Relation between spinal load factors and the high-risk probability of occupational low-back disorder

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Spinal compression is traditionally assumed the principal biomechanical mechanism associated with occupationally related low-back disorders (LBD). However, there is little conclusive evidence demonstrating that compression is related to occupational LBD. The objective of this research was to examine whether axial compression in the lumbar spine can predict the probability that a lifting task should be classified as high risk for LBD. Furthermore, the improvement in predictive ability was examined when analyses include 3-D, dynamic biomechanical factors. Ten experienced warehouse workers transferred 12 pallet loads of boxes in a simulation of warehouse working conditions. Biomechanical estimates of 2-D static and 3-D dynamic spinal compression, shear loads and tissue strains were achieved from the subjects during each lifting exertion. Each lift was also assessed for probability of high LBD risk classification. Regression analyses were performed to examine the relationship between biomechanical and epidemiological factors. Results indicate 2-D static estimates of spinal compression describe $\sim 13\%$ of the probability of high LBD risk variability. Dynamic estimates of spinal compression describe $> 44\%$ of the variability. A multifactor regression model including 3-D spinal loads and tissue strains further improved the predictive ability, but the improvement was not statistically significant. This research demonstrates the biomechanical source of low-back pain is dynamic, multifaceted and multidimensional. Significant improvements in ergonomics assessments can be achieved by examining interactions of dynamic biomechanical factors. Unfortunately, this improved predictive ability is generated at the high cost of computational complexity. However, less realistic biomechanical representations may ignore the injury mechanisms associated with the greater number of workplace injuries. Thus, improved understanding of the dynamic biomechanical interactions influencing the tolerance and injury mechanisms of the spine may permit more accurate assessments of workplace injury factors associated with LBD and reduced incidence of occupationally related low-back pain.

1. Introduction

Spinal compression is traditionally assumed the principal biomechanical mechanism associated with occupationally related low-back disorders (LBD). One of the criteria that the NIOSH lifting guide uses to discriminate between safe and hazardous tasks is based on static estimates of compressive loads on the spine (NIOSH 1981). Consequently, research examining the risk of low-back pain often focuses on axial compressive loads associated with occupational tasks (Schultz and Andersson 1981,

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Freivalds *et al.* 1984). However, epidemiologic studies indicate that other factors, including repetitive twisting or lateral bending and lifting, even with relatively light loads, are significant risk factor for LBD (Kelsey *et al.* 1984, Punnet *et al.* 1991). These findings suggest that spinal shear and torsional loads associated with asymmetric lifting postures may be under-appreciated. Similarly, based on the high correlation between motion dynamics of a task and the risk of LBD (Marras *et al.* 1993), spinal and biomechanical load dynamics may be associated with the mechanism of injury. Nonetheless, traditional ergonomics analyses regarding occupationally related low-back pain focuses predominantly on static estimates of spinal compression. Although assessments of LBD risk have traditionally focused predominantly on axial compression, there is evidence that the biomechanics of low-back pain is significantly more complex than can be represented by static estimates of spinal compression.

Epidemiologic research, examining the association between spinal compression and occupationally related low-back pain, has been unable to demonstrate a strong causative relation between the two. Chaffin and Park (1973) are often referenced when stating the assumption between compression and LBD. However, the authors concluded that LBD was related to 'lifting strength index'. The causative nature of spinal compression on LBD was not demonstrated, but it was hypothesized as a possible mechanism. Herrin *et al.* (1986) examined static estimates of spinal compression as a function of injury rate in 55 industries. Results indicate that tasks associated with compressions > 6500 N had twice the injury rates than tasks with compression < 4500 N. However, injury rates for tasks associated with compression levels between 4500 and 6500 N were nine times greater than tasks with compression > 6500 N. Waters *et al.* (1993) indicate there may have been a typographical error in the representation of the $4500 - 6500$ N data; however, Herrin *et al.* (1986) conclude static estimates of spinal compression account for $< 2\%$ of the injury rate variability, i.e. correlation $r = 0.13$. The NIOSH Revised Lifting Equation (Waters *et al.* 1993) assumes that compression is a causative factor in occupational LBD, citing the work of Chaffin and Park (1973), Herrin *et al.* (1986) and Anderson *et al.* (1985). The latter work developed a static model of spinal load during lifting tasks. No data were provided to demonstrate that compression was related to LBD. Instead the authors concluded a typical lifting exertion may generate spinal loads greater than the NIOSH (1981) recommended guidelines. On the other hand, Punnet *et al.* (1991) concluded that $< 3\%$ of the injuries occurring in an automotive assembly plant demonstrated estimated levels of spinal compression greater than the 3400 N action limit recommended by the NIOSH lifting guide (1981). Hence, there is little conclusive evidence demonstrating that compression is related to occupational LBD. Leamon (1994) states that 'in most cases we simply do not know what causes low back pain disability', indicating the assumption relating occupationally related LBD with spinal compression may be premature. The biomechanical causes of LBD may be highly complex and dynamic, with spinal compression accounting for a small part of the true relationship.

Biomechanical literature indicates that injury solely due to spinal compression is unlikely. Brinkman (1986) demonstrated that compressive loads applied to *in vitro* lumbar vertebra failed to produce clinically relevant injuries unless pre-existing endplate damage was present. Adams *et al.* (1987) stated that compression in the lumbar spine, in the absence of forward bending moments, cannot 'injure the soft tissue without first causing gross damage to the vertebrae'. Thus, tissue failure and

associated LBD is more likely generated by combinations of multidimensional spinal loads. In a theoretical analysis, Shirazi-Adl (1989) found that bending moments and shear forces combined with axial compression significantly increased the risk of injury to the lumbar disc. Yingling *et al*. (1995) demonstrated load rate influences the ultimate strength, stiffness, displacement and failure mechanisms. Clearly, occupationally related low-back pain associated with vertebral tissue injury is unlikely caused by static compression alone.

The objective of this study was to identify some of the multidimensional, dynamic biomechanical factors associated with LBD risk, and to determine whether spinal compression might be a major contributor to injury. Specifically, answers to two questions were sought. Does axial compression in the lumbar spine correlate well with the risk of LBD associated with a prescribed task? Does the inclusion of 3-D, dynamic biomechanical factors more accurately describe the probability of risk than static estimates of compression alone? Identifying possible biomechanical parameters capable of discriminating between safe and hazardous tasks may contribute to the development of more accurate and robust ergonomics analyses and reduced incidence of LBD in the workplace. Therefore, the objective of this study was to examine the relationship between predicted biomechanical load factors on the spine and the probability of high LBD risk.

2. Methods

Ten experienced male warehouse order selectors, 19–49 years of age, were recruited from a local food distribution centre. The weight and stature of the subjects was 80.1 ± 8.4 kg (176 lb) and 180.3 ± 7.1 cm (71 inches) respectively. The subjects' experience as warehouse selectors ranged from 0.25 to 23 years.

To simulate realistic warehouse working conditions, subjects were required to lift boxes ranging from 18.2 kg (40 lb) to 27.3 kg (60 lb) from one pallet to another until the entire pallet load, an average of 47 boxes, was transferred. Twelve pallets of boxes were moved at a frequency of 166 lifts per h, simulating a 'slow' 5-h work day. Dynamic, 3-D trunk motion data were collected using the Lumbar Motion Monitor (LMM) (Marras *et al*. 1993), and integrated myoelectric (EMG) activity of 10 trunk muscles were collected from bipolar surface electrodes (Mirka and Marras 1983) during the depalletizing tasks. Prior to beginning each pallet, a set of 'test/ calibration' exertions were performed while the subjects were standing on a force plate (Bertec 4060A) and with added electrogoniometers to measure the location and orientation of the lumbosacral spine relative to the centre of a force plate. The test exertions were designed to permit data quality assurance and supply calibration data for biomechanical analyses.

Each lifting task was assigned a probability of being classified as high risk for occupationally related LBD. Average high-risk tasks represented industrial jobs with reported LBD incidence rates of 26 injuries per 200 kh. The assessment was achieved from a multiple logistic regression model of dynamic trunk motion parameters and workplace factors (Marras *et al*. 1993). This epidemiologic model was developed from a database developed from on-site measurements over 400 industrial workers, and incorporated factors including the lifting moment, lift rate, multidimensional trunk range of motion and velocities. As subjects' lifted boxes from each region of each pallet, a measure of risk was assigned to that task, i.e. pallet region, by the epidemiologic model, and saved for comparison with a variety of biomechanical parameters.

An EMG-assisted biomechanical model was employed to determine the dynamic spinal loads associated with the lifting exertions (Granata and Marras 1995). The analysis incorporated normalized EMG data, anthropometrically scaled muscle cross-sectional areas and vector directions as well as force-length and force-velocity relations to determine the forces supplied by 10 dynamically co-contracting muscles. Three-dimensional spinal loads were determined from the vector sum of the muscle forces, and trunk moments from the sum of vector products of muscle forces and moment arms. Direct comparison of dynamic trunk moments determined from the force plate data collected during the test exertions, with predicted trunk moments determined from the biomechanical model during those exertions provided subject dependent calibrations and model validation parameters. Model output included peak spinal loads, the rate of change of spinal load, i.e. load rate, posterior tissue strain, e.g. relative length of the posterior ligaments, and posterior tissue strain rate. Modelled strain and strain rate data included representations of the dynamic behaviour of the posterior longitudinal ligament, interspinous/supraspinous ligament equivalent, right and left ligamentum flavum, and right and left intertransverse ligaments (Kirking 1996). Considering the modelled strains and strain rates were nearly identical for the posterior longitudinal ligament and the interspinous/supraspinous ligament, only the interspinous/supraspinous values were reported. Similarly, the intertransverse ligament values represent themselves and the ligamentum flavum. The biomechanical model used to determine these factors is described elsewhere (Granata and Marras 1993, 1995, Marras and Granata 1995, 1996).

For comparison with the dynamic biomechanical data, static compressive forces were computed for each task using the static 2-D method outlined in Chaffin and Andersson (1984). External moments were determined from the product of box weight and moment arm distance from the trunk measured during each lifting exertion. Upper body mass and centre of mass were determined from subject anthropometry and multiplicative coefficients cited in Chaffin and Andersson (1984). The restorative force generated by a single extensor muscle and spinal compression was determined from the muscle moment arm and total trunk moment. Thus, static estimates of spinal compression were determined for comparison with probability of LBD risk and dynamic biomechanical values.

Multiple linear regression analyses were used to examine the association between the probability of a high-risk classification and spine biomechanics factors. Correlations were achieved for all possible models of combined biomechanical factors, including regression models of individual factors. The association between biomechanical parameters and the probability of high-risk classification were evaluated based on the correlation coefficient (r^2). Z-test analyses of the correlation coefficients were performed to identify statistical differences in predictive performance.

3. Results

The correlation between the probability of high risk represented in the simulated warehouse tasks and independent biomechanical factors are presented in tables 1 and 2. Static estimates of compression represented the poorest individual correlation with risk, $r^2 = 0.137$. Results indicate that dynamic spinal compression and ligamentous strain represented the strongest individual correlations with probability of high risk at $r^2 = 0.443 - 0.474$. Dynamic estimates of compression, load rates in all

three dimensions, and strain in the two reported ligaments were significantly ($p<0.05$) better at predicting changes in risk than static estimates of compression, dynamic estimates of shear forces and ligamentous strain rates.

Figure 1 illustrates the probability of high-risk classification as a function of static estimates of spinal compression. As indicated by the scatter of points, the ability to predict a risk probability from a value of statically modelled spinal compression was limited, $r^2 = 0.135$. It is interesting to note that tasks associated with low spinal compression were not necessarily associated with low probability of high-risk classification. However, high spinal compression tended to preclude a low probability of risks.

Dynamic estimates of spinal compression demonstrate a similar discrimination of low probability of risk (figure 2). Thus, epidemiologic model results associated with high spinal compression tasks were always $>50\%$ probability of high-risk classification. Low compression tasks demonstrated both low and high probability of high risk. However, dynamic estimates of spinal compression demonstrated significantly less scatter of the data points ($r^2 = 0.441$) than demonstrated by the static compression (figure 1). Plots of risk probability versus ligamentous strain looked similar to the dynamic compression results.

Combining multidimensional factors and dynamic parameters significantly improved the predictive ability of biomechanical factors to identify the probability of high-risk classification. Table 3 identifies the two best regression models containing two, three and four biomechanical factors each. Also included are the

Table 1. Correlation coefficients (r^2) between peak spinal loads and the probability of a high-risk classification. Dynamic estimates of spinal compression were the strongest individual correlation with probability of risk. Statistical analyses indicate dynamically computed compression, and all load rates predicted the probability of high risk significantly better than shear loads or static estimates of compression.

	Lateral shear	AP shear	Compression
Static load	–	–	0.135
Dynamic load	0.191	0.195	0.441
Load rate	0.343	0.345	0.428

r^2 differing by > 0.09 represent statistically different model performance.

Table 2. Correlation coefficients (r^2) between individual biomechanical factors and the probability of a high-risk classification. Although strain in the modelled ligaments correlated with the probability of high risk better than spinal compression, the differences were not statistically significant at $p<0.05$.

	InterTransv	SuperSpin
Strain	0.477	0.474
Strain rate	0.004	0.002

r^2 differing by > 0.09 represent statistically different model performance.

best regression models containing a spinal compression term. The strongest two-parameter model included modelled estimates of lateral shear load rate and strain in the interspinous/supraspinous ligaments. No biomechanical multiple regression
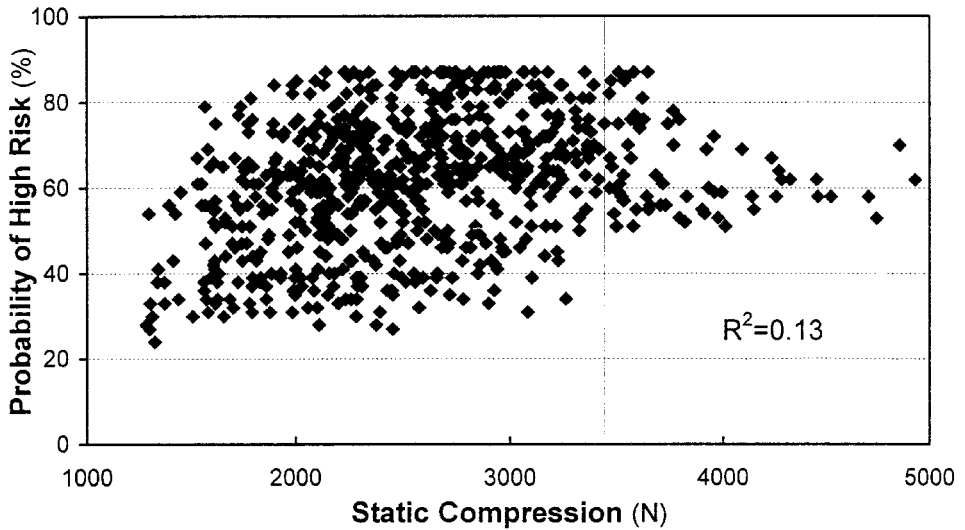

Figure 1. Probability of high-risk classification versus 2-D static estimates of spinal compression. Vertical line identifies the 3400 N compression level corresponding to the 1981 NIOSH AL. Statistical results indicate 2-D static estimates of compression account for < 13.5% of the probability of high-risk variability.
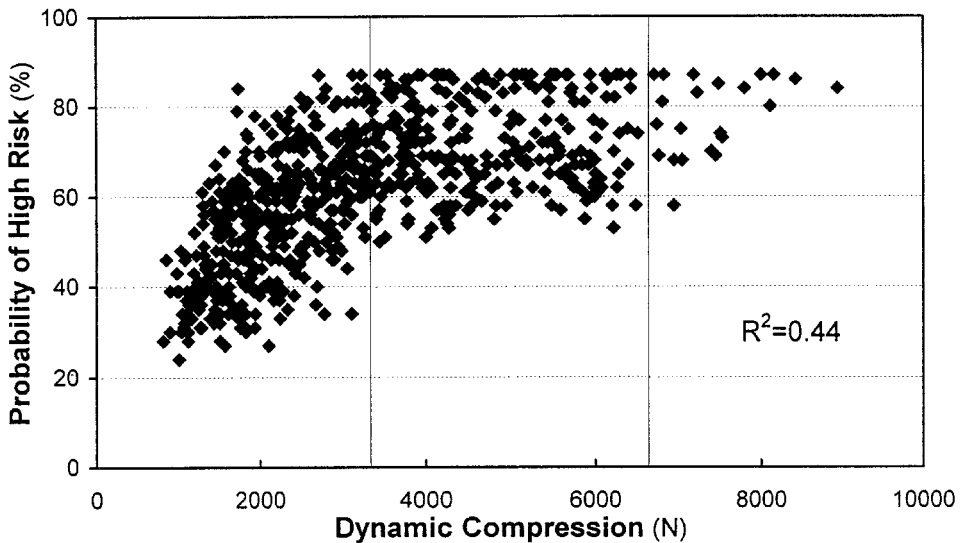

Figure 2. Probability of high-risk classification versus dynamic estimates of spinal compression. Vertical line identifies the 3400 and 6400 N compression level corresponding to the 1981 NIOSH AL and MPL. Statistical results indicate dynamic estimates of compression account for 44% of the probability of high-risk variability.

model predicted the probability of high-risk classification statistically better than any other. However, including multiple biomechanical parameters as identified in table 3 demonstrated significantly improved predictive ability over single factor models (tables 1 and 2). The only exception was the two factor models including lateral shear force and compressive load on the spine. Although they performed significantly better than most single factor models, the r^2 was statistically similar to the univariate

Table 3. Correlation (r^2) between combinations of biomechanical factors and task probability of high risk for LBD. Only the three best regression models in each category are provided.

Regression models	r^2
Two-factor models	
$S_I + F_Z$	0.514
$S_I + F_X$	0.542
$S_S + LR_X$	0.546
Three-factor models	
$F_X + F_Z + S_I$	0.547
$S_I + S_S + F_X$	0.561
$S_S + SR_I + LR_X$	0.562
Four-factor models	
$F_X + F_Y + F_Z + S_I$	0.557
$S_I + S_S + F_X + LR_X$	0.567
$S_I + S_S + F_X + SR_I$	0.573

F_X, lateral shear force; F_Y, AP shear force; F_Z, compressive force; S_I, InterTransv lig. strain; S_S, SuperSpin lig. strain; LR_X, lateral load rate; LR_Y, AP load rate; LR_Z, Compr. load rate; SR_I, InterTransv strain rate; SR_S, SuperSpin strain rate.
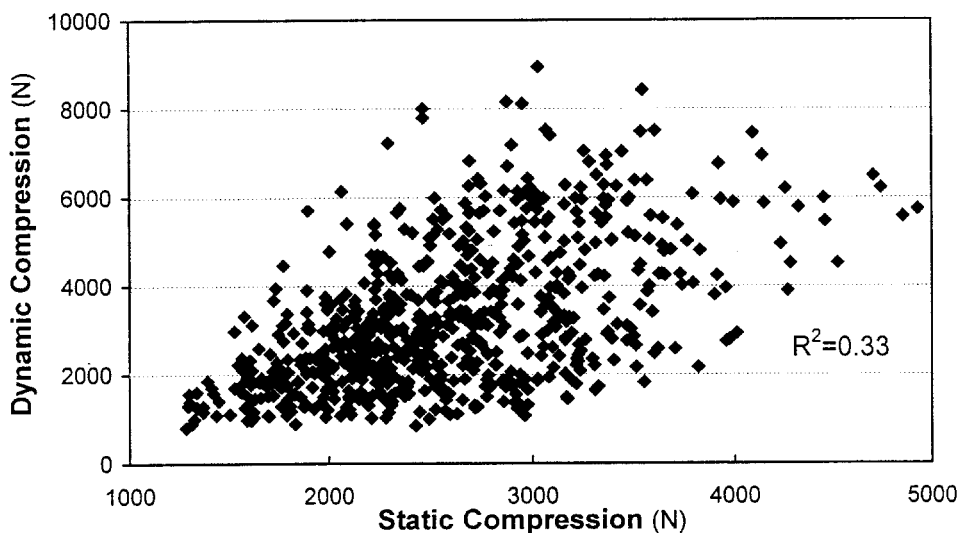

Figure 3. Static estimates of compression versus dynamic estimates of compression. Statistical results indicate 2-D static estimates of spinal compression account for approximately one-third ($r^2 = 0.331$) of the dynamic compression variability.

ligament strains. Predicted probability of risk is best represented by a combination of dynamic spinal loads, tissue strains and either load rates or strain rates. Combinations of biomechanical factors with static compression failed to generate the levels of correlation demonstrated by equivalent models with dynamic spinal load.

4. Discussion

Analyses of LBD risk have traditionally focused on static compressive loads in the spine. However, epidemiologic and biomechanical data suggest dynamic and shear loading parameters may be related to LBD risk (Freivalds *et al.* 1984, Marras and Sommerich 1991a, b, Granata and Marras 1993, 1995, Marras *et al.* 1993, 1995, Marras and Granata 1995, 1996). Similarly, tissue strain, strain rates and torsional loads on the spine may contribute to spinal injury and back pain (Adams *et al.* 1987, Shirazi-Adl 1989, Adams and Dolan 1996). Thus, examination of dynamic, and multidimensional spine biomechanics factors may improve the ability to identify hazardous occupational tasks.

Axial compressive loads on the spine, determined from a fundamental static biomechanical model, were significantly ($p < 0.001$) lower and poorer predictors of the probability of high-risk classification than dynamic spinal compression. The average static compression, 2700 N, was 20% lower than the average dynamic value of 3400 N. Freivalds *et al.* (1984) and McGill and Norman (1985) similarly indicated static analyses under predict dynamic spinal compression by 20–40%. As a single variable, dynamic compression correlated with the results from the epidemiologic risk model at $r^2 = 0.441$, much better than the statically determined compression, $r^2 = 0.137$. This demonstrates the static and dynamic determination of spinal load are largely unrelated, the correlation between the two resulting in an $r^2 = 0.31$. Thus the static and dynamic estimates of spinal compression represent somewhat different aspects of the exertion, as illustrated by the scatter of points when plotting the dynamic estimates of compression versus the static values (figure 3). It is interesting to note that the static estimates of spinal compression in this study described > 13% of the predicted risk variability, compared with the results of Herrin *et al.* (1986) who found only 2% ($r = 0.13$) of the variability was described by static estimates of spinal compression. The differences may be related to the nature of the static compression models. Since the static determination of compression fails to account for the variability associated with lifting dynamics, it becomes difficult for this variable, or combinations including it, to identify relative changes in predicted probability of high risk.

Static and dynamic estimates of spinal compression were able to discriminate between tasks with low probability of risk from those tasks without low probability of risk. Thus, using either static or dynamic estimates of spinal compression, tasks associated with compressions > 3400 N never demonstrated a probability of high-risk classification < 50%. High spinal compressions precluded low probability of high-risk classification. However, a low spinal compression did not preclude high probability of risk. Very high probabilities could be found in association with tasks generating very low compression levels. Hence, compression estimates were able to discriminate low probability tasks, but could not discriminate high probability of risk tasks.

Compressive load rate estimates correlated significantly with the probability of high-risk classification, as well as the dynamic estimates of spinal compression. Comparing dynamic compression force and compressive load rates demonstrate the

two biomechanical factors are highly correlated ($r^2 = 0.88$). Thus, the data describing compressive load rate does not add any more information than compressive force estimates. Similar results were discovered with AP shear force and AP shear load rate ($r^2 = 0.83$). However, lateral shear load rate may add important information to the prediction of LBD risk which is distinct from lateral shear force data ($r^2 = 0.57$).

No single biomechanical factor completely describes the probability of high-risk classification. This indicates the mechanisms of spinal injury require combination of multiple factors to explain the cause of occupational LBD.

The results indicated that a regression model, including multidimensional spinal loads improves the predictive power for LBD risk, when compared with a model including only axial compression. Incorporating spinal compression, lateral shear and AP shear loads into a regression model to estimate the probability of high-risk classification improved the correlation to $r^2 = 0.487$, although the improvement over dynamic compression alone was not statistically significant. The improved performance from multidimensional models agrees with the theoretical assessment of Shirazi-Adl (1989) who concluded that shear forces combined with compression increases the risk of vertebral disc failure. The influence of bending and torsional moments supported by the spine may also increase the risk of injury (Shirazi-Adl *et al.* 1986, Adams *et al.* 1987). In the current study, the relation between passive spinal moment and risk of LBD could not be quantified. However, considering the trend demonstrating greater fidelity when shear loads are included, it is tempting to hypothesize that inclusion of bending and torsional moments would further enhance the predictive power of the biomechanical regression model. Thus, estimates of spinal compression alone cannot fully predict the risk of LBD, because the injury mechanism is influenced by multidimensional loading patterns.

Lifting dynamics may directly influence the load and tolerance of the spine to injury. Analyses of dynamic exertions have demonstrated spinal load increases with velocity and acceleration (Freivalds *et al.* 1984, Granata and Marras 1995). Furthermore, those increased loads must be supported by a spinal column wherein the tissue tolerance may be influenced by the load rate (Yingling *et al.* 1995, Adams and Dolan 1996). There are no prior studies examining the influence of load rate on the risk of LBD, but the results reported here indicate that load rate may adversely affect the safety of a lifting exertion in terms of the probability of high LBD risk classification. However, there was no statistical benefit to examining spinal load rate as opposed to dynamic estimates of spinal compression. Furthermore, only the lateral shear load rate was independent of the computed force trends. Thus, inclusion of lateral shear load rate in a multiple factor model improved the ability to identify high-risk tasks, whereas compressive and AP shear load rate did not appear to contribute added information to the injury mechanism. Passive spine bending moment rate was not examined in this study. Considering the known response of biological tissue to viscous loading, future research might assess whether moment rate offers significant insight into the injury mechanism associated with occupational tasks.

It came as some surprise that the ligamentous strain was a highly significant factor in predicting the probability of high-risk tasks. Adams *et al.* (1994) indicated the ligament tension remains low throughout most of the range of motion, increasing rapidly near the limits of the motion range. Nonetheless, the ligament strains were among the best correlation parameters to identify variation in the probability of risk. Furthermore, the best performing multiple factor models consistently incorporated

ligamentous strain. The results indicate there is an interaction between spinal load and tissue strain that may influence the tolerance of the spine to injury. When attempting to identify risky tasks, the combination of spinal load, specifically lateral shear, and tissue strain generated significantly better response than dynamic estimates of spinal load or tissue strain alone. In their measurements of vertebral mechanics, Adams *et al.* (1994) concluded that sagittal flexion and the associated strain on the disc annulus does not influence the tolerance to compressive failure. However, their research examined the disc and apophyseal joint tissues; they did not examine the ligaments of the motion segments. Moreover, their methods were restricted to sagittal bending and compressive loading. Thus, the interaction between ligamentous strain and shear loading may be a significant factor in the mechanics spinal injury and low-back pain.

The rate at which spinal tissues are strained is related to dynamic motions of the trunk and therefore associated with the risk of LBD. Marras *et al.* (1993, 1995) have documented significantly greater injury rates associated with tasks requiring increased lateral and twisting velocities. Those increased velocities directly affect the strain rate applied to the functional units of the lumbar spine. Adams and Dolan (1996) demonstrated the rate at which the tissues are lengthened influences the resisted bending moments and mechanical performance of the vertebral motion segments. The results reported here indicated that the ligamentous strain rate was unrelated ($r^2 < 0.005$) to the probability of high-risk classification as a single factor model. However, the strain rate helped to improve the performance of three- and four-factor models, although the addition of the third and fourth factor did not give rise to a statistical improvement in performance. It should be noted that only lengthening strain rates were examined, i.e. it was assumed that the rate at which a ligament slackens is biomechanically uninformative. The predominant lengthening strain rates of the posterior tissues will occur during flexion tasks. Therefore, it is not surprising that the extension tasks examined in this study failed to demonstrate a highly significant response from the tissue strain rate. Future research may demonstrate the influence of biomechanical strain rate on LBD risk by examining a complete flexion and extension cycle.

When interpreting these results, it is important to be aware of limitations imposed by the use of an epidemiologic model as a baseline. This study examined the correlation between dynamic biomechanical parameters and predicted probability of LBD risk, which in turn represented the actual risks measured in industry (Marras *et al.* 1993). This method to estimate LBD risk undoubtedly introduced variability into the analyses. However, to achieve the appropriate biomechanical data it was necessary to simulate the warehouse selection environment, requiring an epidemiologic model to estimate probability of high-risk classification. The predictive ability of the epidemiologic model has been determined to be quite high, more than three times better than the NIOSH lifting model. Thus, the data represent an association between biomechanical factors and probability of risk classification, but the precise biomechanical causes of injury remain ambiguous.

Research indicates that psychosocial factors are related to the reporting of low-back injuries. In fact, Hansson (1996) suggested that psychosocial factors may be more effective predictors of occupationally related low-back pain than traditional biomechanical factors. However, traditional biomechanical factors such as static estimates of spinal compression neglect significant influences from spinal dynamics, load rates and biomechanical interactions. When the complex nature of spinal

biomechanics is considered, the potential to identify injury mechanisms and LBD risk may be markedly improved. Bigos *et al*. (1986, 1991) concluded that employees with low approval ratings and who 'hardly ever' enjoyed their jobs were at high risk of subsequently filing a back injury claim. Some (Croft *et al*. 1995) believe symptoms of psychological distress in uninjured workers may predict subsequent reports of low back pain. Others (Hansen *et al*. 1995) have found psychological distress is unrelated to reporting of LBD. It may be likely that these psychosocial factors have biomechanical origin, i.e. the dynamic biomechanics of the lift may change as a result of psychosocial influences. Dionne *et al*. (1995) demonstrated that the relation between educational level and LBD risk is largely explained by occupational factors, i.e. lower educational level was associated with biomechanically more stressful occupations. The results reported here indicate that biomechanical factors can explain > 57% of the variability in the probability of a task being classified as high risk for LBD. Conversely, psychosocial factors may influence the level of pain at which a low-back injury claim is filed. Thus, future analyses of spine biomechanics that include dynamic measures of tissue loads and strains, will permit identification of injury mechanisms, whereas psychosocial factors may indicate when existing injuries are reported.

An under-appreciated aspect of spine biomechanics and associated LBD is the synergistic interaction of various biomechanical factors. The results reported here indicate combinations of dynamic spinal loads and tissue strains best predict the probability of high-risk classification. However, the nature of regression analyses depicts these factors as independent variables. For example, it may be possible that spinal load at peak tissue strain is a better predictor than any combination of independent factors examined in the current study. Adams and Dolan (1996) demonstrate the injury tolerance behaviour of one biomechanical factor may be influenced by the state and history of another. Hence, biomechanical interactions may significantly influence the risk of low-back pain associated with lifting task design.

These analyses indicate the biomechanical source of low-back pain is dynamic, multifaceted and multidimensional. It may be concluded that dynamic spinal compression associated with a task contributes to the probability of high LBD risk. However, significantly improved prediction can be achieved by examining combinations and interactions of biomechanical factors, including multidimensional kinetics and kinematics of the spine. The results agree with the biomechanical literature in that spinal tolerance and material failure has been associated with multidimensional loads on the functional units of the spine. Unfortunately, the improved predictive ability associated with multidimensional, dynamic analyses is generated at the high cost of computational complexity. However, using 1-D static parameters may ignore the injury mechanism associated with the greater number of workplace injuries. Thus, improved understanding of the dynamic biomechanical interactions influencing the tolerance and injury mechanisms of the spine may permit more accurate assessments of workplace injury factors associated with LBD and reduced incidence of occupationally related low-back pain.

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