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# Spine loading at different lumbar levels during pushing and pulling

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As the nature of many materials handling tasks have begun to change from lifting to pushing and pulling, it is important that one understands the biomechanical nature of the risk to which the lumbar spine is exposed. Most previous assessments of push–pull tasks have employed models that may not be sensitive enough to consider the effects of the antagonistic cocontraction occurring during complex pushing and pulling motions in understanding the risk to the spine and the few that have considered the impact of cocontraction only consider spine load at one lumbar level. This study used an electromyography-assisted biomechanical model sensitive to complex motions to assess spine loadings throughout the lumbar spine as 10 males and 10 females pushed and pulled loads at three different handle heights and of three different load magnitudes. Pulling induced greater spine compressive loads than pushing, whereas the reverse was true for shear loads at the different lumbar levels. The results indicate that, under these conditions, anterior–posterior (A/P) shear loads were of sufficient magnitude to be of concern especially at the upper lumbar levels. Pushing and pulling loads equivalent to 20% of body weight appeared to be the limit of acceptable exertions, while pulling at low and medium handle heights (50% and 65% of stature) minimised A/P shear. These findings provide insight to the nature of spine loads and their potential risk to the low back during modern exertions.

**Keywords:** spinal loading; electromyography; push and pull; biomechanical modelling

## 1. Introduction

As the risk of low back pain associated with lifting and carrying has been recognised, many occupational tasks have been converted into pushing or pulling activities and, thus, increasingly expose workers to horizontal applications of force. In addition, as the nature of work changes, pushing and pulling has become more common. Manufacturing globalisation has increased the handling of products prior to distribution. Many products are stored, at least temporarily, in distribution centres, where large amounts of pushing and pulling are common. Patient handling is also often performed by pushing and pulling of patient lift systems. Pushing and pulling activities are also common in delivery tasks (Kelsey 1975, Snook 1978, Frymoyer *et al*. 1983, Damkot *et al*. 1984, Lavender *et al*. 2000).

However, pushing and pulling activities might also be associated with significant risk to the low back (Kelsey 1975, Snook 1978, Frymoyer *et al*. 1983, Damkot *et al*. 1984, National Institute for Occupational Safety and Health 1997, Hoozemans *et al*. 1998). Several sources report that as much as 20% of low back injury claims are associated with pushing and pulling (National Institute for Occupational Safety and Health 1981, Hoozemans *et al*. 1998). A review of low back pain Bureau of Workers' Compensation Claims in Ohio indicates that 27% of claims were related to pushing and pulling activities. Exposure to pushing and pulling activities is increasing in the workplace, yet there is poor understanding of how pushing and pulling may impact biomechanical loading and the subsequent risk of low back pain.

Guidelines and recommendations for pushing and pulling in the workplace have been derived primarily from isometric strength and psychophysical tests (Dempster 1958, Kroemer 1968, Chaffin *et al*. 1983, Snook and Ciriello 1991, Fothergill *et al*. 1992, Kumar 1995, Kumar *et al*. 1995, Imrhan 1999, Resnick and Chaffin 1996). These efforts have resulted in recommendations with considerable variability, yet only address a small number of conditions. When the large variation is considered, along with the inability to control risk to the low back, it is clear that the understanding of low back biomechanics during pushing and pulling is not sufficient for the control of risk.

A limited number of studies have employed biomechanical models in an attempt to calculate forces on the lower lumbar spine (between L4 and S1) during pushing and pulling. Resnick and Chaffin used a 2-D dynamic model with pseudodynamic inertial

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forces from the erector spinae and rectus abdominus muscle groups to examine the pushing and pulling of carts loaded with 45–450 kg (Resnick and Chaffin 1995). Significant spine compression (5000 N) was reported when pushing carts over 225 kg, yet spine shear forces were not reported. Another study employed a 2-D dynamic linked segment model of the upper body with a single extensor muscle to examine pushing and pulling of three different cart types pushing weights from 33–385 kg at different handle heights (de Looze *et al.* 1995). The model predicted compressive forces at L5/S1 from approximately 1600–2800 N, whereas shear forces ranged from approximately 140–320 N. The largest loads occurred while pulling the largest cart with the greatest weight. Another study of refuse collectors used a quasi-static 2-D linked segment model with flexor and extensor equivalent muscles and reported compressive loads at L4/L5 from 605–1445 N and shear loads below 200 N (Schibye *et al.* 2001). A related study examined the effects of pushing and pulling the same two-wheeled refuse container on different surfaces (Laursen and Schibye 2002). They calculated slightly higher compressive loads with a maximum of approximately 1800 N and similar shear loads all below 200 N. All these studies are similar in that static 2-D, single equivalent muscle models were employed that would not be sensitive to the trunk muscle cocontraction expected during dynamic pushing and pulling activities (Lee *et al.* 1989, Andres and Chaffin 1991). Only one evaluation of pushing and pulling could be found that was able to consider the influence of trunk muscle coactivation upon spine loading (Hoozemans *et al.* 2004). However, this study focused primarily upon the portion of the exertion (e.g. initial effort vs. sustained exertion) that yielded the largest loads on the spine. In addition, all previous models have evaluated forces on the lower lumbar spine exclusively.

Models that are static or quasi-static in nature are not able to adequately consider the effects of motion, acceleration or stability goals on the muscle recruitment patterns of the trunk. The quasi-static assumption has been shown to be an ineffective means to represent trunk motion since it cannot predict the trunk muscle recruitment and muscle cocontraction patterns essential to understanding spine loading (Marras and Granata 1997a, Marras 1992, Granata and Marras 1999). Ignoring the cocontraction seen in typical exertions can underestimate compression by 45% and shear by as much as 70% (Granata and Marras 1999).

Historically, spine compression in the lower lumbar spine has been the variable of interest for risk to the low back during work. However, during horizontal force application (pushing and pulling), it is expected that shear forces within the spine increase dramatically due to the application of force in the hands and the reaction of the trunk musculature. Thus, shear may represent the critical measure of risk. Here again, it is expected that static or quasi-static models of spine loading are incapable of accurately assessing shear loads on the spine since they are not able to account for the cocontraction of the torso musculature that define shear loads. Studies examining spine shear tolerances have reported much less tolerance to shear forces than compressive forces. McGill's 1000 N tolerance value is commonly sited as a limit above which there is increased risk of sustaining injury (McGill 1997). Cyron and Hutton report that cyclic shear loading of the neural arch has resulted in tolerance levels between 380 and 760 N (Cyron and Hutton 1978). Hence, it is important that shear forces are accurately and realistically assessed during pushing and pulling if risk of low back disorders is to be assessed.

In order to overcome the problems of assessing the role of muscle cocontraction defining spine loads, biologically assisted models have been developed that directly monitor the involvement of the power producing (and spine loading) trunk muscles during a task. In recent years, electromyography (EMG)-assisted models have been developed and validated in their ability to predict spine loads in three dimensions of loading during dynamic lifting activities (Marras and Sommerich 1991a,b, Granata and Marras 1993, 1995a, Marras and Granata 1995, 1997a,b). However, these EMG-assisted biomechanical models have seldom been used to assess pushing and pulling because they have been optimised for back extension activities as opposed to the flexion and complex loading conditions expected during pushing and pulling.

The objective of this effort was to understand how pushing and pulling activities influence spine loading at the various lumbar spine levels as a function of push–pull task parameters. In order to accomplish this goal it was necessary to employ an EMG-assisted biomechanical model that has been adjusted to respond to horizontally oriented applications of force as is common during pushing and pulling.

## 2. Experimental study

### 2.1. *Approach*

A study was conducted to assess the loads imposed upon the entire lumbar spine during pushing and pulling at three different resistance levels and at three different handle heights. A previously described and validated EMG-assisted biomechanical model was used to assess spine forces.

### 2.2. *Subjects*

A total of 10 males and 10 females were recruited from a university population and served as the subject population in this study. Inclusion criteria required that subjects displayed no prior history of low back pain in order to qualify for study participation. Average (SD) subject age was 24.5 (3.24) years for the males and 22.7 (2.36) years for the females. Mean (SD) subject weight was 72.7 (10.25) kg for males and 57.2 (8.19) kg for females, whereas mean (SD) height was 180.53 (7.32) cm for males and 165.37 (4.32) cm for females. Upon arrival subjects signed consent forms, and the necessary anthropometric measures for the biomechanical model were collected.

### 2.3. *Experimental design*

This study consisted of a two × three × three repeated measures experimental design. Two types of horizontal exertion activities (pushing and pulling), three handle heights (50%, 65% and 80% of subject stature) and three handle force levels (20%, 30% and 40% of subject body weight) served as the independent variables, which yielded 18 combinations of exertions. Conditions were presented in a counterbalanced order. Each of these combinations was repeated four times for each subject. Dependent variables consisted of spine compression, anterior-posterior (A/P) shear and lateral shear at each spinal disc level as predicted by a previously described EMG-assisted biomechanical model (Theado *et al*. 2007). EMG recordings from 10 power producing trunk muscles as well as kinematic and kinetic information about the orientation of the trunk and forces imposed upon the hands were used as inputs to the model.

### 2.4. *Apparatus*

Bipolar surface electrodes were placed over the 10 trunk muscles needed as input to the biomechanical model described previously. EMG data were collected with a Model 12 Neuradata Acquisition System (Grass Technologies West Warwick, RI, USA) at a 1000 Hz collection frequency. The signal was high-pass filtered at 30 Hz, low-pass filtered at 500 Hz and notch filtered at 60 Hz in the hardware. Customised Laboratory Information Management System software developed at the OSU Biodynamics Laboratory collected the EMG data as well as data from all the other equipment through a PCI-6031E Data Acquisition Device (National Instruments, Austin, TX, USA). The EMG data were rectified, averaged using a 40 ms sliding window filter and then normalised relative to values collected during maximum voluntary contractions (MVCs).

### 2.5. *Procedures*

MVCs were obtained by placing subjects in a reference frame, which restricted movement while they performed isometric trunk extension, flexion, right and left lateral bends and right and left twists. Subjects were then fitted with a lumbar motion monitor (LMM) appropriate for their anthropometry.

Subjects were familiarised with a horizontal force pulley system (Figure 1), used as a means to exert pushing and pulling forces, and allowed to practise a number of pushes and pulls until they felt

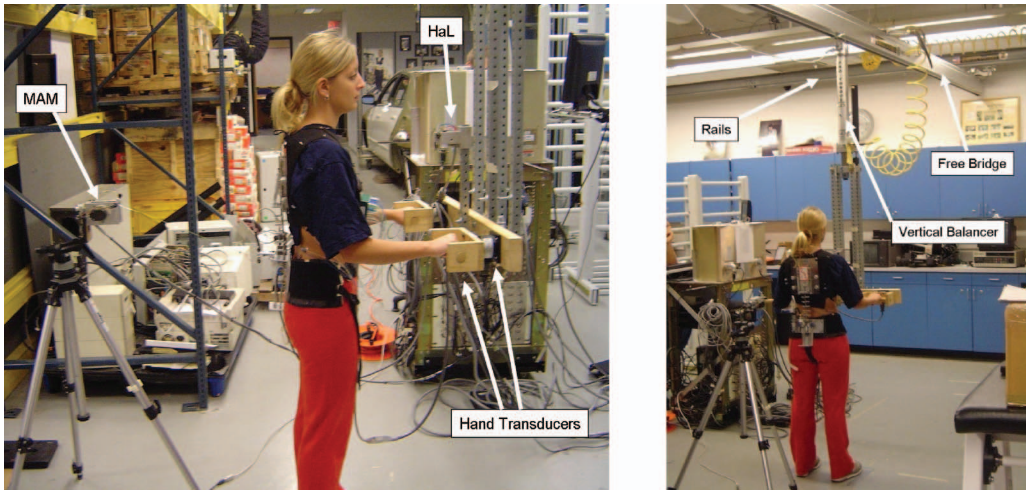


Figure 1. Experimental testing set-up. MAM = moment arm monitor; HaL = handle locator system.

comfortable with the set-up. The horizontal force pulley system consisted of a free horizontal bridge mounted perpendicular to two low friction linear track rails. A vertical balancer mounted to the free bridge supported the handles, transducers and a handle locator system (HaL). An optical encoder tracked the position of the torso in 3-D space moment arm monitor (MAM), whereas another goniometric system (HaL) tracked the position of the shoulders relative to the balancer's handles. The 3-D hand forces and moments at each hand were monitored with Py6–500 force transducers (Bertec Corporation, Columbus, OH, USA) attached to handles. The balancer provided vertical freedom to the handle structure while still allowing specific positions to be maintained. A cable ran from the vertical balancer over a fixed pulley and down to a box that held the weight. This provided a relatively constant horizontal force on the handles. The horizontal force pulley system was designed to be similar to commercially available rail-mounted devices with horizontal and vertical freedom, but in this case modified so as to provide a relatively constant horizontal force level.

Subjects began each exertion with the feet together, standing erect with their hands at their sides in order to record a neutral reference value and to zero the hand transducers. Subjects were then instructed to place their hands around the handles without touching them until instructed to begin the task. At each height, the subjects were instructed to start with a specific initial arm posture at the beginning of each trial in order to remain consistent with the initial conditions in the push–pull model. During the experiment, subjects were instructed to push and pull at a speed that was fast yet comfortable. In addition, subjects were instructed not to twist their body relative to the handle structure any more than was necessary for completion of the task. No restrictions were placed on which foot to step with first or stride length. Subjects were allowed to move their torso as needed, including leaning postures if they chose. Subjects were provided rest periods between each exertion condition.

### 2.6. *Spine load assessment*

The experimental data were used as input to a biologically assisted (EMG-assisted) biomechanical model. The basic logic underlying this model has been described extensively in the literature (Marras and Sommerich 1991a,b, Granata and Marras 1993, 1995b, Marras and Granata 1995, 1997a,b). The model uses kinematic information about the trunk along with EMG information from the trunk musculature to estimate spinal loads, as well as predict the moments imposed on the spine in 3-D space.

The model assumes that the key to understanding tissue loading within the spine is to understand how the internal forces (muscles and ligaments) within the trunk respond to exposure to occupational tasks. Trunk muscle recruitment patterns and antagonistic coactivations have been attributed to not only physical requirements, but are also a function of the organisational or psychosocial environment, training and individual characteristics (Marras *et al.* 2000). Biologically assisted models directly monitor the trunk muscle responses to physical and environmental conditions and use this information as input to a biomechanical model so that the effects of realistic muscle coactivation can be considered in defining tissue loading.

The EMG-assisted dynamic model is unique in that it is person specific in terms of anthropometry (muscle location and size), subject motion (trunk as well as limb motion) and muscle activities. The model structure is multi-dimensional and considers the dynamic response of the modelled subject. Trunk moments and tissue loads are determined by considering the dynamic muscle force vectors and their corresponding trunk muscle moment arms. The model employs 10 muscle equivalent vectors and approximate trunk anatomy and mechanics (Yoo *et al.* 1979, Schultz and Andersson 1981, McGill *et al.* 1988, Dumas *et al.* 1991). Trunk muscle vector orientations have been derived via magnetic resonance imaging (MRI) studies (Jorgensen *et al.* 2001, Marras *et al.* 2001) and scaled via regression equations. Muscle fibres sampled by EMG were originally monitored via intramuscular electrodes (Marras *et al.* 1984) but are now sampled using surface electrodes.

Since complex motions such as pushing and pulling involve flexion moments as well as the extension moments, model adjustments were necessary. Model adjustments include muscle length-force and velocity-force adjustments for torso flexion and extension. These model adjustments have been incorporated into the model and are applicable to both simple lifting exertions as well as complex exertions such as push–pull efforts. Model adjustments also take into account the torso mass and torso angle so that the influence of leaning during pushing and pulling can be considered in spine force estimates. A recent publication describes how the model has been adjusted to accommodate complex pushing and pulling tasks as well as lifting tasks (Theado *et al.* 2007).

The model was also adjusted to incorporate compression and shear calculations at each level of the lumbar spine. In order to accomplish this, it was necessary to predict the vertebral angle of the lumbar vertebrae during instantaneous points in time and determine how trunk loading would influence each vertebrae and disc as a function of time. The LMM was used to derive vertebral angle based upon the findings of Splittstoesser (2001, 2006). This technique is described elsewhere (Knapik 2005). In order to be able to model different types of exertions and to examine different structures in the body in more detail, the current authors have begun using Adams (MSC Software Corporation, Santa Ana, CA, USA) with the LifeMOD (LifeModeler Inc., San Clemente, CA, USA) plug-in. Adams is a motion simulation solution for analysing the complex behaviour of mechanical assemblies. It was possible to embed the existing biomechanical models into the Adams software environment.

### 2.7. *Statistical analyses*

After collecting the data and retrieving results from the push–pull model, the statistical significance of each of the dependent measures was determined. Univariate ANOVA was used to determine statistical significance. Main effects and interactions were assessed for activity (push vs. pull), gender, handle height and hand force level. Subject was used as a blocking factor to account for inter-subject variability.

## 3. Results

Statistically significant differences as a function of push vs. pull activity (activity), gender, handle force, handle height and all two-way interactions with push–pull activity are shown in Table 1 for all lumbar spine disc levels. This table indicates that there are many factors that can influence spine load at the various lumbar spine levels. In general, activity, handle height and hand force level along with the two-way interactions of activity with handle height and hand force influenced spine loading. While gender was also significant, the nature of this relationship was as expected, with males exhibiting greater compression and A/P shear loads most likely due to greater body mass (Marras *et al.* 2002).

The nature of the exertion played a major role in defining spine forces. Figure 2 shows the compression, A/P shear and lateral shear forces as a function of lumbar disc level for pushing vs. pulling. Significant differences in pushing and pulling were observed at all disc levels for lateral shear. However, the magnitudes of these forces were well below a level that would be expected to result in damage. Compression was significantly greater in pulling compared to pushing from L3/L4 to T12/L1. However, here again, the magnitude of these compressive forces would not be expected to lead to tissue damage. A/P shear, however, was of a magnitude that should be of concern for many lumbar levels. Statistically significant differences in A/P shear forces occurred as a function of pushing vs. pulling at all levels except for L5/S1 with pushing resulting in greater force compared to pulling. Of particular concern were the magnitudes of A/P shears occurring between L3/L4 and T12/L1, which

Table 1. Statistical significance summary of experimental conditions upon spine loading.

| | | Activity | Gender | Handle height | Hand force | Activity *gender | Activity* handle force | Activity* handle height |
|---|---|---|---|---|---|---|---|---|
| L5/S1 | Comp | 0.00822 | < 0.0001 | < 0.0001 | < 0.0001 | 0.9701 | < 0.0001 | 0.0005 |
| | A/P | 0.6237 | 0.0015 | 0.0459 | < 0.0001 | 0.1632 | < 0.0001 | < 0.0001 |
| | Lat | 0.0016 | 0.4150 | 0.0052 | < 0.0001 | 0.6976 | 0.0729 | 0.1886 |
| L4/L5 | Comp | 0.0608 | < 0.0001 | < 0.0001 | < 0.0001 | 0.7385 | 0.0052 | 0.0978 |
| | A/P | 0.0016 | 0.1129 | < 0.0001 | < 0.0001 | 0.0751 | < 0.0001 | < 0.0001 |
| | Lat | < 0.0001 | 0.0614 | 0.0035 | < 0.0001 | 0.3311 | 0.0637 | 0.0458 |
| L3/L4 | Comp | 0.0093 | < 0.0001 | 0.0002 | < 0.0001 | 0.4672 | 0.0241 | 0.5206 |
| | A/P | 0.0185 | 0.0016 | < 0.0001 | < 0.0001 | 0.1067 | < 0.0001 | < 0.0001 |
| | Lat | < 0.0001 | 0.0075 | 0.0048 | < 0.0001 | 0.0825 | 0.0574 | 0.0057 |
| L2/L3 | Comp | 0.0013 | < 0.0001 | 0.0003 | < 0.0001 | 0.3279 | 0.0102 | 0.6571 |
| | A/P | 0.0184 | 0.0006 | < 0.0001 | < 0.0001 | 0.1378 | 0.0003 | < 0.0001 |
| | Lat | < 0.0001 | 0.0016 | 0.0040 | < 0.0001 | 0.0394 | 0.1073 | 0.0005 |
| L1/L2 | Comp | 0.0005 | < 0.0001 | 0.0002 | < 0.0001 | 0.2830 | 0.0036 | 0.5767 |
| | A/P | 0.0086 | 0.0004 | < 0.0001 | < 0.0001 | 0.1427 | 0.0023 | < 0.0001 |
| | Lat | < 0.0001 | 0.0009 | 0.0031 | < 0.0001 | 0.0348 | 0.1815 | 0.0002 |
| T12/L1 | Comp | 0.0013 | < 0.0001 | 0.0002 | < 0.0001 | 0.3207 | 0.0035 | 0.4177 |
| | A/P | 0.0018 | 0.0003 | < 0.0001 | < 0.0001 | 0.1995 | 0.0019 | < 0.0001 |
| | Lat | < 0.0001 | 0.0007 | 0.0021 | < 0.0001 | 0.0348 | 0.2253 | 0.0001 |

Note: Activity refers to pushing vs. pulling activities.
The gender, handle height and handle force columns indicate the main effect of these factors on spine loading and their interactions with activity are shown in the last three columns. The influence of these factors on spine compression (Comp), anterior–posterior shear (AP) and lateral shear (Lat) are shown in the rows for the six disc levels of the lumbar spine (L5/S1 through T12/L1).

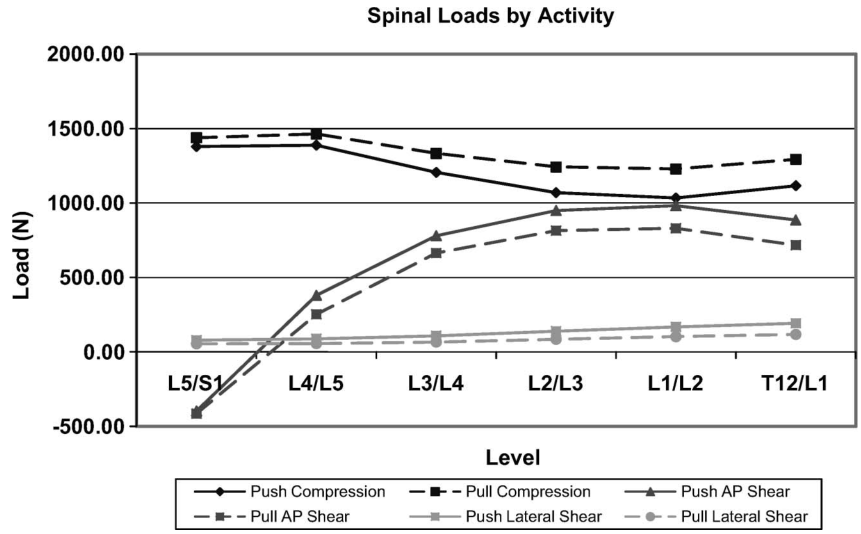


Figure 2. Compression, anterior–posterior (A/P) shear and lateral shear forces at each lumbar disc level as a function of pushing and pulling.



approached shear tolerance levels (Cyron and Hutton 1978, Marras and Granata 1997b, Callaghan and McGill 2001).

Hand force magnitudes had a significant effect on spine loading at all lumbar levels. Compression forces were generally lower during pushing compared to pulling at most hand force levels and, as expected, increased as more force was applied by the hands. Similar trends were observed for lateral shear. However, for both compression and lateral shear, most observations were well below the thresholds for tissue damage (Evans and Lissner 1959). A/P shear forces varied greatly as a function of hand force as shown in Figure 3. Both push and pull hand forces approach or exceed the 1000 N threshold for tissue damage for discs between L3/L4 and T12/L1 when pushing and pulling against forces equivalent to 40% of body weight. Pushing produced greater forces than pulling at disc levels above L3/L4 under these conditions. Pushing loads equivalent to 30% of body weight produced disc A/P shear forces that also approach 1000 N in the discs between L2/L3 and above, whereas pulling forces were generally well below this threshold under the 30% hand force conditions.

Handle height also had a significant influence on spinal loading at all lumbar levels. Compressive forces on the spine were lowest under the medium handle height (65% of body stature) condition for all lumbar levels regardless of whether a push or pull activity was performed. However, pushing always produced lower compressive loads than pulling. The greatest compression forces were observed at the low handle height (50% of stature) condition during pushing and most high handle height conditions (85% of stature) during pulling at most lumbar disc levels. However, as with other conditions, both the compression and lateral shear magnitudes were of a level that should not be of concern for biomechanical loading. Significant differences in A/P shear occurred as a function of pushing vs. pulling at the various disc levels. Figure 4 shows how A/P forces differ as a function of the direction of the exertion and lumbar level. It is interesting to note that the low handle height pushing condition resulted in the greatest A/P shear forces at most lumbar levels, whereas the low handle height pulling conditions resulted in the lowest spine A/P shear forces at most lumbar levels. High height handles pulling and pushing yielded the next greatest A/P shear force levels, respectively, at most lumbar disc levels and exhibited a trend that was the opposite of that for low handle height pushing and pulling.

## 4. Discussion

This study has demonstrated that it is possible to assess risk associated with pushing and pulling activities by employing an EMG-assisted model to account for trunk muscle coactivation in assessing trunk loading. Overall, this study found that there were differences in spine loading as a function of pushing and pulling. The present findings suggest greater compressive loading at all spine levels when pulling compared to pushing. Hoozemans *et al*. (2004) reported similar relationships between pushing and pulling for compression at L5/S1, but the objectives of their study were significantly different than the current study and, thus, not directly comparable. However, the present results suggest that

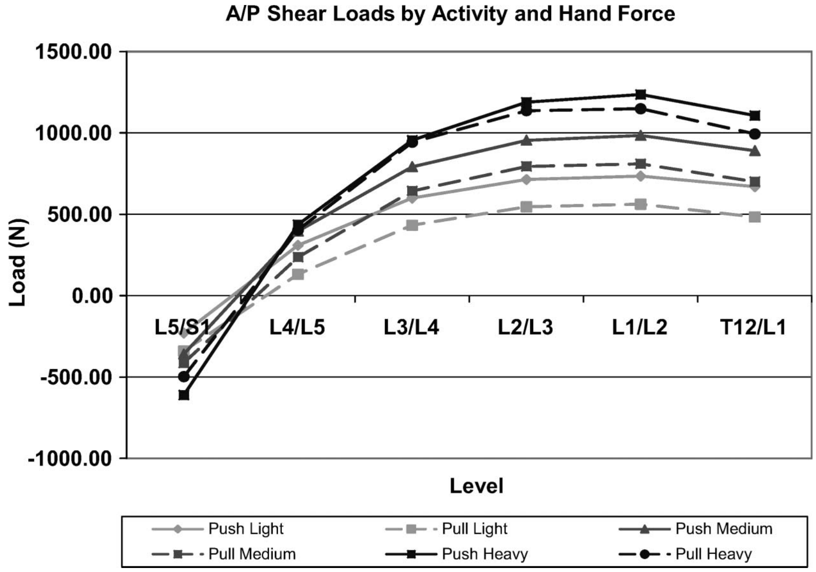


Figure 3. Anterior–posterior (A/P) shear forces as a function of pushing vs. pulling and hand force exertion level (light force = 20% of body weight, medium force = 30% of body weight, and heavy force = 40% of body weight).

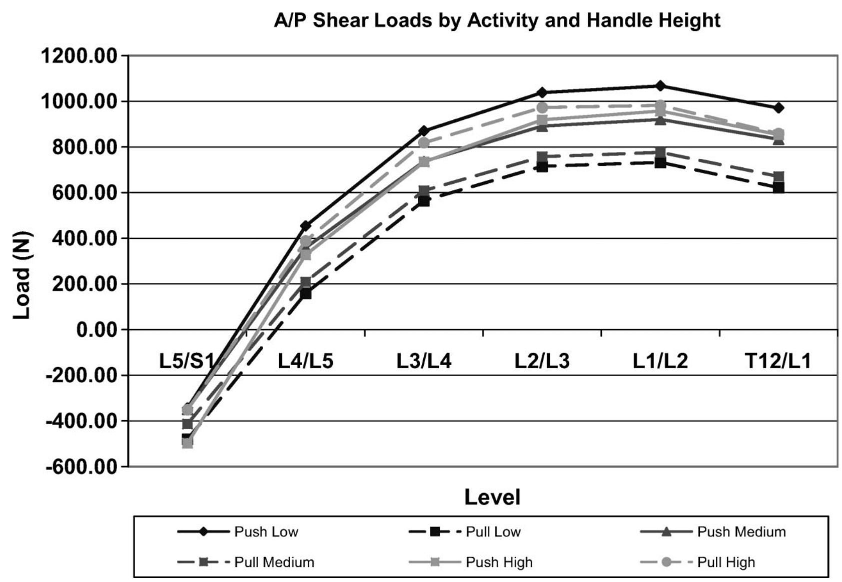


Figure 4. Anterior-posterior (A/P) shear forces as a function of pushing or pulling at different handle heights (low = 50% of stature, medium = 65% of stature, and high = 80% of stature).

compression was generally of a magnitude that would not be expected to result in tissue damage.

Most of the problematic spinal loads appeared to be associated with A/P shear forces. In addition, most of the excessive A/P shear loads occurred at the higher levels of the lumbar spine (L3 and above). These experimental results provided several insights as to the risks associated with pushing and pulling activities. First, in general, pushing activities impose greater potentially risky A/P forces upon the spine than pulling. Pushing imposed up to 23% greater A/P shear forces (on average) compared to pulling. Increases in shear forces were a result of the increased flexor muscle coactivity required by the task. During extension (as in lifting), the large erector spinae muscles can provide much of the power required for a lift. However, during pushing, the flexor muscles that have a much more limited cross-sectional area must generate internal force. In order to generate the required force, much greater coactivations of the muscle flexors are necessary. Since many of the oblique flexor muscles have a large horizontal muscle fibre orientation, these muscles produce significant shear forces. Figure 5 illustrates this point. Note how the more horizontally oriented muscles (oblique muscles) are associated with a horizontal shear force indicated by the disc force vectors.

Another important finding of this study relates to the large difference in A/P shear loads experienced among the lumbar vertebrae. Previous studies assessed the level of load primarily at the lower lumbar vertebrae (i.e. L5/S1) and found little reason for concern. The current study confirms these findings. However, this study represents the first study to assess the loads imposed upon the vertebrae and discs at the upper lumbar levels. It was found that shear forces at these upper levels were two to three times as great as the shear forces in the lower lumbar levels. Part of this difference is due to the curvature of the spine at the lower lumbar levels. L5/S1 is positioned more horizontally in standing positions compared to the upper lumbar levels, which are positioned nearly upright. Thus, L5/S1 forces result in larger compressive loads, which can be withstood by the lower lumbar levels with much greater tolerances than the A/P shear loads experienced at the upper lumbar levels. Hence, the risk to the low back during pushing and pulling appears to be more associated with the lumbar levels at L3 and above.

It should also be noted that these A/P shear forces experienced by the upper lumbar levels are large and of a magnitude that would be expected to produce damage to the discs. However, the forces experienced by the discs themselves would not be expected to be as large as those reported here since a reasonable portion of the A/P shear loads would be expected to be absorbed through bone contact forces within the posterior elements of the spine. The forces reported here would be expected to be problematic when the torso was flexed, which would transfer the majority of the load to the disc. One would expect that much of the lumbar lordosis would be maintained during pushing, which would help control the shear loads experienced by the disc. Nonetheless, significant shear forces are experienced by the lumbar tissues during pushing and should be considered a source of risk.

Second, as expected, the amount of A/P shear force experienced by the spine varies greatly as a function of the magnitude to the load handled. However, this study provides some insight as to how much load is too much from a pushing and pulling perspective. Figure 3 indicates that pushing or pulling loads equivalent to 40% of body weight are problematic, as are pushing loads equivalent to 30% of body weight. Pulling loads equivalent of 30% of body weight appear to be at the margin of safe activities and pushing and pulling loads of 20% of body weight would not be expected to be problematic. Thus, these analyses provide some preliminary biomechanically based limits for occupational tasks.

Third, handle height appears to play an important role in defining A/P shear loading of the spine. Pulling loads at 50–65% of stature appear to expose the spine to the least amount of shear. However, pulling at high heights (80% of stature) greatly increases shears to levels near the highest shear levels observed (pushing at 50% of stature). Thus, the impact of the handle height on shear is highly dependent upon whether one is pushing or pulling. Figure 4 shows that the 50% handle height is associated with both the highest and lowest spine shear forces, depending upon whether one was pushing or pulling. Closer examination of the data indicted that this discrepancy was associated with the vector of force applied to the handles. While it has been assumed that pushing and pulling requires horizontal application of force to the handles, the analyses of applied handle forces indicated that few exertions resulted in truly horizontal applications of force on the handles. While pulling at low levels, subjects tended to lift up on the handles, producing

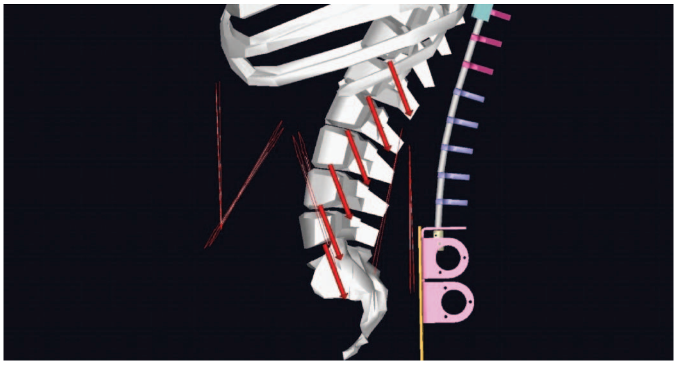

Figure 5. Muscle forces (thin hollow arrows) and resultant disc force vectors at various disc levels (solid arrows) indicating the relationship between the oblique muscles line of action and the shear forces acting on the spine.

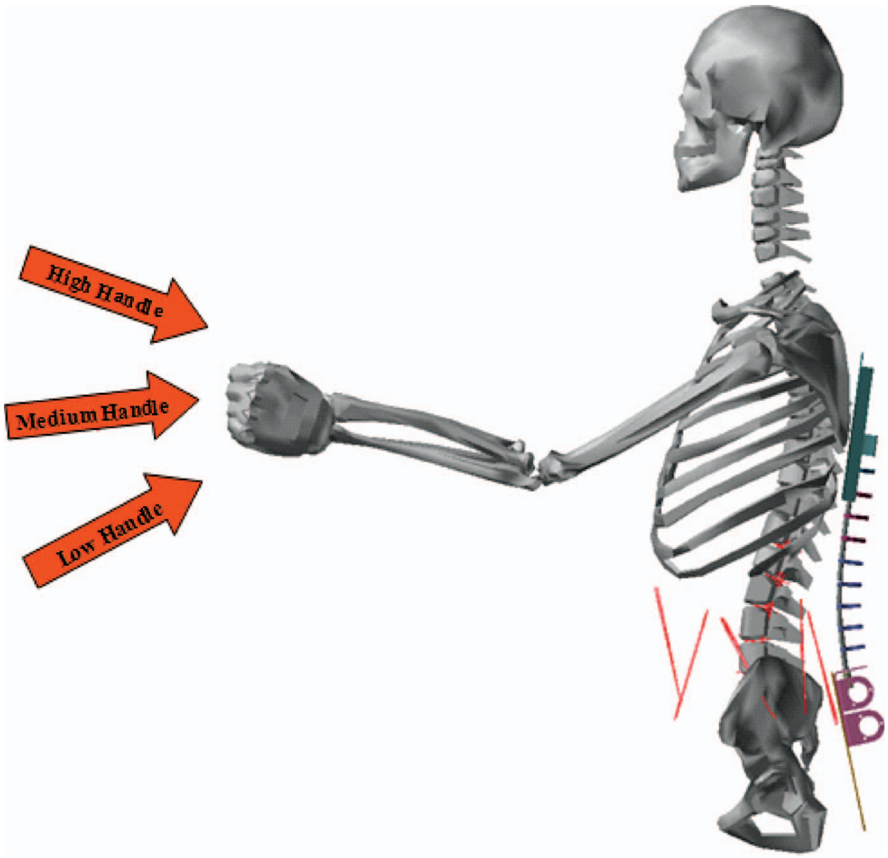


Figure 6. Resultant hand forces as a function of handle height.

more of a lifting action, whereas when pulling at high handle heights, subjects tended to pull down, creating more of a flexion moment about the spine (Figure 6). Hence, pulling at low height levels minimised the A/P shear on the spine.

Potential study limitations should also be noted. First, as with any study these findings are a function of the experimental conditions investigated. The current study explored pushing and pulling directed in a forward or backward (but unrestricted) path from the subject. Thus, these results apply primarily to forward and backward pushing and pulling activities as would be experienced when manipulating a cart or overhead carrier without turning. Further efforts are exploring the impact of floor-based and overhead-based lifting assistance device turning and will be presented in forthcoming publications. Second, the spine forces reported here reflect not only the influence of the object pushed or pulled but also the reaction of the subject. It is possible that another set of subjects would use different inertial forces in their efforts and, thus, different spine loadings would result. However, given the relatively large subject population for a biomechanical study it is felt that these results are representative of the spine loads experienced by inexperienced subjects performing pushing and pulling tasks. Hence, it is felt that this study provides insights to the biomechanics of pushing and pulling.

Collectively, these analyses suggest that pushing and pulling activities are not as intuitive as once thought and the risk to the low back occurs at lumbar levels that have been previously underappreciated. This study provides some insight as to how the body reacts to horizontally applied forces and suggests some important factors that should be considered when assessing risk.

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