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The National Institute for Occupational Safety and Health suggests there is insufficient biomechanical or epidemiological evidence to recommend the use of back belts in industry. From a biomechanical perspective, previous work suggests that lifting belts stiffen the torso, particularly in the frontal and transverse planes. To determine whether lifting belts stiffen the torso and alter the trunk muscle response during a sudden loading event, we tested the hypotheses that (a) lifting belts alter peak muscle activity recorded with electromyography (EMG) during sudden loading and (b) lifting belts have a larger impact on trunk muscle response when sudden loads are applied asymmetric to the torso's midsagittal plane. A sudden load was delivered to 10 men and 10 women without history of low back disorder via a cable attached to a thoracic harness; motion was restricted to the lumbar spine. Results indicate that gender was not a significant factor in this study. The lifting belt reduced the peak normalized EMG of the erector spinae muscles on average by 3% during asymmetric loading, though peak normalized EMG was increased by 2% during symmetric loading. Lifting belts have been shown to slightly reduce peak erector spinae activity during asymmetric sudden loading events in a constrained paradigm; however, the effects of lifting belts are too small to provide effective protection of workers. Actual or potential applications include the assessment of lifting belts as protective devices in workers based on the effects of lifting belts on the trunk muscle activity.

INTRODUCTION

Material handling tasks have been cited as the most frequent cause of work-related low-back injuries (Andersson, 1981; Bigos et al., 1986), and forceful movements or sudden maximal exertions have been associated with the onset of these injuries (Bigos et al., 1986; Magora, 1973; Troup, Martin, & Lloyd, 1981). Sudden maximal exertions often result from slips, falls, or lifting of unstable loads (e.g., a container partially filled with liquid). Relative to expected loading, unexpected loading of the torso has been associated with greater trunk displacement, increased trunk muscle activity (Caldwell et al., 1974; Cresswell, Oddsson, & Thorstensson, 1994; Marras, Rangarajulu, & Lavender, 1987), and faster-onset rate of trunk muscle activity (Lavender et al., 1989; Lavender, Marras, & Miller, 1993; Marras et al., 1987).

Although the effects of lifting belts on spine biomechanics are not fully understood (National Institute for Occupational Safety and Health, Back Belt Working Group, 1994), there is some evidence that lifting belts restrict trunk motion and, hence, may protect the spine during unexpected loading. McGill, Seguin, and Barnett (1994) measured the effect of lifting belts on passive motion of the spine in three planes and reported that the lifting belt increased the passive stiffness of the trunk in the frontal and transverse planes. Additionally,

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Lavender, Thomas, Chang, and Andersson (1995) reported that lifting belts reduce trunk side bending and rotation (twisting) during asymmetric lifting tasks when foot motion is restricted. The reduced motion suggests that a lifting belt increases trunk stiffness, which could be expected to reduce the trunk displacement during a sudden perturbation. We have hypothesized that lifting belts help protect the spine by decreasing the magnitude of trunk muscle response necessary to restore equilibrium to the system following a sudden loading event. Therefore, the purpose of this paper was to investigate the effects of lifting belts on trunk muscular and kinematic responses to a sudden loading event in men and women.

METHODS

Participants

Twenty participants (10 men and 10 women), 20 to 33 years of age, signed an informed consent form approved by two institutional review boards. The participants were screened for a history previous or ongoing low-back pain (LBP). Only individuals with no history of LBP in the last year were allowed to participate in the study.

Experimental Design

A mixed-model design was used in this experiment. Gender was the between-subjects factor. Direction of the applied load (symmetric, asymmetric), trial (three), and the tension of the lifting belt (very loose, tight) were the within-subject factors. Each combination of the experimental conditions was repeated three times (trial) in a randomized sequence with a 1-min rest between trials.

The dependent measures were the trunk kinematics as measured with the Lumbar Motion Monitor (LMM; Chattanooga Corp., Chattanooga, TN) and the normalized surface electromyography (EMG) from the left and right longissimus thoracis (LGT), erector spinae (e.g., iliocostalis, or ERS), external oblique (EXO), and the rectus abdominus (RAB).

Apparatus

Participants stood in a reference frame that allowed them to be secured in an upright position so that motion below the lumbar spine was restricted. The suddenly applied load was delivered via a cable attached to a thoracic harness. The cable was run through pulleys and attached to a bag of lead shot that was dropped 1 m. The weight of the bag was normalized to a value that was 5% of each individual's maximum isometric trunk extensor strength. The applied load was delivered both in the midsagittal plane and in an oblique plane rotated 45° to the right of the midsagittal plane, for the symmetric and asymmetric conditions, respectively (Figure 1). A load cell attached to one of the pulleys was used as an event marker.

Disposable surface EMG electrodes (Nikomed Corp., Hampshire, England) were used for this study. The interelectrode distance was 2 cm. The EMG signals were preamplified (gain of 1000) close to the recording electrodes and fed to the main amplifier via shielded cables. The signals were amplified and rectified with a bandpass frequency range of 15 to 1000 Hz and integrated using a time constant of 30 ms. The integrated signals were sampled at 120 Hz. The raw EMG signals were monitored on a sweep oscilloscope for signal quality.

The trunk position data were obtained with the LMM (a triaxial electrogoniometer) and were collected at 60 Hz utilizing the LMM's software. The LMM attaches to the thoracic spine via a chest harness and to the pelvis at the level of the sacrum with a pelvic harness. The unit weighs approximately 1.4 kg and does not restrict lumbar motion. The reliability of the instrument has been reported by Marras, Fathallah, Miller, Davis, and Mirka (1992).

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 belts, manufactured by Ergodyne (St. Paul, MN), were available in three sizes.

Procedure

Surface EMG electrodes were placed on the skin overlying the muscle bellies of the left and right LGT, ERS, RAB, and EXO muscles. The electrodes were attached at the level of T-10 approximately 4 cm from the midline for the LGT and at the level of L-3 approximately 4 cm from the midline for the ERS. Electrodes for the

RAB were attached at the level of the umbilicus 2 cm from the midline. Placement for the EXO electrodes was at the level of the umbilicus, approximately halfway between the iliac crest and the anterosuperior iliac spine. This is usually 2 cm medially and 2 cm laterally from these respective bony landmarks and rotated 45° from the vertical.

Maximum isometric muscle forces, for the purpose of EMG normalization, were measured by having the participants perform resisted isometric trunk flexion, extension, and rotation.

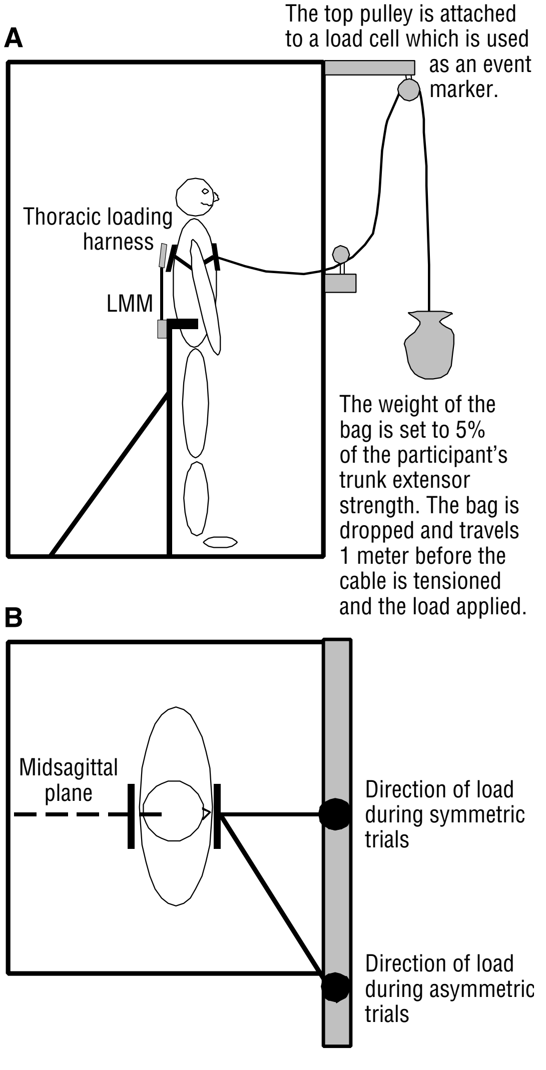

Figure 1. Schematic of the apparatus used to deliver the suddenly applied load. (A) Sagittal view. (B) Top view.

The participant stood in the reference frame with the pelvis firmly secured. A harness attached to the thoracic region was connected to a dynamometer via steel cable. The participant was asked to exert maximal flexion, extension, and rotation forces with his or her trunk while standing in a neutral posture. These tests were repeated at 2-min intervals until the force measured from each muscle group no longer increased and the two greatest trials were within 10% of each other (Caldwell et al., 1974). The maximum extensor force was recorded and used to determine the magnitude of the weight dropped during the suddenly applied load. Baseline or resting EMG values were recorded with the participant standing in a relaxed posture.

Following the maximal exertions, the LMM was attached to the thoracic harness and the stand to which the participant's pelvis was secured (Figure 1). Closed-cell foam padding (7.5 mm thick) was placed around all the EMG electrodes to prevent compression from the thoracic harness and the lifting belt.

The load was applied either symmetric or asymmetric to the torso's midsagittal plane. During symmetric loading, the cable was attached to the midpoint of the thoracic loading harness and run through two pulleys in the midsagittal plane. During asymmetric loading, the two pulleys were set 45° to the right of the midsagittal plane. The application of the sudden load during the asymmetric condition therefore created a moment that had forward flexion, right lateral flexion, and right rotational components. In half the sudden loadings the lifting belt was worn loosely (enough space to slide a hand between the belt and the abdomen). In the other half of the sudden loadings, participants were instructed to stretch the elastic component of the lifting belt as tight as possible. Visual and auditory cues of the applied load were masked with a blindfold and a noise generator. Trunk kinematics and EMG data were collected for 1 s prior to the release of the load and for 2 s after the suddenly applied load.

Data Treatment

The integrated EMG (IEMG) data were normalized for each participant with respect to the EMG data collected during maximal trunk exertions and to the resting EMG levels according to the following equation:

$$NEMG(i,j,k) = [IEMG(i,j,k) - REST(i)] / [MAX(i) - REST(i)], \quad (1)$$

where i = Muscles 1 through 8, j = Experimental Conditions 1 through 4, k = Trials 1 through 3, $NEMG(i,j,k)$ = the normalized EMG for muscle i in condition j in trial k, $IEMG(i,j,k)$ = the current integrated EMG value for muscle i in condition j in trial k, $REST(i)$ = the minimum resting IEMG value for muscle i for relaxed standing, and $MAX(i)$ = the maximum IEMG value from muscle i during the maximal isometric exertion.

The onset activity for each muscle was determined by visual inspection of the NEMG signal. The area under the NEMG signal from muscle onset to the onset of the sudden load was calculated by Equation 2. This value was used as a measure of the muscle activity prior to the onset of the sudden load:

$$\text{Pre-Load Area } (i,j,k) = \sum_{m}^{SAL} NEMG\,(i,j,k), \quad (2)$$

where i = Muscles 1 through 8, j = Experimental Conditions 1 through 4, k = Trials 1 through 3, $NEMG(i,j,k)$ = the normalized EMG for muscle i in condition j in trial k, m = sample coinciding with the onset of activity in muscle i, and SL = sample coinciding with the onset of the sudden load.

The peak EMG values and the area of NEMG activity prior to the suddenly applied load were used in this investigation. Trunk position data were obtained from the LMM. The position data were smoothed with a 3-point moving average after visual inspection. A 3-point average was selected in order to reduce the noise in the position data without masking maximal position changes. This method results in an effective cutoff frequency of approximately 12.5 Hz. The second central point difference method was used to calculate the velocity. The same procedure was repeated on the velocity data to determine acceleration (Lanczos, 1988, chapter 5).

Two four-way multivariate analysis of variance (MANOVA) procedures were used to analyze the peak NEMG and the area of NEMG activity prior to the suddenly applied load. Three additional four-way MANOVA procedures were utilized in the analyses of trunk kinematic data in the frontal, sagittal, and horizontal planes. Significant MANOVA findings were followed up with univariate four-way, mixed-model analysis of variance (ANOVA) procedures on individual muscles or kinematic variables.

RESULTS

Analysis of Peak Normalized EMG

Multivariate analysis of variance on the eight peak EMG measures revealed an interaction of belt and symmetry $F(8, 208) = 2.6$, $p < .01$, but no other belt effects were found. This indicates that the peak EMG response attributable to the belt was independent of gender and did not change with repeated trials. Univariate analyses of the peak normalized EMG of the eight trunk muscles tested showed a significant interaction of the belt condition and the direction of the applied load only for the left erector spinae, $F(1, 18) = 8.67$, $p < .01$, and right erector spinae, $F(1, 18) = 6.33$, $p < .01$ (Figure 2a and 2b). Analysis of the simple effects of this interaction revealed that when the applied load was symmetric to the midsagittal plane, the peak NEMG of the left erector spinae muscle was, on average, 37.8% of maximal voluntary contraction (MVC) with the lifting belt and 34.9% MVC without the belt, $F(1, 18) = 4.18$, $p < .055$. In contrast, during asymmetric loading the peak NEMG for the left erector spinae (contralateral muscle) were on average 33.1% MVC when the lifting belt was tensioned and 36.4% MVC when the belt was loose, $F(1, 18) = 5.01$, $p < .036$ (Figure 2a). The right (ipsilateral) erector spinae peak responses were 35.0% MVC with the lifting belt tensioned and 33.3% MVC when the belt was loosened during symmetric loading, $F(1, 18) = 2.01$, $p > .1$ (Figure 2b). During asymmetric loading conditions, the right erector spinae peak responses were 23.5% MVC with the lifting belt tensioned and 26.0% MVC when the belt was loosened, $F(1, 18) = 2.84$, $p > .1$ (Figure 2b).

Analysis of Preload Area of Normalized EMG

The multivariate analysis showed that the preload area of normalized EMG, which is a

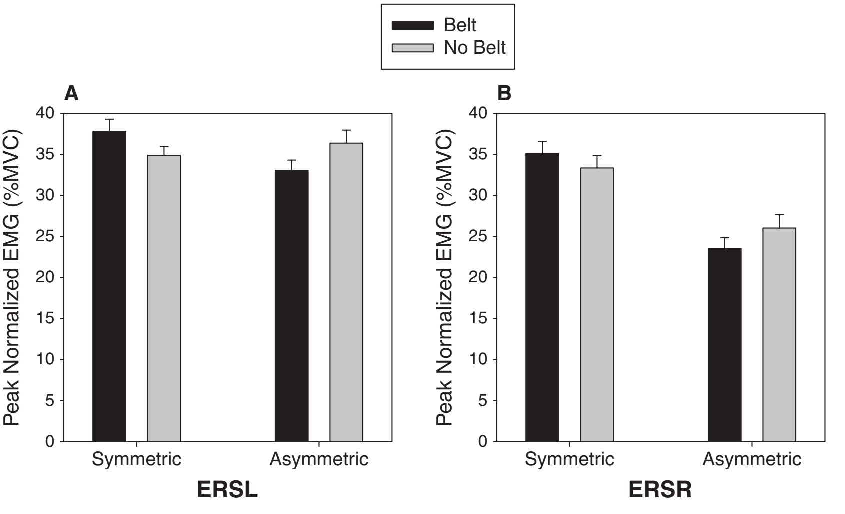

Figure 2. The peak normalized EMG response of the (A) left erector spinae and (B) right erector spinae muscle averaged over participants as a function of the belt and direction of applied load. Error bars represent one standard error of measure.

measure of muscle activity prior to the onset of the impending load, was not significantly affected by the lifting belt or its interactions with direction of the applied load, gender, or trial.

Analysis of Trunk Kinematics

Multivariate analysis of the kinematic data revealed a main effect of the lifting belt for frontal plane trunk motion, $F(5, 208) = 126$, $p < .01$. Trunk motions in the sagittal or transverse planes were not affected by the lifting belt. Although univariate analysis of variance revealed a statistically significant difference in the peak displacement and velocity of the trunk in the frontal plane, $F(1, 18) = 75.4$, $p < .01$, $F(1, 18)$, $p < .01$, the reduction in position was, on average, 0.5° (from 7.3° to 6.7°) when the lifting belt was worn. The reduction in velocity was 2.0°/s (from 28.9°/s to 27.1°/s) when the belt was worn.

DISCUSSION

Sudden unexpected loading has been identified as a significant risk factor for developing costly low-back injuries (Bigos et al., 1986; Magora, 1973; Troup et al., 1981). Perhaps this is attributable to the increased stresses placed on the spinal structures from increased trunk muscle contractions. Given the work of McGill et al. (1994) and Lavender et al. (1995), who reported that lifting belts limit motion of the torso in the frontal and transverse planes, we had hypothesized that if wearing a lifting belt could stiffen the torso, then the trunk muscle activity required to restore equilibrium to the body during an asymmetric sudden loading event would be reduced. This is supported by data suggesting that wearing a lifting belt reduced the frontal plane trunk displacement by 0.5° on average and reduced the peak NEMG of the left erector spinae muscles on average by 3.3% MVC during asymmetric loading (Figure 2b). In contrast, during symmetric loading, there was no effect of the lifting belt on sagittal plane trunk displacement, and the peak NEMG of the left erector spinae activity increased on average by 2.9% MVC when the lifting belt was worn (Figure 2a).

An increase in the peak muscle response during symmetric loading was an unexpected result and is not consistent with other investigations on the effect of lifting belts and erector spinae muscle activity during symmetric tasks (Magnusson, Pope, & Hansson, 1996). Perhaps the increase in response of the erector spinae to symmetric loading is attributable to a change in recruitment strategy in the posterior trunk muscles. Lavender, Shakeel, Andersson, and Thomas (in press) showed that lateral bending was reduced in free-standing participants experiencing an unexpected loading when the belt was tensioned, thereby suggesting greater lateral stability. In the current study, a perceived increase in lateral stability may have led to a shift in load sharing from more laterally placed muscles (e.g., quadratus lumborum) to components of the erector spinae, possibly explaining the small increase in the peak NEMG of the erector spinae in the symmetric loading condition.

In this investigation, the effect of lifting belts on the peak response of the longissimus thoracis, rectus abdominus, and external oblique were also measured. The lifting belts used in this study covered from the top of the sacrum to the first lumbar vertebrae and did not cover the belly of the longissimus thoracis muscles. The activity of these muscles was measured because of a concern that the bending stress from the applied load would be displaced to the thoracic column and increase demand on the longissimus. However, the lifting belt had no effect on the peak NEMG of the longissimus thoracis muscle. The peak NEMG of the four abdominal muscles was also not affected by the lifting belt, which is consistent with McGill, Norman, and Sharratt (1990), who reported no change in abdominal EMG activity during lifting tasks in which a lifting belt was used.

The lifting belt had no effect on the preload area of NEMG in this paradigm of unexpected sudden loading. This finding suggests that the participants had no sense of increased trunk stability attributable to the lifting because they did not alter their preload preparation strategy. However, no specific questions were asked to determine the participants' perception of trunk stability.

Several studies on lifting belts have included female participants (McGill et al., 1990; Reyna, Leggett, Kenney, Holmes, & Mooney, 1995; Sullivan & Mayhew, 1996). Sullivan and Mayhew reported that only their male participants had increased isometric force production attributable to lifting belts, whereas McGill et al. found that female participants had less trunk stiffness attributable to the belt during forward flexion and left side bending. We found no interaction of gender and the lifting belt conditions, which may result in part from the experimental design. The participants were constrained so that motion was available only from the lumbar spine. This constraint significantly reduces the available degrees of freedom and postural response strategies to attenuate the forces during a suddenly applied load, thereby masking potential interactions of gender and the lifting belt.

Clearly, one limitation of this study was that the method of load application was different from those experienced in a material handling task. Ultimately the loads experienced in an actual task result in bending and torsional moments on the spine. It is these moments that we sought to simulate in this investigation. We also recognize that during many material handling situations, the biomechanical system is preloaded prior to the onset of a perturbing event, and so further investigation is necessary to determine the belt's effectiveness under these conditions.

Another limitation of this study is that the sampling rate of 60 Hz for the LMM may have been too low for a sudden loading event, possibly masking the true peak trunk displacement and thus introducing Type II error. However, we believe the frequency response of the trunk was low enough that 60 Hz sampling rate was sufficient. In the short run, the data presented here provide insight into how the body uses and adapts to a hypothetical external support in response to a suddenly applied bending moment.

CONCLUSION

There are conflicting reports on the biomechanical effects of lifting belts. Our data suggest that during asymmetric unexpected loading events, lifting belts can minimally reduce lateral bending motion and contralateral erector spinae activity. However, when the unexpected load was symmetric to the midsagittal plane, erector spinae activity increased when the belt was worn. Given our data, it would appear that the biomechanical effect of the belt during sudden loading is small and situationally dependent.

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