures has been quantified in very few dynamic lifting studies. In most studies, only the compressive loads are reported in the findings, without quantification of the shear forces. In this study, complex three-dimensional loading was assessed quantitatively *in vivo*. The study is unique because it is designed to account not only for the effects of dynamic motion variables and inertial forces on the net



(a) Symmetric lifts; high weight



(b) Asymmetric lifts; high weight

Figure 6. Compression and lateral shear bivariate cumulative distribution functions for the high weight symmetric (a) and asymmetric (b) lifts.

Table 2. MANOVA and ANOVA for Maximum Spinal Loading Rates: Rate of Compression Rate (R_COMP), Rate of Lateral Shear (R_LATSHR), and Rate of Anterior Shear (R_ANTSHR)—Type I Error Probabilities Shown Only for Significant Effects

		ANOVA		
Condition	MANOVA	R_COMP	R_LATSHR	R_ANTSHR
Lift type (L)	< 0.001	< 0.001	< 0.001	NS
Weight (W)	< 0.001	< 0.001	< 0.001	< 0.001
Speed (S)	< 0.001	< 0.001	< 0.001	< 0.001
L × W	< 0.001	< 0.001	< 0.001	NS
L × S	< 0.001	NS	< 0.01	NS
W × S	NS	NS	NS	NS
L × W × S	NS	NS	NS	NS

NS = not significant.

external moment, but also the effects on the internal loading and muscular recruitment parameters. In most three-dimensional biomechanical models (*e.g.*, the inverse dynamics approach) the external moments are computed. However, electromyogram-assisted models are the only class of models that can incorporate the intersubject and intertrial variability on the internal loading of the spine. In results of prior studies,[25,37,40] substantial influence of the trunk velocity and acceleration profiles on the recruitment strategies solicited by the nervous system under similar external moment conditions have been observed. Therefore, this added sensitivity of the model allows the projection of the complete effects of workplace factors and dynamic motion profiles on the spinal loads and their rates. Ultimately, this approach allows the identification of loading patterns that

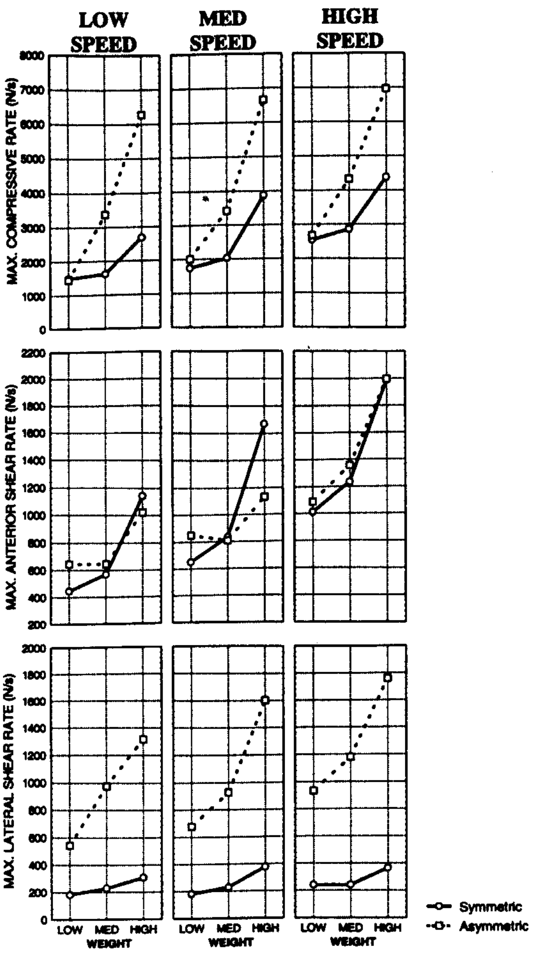


Figure 7. Average maximum loading rate in all spinal loading directions under each weight–speed combination for the symmetric and asymmetric lifts. Note that the high weight–high speed asymmetric condition is considered a high risk situation, whereas the low weight–low speed symmetric condition would be considered a low risk situation.

expose the spinal structures to loading levels that may exceed their tolerance limits and could lead to injury. These situations could be identified and abated through proper workplace design. In sum, the level of detail provided by the current approach allows determination of the magnitudes and temporal occurrences of combined loading and assessment of the sensitivity of these loading patterns to workplace characteristics.

Combined Loading

The complex lifting conditions (asymmetric dynamic lifts) facilitated further understanding of the unique patterns of complex motions observed in groups of elevated risk in industrial settings.[17] Asymmetric lifting conditions in general, and particularly those that had a combination of high weight and high lifting speed (velocity) exhibited pronounced shifts in their combined compressive and shear forces distributions. The shift in the distributions was always observed toward elevated magnitudes of combined loading (compression–anterior shear; compression–lateral shear). This indicates that, for a given weight lifted, the manner by which the worker performs the lifting task can have a dramatic effect on the nature of the loading patterns on the spine. This may be a key element in explaining the mounting epidemiologic evidence that implicates asymmetric lifting in LBDs. Unlike symmetric lifts, during asymmetric lifting, the person is subjected to substantial levels of simultaneously occurring compressive and shear forces for a good portion of the lift. These types of loads may well surpass the spinal tissue tolerance and could lead to injury. There is enough empirical evidence to support this hypothesis.

Shirazi–Adl et al,[52] using finite element modeling, found that the disc anulus was not vulnerable under pure compressive loading. Duncan and Ahmed[13] have shown that with an intact facet joint, axial rotation does not impose unusual stress on the structure. However, simultaneous loading in multiple directions (combined loading) may constitute a key factor in explaining the failure mechanism of spinal structures. The results of Gunzburg et al's[26] *in vivo* experiment indicated that the axial range of motion of the lumbar functional units decreased with flexion, suggesting that the facets restricted the axial rotation. This was hypothesized to be that the bracing action of muscular activities caused more load to be carried by the facets. The altered muscular activation during complex loading[18,40] may alter the internal load sharing within the elements of the functional spinal units,[31,47] and therefore, may augment the risk of injury.[53–55] Shirazi–Adl et al[53] have shown that when lateral bending and twisting occur simultaneously, the disc fiber strain increases markedly. Shirazi–Adl[54] has also shown that when axial torque is combined with compression, the disc fibers located at the posterolateral and posterior locations of the spine become more vulnerable when combined with bending moments. In addition, in reporting results of a more recent study, Shirazi–Adl[55] noted that when lateral bending and twisting occur during lifting, facet compressive and shear contact loads increase significantly. This may increase the risk of facet injury and degeneration.

Therefore, when performing industrial tasks that involve complex motions, such as those generated during asymmetric lifting, it is suspected that the worker may be at an increase risk of back injury. Quantifying and reducing the percentage of time the worker is exposed to combined loading may provide an effective means of minimizing the risk of back injury during lifting activities.

Loading Rate

Assessing load rate *in vivo* could further improve the assessment of combined loading by providing additional insight into the effects of dynamic factors on the complex patterns of spinal loading. Loading rate in the lateral direction combined with compressive loading rate seemed to provide a key element in distinguishing between the symmetric and the complex asymmetric lifts.

For example, under low weight conditions, both the compressive and anterior shear rates were comparable between the symmetric and asymmetric lifts. However, the lateral shear loading rate was substantially elevated in the asymmetric conditions. Furthermore, when the load increased, compressive loading rates were significantly increased for the asymmetric lifts in combination with the elevated lateral shear loading rates. Thus, shear loading rate increases much more with asymmetric lifts and is increased greatly with greater weight of the lift. This may explain why the complex patterns of simultaneously occurring lateral and twisting velocities observed in industrial tasks were specific to medium and high risk groups.[17] These complex dynamic motions of the trunk seem to generate substantial combined loading rates (and magnitudes), which pose greater risk to the spinal structures. It is the combination of spinal force magnitudes and their temporal patterns (*i.e.*, loading rates) that may provide the distinguishing factor in explaining the risk imposed on the spine during manual work. However, comparing loading levels to spinal tissue tolerance obtained under simple, nondynamic *in vitro* conditions may not be suitable (especially in that very few risky conditions in this study exceeded these simple tissue tolerances). There seems to be enough empirical evidence to support such a notion.[11,46,59,66,67,69]

The response of biomaterials in general and components of the spinal structure in particular is time dependent. This time dependency has been described in terms of viscoelastic (flow-independent) or poroelastic (flow-dependent) constitutive laws.[3,6,30,34] Therefore, given such realization, it is essential to quantify both the magnitude and the rate of loading to perform a proper risk assessment. Stiffness of viscoelastic materials increases at higher rates of loading, which in turn may affect the failure mechanism of the material. The load-sharing patterns within the passive elements of the spine is significantly affected by the magnitudes, combination, and rates of the external loads. This complex behavior is believed to be caused by the heterogeneity of materials, nonlinear material properties, geometric nonlinearity (caused by large deformations), and time-dependent material properties. Therefore, it has been suggested that the mechanisms of failure under different loading conditions (magnitudes and rates) would be different.

The viscoelastic response of the nucleus and anulus fibrosus may significantly affect the stress–strain distribution within the passive elements. Using a viscoelastic finite element model (FEM) of the L2–L3 motion segment, Wang et al[62,63] have indicated that higher stresses in anulus fibers were experienced at the fastest loading rates during complex (asymmetric) loading. Their results indicated that the highest stress in fibers was obtained at lower fiber strain, and lower overall deformation of the motion segment at the higher loading rate. In other words, the ultimate stress can be exceeded at faster loading conditions even under lower angular deformation. This suggests that risk assessment based on pure angular position can lead to erroneous conclusions. The current study results, coupled with the results in FEM studies of Wang et al,[62,63] indicate the need for new experimental studies to investigate tissue tolerance under dynamic, complex loading conditions. More complete appreciation of the current study and its results can be gained once such a database is developed, whereas the specifications for future *in vitro* studies can be drawn from the results of this experimental study as well as those of industrial surveillance studies.[41,42,17]

Note that this experimental study was focused on assessing complex spinal loading observed during dynamic lifting tasks similar to those observed in a companion field study.[17] In that study, the role of complex trunk motions in relation to the risk of occupational LBD risk was examined.

Limitations

This study had several limitations that should be enumerated. The model used was mainly driven or assisted by the electromyogram signals of different trunk muscles. Therefore, the quality of these signals could greatly affect the performance of the model. The investigator insured proper preparation and application technique of the electrodes, along with a systematic and frequent quality check on all collected electromyogram signals. In addition, the role of passive tissues was not directly examined in the current model. However, because the subjects' maximum sagittal flexion did not exceed 45°, the contribution of passive tissues is expected to be minimal.[44] The model did not include all muscles that may contribute to internal moments. Instead, only the major load-producing muscles were included. Given the generally good agreement between the predicted and measured moments, this assumption seems to hold well. The study, for practical reasons, was limited in the scope of the conditions and types of lifts assessed. Further studies could expand on this study and include a wider range of situations. Lastly, this study was focused on assessing the effects of physical factors on spinal loading patterns; other factors, such as psychosocial factors, may alter or affect these patterns. Further research is needed to examine these potential factors.

■ Conclusions

Biomechanical and epidemiologic literature has identified complex motion and loading of the spine as subjecting the structure to potentially risky conditions. Acknowledging such potential risk, this study was designed to explore, in detail, the variation in complex three-dimensional spinal loading during various dynamic lifting conditions. More specifically, the results of this study have shown that:

- Investigating only one component of the net resultant spinal forces (*i.e.*, compression) and ignoring the other components may lead to erroneous judgment regarding the risk associated with manual materials handling activities.

• The complex dynamic lifting conditions similar to those observed in industrial risky situations generated complex patterns of spinal loading. The combination of compressive and lateral shear loading patterns (magnitudes and rates) seemed to be a key in distinguishing these conditions that may put the spinal structure at risk.

• There is a need to investigate further the tolerance of spinal tissue in dynamic, complex loading conditions.

• This approach provides a better understanding of the mechanical loading imposed on the spine under realistic biomechanical conditions. Future research should be focused on coupling this information with biologic responses to complex loading and, ultimately, should assess the risk of the development of low back disorders.

Acknowledgments

The authors thank Dr. Kevin Granata for his invaluable assistance in the experimental phase of this study.

References

1. Adams MA, Dolan P. Recent advances in lumbar spinal mechanics and their clinical significance. Clin Biomech 1995; 10:3–19.
2. Adams MA, Hutton WC. The relevance of torsion to the mechanical derangement of the lumbar spine. Spine 1981;6: 241–248.
3. Adams, MA, Dolan P. Time dependent changes in the lumbar spine resistance to bending. Clin Biomech 1996;11:194–200.
4. Andersson GB. Epidemiologic aspects of low back pain in industry. Spine 1981;6:53–60.
5. Andersson GB. The epidemiology of spinal disorders. In: Frymoyer JW, et. The Adult Spine: Principles and Practice. New York:Raven Press, 1991:107–46.
6. Best BA, Guilak F, Setton LA, et al. Compressive mechanical properties of the human anulus fibrosus and their relationship to biochemical composition. Spine 1994;19:212–21.
7. Bigos S, Spengler D, Martin N, et al. Back injuries in industry: a retrospective study. II: Injury factors. Spine 1986;11:246–51.
8. Broberg K. On the mechanical behavior of intervertebral discs. Spine 1981;8:151–65.
9. Caldwell LS, Chaffin DB, Dukes–Dobos FN, et al. A proposed standard procedure for static muscle strength testing. American Industrial Hygiene Association Journal 1974;35: 201–12.
10. Cholewicki J, McGill SM, Norman RW. Comparison of muscle forces and joint load from an optimization and EMG assisted lumbar spine model: Toward development of a hybrid approach. J Biomech 1995;28:321–31.
11. Danto MI, Woo SLY. The mechanical properties of skeletally mature rabbit anterior cruciate ligament and patellar tendon over a range of strain rates. J Orth Res 1993;11:58–67.
12. Dolan P, Earley M, Adams MA. Bending and compressive stresses acting on the lumbar spine during lifting activities. J Biomechan 1994;27:1237–48.
13. Duncan NA, Ahmed AM. The role of axial rotation in the etiology of unilateral disc prolapse: An experimental and finite-element analysis. Spine 1991;16:1089–98.
14. Ekholm J, Arborelius U, Nemeth G. The load on the lumbo-sacral joint and trunk muscle activity during lifting. Ergonomics 1982;25:146–61.
15. Fasano G, Franceschini A. A multidimensional version of the Kolmogorov–Smirnov test. Monthly Notes of the Royal Astronomical Society 1987;255:155–70.
16. Fathallah, FA, Marras, WS, Parnianpour, M, Granata, KP. A method for measuring external spinal loads during unconstrained free-dynamic lifting. J Biomech 1997;30:975–8.
17. Fathallah FA, Marras WS, Parnianpour M. The role of complex simultaneous trunk motions in the risk of occupationally-related low back disorders. Spine (in press).
18. Fathallah FA. Coupled spine motions, spine loading and risk of occupational related low back disorders. Ohio State University, Columbus, Ohio, 1995. Dissertation.
19. Freivalds A, Chaffin D, Garg A, Lee KS. A dynamic evaluation of lifting maximum acceptable loads. J Biomech 1984;17: 252–62.
20. Frigo C. Three-dimensional model for studying the dynamic loads on the spine during lifting. Clin Biomechan 1990; 5:143–52.
21. Garg A, Chaffin DB, Freivalds A. Biomechanical stress from manual load lifting a static vs dynamic evaluation. IIE Trans 1982;14:272–81.
22. Gordon SJ, Kink HY, Mayer PJ, et al. Mechanism of disc rupture. A preliminary report. Spine 1991;16:450–6.
23. Granata KP, Marras WS. An EMG-assisted biodynamic lifting model simulating spinal loads during asymmetric trunk extensions. J Biomech 1993;26:1429–38.
24. Granata KP, Marras WS. An EMG-assisted model of trunk loading during free-dynamic lifting. J Biomech 1995;28:1309–17.
25. Granata KP, Marras WS. The influence of trunk muscle coactivity on dynamic spinal loads. Spine 1995;20:913–9.
26. Gunzburg R, Hutton WC, Fraser R. Axial rotation of the lumbar spine and the effect of flexion: An in vitro and in vivo biomechanical study. Spine 1991;16:22–8.
27. Hollbrook TL, Grazier K, Kelsey JL, Stauffer RN. The frequency of occurrence, impact and cost of selected musculoskeletal conditions in the United States. Proceedings of the American Academy of Orthopeadic Surgeons Chicago, Illinois, 1984. 24–25, 1984.
28. Jager M, Luttman A. Biomechanical analysis and assessment of lumbar tress during load lifting using a dynamic 19-segment human model. Ergonomics 1989;32:93–112.
29. Kelsey JL, Githens PB, White AA, et al. An epidemiological study of lifting and twisting on the job and risk for acute prolapsed lumbar intervertebral disc. J Orthop Res. 1984;2:61–6.
30. Laible JP, Pflaster DS. A poroelastic-swelling finite element model with application to intervertebral disc. Spine 1993;18: 659–70.
31. Lin HS, Liu YK, Adams KH. Mechanical response of the lumbar intervertebral joint under physiological (complex) loading. J Bone Joint Surg [Am] 1978;60:41–55.
32. Liskenen T, Stalhammar H, Kuorkina I, Troup J. The effect of inertial factors on spinal stress when lifting. Engineering in Medicine 1983;12:87–9.
33. Magora A. Investigation of the relation between low back pain and occupation: 4. Physical requirements: Bending, rota-

tion, reaching and sudden maximal effort. Scand J Rehabil Med 1973;5:186–90.

34. Maki AF. The apparent viscoelastic behavior of articular cartilage—The contribution from the intrinsic matrix of viscoelasticity and interstitial fluid flows. J Biomech Eng 1986; 108:123–30.

35. Marras WS, Reilly CH. Networks of internal trunk-loading activities under controlled trunk-motion conditions. Spine 1988;13:661–7.

36. Marras WS. Guidelines: Industrial electromyography (EMG). Int J Ind Ergonomics 1990;6:89–93.

37. Marras WS Mirka GA. Muscle activities during asymmetric trunk angular accelerations. J Orthop Res 1990;8:824–32.

38. Marras WS, Sommerich CM. A three-dimensional motion model of loads on the lumbar spine: I. Model structure. Hum Factors 1991;32:123–37.

39. Marras WS, Sommerich CM. A three-dimensional motion model of loads on the lumbar spine: II. Model validation. Hum Factors 1991;32:139–49.

40. Marras WS, Mirka GA. A comprehensive evaluation of trunk response to asymmetric trunk motion. Spine 1992;17: 318–326.

41. Marras WS, Lavender SA, Leurgans SE, et al. The role of dynamic three-dimensional trunk motion in occupational related low back disorders: The effects of workplace factors, trunk position and trunk motion characteristics on risk of injury. Spine 1993;18:617–28.

42. Marras WS, Lavender SA, Leurgans SE, et al. Biomechanical risk factors for occupationally related low back disorders. Ergonomics 1995;38:377–410.

43. Marras WS, Granata KP. A biomechanical assessment and model of axial twisting in the thoraco-lumbar spine. Spine 1995;20:1440–51.

44. McGill SM, Norman RW. Partitioning of the L4–L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine 1986;11:666–78.

45. McGill, SM, Norman, RW. Effects of an anatomically detailed erector spinae model on L4–S1 disc compression and shear. J Biomech 1987;20:591–600.

46. Newman P, Keller TS, Ekstrom L, Hansson T. Effects of strain rate and bone mineral on the structural properties of the human anterior longitudinal ligament. Spine 1994;19:205–11.

47. Parsad P, King AI. An experimentally validated dynamic model of the spine. J Appl Mech 1974;41:546–50.

48. Pope MH. Risk indicators in low back pain. Ann Med 1989;21:387–92.

49. Pope MH, Andersson GBJ, Frymoyer, JW, Chaffin DB. Occupational Low Back Pain: Assessment, Treatment, and Prevention. Chicago: Mosby Year Book

50. Praemer A, Furner S, Rice, DP. Musculoskeletal conditions in the United States. Park Ridge, IL: American Academy of Orthopeadic Surgeons, 1992:23–33.

51. Reid, JG, Costigan, PA. Trunk muscle balance and muscular force. Spine 1987;12:783–6.

52. Shirazi-Adl A, Shrivastava SC, Ahmed AM. Stress analysis of the lumbar disc-body unit in compression: A three-dimensional nonlinear finite element study. Spine 1984;9:914–27.

53. Shirazi-Adl A, Ahmed AM, Shrivastava SC. Mechanical response of a lumbar segment in axial torque alone and combined with compression. Spine 1986;11:914–27.

54. Shirazi-Adl A. Strain in fibers of a lumbar disc: Analysis of the role of lifting in producing disc prolapse. Spine 1989;14: 96–103.

55. Shirazi-Adl A. Finite-element evaluation of contact loads on facets of an L2–L3 lumbar segment in complex loads. Spine 1991;16:533–41.

56. Shirazi-Adl A. Biomechanics of the lumbar spine in sagittal/lateral moments. Spine 1994;19:914–27.

57. Snook SH. Low back pain in industry. In: White AA, Gordon SL, eds. Symposium on Idiopathic Low Back Pain. St. Louis: CV Mosby, 1982.

58. Spengler DM, Bigos SJ, Martin BA, Zeh J, Lloyd F, Nachemson A. Back injuries in industry: A retrospective study. I. Overview and cost analysis. Spine 1986;11:241–5.

59. Tran NT, Watson NA, Tencer AF, Ching RP, Anderson PA. Mechanism of burst fracture in the thoracolumbar spine. Spine 1995;20:1984–8.

60. U.S. Department of Labor. Back injuries—nation's number one workplace safety problem. Fact sheet No. OSHA 89-09, Washington, DC: Occupational Safety and Health Administration, 1989.

61. Videman T, Nurminen M, Troup TDG. Lumbar spine pathology in cadaveric material in relation to history of back pain, occupation, and physical loading. Spine 1990;15:728–40.

62. Wang JL, Shirazi-Adl A, Engin AE, Li S, Parnianpour M. A model study of the effect of rate loading on the internal stress and strain of an L2–L3 lumbar motion segment. 1995 Advances in Bioengineering 1995;31:19–20.

63. Wang JL, Shirazi-Adl A, Engin AE, Li S, Parnianpour M. The viscoelastic model development of anulus fiber for a cumulative trauma model of low back injury. 1995 Advances in Bioengineering 1995;31:21–2.

64. Webster BS, Snook SH. The cost of 1989 workers' compensation low back pain claims. Spine 1994;19:1111–6.

65. Weis-Fogh, T, Alexander, RM. The sustained power output from striated muscle. Scale effects in locomotion. London: Academic Press, 1977:511–25.

66. Wright T, Hayes W. Tensile testing of bone over a wide range of strain rate: Effect of strain rate, microstructure and density. Med Biol Eng 1976;14:671–9.

67. Yahia LH, Audet J, Drouin G. Rheological properties of the human lumbar spine ligaments. J Biomed Eng 1991;13: 399–406.

68. Yingling VR, Callaghan JP, McGill SM. The effect of load rate on the mechanical properties of porcine spinal motion segments. Proceedings of the 19th Annual Meeting of the American Society of Biomechanics; Palo Alto, California, October, 1995. Palo Alto: American Society of Biomechanics, 1995.

69. Yogandandan N, Pintar F, Ray G, Sances A Jr. Stiffness and strain energy criteria to evaluate the threshold of injury to an intervertebral joint. J Biomech 1989;22:135–42.

Address reprint requests to

Fadi Fathallah, PhD
Liberty Mutual Research Center for Safety & Health
71 Frankland Road
Hopkinton, MA 01748