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Evaluation of spinal loading during lowering and lifting

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Abstract

Objective. To estimate the three-dimensional spinal loads during various lifting and lowering tasks.
Design. The *in vivo* measurements of the trunk dynamics, moments, and myoelectric activity were used as inputs into an electromyographic-assisted model used to predict the three-dimensional spinal loads.
Background. Previous studies of eccentric motions have investigated muscle activity, trunk strength, and trunk moments. A void in the body of knowledge exists in that none of these studies investigated spinal loading.
Methods. Ten subjects lifted (40° of flexion to 0°) and lowered (0° of flexion to 40°) boxes while positioned in a structure that restrained the pelvis and hips. The tasks were performed under isokinetic trunk velocities of 5, 10, 20, 40, and 80 deg s^{-1} while holding a box with weights of 9.1, 18.2, and 27.3 kg.
Results. Lowering strength was found to be 56% greater than lifting strength. The lowering tasks produced significantly higher compression forces but lower anterior–posterior shear forces than the lifting tasks. The differences in the spinal loads produced by the two lifting tasks were attributed to differences in coactivity and unequal lifting moments (i.e. holding the box farther away from the body).
Conclusions. The nature of the spinal loads that occur during lowering and lifting were significantly different. The difference in spinal loads may be explained by different lifting styles.

Relevance

This study revealed the importance of investigating lowering as well as lifting since these types of motions result in drastically different EMG–torque relationships and, ultimately, different spinal loading patterns. Furthermore, this study indicates the importance of taking into account differences in lifting style (trunk moments) and the coactivity of the trunk muscles when estimating loads on the spine. © 1998 Elsevier Science Ltd. All rights reserved

Key words: Lowering, coactivity, spinal loading, EMG-assisted model, isokinetic velocity

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Introduction

Whereas research exploring various lifting tasks is extensive, there has been limited investigation into lowering tasks. Some evidence exists that the body reacts differently during these two types of lifting. Henriksson *et al.*[1] found that the perceived exertion for lowering tasks was less than that for lifting tasks. Also, physiological factors such as heart rate and oxygen consumption were lower for lowering than lifting. Other researchers have found that the muscle activity of the back extensor muscles was lower for lowering tasks than for lifting tasks despite the generation of greater external torque[2–5]. Further evidence that the muscle force generation capability is

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different under the two motions is the large difference in strength between lifting and lowering. Marras and Mirka[3] found that the strength for lowering was greater than lifting when the trunk was flexed forward. These results were independent of trunk asymmetry. Further, the trunk strength increased as the lowering trunk velocity increased. Reid and Costigan[6] found that lowering strength was about 20% higher than lifting strength for the trunk extensors. Other researchers have seen larger strength for lowering than lifting[7].

Several researchers have used a rigid link model to estimate the trunk moments imposed on the spine in the evaluation of lowering and lifting exertions. Differences in moments imposed on the spine would indicate that different lifting styles were used during the lowering and lifting motions. De Looze *et al.*[5] used a two-dimensional dynamic linked-segment model to estimate joint moments about the hip, ankle, knee, and L5/S1 joint. These authors found that the external moments were slightly larger for lifting compared with lowering. Similar results were also found by Gagnon and Smyth[8]. However, Gagnon and Gagnon[9] found the trunk moments during lowering and lifting were equivalent under similar dynamic conditions.

The above research has emphasized some underlying differences between lowering and lifting. In general, individuals were stronger but have lower muscle activities during lowering. This would indicate that the capability of the trunk muscles to generate force is different under the two lifting conditions. However, none of these studies investigated the resulting spinal loads under lowering and lifting. One would expect that the nature of spinal loading would be different since the muscle recruitment patterns would change.

This study made adjustments to the dynamic electromyographic (EMG)-assisted model developed at the Ohio State University (OSU) over the past decade[10–17]. This model predicts the three-dimensional spinal loads at the lumbro-sacral joint (L5/S1) (shear and compression). The model, which incorporates the ten major trunk muscles, accounts for the change in coactivity that would be expected with eccentric lifting. Thus, more realistic estimates of spinal loads are expected. The OSU model has traditionally been validated for concentric lifting. In order to use the model for eccentric tasks, the empirical relationships needed to adjust the model for lowering motion were determined. Adjustments to the model included the development of length–strength and force–velocity relationships needed to calculate spinal loading accurately during lowering exertions. The length–strength and force–velocity relationships for lifting used in the model were computed by minimizing the average variation in predicted muscle gain as a function of muscle length and velocity[15,18]. Similar techniques were used in this study to develop these relationships for lowering motions.

Thus, there were two objectives of this study: first, to develop the empirical relationships for length–strength and force–velocity under lowering to be incorporated into the model; second, to interpret the resulting spinal loading during controlled lowering and lifting.

The investigation of the lowering and lifting tasks was limited to the active range of the extensor trunk muscles.

Model adjustment and performance

In the EMG-assisted model, the three-dimensional spinal loads are predicted by the vector summation of the forces of the ten trunk muscles. The estimation of muscle force is a function of the cross-sectional area, muscle gain, normalized muscle activity, muscle length and muscle velocity. The model utilizes the following equation to estimate the muscle forces:

Muscle force

$$= \left[\text{cross-sectional area} \times \text{gain} \times \frac{\text{EMG}}{\text{EMG}_{\text{max}}} \times f(\text{length–strength}) \times f(\text{force–velocity}) \right]$$

In this equation, the muscle gain is an estimate of the stress capacity of the muscle. The third term is the measured muscle activity normalized to maximum voluntary muscular activity. The last two terms are the empirical relationships that define the EMG–force relationship as a function of length and velocity. Researchers have determined that length and velocity change the maximum muscle force[19–22]. Thus, for any valid EMG-assisted spinal load model, the length–strength and force–velocity factors need to be developed.

The estimation of spinal loading during lowering requires the development of the empirical relationships between muscle torque and muscle length and velocity. In order to do this, the method used by Granata and Marras[15] and Raschke and Chaffin[18] was employed which minimized the average variation in the ratio of external to internal torque as a function of muscle length and velocity. Since, according to the logic, the internal moments need to be equal to the external moments, the resulting relationship mapped out from these ratios would then describe the length–strength characteristic of the extensor muscles. A similar technique was performed to estimate the force–velocity relationship of the trunk muscles.

The length–strength relationship for lowering was found to be very similar to that found by Granata and

Marras[15] for lifting exertions (Figure 1). In both studies, the length–strength (L–S) relationship was described as a third-degree polynomial as determined by regression techniques. The resting length L_0 for the erector spinae muscles is the sagittal flexion position at which the muscle has the greatest potential for force generation, approximately 20° of sagittal flexion[3,23]. The differences between the estimated L–S curves were found for extreme lengths of the muscle resulting from slightly different fits of the regression curves.

The force–velocity (F–V) modulation for the lifting tasks was described as an exponential function, as reported by others[15,21,22,24]. Conversely, the lowering force–velocity modulation was found to be a constant at 1.56 (Figure 2). These relationships were the 'best' fit equations derived from regression methods. Whereas Sutarno and McGill[24] found a similar F–V relationship for low level forces, they found that the F–V relationship actually decreased with faster lowering velocities when exerting at higher loads. The differences between the relationships in the two studies might have resulted from the methods used to derive the relationships. In the current study, the F–V relationship was derived from the muscle forces, whereas Sutarno and McGill[24] based their F–V curves on the EMG of the right erector spinae muscle.

The value of gain for an individual should remain constant under both lowering and lifting exertions; thus, the F–V constant was set to the ratio of lifting and lowering gains (averaged over all subjects). This ratio indicates that individuals were stronger during lowering than during lifting (56% more).

To determine the validity of the new length–strength and force–velocity modulations, the EMG-assisted model was exercised for both lowering and lifting. Model fidelity is indicated by the data in Table 1. The model performance was evaluated via gain, r^2, and average absolute error (AAE). Significant differences between lifting and lowering exertions were indicated ($p \leq 0.05$). The gains between the two types of lifts were almost identical. For both lifting conditions, the gains were within physiological limits ($<100\ \mathrm{N\,cm}^{-2}$)[6,25]. The r^2 and AAE indicate how well the model predicted the internal moments compared with the measured external trunk moments. The r^2 was actually better for the lowering exertions (0.95) than the lifting tasks (0.88). The value of r^2 indicates how well the measured and predicted moment variabilities coincide. Additionally, the lowering tasks had a lower AAE than the lifting tasks. The AAE, which indicates the magnitude of the difference between the measured and predicted moments, was more impressive when normalized as a function of moment. Under both conditions, the AAE was less than 10% of maximum sagittal trunk moment. Although both the r^2 and AAE values were better for lowering motions, the model performed well under both conditions.

Further validation was completed by evaluating data from an independent source: in this case, kinematic and EMG data from Fathallah[26] were

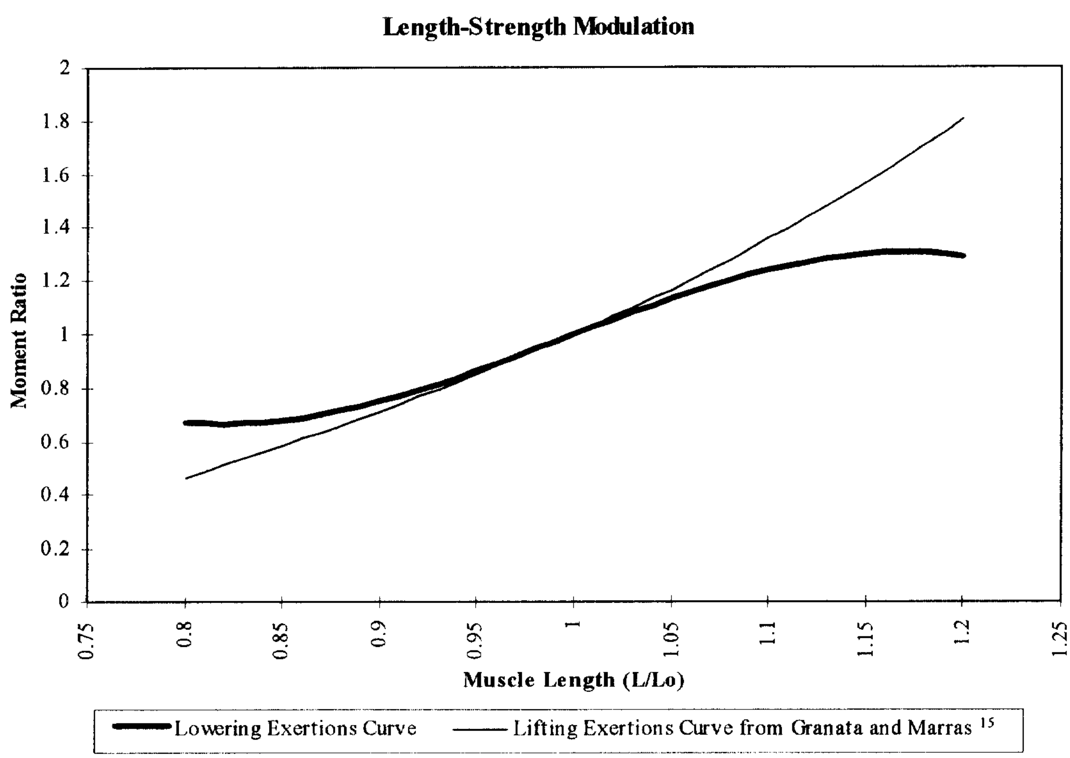

Figure 1. The length–strength modulation found for lowering and lifting exertions.

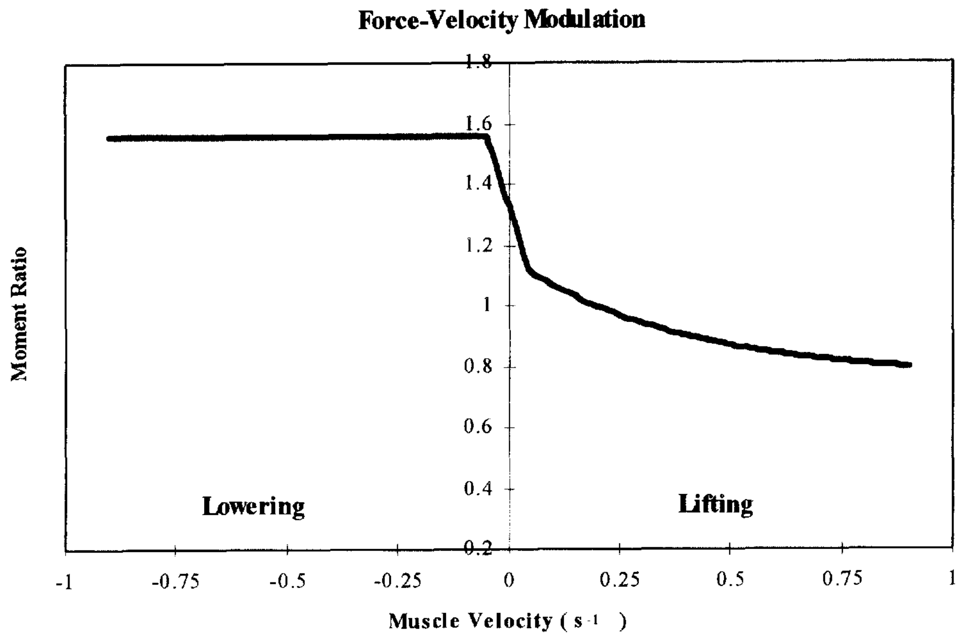

Figure 2. The force–velocity modulation found for lowering and lifting exertions.

analyzed. The lifts were sagittally symmetric lifts performed under free-standing conditions at various speeds and weights. The lifts required that the subjects start in a sagittally flexed position (approximately 40° of flexion), lifted to the upright position, pause for approximately 1 s and then lower to the sagittally flexed position. The 11 subjects were male students with no history of back problems and ages ranging from 23 to 38 years. For further details on the subjects, the collection of data and the protocol of the study refer to Fathallah[26].

The EMG-assisted model with the new L–S and F–V modulation factors was used to evaluate the lifts in the study just described[26]. The gain values were found to be well within the physiological limits, with a low standard deviation between the subjects (mean = 41.8, sd = 12.0). An average $r^2 = 0.91$ indicated that the internal moments fitted to the external moments accurately. This was further supported by the low standard deviation (sd = 0.05) around that mean. The average absolute error averaged about 11 N m (sd = 4.39) or 7.23% of the maximum sagittal moment. These results indicate that the model performed well during tasks which contained both lowering and lifting motions. Based on the above results for exertions containing both lowering and lifting, the length–strength and velocity modulations in the EMG-assisted model were considered valid.

Methods

Subjects

Ten male students with no prior history of low back pain volunteered to participate in the study. The ages of the subjects ranged from 22 to 34 years. The subjects had a mean (sd) height, weight, waist depth, and waist breadth of 181.0 cm (6.6), 79.3 kg (12.6), 21.6 cm (2.9), and 29.7 cm (3.2) respectively. These anthropometric measurements were used in the EMG-assisted model.

Experimental design

The lowering study was a three-way, within-subject design. The independent variables were chosen to include a range of workplace features that might be

Table 1. The model outputs for the lowering and lifting exertions in the study

Model output	Lowering exertions		Lifting exertions	
	Mean	(sd)	Mean	(sd)
Subject gain	46.95	(22.93)	45.00	(19.93)
r^2*	0.95	(0.09)	0.88	(0.19)
AAE* (N m)	7.75	(4.74)	9.28	(6.36)
AAE of maximum sagittal moment (%)	5.52	(3.37)	8.21	(5.62)

*Indicates a significant difference at $p \leq 0.05$.

expected in industry. These included the following: box weight, isokinetic trunk velocity, and type of lifting task (lifting versus lowering). In order to account for variability between the subjects, subjects were used as a random effect. The lifting tasks performed were lowering (eccentric) and lifting (concentric). The box weights used in this study were 20 lb (9.1 kg), 40 lb (18.2 kg), and 60 lb (27.3 kg). The trunk velocities were 5, 10, 20, 40 and 80 deg s^{-1}. These weights and velocities were chosen to reflect the values commonly found in industry[27]. The trunk velocities were performed under isokinetic conditions. The subjects controlled their velocity by monitoring a computer display of their instantaneous velocity.

The dependent variables used in this experiment consisted of maximum spinal loads. The spinal loads computed were compression, anterior–posterior (A–P) shear and lateral shear forces about the L5/S1 junction. These loads were computed using the OSU EMG-assisted biomechanical model developed at the Ohio State Biodynamics Lab described previously.

Apparatus

The Lumbar Motion Monitor (LMM) (Chattanooga Group™, Inc., Hixson, USA) was used to collect the trunk motion variables. The LMM is essentially an exoskeleton of the spine in the form of a triaxial electro-goniometer that measures instantaneous three-dimensional position, velocity, and acceleration of the trunk (see Figure 3). The light-weight design of the LMM allowed the data to be collected with minimal obstruction to the subject's movements. The LMM measured the angular movements between the pelvis and the tenth thoracic vertebra. For more information on the design, accuracy, and application of the LMM, refer to Marras *et al.*[28].

EMG activity was monitored through the use of bi-polar electrodes spaced approximately 3 cm apart at the ten major trunk muscle sites. The ten muscles of interest were: right and left erector spinae; right and left latissimus dorsi; right and left internal obliques; right and left external obliques; right and left rectus abdominis. For the standard locations of the electrode placement for these muscles, refer to Mirka and Marras[29]. The raw EMG signals were pre-amplified, high-passed filtered at 30 Hz, low-passed filtered at 1000 Hz, rectified, and integrated via a 20 ms sliding window hardware filter.

A force plate (Bertec™ 4060A, Worthington, USA) was used to measure the kinetic variables of the lifts. The subject was positioned into a pelvic support structure (PSS) that was attached to the force plate. The PSS restrained the subject's pelvis and hips in a fixed position (see Figure 4). Also, the position of L5/S1 relative to the center of the force plate remained constant for the entire experiment. The PSS permitted the trunk moments to be predicted accurately[30].

All signals from the above equipment were collected simultaneously through a customized Windows™-based software developed in the Biodynamics Laboratory. The signals were collected at 100 Hz and recorded on a 486 portable computer via an analog-to-digital (A/D) board. The data were saved by the computer for subsequent analysis.

An additional computer was used to display the instantaneous velocity recorded by the LMM in real time. The signal was transferred from the LMM to the computer through an A/D board and converted into velocity by customized software. The computer monitor was positioned directly in front of the PSS. The signals of the LMM, force plate, and EMG electrodes were input to the A/D board on the data collection rack positioned behind the PSS.

Procedure

Upon arriving at the Biodynamics Laboratory, subjects were provided with a brief description of the study and experimental protocol. Subjects read and signed a consent form. Next, anthropometric measurements were taken. The surface electrodes were then applied to the skin, using proper place-

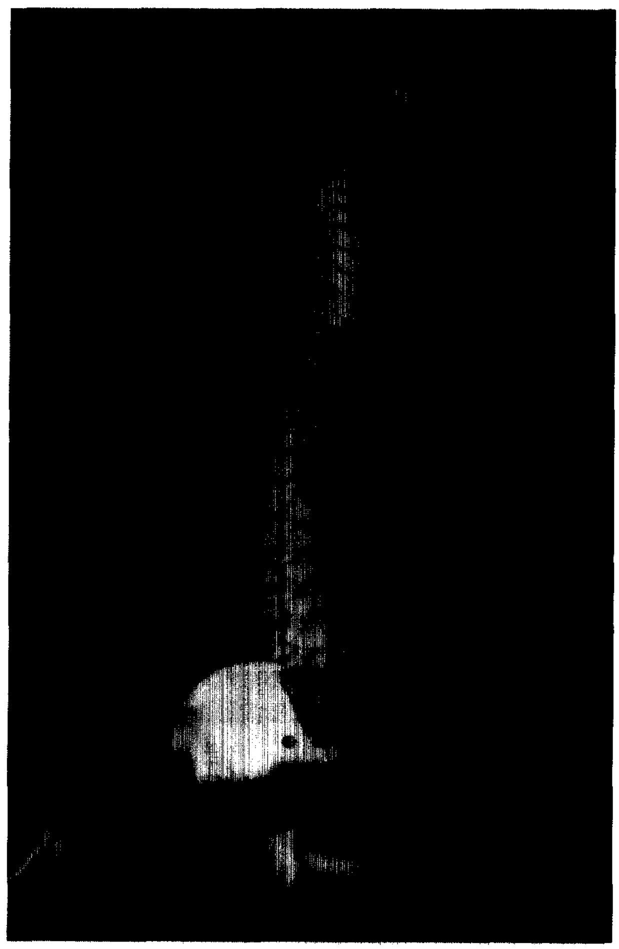

Figure 3. The LMM.

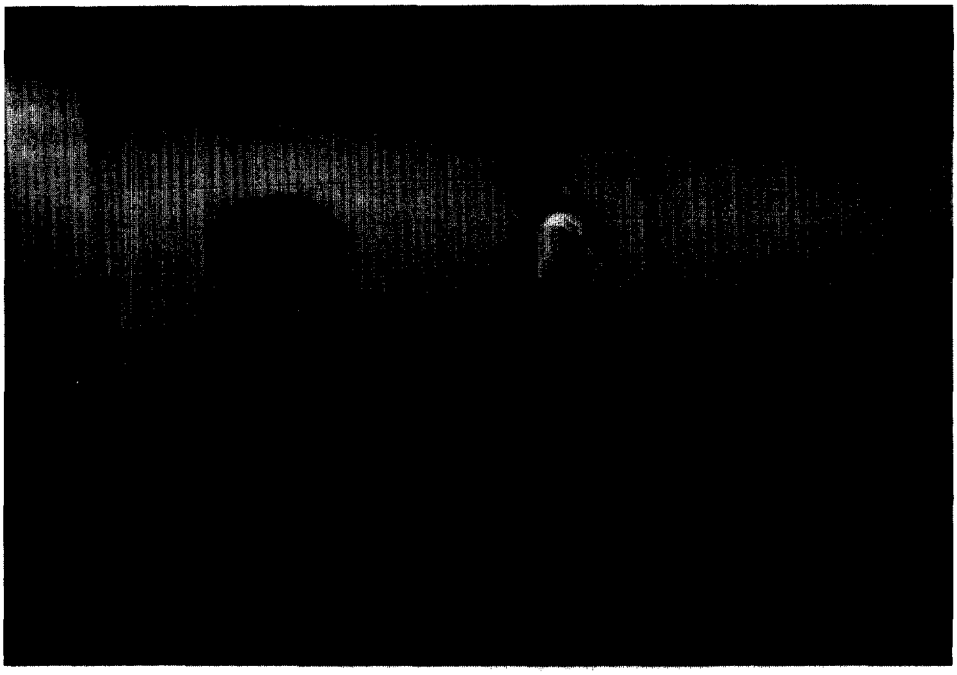

Figure 4. A subject performing an exertion for a given velocity, while positioned in the pelvic support structure.

ment procedures to sample the muscles of interest. The impedance at each electrode site was kept below 1 MΩ. The subject was then placed into a rigid structure that allowed maximum exertions to be performed. After each maximum exertion, 2 min of rest was given, in accordance with strength testing procedures[31].

After the subjects were positioned into the PSS and the LMM was attached to their back, neutrals of the LMM and force plate were collected. These neutrals were collected in order to determine the initial position of the LMM for each subject standing erect, along with the initial readings of a zeroed force plate. The subjects performed the lifting and lowering exertions. The lifts were completely randomized with respect to the independent variables. Velocities were controlled by the subject by following a trace through a given region displayed on the computer screen while attempting to keep their trunk position within defined 'tolerance'. All subjects were allowed to practice at the different velocities until they were able to remain within the tolerances. If the subject's trace fell outside the tolerance levels, the lift was repeated. A 3% tolerance was used by displaying two lines that were 1.5% above and below the target velocity. For the concentric lifts, the subjects started the lift bending forward at 40° of flexion and ended in an upright position. Conversely, the lowering exertions started in the upright position and ended when the subjects were at 40° of flexion. The computer screen displayed lines with a positive slope for the lifting exertions and a negative slope for the lowering exertions. Figure 4 shows a subject performing an experimental exertion while watching the display on the computer screen.

Data analysis

The kinematic, kinetic, and EMG data were used as inputs in the EMG-assisted spinal load model. The kinematic variables were measured and recorded by the LMM. Customized software converted the voltages into the respective angles, velocities, and accelerations of the trunk. The EMG data were normalized with respect to the maximum muscle activity of each muscle (obtained at the beginning of the study) with respect to length and velocity. The kinematic, kinetic, and EMG data were imported into the EMG-assisted model to calculate spinal forces and moments on the lumbosacral joint.

For all of the dependent variables, descriptive statistics were computed. These descriptive statistics included means, standard deviations, and maximum values. Graphical representations were utilized to help display various relationships. Analysis of variance (ANOVA) statistical analyses then were performed on all the dependent variables. For all significant independent variables, post-hoc analyses, in the form of Tukey multiple pairwise comparisons were performed to determine the source of the significant effect(s).

Results

Once the modulation factors were validated, the model was employed to evaluate the spinal loads. The ANOVA indicated that several main effects and interactions were significant (Table 2). The variable 'Task' refers to the type of lifting being performed (lowering versus lifting motion). All the main effects were significant for maximum A–P shear and compression forces. The Weight W main effect was found to be significant for all three spinal loads. The Weight–Task interaction ($W \times T$) was significant for maximum lateral and A–P shear forces. The Velocity–Task interaction ($V \times T$) was significant for A–P shear and compression forces. The Weight–Velocity interaction ($W \times V$) and the three-way interaction ($W \times V \times T$) were not significant for any of the dependent variables.

The type of lifting motion was found to effect both A–P shear and compression forces (Figure 5). Lowering exertions were found to have significantly lower A–P shear force than lifting tasks. The A–P shear forces for lowering and lifting exertions were 680.4 N (SD = 161.4) and 815.1 N (SD = 281.4) respectively. Thus, the difference between the two lifting conditions was about 135 N. During lowering motions, the maximum compression forces were greater than for lifting exertions. The lowering conditions were found to have 3269.1 N (SD = 843.3) of compression force, whereas the lifting exertions had 2665.2 N (SD = 719.6) of compression force. The difference in compression loading between lowering and lifting was close to 600 N.

Table 2. Summary of significant effects for the EMG-assisted model output in the study (values indicate p-values)

Effect	Maximum lateral force	Maximum A–P shear force	Maximum compression force
Task (T)†	0.83	0.01*	0.0004*
Weight (W)	0.002*	0.0001*	0.0001*
Velocity (V)	0.09	0.01*	0.0001*
$W \times T$	0.02*	0.001*	0.29
$V \times T$	0.88	0.003*	0.01
$W \times V$	0.65	0.14	0.31
$W \times V \times T$	0.33	0.07	0.20

*Indicates significant at $p \leq 0.05$.
†Task indicates lowering versus lifting.

Figure 5 also indicates spine tolerance limits for shear forces (1000 N)[32] and compression forces (3400 N)[33]. These are the limits of force at which it is believed the spine begins to suffer micro-fractures in the vertebral endplates and tears to the annulus fibrosis. On average, the lowering conditions approached the compression tolerance limits, whereas the lifting tasks remained well below both tolerance limits.

As expected, the spinal forces increased in all three dimensions when the weight of lift increased. The $W \times T$ interaction for A–P shear forces indicated that the differences between the weight levels depended upon the lifting motion. The difference in A–P shear forces between the three weights was larger for the lifting tasks than during the lowering conditions (Figure 6). Also, a lifting weight condition was found to be equivalent to the next higher weight level under lowering motion (e.g. the 18.2 kg-lifting condition was equal in A–P shear to the 27.3 kg-lowering condition). The 27.3 kg-lifting exertions approached the spinal tolerance for shear forces.

There was also a larger difference in lateral shear forces between the 18.2 and 27.3 kg conditions than between the 18.2 and 9.1 kg exertions for lowering conditions. The opposite was true for the lifting tasks. Additionally, the lateral shear force magnitudes were minor, compared with the A–P shear forces, with

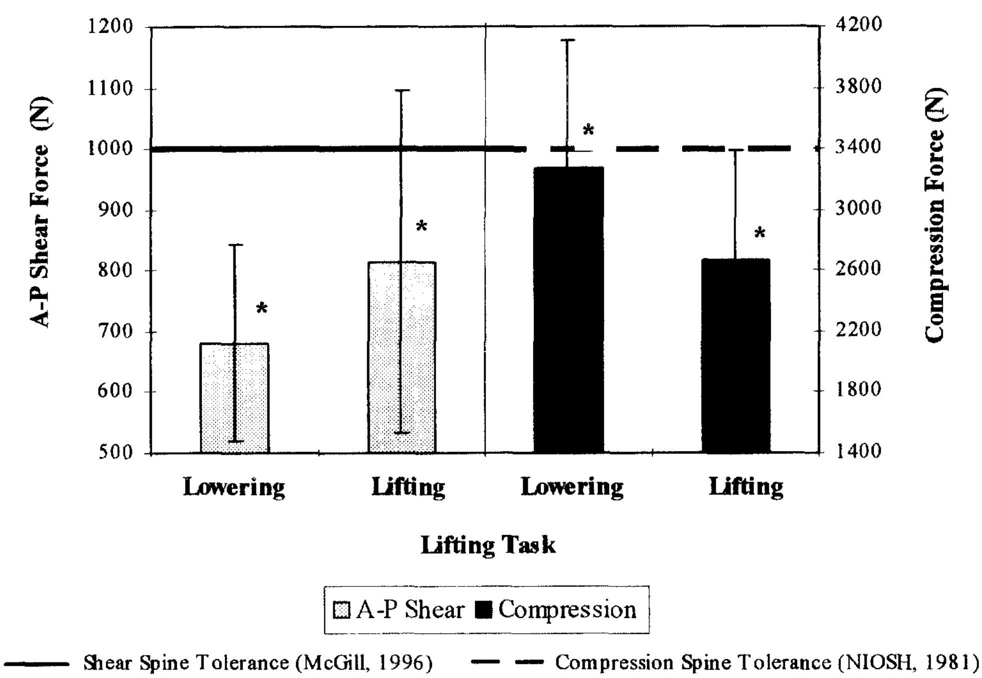

Figure 5. Maximum A–P shear and compression force as a function of lifting task.

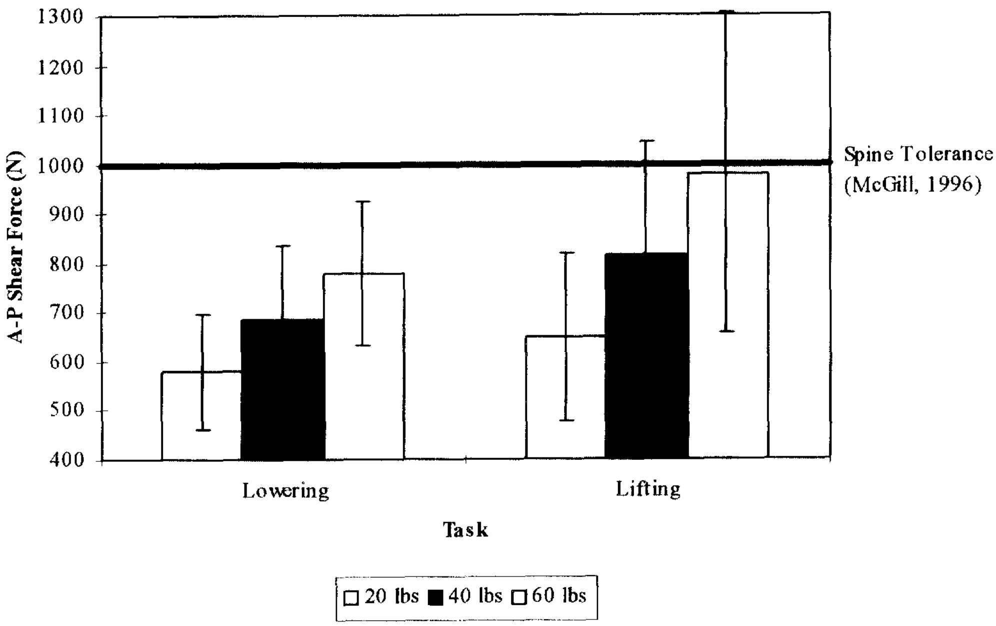

Figure 6. Maximum A–P shear force as a function of box weight and lifting task.

average values of less than 150 N. These smaller forces resulted from the lifting motion being sagittally symmetric.

Overall, the 5 deg s^{-1} velocity condition was found to be significantly lower then the other velocities for both A–P shear and compression forces. The $V \times T$ interactions for A–P shear and compression forces indicated that the differences between the velocities depended upon the type of lifting being performed. Under lowering motions there was no difference between the five velocities (Figure 7). On the other hand, the lifting tasks produced increases in A–P shear force with increases in velocity. Although none of the conditions, on average, approached the spinal tolerances for shear force, the lifting tasks had more exertions exceeding the tolerance limits, as indicated by the larger standard deviations. An increase in velocity was found to increase the compression forces during the lowering exertions (Figure 8). The only velocity found to be significantly different under the lifting conditions was the 5 deg s^{-1} velocity which was smaller. Furthermore, many of the lowering velocity

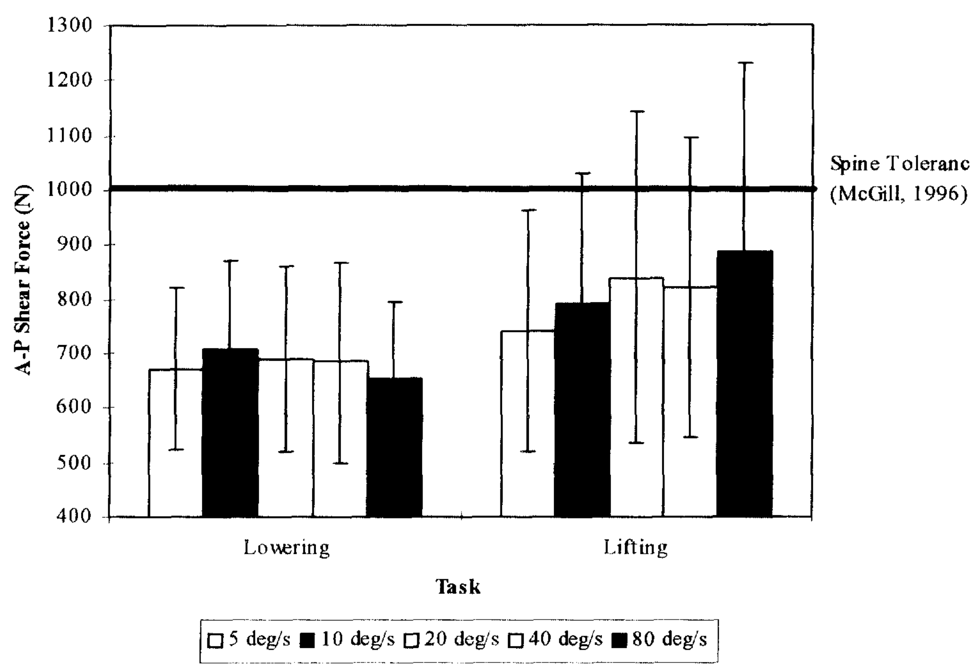

Figure 7. Maximum A–P shear force as a function of lifting velocity and lifting task.

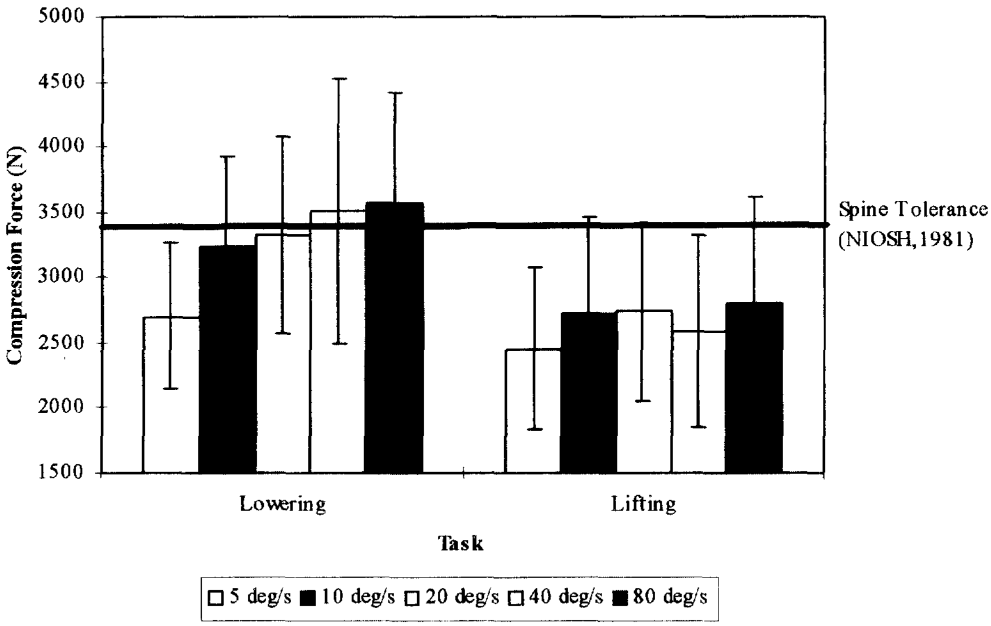

Figure 8. Maximum compression force as a function of lifting velocity and lifting task.

conditions, on average, were at or above the spinal tolerances for compression. All velocity conditions for lifting remained below the tolerance limits.

Discussion

Previous EMG-assisted models have been limited to lifting motions. However, in industrial settings, workers perform lowering movements as often as lifting tasks. Drury *et al.*[34] found that 52% of all box-handling tasks were lowering in nature. Thus, an accurate understanding of the EMG–force relationships under lowering motions must be completed, in order to evaluate completely the loading on the spine. Marras and Sommerich[12,13] and Granata and Marras[14,15] incorporated physiologic, length–strength and force–velocity relationships into the EMG-assisted model. These modulation factors were empirically derived by minimizing the variability in gain as a function of length and velocity. In the present study, similar relationships were derived for lowering using the same methodology developed by Granata and Marras[15].

The length–strength modulation found during lowering conditions had a similar relationship to that reported previously by Granata and Marras[15]. The modulation factors for the two types of motion were found to be different at the more extreme lengths of the muscle. One possible explanation for this difference is that the present study remained in a predominantly active region (e.g. 0 to 40° flexion) and, therefore, the effects of the passive tissue on the length–strength curve were not present. The elimination of the passive region of the lift would have under-predicted the length–strength curve when the muscle is at a length approximately 1.2 to 1.5 times the resting length. The resting length for the erector spinae muscles correspond to approximately 20° of sagittal flexion, which, by definition, would be the position where the muscle has the most potential for force generation[3,23]. This would result in different fits of the regression curves used to predict the length–strength modulations for both lifting conditions. Thus, since both length–strength curves are similar, either relationship would be appropriate for both lifting as well as lowering trunk motions.

The force–velocity modulation for lowering was found to be drastically different to the relation found by Granata and Marras[15] for lifting. The lowering relationship for force–velocity was found to be constant across all velocities, whereas the lifting modulation was found to be an exponential function which is represented by the Hill equation[20]. Since the modulation factors were not the same for the two types of lifting, this indicates that there is a difference in the internal mechanism for generating muscle force during the lowering and lifting. It is also interesting to note that the force–velocity modulation factor for lowering was determined to be 1.56, indicating that the subjects were, on average, 56% stronger.

It has been commonly found that the lowering strength is greater than lifting strength[6,7,35]. Reid and Costigan[6] found that the lowering strength was about 20% greater than for lifting. Marras and Mirka[35] verified this result but found it was only true for specific flexed postures. The differences in the strength estimates could be directly related to how the parameters were determined. However, there were several differences in the experimental designs and protocols. In the present study, the LMM was used to provide feedback to the subjects who then controlled the isokinetic velocities, whereas both the other studies used a trunk dynamometer Kin/Com™ to control the motion of the trunk. Additionally, Reid and Costigan[6] used a strap around the chest which resulted in the applied torque at L5/S1 being non-constant. Another difference between the Reid and Costigan[6] study and the present study was the use of different postures during the exertion. The present study had the subjects perform the exertions in an upright posture, whereas Reid and Costigan[6] had the subjects seated with their hips and knees flexed.

After the new lowering modulation factors were incorporated into the EMG-assisted model, the model performed well under both lifting and lowering exertions, as seen by stable gains, low r^2 and low AAEs. Good model performance indicates that external moments were accurately predicted and is an indirect validation of spinal loads. The lifting tasks were found to have lower compression forces but larger A–P shear forces than the lowering conditions. In addition, the lowering tasks approached the compression spinal tolerance limits, whereas the lifting exertions were well below both tolerances. Thus, the lowering conditions were found to put the individual at higher risk of a lower-back disorder.

The difference between the two types of task might have possibly resulted from either changes in external or internal circumstances. Changes in the external factors would result in different moments imposed on the spine resulting from altering the lifting style, that is increased flexion, higher velocities, and lifting the box farther away from the body. Different levels of coactivity of the trunk muscles would also contribute to higher spinal loads[36].

Upon evaluation of the sagittal trunk moments, it was found that the lowering exertions had, on average, significantly larger moments than the lifting tasks. The sagittal trunk moments were 140.5 N m for lowering tasks and 113.1 N m for lifting tasks (a difference of 27.4 N m). Thus, the subject seemed to lift the box farther away from the body during the lowering exertions. Other researchers have found lifting to result in higher trunk moments[5,8]. The difference between the studies might have resulted from different methods of controlling the exertion. In the current study the subjects were locked into the PSS, which eliminated almost all motion in the hips and pelvis and allowed the trunk motion to be easily controlled. For the other two studies the subjects were able to move the lower body, which possibly resulted in the individuals adopting a different lifting style.

In general, an increase in velocity was found to increase the spinal loading for both lifting and lowering exertions. For the lowering exertions, the compression forces were found to increase with faster velocities, whereas the A–P shear forces were found to increase with faster lifting velocities. Many researchers have found similar results for various lifting exertions[13,15,25]. The results for the lowering exertions seem somewhat counter-intuitive, since the lowering tasks would have gravity to assist. However, the level of coactivity of the trunk muscles was found to increase with faster velocities. This increase in coactivity would have resulted in direct increases in compression. Additionally, the increase in coactivity may indicate an increase in guarding against an injury.

Next, several possible limitations of this study must be addressed. First, the velocities were limited to isokinetic velocities. However, Marras and Mirka[29,37] found little influence of acceleration on muscle activity, thus indicating a limited effect on the results. Also, while it is widely known that acceleration is directly related to the force generated by the muscles, the experimental protocol already accounts for this effect by measuring the resulting dynamic moments generated during the various exertions. Second, the subjects performed the lifting and lowering exertions under sagittally symmetric conditions. The addition of asymmetry could possibly alter the empirical relationships used to estimate the muscle forces and spinal loads. Third, there was no contribution of the passive elements such as ligaments, tendons, and passive components of the muscles. Again, this would have had a limited effect on the results since the exertions remained in the active range of the muscles (sagittal flexion less than 45°). A forthcoming study will address this issue. Fourth, the exertions performed in this study were not completely free-dynamic, that is the lifting style could have been influenced by the restriction of the hips and pelvis. Future research should investigate lowering and lifting under completely free-dynamic conditions.

Conclusions

The present study has delineated the significance of lowering motion compared with lifting. These types of motion result in drastically different EMG relationships as a function of velocity, whereas the length–strength relationship was found to be similar during both lowering and lifting. When adjusting the EMG-assisted model for lowering motion, the model performed well under both lifting and lowering conditions. The lowering strength was found to be more than 50% greater than lifting; however, the compression forces were higher during lowering exertions (approaching spinal load tolerance), whereas the A–P shear forces were smaller. These findings help to better understand the spine loading and subsequent risk associated with dynamic materials handling.

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