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Effect of foot movement and an elastic lumbar back support on spinal loading during free-dynamic symmetric and asymmetric lifting exertions

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The aim of this study was to assess the effect of an elastic lumbar back support on spinal loading and trunk, hip and knee kinematics while allowing subjects to move their feet during lifting exertions. Predicted spinal forces and moments about the L5/S1 intervertebral disc from a three-dimensional EMG-assisted biomechanical model, trunk position, velocities and accelerations, and hip and knee angles were evaluated as a function of wearing an elastic lumbar back support, while lifting two different box weights (13.6 and 22.7 kg) from two different heights (knee and 10 cm above knee height), and from two different asymmetries at the start of the lift (sagittally symmetric and 60° asymmetry). Subjects were allowed to lift using any lifting style they preferred, and were allowed to move their feet during the lifting exertion. Wearing a lumbar back support resulted in no significant differences for any measure of spinal loading as compared with the no-back support condition. However, wearing a lumbar back support resulted in a modest but significant decrease in the maximum sagittal flexion angle (36.5 to 32.7°), as well as reduction in the sagittal trunk extension velocity (47.2 to 40.2° s^{-1}). Thus, the use of the elastic lumbar back support provided no protective effect regarding spinal loading when individuals were allowed to move their feet during a lifting exertion.

1. Introduction

To combat the high incidence of low back disorders (LBD) for occupational tasks that involve manual material handling (MMH), the use of industrial lumbar back supports (i.e. back supports) has become very common. The US Occupational Safety and Health Administration (OSHA), however, does not consider back supports to be personal protective equipment (PPE) for the prevention of LBD (USDOL 1990), and in a recent review of the literature, the US National Institute for Occupational Safety and Health (NIOSH) concluded that back supports do not prevent injuries to healthy workers and should not be considered PPE (USDHHS 1994).

The results of epidemiological research on back supports have been mixed regarding their protective effect (Reddell *et al.* 1992, Mitchell *et al.* 1994, Kraus *et al.* 1996, van Poppel *et al.* 1998). The studies that have assessed spinal loading during back support use for the most part have used unrealistic tasks. Several studies have used squat lifts with near maximum weights to assess the effect on spinal loading

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(McGill *et al.* 1990, Lander *et al.* 1990, Woodhouse *et al.* 1995). In our previous study (Granata *et al.* 1997), much lighter weights were used, as well as an asymmetric lift component in the lifting task. Our previous study, as well as others assessing spinal loading, were performed with the subjects standing on a force plate without moving their feet, but ours was the only study to find a significant decrease in spinal loading for some subjects when wearing a back support, and this was for only one of three back support types studied (elastic back support).

In our previous study, it was hypothesized that elastic back supports might be beneficial because the support connects the pelvis and the thorax region of the trunk. This would reduce trunk muscle coactivity and result in less spinal loading. The other back supports tested were much shorter and did not connect the thorax with the pelvis. Although slight reductions in predicted compression force and anterior-posterior shear force on the L5/S1 intervertebral disc were found when wearing an elastic back support, this may have been influenced by the nature of the experimental task, where subjects were not allowed to move their feet during the lifting exertions. This is an unrealistic task, as many MMH tasks allow individuals to move their feet.

It is hypothesized that the unrealistic nature of the task in our previous study may have influenced the observed effect on predicted spinal loading. Thus, the objective of this study was to assess spinal loading and trunk kinematics as a function of wearing and not wearing an elastic lumbar back support, with subjects lifting in a free-dynamic nature while being allowed to move their feet.

2. Methods

2.1. *Approach*

Subjects in this study performed free-dynamic lifting tasks, lifting two different weights from two different heights, both symmetrically and asymmetrically. All of these conditions were performed with and without an elastic lumbar back support. Trunk, hip, and knee angles and trunk dynamics were recorded during each of the exertions, and spinal loading was predicted through the use of a thoroughly explored dynamic three-dimensional EMG-assisted biomechanical model (Granata *et al.* 1999, Marras *et al.* 1999).

2.2. *Subjects*

Twenty male subjects participated in this study. Their ages ranged from 21 to 29 years (mean 22.8; SD 1.8), height from 165.1 to 194.4 cm (mean 179.0; SD 8.8), and weight from 59.9 to 109.8 kg (mean 75.6; SD 13.5). None of the subjects experienced low back pain at the time of the study, nor did any report a history of activity limiting chronic LBD.

2.3. *Apparatus*

The trunk kinematics (i.e. angle, velocity and acceleration) in three planes were measured by a lumbar motion monitor (LMM). This lightweight device ($\sim$1.4 kg) is an exoskeleton, which is attached to the posterior aspect of the trunk (Marras *et al.* 1992), and produces very little interference with lifting activities. Hip and knee angles were measured using electrogoniometers, with voltages converted to angles using calibration equations. The hip monitor consisted of a thin non-flexible rod that extended from the LMM to a cuff attached at mid-thigh. The rod was aligned with the longitudinal axis of the right thigh, and measured the angle of the right hip in the coronal and sagittal plane through changes in voltage of the two potentiometers.

Standing upright corresponded to 0° and positive values were assigned to hip flexion angles; hip abduction angles in the coronal plane were assigned positive angles, whereas hip adduction angles in the coronal plane were assigned negative angles. The angle of the right knee was measured using a uniaxial potentiometer, which was attached to a knee brace. The centre of the knee monitor was placed at the approximate axis of rotation of the knee. Knee angle was measured only in the sagittal plane (knee flexion), with a fully extended knee corresponding to 180°, and knee flexion angle defined as the included angle between the upper and lower leg.

Electromyographic activity from the trunk muscles was measured using Ag-AgCl surface electrodes (4 mm diameter) in a bipolar configuration. The bipolar electrodes were spaced 3 cm apart over the muscle in the direction of the line of action for the muscles, following the methods of Mirka and Marras (1993). The electrodes were connected to preamplifiers located close to the body, where they were preamplified, high- and low-pass filtered at 30 and 1000 Hz respectively, rectified and integrated via a 20 ms sliding window hardware filter. The trunk muscles sampled included the right and left pairs of the latissimus dorsi, erector spinae, rectus abdominis, external oblique, and the internal oblique. To determine the subject specific muscle gain to predict muscle force in the EMG-assisted biomechanical model, an L5/S1 locator and a force plate (Bertec, Worthington, OH, USA) were used (Fathallah *et al.* 1997). The force plate measured ground reaction forces and moments in the three planes during lifting calibration exertions necessary for the EMG-assisted biomechanical model. The L5/S1 locator consisted of electrogoniometers that track the L5/S1 position in three-dimensional space, and allowed the forces and moments measured on the force plate to be rotated and translated to the position of the L5/S1 intervertebral disc.

The signals from all equipment (i.e. force plate, L5/S1 locator, EMG electrodes, LMM, hip and knee electrogoniometers) were collected at 100 Hz using an analogue-to-digital converter.

The dimensions of the wooden box to be lifted were $22.0 \times 30.5 \times 28.5$ cm ($h \times w \times d$), with the centre of the handles 12 cm below the top of the box, on both sides of the box.

Finally, the back support was a nylon elastic support with suspenders (Chattanooga Group, Inc., Chattanooga, TN, USA). This support consisted of primary panel flaps, which overlap each other loosely across the abdomen to hold the support in place, and elastic flaps tensioned anteriorly across the abdomen and attached via Velcro to the primary panel flaps to tighten the support. Two sizes were used in this study (i.e. medium and small), in which all subjects were accommodated.

2.4. *Experimental design*

The experimental design for this study was a four-way repeated measures statistical design. Each subject, therefore, was exposed to each combination of the four independent variables. The independent variables in this study corresponded to those in our previous study (Granata *et al.* 1997). They consisted of a back support condition (with and without the back support); beginning height of the lift (handles of the box at knee height, and handles 10 cm above knee height); direction of the starting position of the box with respect to the subject (0° asymmetry—sagittally symmetric, and 60° asymmetric to the right of the subject); and box weight (13.6 and 22.7 kg). The dependent variables included: (1) the predicted maximum spinal forces on the L5/S1 intervertebral disc (i.e. compression force, anterior/posterior and lateral shear forces); (2) predicted maximum moments about the L5/S1 intervertebral disc in the three planes; (3) trunk kinematics (position, velocity, acceleration) in the three planes; (4) the maximum right hip angle in the sagittal and coronal plane; and (5) the maximum sagittal plane right knee angle.

2.5. *Experimental task*

The task consisted of lifting a box using any style of lifting the subject preferred. Subjects were instructed to imagine they were working on an assembly line, where their job was to lift a box from two different origins and lift to a height such that the handles were at about elbow height, simulating a conveyor height. With the subjects not being restrained to the force plate (i.e. allowed to step freely), each lifting motion started with the feet at the same starting location, and each lift ended with the feet at the same ending location, irrespective of the beginning lift height, weight of the load and asymmetric angle of the origin of lift. Thus, for each lift, the subject would step toward the load (either 65 cm directly in front of the feet of the subject, or 65 cm away at 60° to the right) (figure 1), and lift the box such that the handles were at elbow height, while ending with the feet at the ending position, 65 cm directly in front of the starting location (figure 2). After each lift, at least 30 s of rest was allowed until the initiation of the next lift. The presentation of the back support condition was counterbalanced among the subjects, and all combinations of the beginning height, weight of the load and direction of the lift were randomly presented within each back support condition, with two replications per combination.

2.6. *Procedure*

After arrival to the laboratory, the subjects were briefed on the study, read and signed an informed consent form, followed by the recording of anthropometric measures. The subjects' skin was then prepared for application of the EMG electrodes over the ten trunk muscles of interest (Marras 1990, USDHHS 1992, Mirka and Marras 1993), keeping the skin resistance < 500 kΩ. Foam spacers with circular cut-outs were then placed over the electrodes to protect against contact distortions from the back support. Following the application of the electrodes, maximum voluntary contractions (MVC) of the trunk muscles were then performed to be used for EMG normalization of the experimental tasks. Isometric MVC were elicited in six directions: trunk extension with the trunk flexed 20°, and trunk flexion, right and left lateral bending, and right and left trunk twisting performed in an upright standing posture, with 2 min of rest between each MVC to reduce the effect of fatigue (Caldwell *et al.* 1974). The subjects were stabilized about the hip to isolate the exertions to the trunk, as well as stabilized about the chest and shoulder.

The EMG-assisted biomechanical model used to estimate spinal loading requires calibration exertions using a force plate and an L5/S1 locator to determine subject-specific muscle gain. A set of calibration exertions was performed before each of the back support conditions.

The LMM was attached to a harness strapped around the upper torso and to an orthoplast mould around the hips. The orthoplast hip mould is placed on the subject such that the top of the mould is located near the L5/S1 level of the spine, which then extends caudally. Thus, the LMM attachment system does not cover the lumbar area and does not act as a back support itself. For the back support condition, the back support was applied first, with the elastic straps tensioned to 44.5 N as in Granata *et al.* (1997), followed by the application of the LMM upper torso harness, orthoplast hip mould, and the LMM.

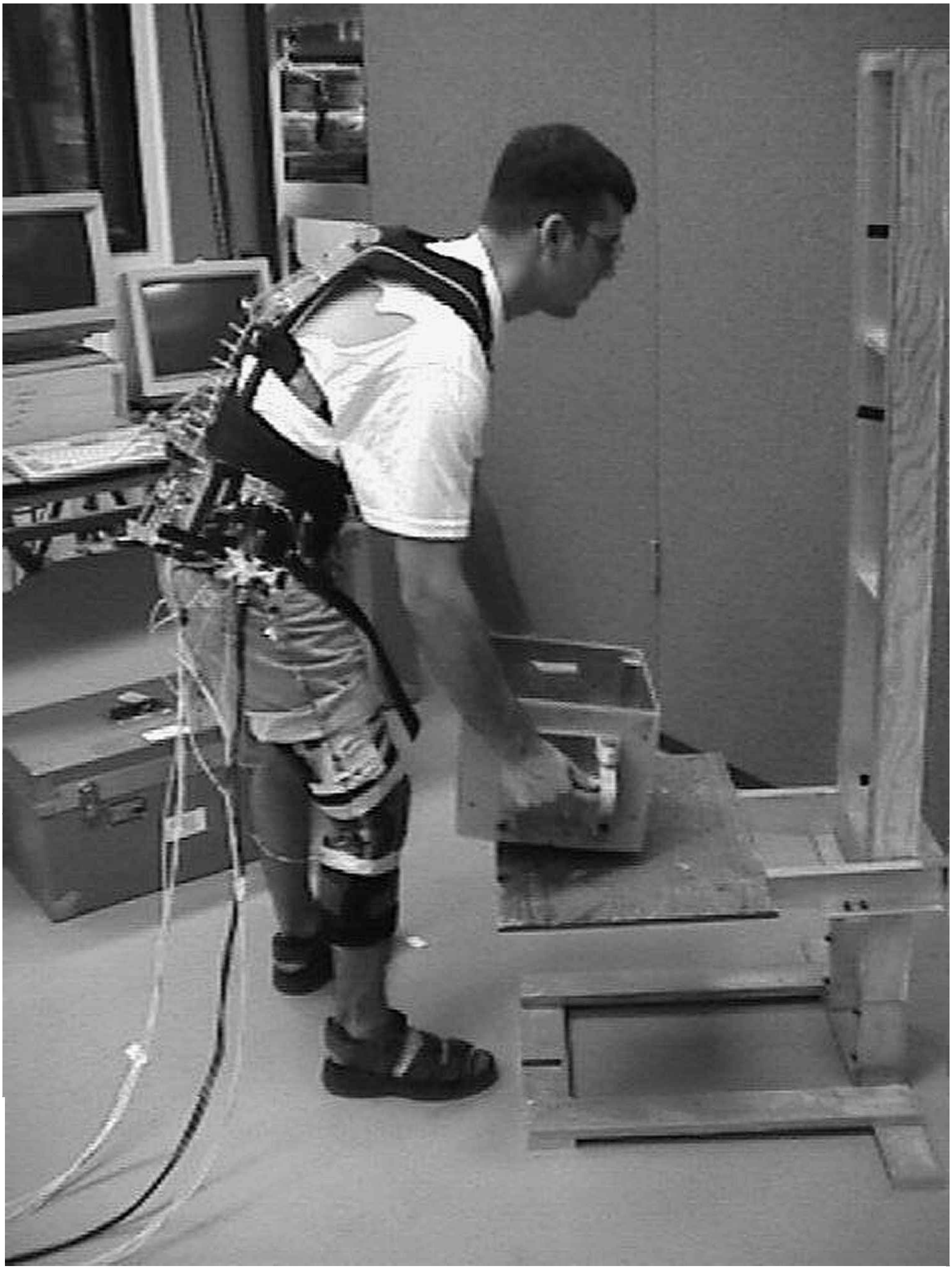

Figure 1. Subject shown during the beginning of the 60° asymmetric lifting exertion from knee height.

2.7. *Data analysis and biomechanical model*

Instantaneous trunk, hip and knee angles for each exertion were determined via calibration equations converting voltage to angles using custom conversion software. All dynamic data, including kinetics, kinematics and EMG were smoothed via a 10 Hz Hanning weighted time domain filter within an EMG-assisted biomechanical

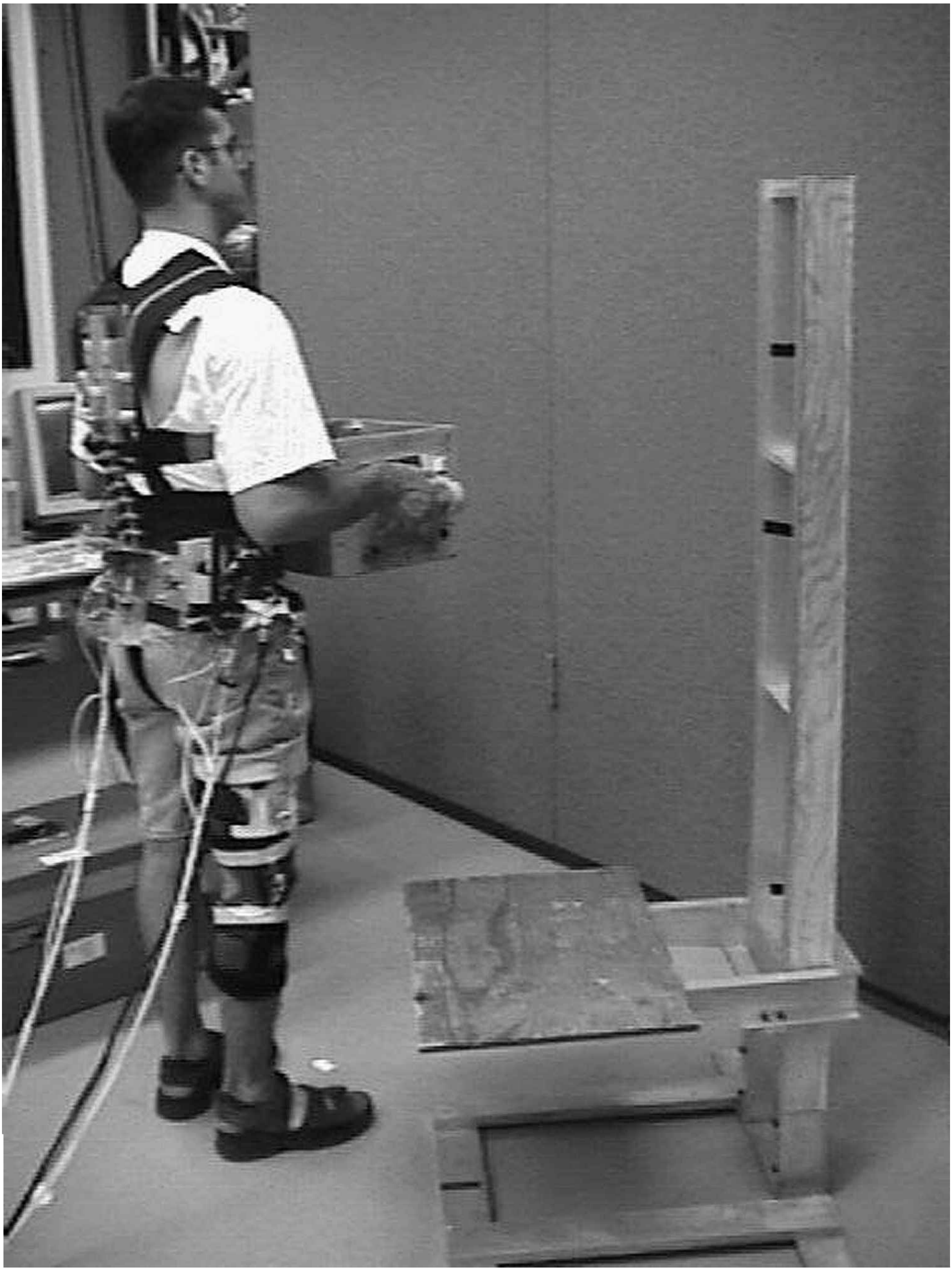

Figure 2. Subject shown at the ending position of a lifting exertion.

model (Granata and Marras 1995a). Trunk velocities and accelerations were derived from the instantaneous trunk position data.

The EMG and trunk kinematic data were imported into an EMG-assisted spinal loading model that has been developed over the last decade in the Biodynamics Laboratory (Marras and Reilly 1988, Reilly and Marras 1989, Marras and Sommerich 1991a,b, Granata and Marras 1993, 1995a, b, Marras and Granata

1995, 1997a,b, Granata *et al.* 1999, Marras *et al.* 1999). The model has been thoroughly evaluated under forward trunk bending (Marras and Sommerich 1991a, b, Granata and Marras 1993, 1995a), trunk twisting (Marras and Granata 1995), and lateral trunk bending (Marras and Granata 1997a) motions. Generally, the EMG-assisted biomechanical model measures EMG activity to predict muscle forces acting on the L5/S1 intervertebral disc. Given the trunk geometry (moment-arms and muscle cross-sectional area derived from subject anthropometry), the model evaluates instantaneous spinal loading by summing the major muscle groups' force contributions in each direction (compression and lateral and anterior/posterior shear). Muscle force is assessed by considering the relative amount of EMG activity (percentage of maximum) in a muscle and multiplying this value by the cross-sectional area of the muscle and the muscle gain (force per unit area). Both the muscle length–strength and force–velocity relations then modulate muscle force. Individual muscle gain for each back support condition was determined using the equipment described in Fathallah *et al.* (1997). Subjects performed five calibration exertions, lifting a 22.7 kg load from the knee height to elbow height, keeping the legs relatively straight while standing on a force plate. The measured forces and moments were translated and rotated from the centre of the force plate to the L5/S1 (Fathallah *et al.* 1997). The predicted internal moments at L5/S1 were then adjusted to equal the external moments through the use of the predicted gain factor. This gain factor was then used to estimate the muscle forces and internal moments for the experimental task, which allowed the subjects to move without being restricted to a force plate.

The performance of the biomechanical model was assessed for each of the back support conditions via several measures. However, since the experimental tasks were performed without the subjects standing on a force plate (i.e. 'open-loop'), the model performance was assessed from the calibration trials that were performed to estimate the individual specific muscle gain. First, the trunk moment predicted by the model was compared with the measured moment. This comparison is made by evaluating the r^2 statistic, which assessed the trend in the changing applied moment. Second, the average absolute error (AAE) between the predicted moment and measured moment was assessed for the magnitude of the difference. And third, the muscle gain was assessed to ensure that the gain was physiologically reasonable. To be physiologically valid, the predicted gain must fall within the range 30 – 100 N/cm^2 (Weis-Fogh and Alexander 1977, Reid and Costigan 1987).

2.8. *Statistical analysis*

Descriptive statistics were generated, consisting of the maximum values for each exertion for the trunk kinematics (i.e. angle, velocity and acceleration), maximum hip and knee angles, and spinal forces and moments in each of the three planes, averaged across all 20 subjects. The effect of the independent variables on the dependent variables was assessed via a four-way repeated measures analysis of variance (ANOVA). Significant effects were assessed using Tukey *post-hoc* comparisons. All statistical analyses were performed by SAS statistical software (Cary, NC, USA), while controlling for a test-wise Type I error using $\alpha = 0.05$.

3. Results

3.1. *Model performance and spinal loading*

As shown in table 1, the mean and median muscle gains across all subjects were very similar between the back support conditions and within the physiological range, and the r^2's and AAE for both back support conditions were adequate. These measures indicated that the biomechanical model performed extremely well during the calibration phase of the experiment, as well as increasing confidence in the spinal loading predictions during the experimental tasks.

The effect of wearing a back support resulted in no significant differences for any of the spinal loading variables (tables 2 and 3). Only the twist moment resulted in a significant decrease as a function of wearing the back support. None of the interactions between the back support condition and the other three independent variables resulted in significant differences in spinal loading.

3.2. *Trunk kinematics*

Wearing a back support resulted in significant decreases in trunk angle, velocity and acceleration (table 4). Peak trunk bending in the sagittal plane decreased 3.8° from 36.5 to 32.7° (table 5), and trunk rotation in the transverse plane decreased from 3.9 to 1.6° as a function of wearing the back support. Trunk bending in the coronal plane changed as a function of asymmetric angle of the load position and of whether the subject was wearing a back support. For the sagittally symmetric lifts (0° asymmetry), trunk bending decreased slightly from 1.4 to 1.0° when not wearing and wearing a back support respectively, whereas, for the 60° asymmetric lifting condition, the peak trunk angles in the coronal plane decreased from 3.9° when not wearing a back support to 2.2° when wearing a back support. Wearing a back support also decreased the trunk extension velocity and acceleration in the sagittal plane (14.8 and 11.6% decrease respectively), as well as decreasing of twisting velocity by 1.7° s^{-1} (table 5).

Table 1. Biomechanical model performance resulting from the calibration exertions as a function of the back support condition.

	No back support			Back support		
Model evaluation Variable	Gain (N/cm^2)	r^2	AAE (N)	Gain (N/cm^2)	r^2	AAE (N)
Mean	56.0	0.84	23.6	62.9	0.85	21.6
Median	53.4	0.86	19.5	59.0	0.88	20.4
SD	21.5	0.09	13.2	24.0	0.10	9.5

Table 2. Analysis of variance (ANOVA) results (p) of independent variables for spinal loading. Italic cells represent significant differences at the $p \leq 0.05$ level.

	Moment				Force on L_5/S_1		
Independent variable	Sagittal plane	Coronal (lateral) plane	Transverse (twist) plane	Resultant	Lateral shear force	A/P shear force	Compression force
Back support (B)	0.8932	0.2612	*0.0201*	0.3626	0.5230	0.5029	0.4287
Asymmetery (A)	0.0860	*0.0001*	*0.0001*	*0.0035*	*0.0004*	0.2075	0.2030
Weight (W)	*0.0001*	*0.0006*	*0.0001*	*0.0001*	*0.0001*	*0.0001*	*0.0001*
Height (H)	*0.0001*	0.0893	0.0654	*0.0001*	0.8403	*0.0451*	*0.0001*
B × A	0.6528	0.2440	0.0918	0.1024	0.7417	0.9812	0.7374
B × W	0.3526	0.7960	0.7114	0.5775	0.7121	0.5508	0.3862
B × H	0.0924	0.5345	0.2940	0.1064	0.3936	0.1817	0.0745

Table 3. Descriptive statistics (mean and SD) for spinal loading variables. Italic cells represent significant differences at the $p \leq 0.05$ level.

Independent variable		Sagittal moment (Nm)	Lateral moment (Nm)	Twist moment (Nm)	Resultant moment (Nm)	Lateral shear force (N)	A/P shear force (N)	Compression force (N)
Back support	No	199.6	50.6	*65.5*	220.7	306.6	811.4	4269.2
		(55.4)	(28.9)	*(43.5)*	(61.6)	(186.6)	(228.7)	(1046.9)
	Yes	198.7	46.5	*51.8*	214.2	321.7	833.2	4164.3
		(52.8)	(26.7)	*(37.0)*	(57.4)	(197.4)	(215.0)	(1058.7)
Asymmetry (°)	0	201.8	*37.1*	*44.3*	*212.0*	*263.4*	816.6	4257.8
		(55.9)	*(19.5)*	*(27.8)*	*(58.3)*	*(145.9)*	(218.1)	(1109.1)
	60	196.6	*60.0*	*72.9*	*222.9*	*364.9*	827.9	4175.7
		(52.1)	*(30.2)*	*(46.6)*	*(60.5)*	*(217.8)*	(226.2)	(994.5)
Weight (kg)	13.6	*180.0*	*45.2*	*52.9*	*197.0*	*279.7*	*751.4*	*3830.9*
		(43.6)	*(26.4)*	*(37.0)*	*(49.0)*	*(167.7)*	*(187.1)*	*(836.2)*
	22.7	*218.3*	*51.9*	*64.4*	*237.9*	*348.6*	*893.1*	*4602.6*
		(56.7)	*(28.9)*	*(43.9)*	*(62.3)*	*(208.3)*	*(231.7)*	*(1106.3)*
Height	Knee	*210.5*	50.1	61.0	*229.0*	313.0	*829.4*	*4492.1*
		(57.3)	(28.2)	(42.2)	*(62.8)*	(190.2)	*(226.4)*	*(1091.4)*
	> Knee	*187.8*	47.0	56.3	*205.9*	315.2	*815.1*	*3941.5*
		(48.1)	(27.5)	(39.6)	*(53.9)*	(194.1)	*(217.8)*	*(937.8)*

3.3. *Hip and knee angles*

As shown in tables 6 and 7, wearing a back support resulted in a significant increase in the peak hip flexion angle (41.5 to 44.2°). This increase in peak hip angle was offset by a non-significant 2.1° decrease in the peak knee flexion angle (25.7 to 23.6°).

4. Discussion

The major findings of this study include the lack of significant differences in spinal loading, the increase in the peak hip angle in the sagittal plane as a function of wearing the back support, and differences in trunk kinematics observed between the back support and no back support conditions.

The use of the elastic back support had a non-significant influence on the magnitudes of the spinal loading variables typically used to assess the risk of LBD. Similar to the findings in our previous study, individual differences were present. The anterior/posterior (A/P) shear force and compression force increased in 12 and nine subjects respectively, when wearing the back support versus not wearing the back support. The largest increase and decrease in predicted compression force experienced by individual subjects when wearing the back support were 1091.0 and 1182.2 N respectively. Similarly, the largest increase and decrease in predicted A/P shear force were 280.1 and 260.5 N respectively when wearing the back support. Thus, although individual differences were present and the overall effect was non-significant, clearly the variability in spinal loading resulting from wearing a back support may place certain individuals at higher risk of LBD than when not wearing a back support. These findings also suggest that the slight decreases in spinal loading found in Granata *et al.* (1997) were influenced by the nature of the experimental task. The main difference between these two studies is that in our previous study the subjects' feet were stationary on a force plate and were not allowed to move, whereas in the current study, the subjects were allowed to move and step during the lifting exertions. Thus, allowing an individual to move their feet during a lifting exertion

Table 4. Analysis of variance (ANOVA) results (p) of independent variables on trunk angle, velocity and acceleration. Italic cells represent significant differences at the $p \leqslant 0.05$ level.

Independent variable	Trunk angle: Sagittal plane	Trunk angle: Coronal (lateral) plane	Trunk angle: Transverse (twist) plane	Trunk velocity: Sagittal plane	Trunk velocity: Coronal (lateral) plane	Trunk velocity: Transverse (twist) plane	Trunk acceleration: Sagittal plane	Trunk acceleration: Coronal (lateral) plane	Trunk acceleration: Transverse (twist) plane
Back support (B)	*0.0031*	0.2261	*0.0409*	*0.0001*	0.8537	*0.0302*	*0.0013*	0.8245	0.6896
Asymmetry (A)	0.7521	*0.0001*	*0.0010*	*0.0001*	0.1255	0.3349	*0.0014*	0.4626	0.0528
Weight (W)	*0.0006*	0.1413	0.1119	*0.0031*	0.0669	0.7751	*0.0001*	0.1602	0.7021
Height (H)	*0.0001*	0.4823	*0.0058*	*0.0001*	0.8201	0.0978	*0.0004*	0.6969	0.4495
B × A	0.7119	*0.0022*	0.1410	0.2728	0.3189	0.1075	0.2305	0.5732	0.6805
B × W	0.1471	0.9137	0.6864	0.5425	0.3205	0.9608	0.8516	0.8750	0.7435
B × H	0.2601	0.0891	0.1911	0.7832	0.8261	0.9231	0.7481	0.6550	0.4628

Table 5. Descriptive statistics (mean and SD) for trunk angle and motion variables. Italic cells represent significant differences at the $p \leqslant 0.05$ level. The sagittal angle represents forward trunk flexion from upright neutral. Lateral trunk bending to the right is assigned a positive value, and lateral trunk bending to the left is assigned a negative value. Twisting to the right is assigned a negative value, and twisting to the left is assigned a positive value.

Independent variable		Sagittal angle (°)	Lateral angle (°)	Twist angle (°)	Sagittal velocity (°/s)	Lateral velocity (°/s)	Twist velocity (°/s)	Sagittal acceleral (°/s^2)	Lateral acceleral (°/s^2)	Twist acceleral (°/s^2)
Back	No	*36.5*	2.7	*– 3.9*	*47.2*	– 1.4	*1.2*	*104.8*	2.5	1.1
Support		*(6.8)*	(3.8)	*(4.7)*	*(14.8)*	(8.5)	*(7.0)*	*(39.4)*	(35.2)	(24.8)
	Yes	*32.7*	1.6	*– 1.6*	*40.2*	– 1.6	*– 0.5*	*92.7*	3.0	0.6
		(7.0)	(3.7)	*(3.5)*	*(13.5)*	(6.9)	*(4.8)*	*(34.6)*	(28.0)	(17.7)
Asymmetry	0	34.7	*1.2*	*– 1.4*	*47.9*	– 0.7	– 0.3	*105.9*	4.2	– 1.2
(°)		(7.2)	*(3.4)*	*(3.8)*	*(14.9)*	(6.9)	(4.5)	*(39.8)*	(26.1)	(16.6)
	60	34.5	*3.1*	*– 4.1*	*39.5*	– 2.3	1.0	*91.5*	1.3	2.9
		(7.1)	*(3.9)*	*(4.4)*	*(12.9)*	(8.5)	(7.2)	*(33.7)*	(36.6)	(25.4)
Weight	13.6	*34.3*	*2.2*	– 2.9	*44.9*	– 1.9	0.3	*105.5*	0.0	0.5
(kg)		*(6.9)*	(3.7)	(4.3)	*(14.5)*	(7.6)	(5.7)	*(37.9)*	(30.8)	(20.5)
	22.7	*34.9*	2.0	*– 2.6*	*42.5*	– 1.1	0.4	*91.9*	5.5	1.2
		(7.3)	(3.9)	*(4.3)*	*(14.6)*	(7.9)	(6.4)	*(36.1)*	(32.6)	(22.6)
Height	Knee	*37.2*	2.2	*– 2.6*	*46.6*	– 1.5	0.8	*102.5*	3.3	0.3
		(6.9)	(3.9)	*(4.3)*	*(14.8)*	(7.8)	(6.2)	*(37.3)*	(32.0)	(22.4)
	> Knee	*32.1*	2.1	*– 2.9*	*40.7*	– 1.6	0.1	*94.9*	2.2	1.4
		(6.5)	(3.6)	*(4.3)*	*(13.8)*	(7.7)	(5.8)	*(37.5)*	(31.6)	(20.7)

Table 6. Hip and knee angle variable p. Italic cells represent significant differences at the $p \leqslant 0.05$ level.

Independent variable	Maximum hip flexion angle in the sagittal plane	Maximum hip angle in the coronal plane	Maximum knee flexion angle in the sagittal plane
Back support (B)	*0.0493*	0.9761	0.3002
Asymmetry (A)	*0.0002*	*0.0071*	0.6970
Weight (W)	*0.0001*	0.1437	*0.0002*
Height (H)	*0.0001*	0.0528	*0.0001*
B × A	0.2331	0.4860	0.8143
B × W	0.8634	0.1777	0.5224
B × H	0.7864	0.9252	0.2797

Table 7. Descriptive statistics (mean and SD) for hip and knee angle variables. Italic cells represent significant differences at the $p \leqslant 0.05$ level. For hip angles in the coronal plane, positive angles correspond to hip abduction, and negative angles correspond to hip adduction in the coronal plane. The knee flexion angle is defined as the included angle between the lower leg and upper leg.

Independent variable		Maximum hip flexion angle in Sagittal plane	Maximum hip angle in coronal plane	Maximum knee flexion angle in sagittal plane
Back support	No	*41.5*	– 3.0	154.3
		(11.0)	(3.5)	(12.2)
	Yes	*44.2*	– 3.0	156.4
		(12.0)	(2.5)	(12.8)
Asymmetry (°)	0	*41.5*	*– 2.4*	155.6
		(11.2)	*(2.6)*	(13.4)
	60	*44.3*	*– 3.6*	155.1
		(11.9)	*(3.4)*	(11.6)
Weight (kg)	13.6	*42.0*	– 3.1	*156.3*
		(11.6)	(3.1)	*(12.6)*
	22.7	*43.8*	– 2.9	*154.4*
		(11.5)	(3.0)	*(12.5)*
Height	Knee	*45.8*	– 2.8	*153.6*
		(11.5)	(3.1)	*(12.9)*
	> Knee	*39.9*	– 3.2	*157.2*
		(10.9)	(3.1)	*(11.9)*

may negate any positive effect the back support may have had on spinal loading via reductions in trunk muscle coactivity.

Another indicator of risk of occupational LBD is the peak sagittal moment, where increases in the peak sagittal moment have been associated with an elevated incidence rate of LBD (Chaffin and Park 1973, Marras *et al.* 1993, 1995). Consistent with the findings in our previous study, the peak sagittal moment in the present study was not influenced by the use of an elastic back support as compared with the no-back support condition. In fact, the peak moment in the sagittal plane was almost identical in both of the back support conditions (with and without). Although intra-abdominal pressure (IAP) was not measured in this study, if it is assumed that IAP increased as a result of wearing the back support, then the hypothesis of IAP generating an extensor moment and relieving the extensor muscles from producing the full amount of the moment (Morris *et al.* 1961) is not supported, consistent with the findings that increases in IAP had very little beneficial mechanical effect on spinal loading (Marras and Mirka 1996). This is further indicated by the lack of significant decrease in compression and shear forces on the L5/S1 intervertebral disc. Thus, the elastic back support did not have a beneficial biomechanical effect regarding spinal loading.

A possible beneficial effect from the use of back supports, as found in this study and consistent with the findings of others (Lavender *et al.* 1995, Granata *et al.* 1997, Sparto *et al.* 1998), was the reduction in trunk kinematics. As indicated by Marras *et al.* (1993, 1995), increases in forward trunk bending in the sagittal plane and trunk velocities in the transverse and lateral planes were associated with an elevated risk of LBD in industry. The use of an elastic back support in this study resulted in significant decreases in trunk position and velocities in both the sagittal and transverse plane. Granata *et al.* (1997) found decreases in both trunk position and extension velocity in the sagittal plane, as did Sparto *et al.* (1998). Caution is warranted, however, in extrapolating the decreases in kinematics observed from these studies as an indication of reduction of risk of LBD. The epidemiological study by Marras *et al.* (1993, 1995) was performed on industrial workers not wearing back supports of any kind. Thus, the motions observed in studies using back supports may not be applicable to those observed in Marras *et al.* (1993, 1995) when assessing risk of LBD, as back supports may influence other mechanisms of LBD independent of those attributable solely to trunk motions.

The observed 3.8° decrease in the peak trunk bending in the sagittal plane, and thus any potential benefit derived from wearing a back support, however, may be offset by a 2.7° increase in the peak hip angle in the sagittal plane. Although these small changes may have a negligible biomechanical effect, this pattern is consistent with results from other studies (Sparto *et al.* 1998, Granata *et al.* 1997), where tradeoffs for angles and velocities were observed between the trunk and the hip and pelvis as a result of wearing an elastic back support. Furthermore, Sparto *et al.* (1998) observed a decrease in mechanical work about the trunk, with an increase in mechanical work about the hip. Collectively, the findings from these studies may indicate that any changes in spinal loading which may have occurred as a result of a more upright trunk or decreases in trunk motion may result in an increased loading on the hip, and thus a transfer of a potential musculoskeletal problem from one joint to another.

At first glance, one would expect that the collective influence of the changes in trunk kinematics would have resulted in decreased spinal loading. Changes in trunk extension and twist velocity have been observed to influence antagonistic muscle coactivity, which in turn, results in changes in compression on the L5/S1 joint (Marras and Mirka 1992, Granata and Marras 1995b, Marras and Granata 1995). Similarly, decreases in sagittal trunk flexion would result in decreased extension moments, which in turn would result in a decrease in spinal loading. Although decreases in trunk kinematics when wearing the elastic back support were observed, the sagittal extension moment and compression force did not decrease. This suggests that a redistribution of the muscle loading magnitudes may have occurred as a result of wearing the back support, as was indicated in our previous study (Granata *et al.* 1997).

The changes in measures of trunk kinematics and spinal loading resulting from using a back support can be contrasted against the results of trunk kinematics and spinal loading from changes in the other independent variables. Decreasing the weight from 22.7 to 13.6 kg resulted in a 5.7% increase in sagittal trunk extension velocity, and a 16.7, 17.5 and 17.5% decrease in compression force, A/P shear force and sagittal extension moment respectively. Similarly, increasing the height of the beginning of the lift from the knee to 10 cm above the knee resulted in a 13.7% decrease in the maximum trunk flexion angle in the sagittal plane, a 12.7% decrease in the trunk extension velocity in the sagittal plane, and a 12.2 and 10.8% decrease in compression force and sagittal extension moment respectively. Thus, although the changes in trunk kinematics were similar when wearing the back support as when decreasing the weight or raising the beginning height of the lift, the resulting decrease in compression force and trunk extension moment in the sagittal plane were much greater when changing the weight or the beginning height of the lift, as opposed to that experienced when wearing a back support.

The results of this study may be more applicable to an industrial setting than previous laboratory studies. Many of the previous studies that assessed spinal loading were performed with the subjects restricted to a force plate, thus not allowing the subjects to move. It is questionable whether the results from these prior studies can be generalized to activities in the workplace, as a majority of MMH tasks are performed when moving the feet. In a cross-sectional database on MMH activities of over 400 manufacturing jobs collected by Marras *et al.* (1993, 1995), a random survey of 100 jobs involving lifting indicated that 87% of the these jobs involved movement of the feet when handling material. Additionally, the loads investigated in previous studies were all extremely high. Lander *et al.* (1990, 1992) investigated squat lifts using loads of 90% of the subject's one time maximum squat lift, where the mean maximum was 128 kg. McGill *et al.* (1990) investigated squat lifts using loads between 72 and 91 kg; and Harman *et al.* (1989) also used loads that were 90% of a one time maximum dead lift, where the mean maximum across subjects was 143 kg. However, Marras *et al.* (1993, 1995) found that the mean maximum load handled in high-risk jobs (incidence rate of 12.0 per 100 persons per year, with a mean of 26.3) was 10.6 kg. Thus, the extremely high magnitudes of loads in the previous studies are not reflective of the weights of the loads typically handled in industry.

The results of this study should be interpreted with consideration of several limitations. First, this was a controlled laboratory study, thus the task is not a mirror image of those found in industry, where there may be time pressures to keep up the pace. Every attempt was made, however, to investigate a realistic task, including load weights that are more reflective of those in industry, allowing subjects to move their feet, and allowing them to utilize a freestyle lifting procedure. Second, the subjects used in this study were young college aged males, who were not experienced in MMH activities. Thus, it remains to be seen if these results would apply to a female population or to an older population who may be more experienced in MMH tasks. Third, although there are a multitude of back supports currently on the market, only one style of back support was investigated in this study (an elastic lumbar support), and results may be different for other styles of back supports, such as those that are more rigid. However, this style of back support was chosen, as it is the most popular style used in industry, and in previous studies, has been shown to influence spinal loads when compared with other styles of back supports.

5. Conclusions

Wearing an elastic lumbar back support during a realistic manual materials handling task where subjects were allowed to move their feet during the lifting exertion resulted in no difference in spinal loading about the L5/S1 intervertebral disc as compared with exertions without an elastic lumbar back support. Thus, it is concluded that during manual materials handling activities where individuals can move their feet, lumbar back supports do not decrease the risk of LBD resulting from loading on the low back.

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