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Effects of a Lifting Belt on Spine Moments and Muscle Recruitments After Unexpected Sudden Loading

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Study Design. Ten men and eight women participated in a repeated-measures experiment in which sudden loads were applied unexpectedly to a container held in the hands. Three independent variables were investigated: lifting belt use, preload, and load symmetry.

Objectives. To determine whether a lifting belt would help protect the spine in sudden symmetric and asymmetric loading situations.

Summary of Background Data. Unexpected loading events have long been associated with the onset of back pain. Based on work showing that lifting belts restrict motion of the torso, the hypothesis was that a lifting belt would stiffen the spine, thereby protecting its supporting tissues.

Methods. A weight, equal to 7.5% of the subjects' trunk extension force, was allowed to fall 1 m before the bottom of a box held by blindfolded subjects was pulled. Kinetic and kinematic data, obtained from two force plates and a magnetic motion measurement system, were used in a three-dimensional, dynamic, linked-segment biomechanical model to calculate spine moments. Electromyogram data were simultaneously obtained from eight trunk muscles.

Results. The belt reduced the forward bending of the spine during the symmetric loadings. In the men, the belt also reduced the forward flexion moment acting on the spine. The belt restricted lateral bending in the women and men, when the box was preloaded. The peak electromyogram amplitudes from posterior contralateral erector spinae and latissimus dorsi muscles increased during the asymmetric loadings, whereas three ipsilateral muscles were less active.

Conclusions. The conflicting moment and electromyographic results, combined with the influence of load symmetry, preload, and gender make the benefits of the lifting belt difficult to delineate. Although the data support the hypothesis that the belt stiffens the torso's response to sudden loading, the effects are small, and considerable individual differences exist. The findings show that during unexpected sudden loading, a belt may reduce the net external moment loading. At the same time the belt appears to alter the muscle response strategy so that the belt's overall effect on an individual's safety is hard to determine. [Key words: back support, ergonomics, lifting belt, low back pain, sudden loading] **Spine 2000;25:1569–1578**

Magora[22] reported an epidemiologic link between the frequency of sudden maximal efforts, especially when unexpected, and the occurrence of occupational low back pain. Similarly, Manning et al[23] determined that 66% of the back injuries recorded in an industrial setting were preceded by some type of underfoot accident. Often these injuries were initiated by slips without falls. This suggests an injury scenario in which the neuromuscular system overreacts to an unanticipated event, perhaps in a poorly coordinated manner, and in the process damages tissue. A similar scenario can be expected to occur when a unexpected sudden load is imposed on the body.

When adequate warning is available before anticipated involuntary motions, muscles are pretensed.[2,3,6,16] Through the synergistic activation of antagonistic muscles the stiffness of the musculoskeletal response increases.[12] This stiffening response reduces the amplitude and the time delay of the musculoskeletal response.[30] Thus, tissue strain is reduced, and the disturbance to postural stability is minimized.

When the perturbation is unexpected, or occurs in the absence of visual, auditory, or temporal cues, the system is unexpectedly loaded, and a startle response is generated to which the body overreacts while attempting to preserve its stability.[11] Several investigators have described the trunk muscle recruitments in response to unexpected loads applied to the hands,[18,20,24] loads applied to the torso directly,[4,6,26,28] and during impending falls.[27] In sum, these unexpected perturbations lead to a rapid onset and high peak amplitudes in the measured electromyographic activities, confirming the proposed injury scenario.

By the definition of these investigators, unexpected loading events occur when the body is completely or at least partially unprepared for the impending load, thereby preventing any stiffening reaction in the torso from occurring. Findings in recent research on lifting belts (also called back belts or back supports) show that the torso stiffness in the frontal and transverse planes is increased when an elastic belt wrapped around the abdomen is tensioned.[10,17,25] It was theorized that a lifting belt by limiting the torso's motion could reduce the potential for back injury initiated by unexpected sudden perturbations. This would be of particular interest for those handling objects that are prone to shifting or those who work where there are slippery footing conditions.

The following specific hypotheses were tested:

- A lifting belt reduces torso motion and external moments when sagittally symmetric and asymmetric unexpected sudden loads are applied to the hands.

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• A lifting belt reduces the peak electromyographic activity of the major trunk muscles supporting the torso after unexpected sudden loading.
• During unexpected sudden loading, the effectiveness of a lifting belt in minimizing the muscular loading of the torso is dependent on the pre-perturbation loading of the system.

■ Methods

Experimental Design. To investigate the effectiveness of lifting belts during unexpected sudden loading, other factors, potentially interactive, were included in a repeated-measures experimental design. Thus, the two lifting belt conditions (tensioned or not tensioned), were repeated for subjects as they received unexpected sudden loads that were symmetric and asymmetric with respect to the torso's midsagittal plane, and with and without preloading of the box the subjects were holding. In all, each volunteer participated in eight trials, presented in a randomized sequence corresponding to the eight combinations of the three two-level independent variables.

The dependent measures comprised surface electromyogram from eight trunk muscles, postural changes in 12 body segments, the applied force, and external moments at L5–S1, the knees, and the hips. The electromyogram data were obtained bilaterally from the following muscle groups: latissimus dorsi (right [LATR] and left [LATL]), erector spinae (right [ERSR] and left [ERSL]), external oblique (right [EXOR] and left [EXOL]) and rectus abdominis (right [RABR] and left [RABL]).

Subjects. The subjects were recruited from the medical center and surrounding educational facilities. Ten men and eight women between 22 and 47 years of age participated in the experiment. Mean height and weight of the men was 1.76 ± 0.072 m and 74.3 ± 10.5 kg and of the women was 1.63 ± 0.076 m and 58.8 ± 7.1 kg, respectively. The subjects were screened for history of back pain. Only persons with no back pain during the past year were permitted to participate. All volunteers signed an informed consent approved by the institution.

Apparatus. Subjects stood on two solid wooden blocks (23.5 × 16 × 14 in.) placed over two force plates. The force platforms (Bertec Corp., Worthington, OH) provided the ground reaction force vector, moment vector, and center of pressure under each foot. Surrounding the two wooden blocks was a wooden platform 14 in. high. This platform served to mount the loading apparatus constructed using polyvinyl chloride (PVC) tubing (Figure 1). The PVC loading apparatus was used to position the two pulleys used in the application of the sudden load. A Kevlar (no-stretch) line tied to the bottom of the plastic box held by the subject was connected to a force gauge and weight by the two pulleys. Two equally weighted bags of lead shot were used for the sudden load and the preload. The weight of each bag was normalized to each subject's maximum voluntary extension force and weighed 7.5% of this value (actual range, 3–7 kg). The sudden load was applied by allowing one bag of lead shot to fall 1 m, at which point the line become taught, and the force was transmitted to the bottom of the box through the force gauge. The force gauge measured the impulse force and also served as an event marker within the data stream. The bag used in the preload trials was also suspended from the line coming from the bottom of the box.

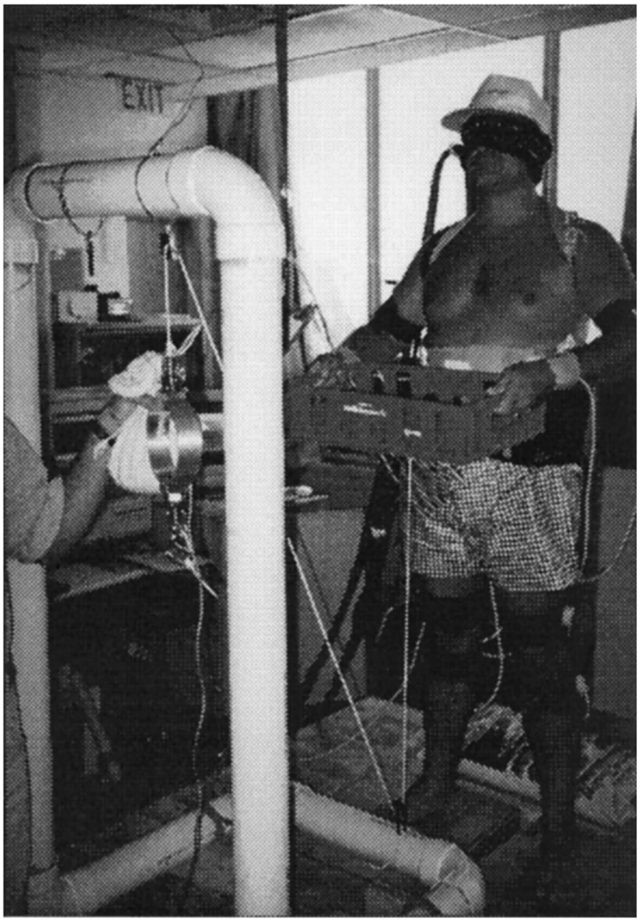

Figure 1. The loading apparatus. Weights were allowed to fall 1 m before the kevlar line was tensioned and the box held in the subject's hands was pulled downward.

An electromagnetic tracking system (Motion Monitor; Innovative Sports Training, Chicago, IL), was used to obtain kinematic data. Twelve sensors were sampled simultaneously with the 21 analog channels containing electromyogram, force plate, and dynamometer data. The transmitter was placed 111 cm above the floor and 62 cm behind the two force platforms. The sensors were positioned on the head, at T1, at L1, on top of S1, and on the left and right upper arms, forearms, thighs, and shanks. Sensors were connected by cable to the data collection computer. The sensors were used within a calibrated region that measured 137 × 91 × 183 cm. Mean dynamic error within the measurement space averaged 0.7 ± 0.3 cm.

Eight active bipolar surface electrodes (Medical Devices, Inc., St. Paul, MN, USA) were placed on trunk muscles to obtain the electromyographic signals. Each electrode had a gain of 10 and had an interelectrode spacing of 1 cm. A preamplifier (gain of 1000) amplified the signals before filtering. A 144-Hz notch filter and a 480-Hz low-pass filter were used to remove the noise introduced by the magnetic field into the electromyogram signal. The signal was again amplified (gain = 3), rectified, and integrated (time constant, 50 msec) before it was passed to the analog-to-digital boards.

A Lido back machine (Loredan Biomedical, Davis, CA, USA) was used to measure each subject's trunk extension strength and to obtain maximum voluntary contractions from the eight muscles sampled. The extension strength was used to normalize the weight, as described, applied during the sudden loading to each subject's capacity.

The lifting belt used in this study was constructed of webbed material covered by an elastic band 17 cm wide that stretched anteriorly and attached with Velcro. The lifting belts (Ergodyne, St. Paul, MN), were available in three sizes. In the two lifting belt conditions the elastic was either tensioned or completely slack.

Procedure. Each subject's session was initiated with the signing of the informed consent documents and the acquisition of weight, height, and age data. Surface electrodes were placed on the skin in line with the muscle's line of action using surgical tape at the following bilateral sites:

- Latissimus dorsi: T7 over the belly of the muscle and at T10 approximately 2 cm lateral to the midline.
- Erector spinae: at L3–L4 approximately 5 cm lateral to the midline.
- External oblique: along the line between the superior iliac spine and the umbilicus approximately 5 cm medial to the superior iliac spine.
- Rectus abdominis: at the level of the umbilicus 2 cm lateral to the midline.

Maximum voluntary exertions were performed in the Lido Back machine. The subjects were asked to flex, extend, twist right, and twist left at maximum ability. The machine was set to its lowest isokinetic velocity (5 deg/sec). The electromyogram was sampled as the trunk was flexed or extended approximately 15°. The maximum extension trunk strength was recorded and used to normalize the weight and preload used in sudden-loading trials. The baseline electromyogram was taken in the in measurement space within the magnetic field as the subject stood in a relaxed posture.

The sensors from the motion monitor were placed securely using Velcro and self-adhesive wraps on the following 12 rigid body parts: lateral side of the left and right shank, lateral side of the left and right thigh, top of the sacrum, over the spinous process for L1, over the spinous process for T1, on the lateral side of the left and right upper arms, on the lateral side at the distal end of the left and right forearms, and on the back of the head, by using an adjustable cap.

Closed-cell foam pads 7 mm thick were taped around the electrodes to prevent the lifting belt from exerting direct pressure on the electrodes. A lifting belt sized for the subject was placed loosely around the waist. The belt was tensioned only during the trials that required the lifting belt; otherwise, it was kept loose enough that a hand could be easily slipped between the belt and subject's abdomen.

Temporal conditions were controlled by a blindfold used to block visual cues and by a noise generator used to mask auditory cues. Before each trial, the subject was instructed to put on the blindfold and was instructed about belt tension and symmetry conditions. The subject was instructed to hold the box by grasping the handles. The box was held close to the body so that during sagittally symmetric trials the shoulders were neither flexed nor extended, and the elbows were flexed 90°. The box was positioned at the same distance in front of the spine during the asymmetric trials, but it was shifted laterally approximately 25 cm. Once the box was positioned, the preload was applied if required for the given trial. The weight used for the sudden load was then raised 1 m and was allowed to fall freely after a variable interval of 3 to 8 seconds. Data were recorded by the computer for the 3 seconds before and the 1 second after the dynamometer signal (indicating the loading had occurred).

Data Treatment. The integrated electromyograms were normalized according to each subject's maximum voluntary contraction obtained from the exertions performed while in the Lido machine. The peak normalized electromyogram values after the load onset were obtained from the following equation:

$$\mathrm{NEMG}_{i,j} = (\mathrm{IEMG}_{i,j} - \mathrm{REST}_i)/(\mathrm{MAX}_i - \mathrm{REST}_i)$$

where i is muscles 1 through 8, j is experimental conditions 1 through 8, $\mathrm{NEMG}_{i,j}$ is the normalized electromyogram for muscle i in condition j, $\mathrm{IEMG}_{i,j}$ is the current integrated electromyogram for muscle i in condition j, REST_i is the minimum resting integrated electromyogram value for muscle i for the relaxed standing measure, and MAX_i is the maximum integrated electromyogram value from muscle i during any of the maximal isometric exertions.

Kinetic and kinematic data were used to compute the external moments on the spine using a linked-segment model. The three-dimensional model comprised seven rigid bodies: two feet, two shanks, two thighs, and the pelvis. Joint centers were obtained from marker locations for the knees and ankles by using half the associated breadth measurement. The hip joint center was positioned medially at 25% of the hip breadth measure. L5–S1 was assumed to be in the midsagittal plane 9.04 cm from the L5 spinous process along a vector directed anteriorly and angled 6° below the horizontal.[5]

The position data were low-pass filtered (4 Hz) using a forward and backward Butterworth filter. Filtered position data were used to compute the angular orientations of the body segments using the projection-angle method. A 5-point numerical differentiation that uses the best fit fourth-order polynomial was used to obtain the angular velocity.[14] The same 5-point numerical differentiation technique was then used to obtain the angular acceleration for each body segment using the velocity data derived in the previous step.

Each body segment was modeled as a rigid body wherein the shank, thigh, and pelvis were idealized as being symmetric around their principal axes (*i.e.*, a slender rod), and had local coordinate systems chosen to coincide with the principle axes of inertia. The angular velocity and acceleration around the longitudinal axis of each segment were considered negligible. Body segment inertial properties were approximated using the radius of gyration data published by Webb Associates.[29] The center of mass and the mass distributions of each body segment were approximated by data obtained from the literature.[9,31] The foot was assumed to have no mass. The moments producing flexion and extension were predicted at the knee, hip, and L5–S1 joints and were obtained using the methodology published by Andriacchi et al.[1] The peak value of each directional component of the L5–S1 moment vector that occurred after the loading was extracted for analysis.

Statistical Analysis. Repeated-measures analyses of variance (ANOVAs) were conducted for the peak normalized electromyogram responses, the peak magnitudes of postural change, and the peak components of the L5–S1 reaction moment. The analyses included three within-subject factors (belt condition,

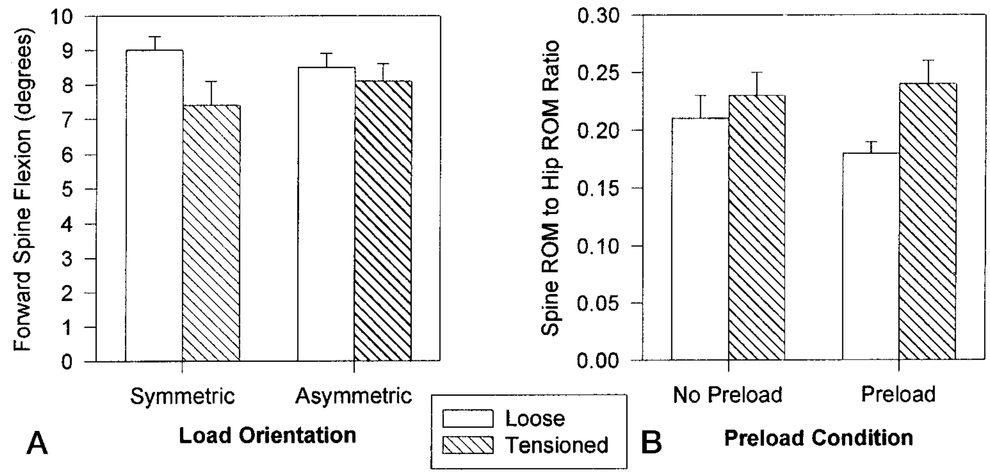

Figure 2. Mean spine flexion during the sudden, unexpected loadings as a function of the load symmetry and the lifting belt (**A**) and the ratio of spine motion to hip motion (**B**).

preload condition, and loading symmetry) plus the between-subjects factor, gender. Significant effects ($P < 0.05$) and trends ($P < 0.07$) are reported in the results. Additional ANOVAs and multiple-comparisons procedures were used to understand the interactions between the independent variables. Because this research is exploratory, the decision was made not to apply a correction against Type I error but instead to present the outcome of each statistical test performed so that the reader could evaluate the its significance.

■ Results

Kinematic Analysis

The forward-bending motion of the torso was significantly reduced when the belt was tensioned but only during the symmetric loadings (F = 15.72; *df* = 1,16; $P = 0.001$). On average, the decrease with the lifting belt tensioned during the symmetric loadings was 1.6°, or 17% of the sagittal plane motion measured in the thoracic and lumbar spine (Figure 2A). It is interesting to note that although the hip flexion did not change with the lifting belt, the average magnitude of this motion in both hips was approximately 20°. An analysis of the ratio of forward spine flexion to bilateral hip flexion showed that although the lifting belt significantly decreased the spine motion relative to the hip motion (F = 5.72; *df* = 1,16; $P = 0.029$), the effect was also dependent on the preload and symmetry conditions (F = 4.83; 1,16; $P = 0.043$). In essence, this interaction highlighted that the spine-to-hip flexion ratio was reduced during the sagittally symmetric preload trials with the belt tensioned (Figure 2B). This means that the reduction in total hip and spine motion that occurred during the preload trials with the belt tensioned was due to reduced spine motion in the spine, rather than to reduced hip motion, whereas during the preload trials, motions at the spine and hips decreased by approximately equal proportions with the belt loose.

The lateral flexion of the spine was significantly reduced by tensioning the lifting belt (F = 12.8; *df* = 1,16; $P = 0.003$), although the belt effect was also dependent on the preload condition and the subjects' gender (F = 10.7; *df* = 1,16; $P = 0.005$). Results of further analysis of the data displayed in Figure 3 showed that the lateral

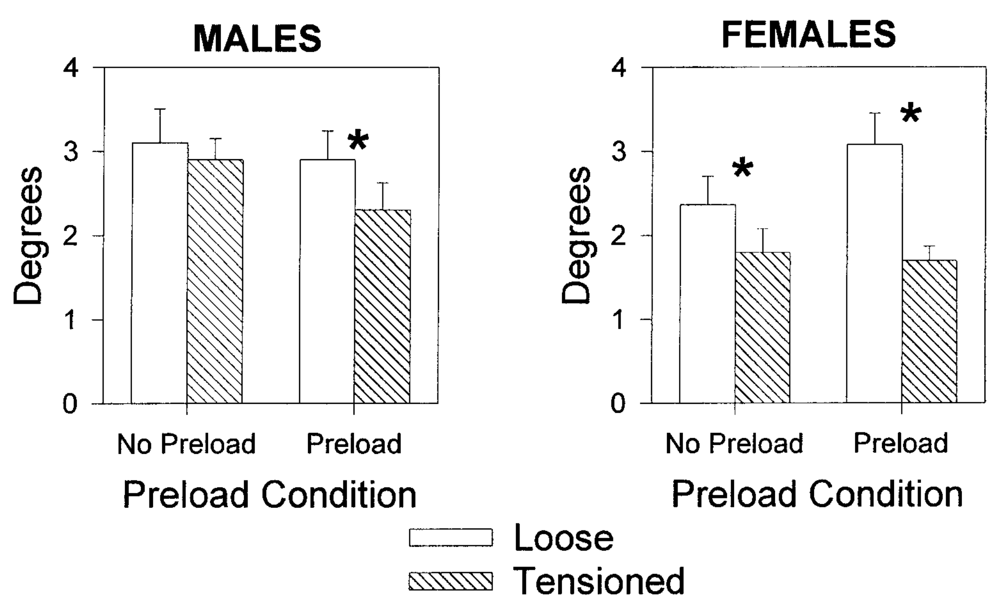

Figure 3. The lateral flexion of the spine as a function of the lifting belt, the preload condition, and the subject's gender (* $P <$ 0.05).

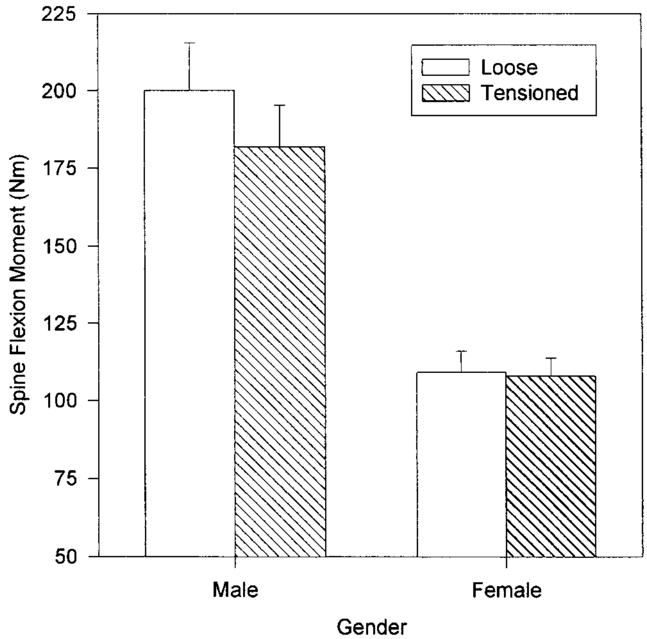

Figure 4. Mean forward spine flexion moment on the spine during unexpected sudden loadings as a function of the subject's gender and the belt's tension.

bending was reduced in the preload conditions in both genders when the belt was tensioned (F = 8.39; *df* = 1,16; *P* = 0.011). During the conditions without the preload, the belt reduced lateral bending only in the women. The twisting motions of the spine were unaffected by the lifting belt, even during the asymmetric loadings.

Kinetic Analysis

The forces measured by the strain gauge placed in series with the load averaged 387 ± 67 N for men and 322 ± 57.8 N for women. Statistical analysis of the force data indicated a significant belt-by-preload interaction (F = 8.30; *df* = 1,16; *P* = 0.011). Essentially, without preload there was no change in the applied force due to belt tension. With preload, the force significantly increased (F = 8.42; *df* = 1,16; *P* = 0.010) from 323 ± 40 N to 338 ± 44 N as the belt was tensioned.

External moments were calculated using the kinematic and the ground reaction force data in a three-dimensional, dynamic, linked-segment model. Statistical analyses of the peak localized joint moments indicated significant decreases in the forward flexion moments at the spine and the right hip when the lifting belt was tensioned, although this occurred only in the men (F = 7.89; *df* = 1,16; *P* = 0.013; F = 4.58; *df* = 1,16; *P* = 0.049). Figure 4 shows the spine moments in the men decreased from a mean of 200 ± 49 Nm to 181 ± 43 Nm, a 9% change. The change in the right hip flexion moment in the men decreased 12 Nm, also a 9% change. There were no significant changes in the lateral bending or twisting moments at the spine that were associated with the lifting belt.

Electromyographic Analysis

The top part of Table 1 shows the muscles significantly affected by the use of the lifting belt, either by itself or in combination with other factors (interactive effects). The lifting belt affected the peak activation levels of both erector spinae muscles; however, this was also dependent on the symmetry of the loading, and for the ERSL this effect was also dependent on the gender of the subject. Figure 5A shows that tensioning the belt yielded greater peak activity in the ERSL in the symmetric conditions for both genders, although this increase was nonsignificant in *post hoc* tests ($P > 0.10$). As would be expected, during asymmetric conditions, this contralateral muscle showed increased activation, but the belt further increased the peak response in the men (F = 5.93; *df* = 1,9; *P* = 0.038). The response of the ERSL in the women was essentially unaffected by the belt tension during the asymmetric loadings. Meanwhile, the ipsilateral ERSR peak response (Figure 5B) was reduced during the asymmetric loading, but more so when the lifting belt was tensioned (F = 5.14; *df* = 1,14; *P* = 0.038). The increase in the ERSR response shown in Figure 5B during the symmetric loadings was not statistically significant.

With regard to the latissimus dorsi muscles, only the LATL was significantly affected by the lifting belt (Table 1), but the response of this muscle was also dependent on the gender of the subject, and independent of gender, was dependent on the preload condition. The peak response of the LATL in the men was slightly larger with the belt tensioned (F = 4.63; *df* = 1,9; *P* = 0.060), whereas the response in the women was not significantly affected by the lifting belt. Independent of gender, the response of the LATL muscle to the lifting belt was dependent on the symmetry of the loading and the preload conditions. Although the response of this muscle was elevated during the asymmetric loadings relative to the symmetric loading and during the nonpreload conditions, *post hoc* tests did not identify a significant differential response of this muscle across conditions that was due to the lifting belt.

The lifting belt significantly affected three of the four anterior muscles sampled. Only the response of the RABL was unchanged by the belt. The peak activities in the RABR and the EXOR, although relatively small in absolute terms, were reduced by 41% and 19%, respectively (Figure 6).

The peak response of the EXOL to the lifting belt was also dependent on the symmetry of the loading and the gender of the subject (Table 1). During the symmetric loadings, the belt had little impact on EXOL response. Asymmetric loadings, although leading to a larger response in this contralateral muscle, showed gender-specific responses to the lifting belt. The *post hoc* test results indicate that the women had a lower peak EXOL response with the belt tensioned during the asymmetric loadings (F = 3.92; *df* = 1,7; *P* = 0.088).

Table 1. The Results From the Repeated Measures Analysis of Variance for Each Dependent Variable. Only the Main and Interactive (X) Effects That Involved the Belt and Were Significant for at Least One of the Dependent Variables are Shown

Dependent Variable		Effect						
		B	BXS	BXP	BXG	BXSXP	BXSXG	BXPXG
	F						5.56	
ERSL	df						1,16	
	p						0.032	
	F		5.14					
ERSR	df		1,14					
	p		0.040					
	F							
LATR	df							
	p							
	F				7.17	8.66		
LATL	df				1,16	1,16		
	p				0.017	0.010		
	F				4.29		4.72	
EXOL	df				1,15		1,15	
	p				0.056		0.046	
	F	6.00					5.56	
EXOR	df	1,16					1,16	
	p	0.026					0.032	
	F							
RABL	df							
	p							
	F	6.92						
RABR	df	1,16						
	p	0.018						
Forward	F	12.47	7.15					
Spine	df	1,16	1,16					
Flexion	p	0.003	0.017					
Lateral	F	12.88						10.69
Spine	df	1,16						1,16
Flexion	p	0.003						0.005
Spine	F							
Twist	df							
	p							
Spine	F	10.03			7.89			
Flexion	df	1,16			1,16			
Moment	p	0.006			0.013			
Spine	F							
Lateral	df							
Moment	p							
Spine	F							
Twisting	df							
Moment	p							
Applied	F			8.30				
Force	df			1,16				
	p			0.011				

F = F-Test values; df = degrees of freedom; p = probability associated with each effect; B = belt; S = symmetry; P = preload; G = gender; ERSL = erector spinae left; ERSR = erector spinae right; LATR = latissimus dorsi right; LATL = latissimus dorsi left; EXOL = external oblique left; EXOR = external oblique right; RABL = rectus abdominis left; RABR = rectus abdominis right.

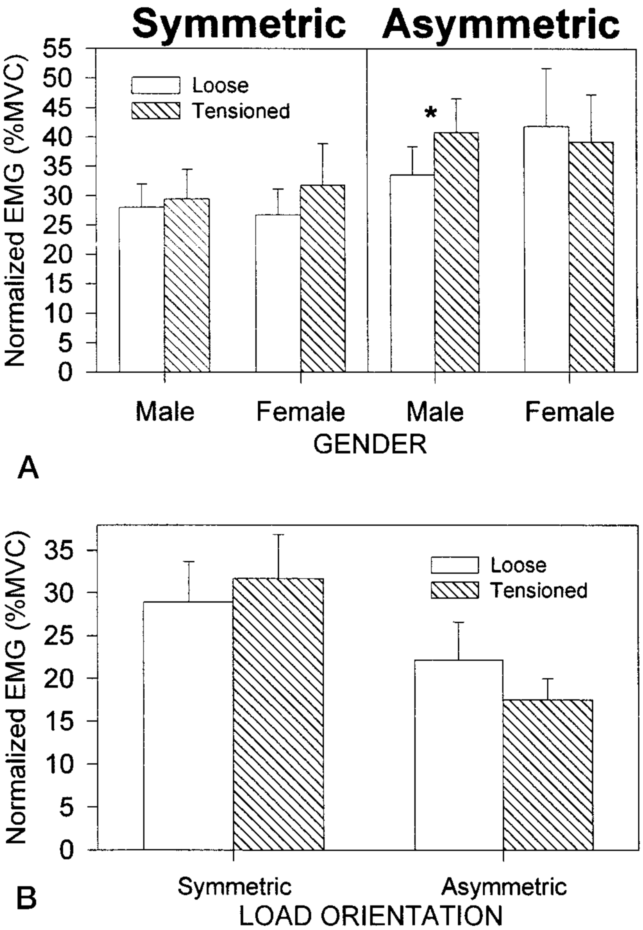

Figure 5. The peak response of the *left erector spinae* as a function of the lifting belt, load symmetry, and gender (**A**). The peak response of the *right erector spinae* as a function of the lifting belt and the load symmetry (**B**).

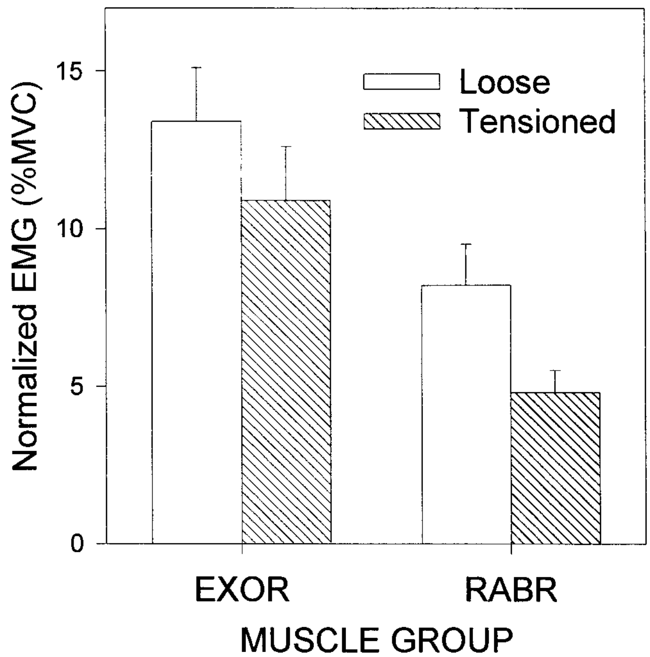

Figure 6. The peak activities from the right rectus abdominis and the right external oblique after sudden loading.

■ Discussion

The data show that the effects of the lifting belt on spinal loading during sudden perturbations are more complex than was hypothesized. The belt's biomechanical impact, although not large, was observable. That the belt's effect on the spine moments, trunk motions, and muscle recruitments was dependent on the subject's gender, the preload condition, and load orientation, forces reconsideration of the simplicity of the original hypothesis.

Support for the stiffening hypothesis comes from the reduced forward bending in the spine. Although the statistically significant reductions occurred only with the symmetric loadings, the asymmetric loading showed a similar trend. The reduced forward bending with the lifting belt is consistent with data reported by others.[10,21] Magnusson et al,[21] while measuring trunk motion in a small sample of subjects performing a lifting task, found less forward bending of the torso with a lifting belt than without one. Granata et al[10] reported that the subjects wearing the elastic and leather belts showed less trunk flexion than in the no-belt condition. However, at the same time, these authors found the pelvic flexion increased with these same belts. Results of previous work from the the current authors' laboratory, which quantified triaxial trunk motions when lifting with and without a belt showed that the belt had no effect on sagittal plane spine motions. Others have found similar trunk motion results when studying the passive stiffness changes in the torso with and without a belt,[25] which may be the data most pertinent to the present experiment.

It is interesting to note the belt's impact on the total trunk motion during the sagittally symmetric loadings was also dependent on the preload condition, both in the magnitude of the total angular excursion and in the relative changes in hip and spine rotations. In the absence of preload, there was more trunk motion than with preload. However, the total motion was reduced in the nonpreloaded conditions with the belt. But under these conditions there was no change in the relative amounts of pelvic (hip) flexion and spine flexion. In contrast, when the system was preloaded, the amount of spine flexion relative to pelvic flexion decreased with the belt, whereas the total trunk motion was unchanged.

The findings of increased lateral stiffness with the belt are consistent with the passive bending resistance measured by McGill et al,[25] the reduced lateral bending as individuals lifted observed by Lavender et al,[17] and the asymmetric lifting results of Granata et al.[10] That there was no statistically significant interaction effect in the current study between the belt tension condition and the load symmetry condition shows that the belt's effectiveness in controlling lateral bending motions may not have differed between the symmetric and asymmetric loadings. Moreover, the results show that the symmetric loads resulted in more than just sagittal plane motions, indicating the destabilizing effects of the applied perturbation. The lateral bending was minimized when the box was preloaded and the lifting belt was tensioned, indicating that the perturbation's effect on stability was mitigated if the muscles were already actively contracting. This response is consistent with results in earlier studies showing less motion and lower spine compression forces when the onset of sudden perturbations can be anticipated.[16,18]

No changes were found in the twisting motion with the lifting belt—perhaps in part because the twisting motions were quite small in this loading paradigm, even during the asymmetric loadings. Under these asymmetric conditions the applied load would initially create only a forward and lateral flexion moment. Only after the trunk and pelvis flexed forward would the forces acting on the hands result in a shift from lateral bending to axial rotation moments, and therefore, twisting motions. It is possible that because the twisting moments would be secondary to the sagittal plane postural changes, the timing may have been such that the muscles activated during the initial impulse served to minimize the axial rotation of the torso.

The belt, most likely by damping the spine flexion, reduced the flexion moment calculated at L5–S1. It is interesting that this effect only occurred in the men. There are two issues that could have contributed to this finding. First, anatomic differences in the shape of the male and female pelvis could have affected the belt's ability to splint the gap between the pelvis and the rib cage, and thus, its effectiveness in controlling spine motions. If this were true a significant belt-by-gender interaction effect would be expected in the spine's forward flexion motion. Second, the measured moments in the men were nearly twice the magnitude of those measured in the women, largely because of the normalization procedure in which the magnitude of the sudden load was based on the trunk extension strength. Previous studies of trunk extension strength indicate that women have less strength when adjusting for body mass, and therefore, relative to men, would have been subjected to smaller postural perturbations with the normalization procedure. This may account for the fact that the men flexed their spines approximately 28% farther than the women ($P < 0.01$) in response to the loadings. Thus the men, in addition to the large moments stemming from the larger sudden load magnitudes, would have experienced greater moments due to the trunk's orientation and motion. In other words, the moments were already relatively small for the women, thereby allowing little room for variation due to the lifting belt.

Extrapolating the significance of the 9% change in the moments experienced by the men outside the loading paradigm in the current study is challenging at best. However, it should be recognized that a loading force equivalent to the mean loading of 387 N in the men was significant and could be generated under a variety of real-world conditions. The resultant 200-Nm moment experienced by the men is similar to that found when men lift 11–12-kg boxes from the floor.[8,15] It should be recognized that in unanticipated loading, this same peak moment is likely to occur much faster and with potentially unprepared muscles, showing that even a 9% decrease could be beneficial when it comes to preventing some injuries.

Muscle Strategy

The change in muscle activations indicates that a belt alters the underlying strategy used by the body when dealing with potentially destabilizing perturbations. Given that the overall hypothesis tested in this study was that the lifting belt would stiffen the torso and, in so doing, would protect the torso, a reduction in the trunk's primary agonist muscles responsible for the perturbation recovery process would be expected. Therefore, the greater peak muscle response in the LATL, ERSL, and EXOL was unexpected. At the same time, the overall decrease in electromyogram activity observed in the EXOR, RABR, and ERSL (asymmetric trials only) with the belt tension could be viewed as a reduction in the antagonistic cocontraction response. Similar reductions in the cocontraction response have been reported by Granata et al[10] in subjects using an elastic belt during a lifting task. Thus, the belt may have been perceived by subjects as protective, which in turn resulted in reduced use of antagonistic muscles normally recruited for stabilization purposes.[13,19] Greater agonistic muscle force, combined with less antagonistic force would result in a greater deceleration of the trunk after the onset of the loading, which reduces the motion, which in turn, lowers the peak moment. This analysis therefore shows that the belt's effect may be less biomechanical and more psychological, in that it changes muscle response strategies either before or during the loading. However, although the spine moment results indicate that the belt may be beneficial, at least for men, the greater peak muscle activities in the ERSL and LATL show that there may be increased potential for a muscle overexertion injury with the belt.

Limitations

There are limitations of this study to consider while interpreting the findings. First, there was no correction for the postural change in the electromyogram signal; however, given that the differences between postures between belt conditions were relatively small, it is unlikely that the postural change significantly affected the comparisons. In addition, because of the limited duration of actual belt use in this study (<1 hour), it should be recognized that any of the psychological or proprioceptive effects altering muscle recruitments could change with extended use. Moreover, given that the control condition was not the absence of a belt, but rather a loose belt, if over time a person becomes lax about tensioning the belt, there would clearly be no biomechanical benefits.

Conclusions

The results of this study do not provide an easy answer regarding the benefits of the lifting belt in a sudden-loading paradigm. In part, this is because of the interactions found between the belt, the symmetry of the loading event, and the gender of the individual subjected to the loading. At first glance, the data support the hypothesis that the belt stiffens the torso's response to sudden loading. Clearly the effects are small and the individual differences are large. These findings indicate that during unexpected sudden loadings, a belt may reduce the net external moment loading; however, it may also change the internal responses in a manner that raises considerable question about the belt's overall effect. Although further research is needed to understand the individual differences in response to the perturbation, it is doubtful that the findings from this finer level of analysis will be of use to those making decisions regarding the efficacy of using lifting belts in the workplace.

Key Points

- During unexpected sudden loadings with a tensioned elastic-type lifting belt there was less forward bending of the torso.
- There was a reduction in the external forward bending moment in the men.
- There were changes in the muscle cocontraction so that the antagonistic muscle activities were reduced, whereas the activation of the primary agonist muscle in the back increased.

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Point of View

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All clinicians who treat patients with low back problems must manage injuries due to sudden unexpected loading. Furthermore, we are frequently asked by our patients if there is a proven rationale for the use of protective corsets (lifting belts). The authors have performed a valuable service to patients and clinicians by clearly investigating (with a repeated-measure experiment) the effects of a lifting belt on both spine moments and muscle recruitments. Imposing loads in a sudden unexpected manner has made their study especially helpful and applicable to clinical situations.

Because of the skepticism of clinicians regarding the protective effects of a passively applied apparatus such as a lifting belt, a carefully controlled and clearly conducted epxeriment using human subjects is mandatory in reaching any conclusions, favorable or unfavorable, which are clinically valid. Therefore, based upon the authors' thorough methodology and rigorous data analysis, we can accept as valid their conclusions that the belt reduces forward-bending of the spine during symmetrical loading, and that in males it decreases the forward flexion moment acting on the spine. But we must inform our patients that the belt's stiffening effect is a small one, and that, not surprisingly, individual variations are large, especially if the belt is worn loosely.

Furthermore, the reduction in use of paraspinous muscles that normally stabilize the spine in response to sudden loading (EMG data) implies that patients may regard the belt as more protective than it actually is. This realization brings the clinician full circle back to his or her original skepticism, and to the suspicion that the benefits of the belt may be more psychological than biomechanical. Because of the authors' valuable and clearly reported study, this clinician will tell his patients that a snugly applied lifting belt may be helpful in decreasing forward-bending of the spine in response to sudden unexpected loading, but that it is no substitute for constant attention to lifting guidelines such as postural precautions, load weight, and lifting frequency. Nor is the belt a substitute for the maintenance of optimum tone and strength of paraspinous and torso musculature.