DEPARTMENT OF HEALTH AND HUMAN SERVICES

Public Health Service
Centers for Disease Control
and Prevention (CDC)

Memorandum

Date: March 27, 2001

From: Roy M. Fleming, Sc.D., Director, Research Grants Program RMF
Office of Extramural Programs, NIOSH, D30

Subject: Final Report Submitted for Entry into NTIS for Grant 5 R03 OH003339-02.

To: William D. Bennett
Data Systems Team, Information Resources Branch, EID, NIOSH, P03/C18

The attached final report has been received from the principal investigator on the subject NIOSH grant. If this document is forwarded to the National Technical Information Service, please let us know when a document number is known so that we can inform anyone who inquires about this final report.

Any publications that are included with this report are highlighted on the list below.

Attachment
cc: Sherri Diana, EID, P03/C13

List of Publications

Thomas JS, Lavender SA, Corcos DM, Anderson GBJ: Trunk Kinematics and Trunk Muscle Activity during aA Rapidly Applied Load. Journal of Electromyography and Kinesiology, accepted, 1998

Title: Do Lifting Belts Protect the Back During Sudden Loading
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Award Number: 5 R03 OH003339-02
Start & End Date: 9/1/1995–8/31/1997
Total Project Cost: $72,597
Program Area: Musculoskeletal Disorders: Low Back
Key Words:

Abstract:
The biomechanical role lifting belts play in the prevention of low-back injury has been the subject of considerable debate. Recent studies have shown that lifting belts restrict trunk motion in the frontal and transverse planes during passive motion tests and when lifting. It is theorized that through this restriction of trunk motion, or in other words the stiffening of the torso, that lifting belts may protect the back from injury, particularly when the body is subjected to sudden unexpected loads. Epidemiological studies have indicated that sudden unexpected events, whether they consist of a sudden load imposed upon the body or a rapid trunk motion during a slip, are frequently related to the onset of low-back disorder (LBD's). Biomechanically, these events create large internal loadings on the spine and it's supporting structures as the muscles attempt to null the perturbation. It was hypothesized that the stiffening effects of the lifting belt may protect workers' backs from the extreme loads encountered during unanticipated loading events. Thus, the objective of the two studies was to determine whether lifting belts protect individuals exposed to sudden loading of the torso. The approach taken in both studies was to simulate sudden loading by rapidly applying a load either directly to the torso (experiment 1) or via a container held in the hands (experiment 2). In the "unexpected" trials within each study, the subjects were blindfolded and auditory cues were masked so that the temporal onset of the loading could not be determined. In half the trials, a lifting was belt was tensioned versus the remaining half of the trials where the same belt was extremely loose. Half of the trials were symmetric about the torso's mid-sagittal plane, and in half it was asymmetric (45 degrees). In each study, 8 trunk muscles were sampled with surface electromyography (EMG) prior to and during the sudden loading. In the first experiment, the subject's pelvis was fixed to a reference frame structure and the loads were applied directly to the torso. This allowed for the isolation of the trunk response to sudden loading independent of other body segments. In this study, 20 subjects, 10 male and 10 female, experienced 24 sudden loads (3 trials of each combination of the belt, expectancy, and asymmetry conditions). When the unexpected trials were examined, the benefits of the lifting belt were only apparent during the asymmetric loading conditions. On average, the normalized left and right erector spinae EMG decreased from 31 to 28 percent of the maximum voluntary contraction level (MVC), respectively. Kinematic changes measured with a Lumbar Motion Monitor (LMM), with the exception of a slight decrease in lateral bending ($p<.05$), were not existent when the belt was tensioned. No changes were observed in the EMG or the trunk kinematics prior to the unexpected loading with the lifting belt tensioned.

In the second unexpected loading study, free-standing subjects were loaded by rapidly applying a force to a container held in the hands. The 10 male and 8 female subjects participated in 12 loading trials that investigated all combination of the three independent variables: belt use, preload, and load symmetry. The applied force was normalized to each subject's isokinetic trunk strength. Kinetic and kinematic data were obtained using two forceplates combined with an electro-magnetic motion measurement system (The Motion Monitors). These data allowed for the determination of postural changes and the computation of moments using a bottom-up model. EMG data were obtained from eight trunk muscles following the onset of the sudden unexpected load. In this study, the belt reduced forward bending of the spine during symmetric unexpected loadings. In females, the belt slightly reduced the lateral bending of the spine. In males, the belt reduced the flexion moment acting on the spine. Little change was found in the peak EMG signals from the posterior muscles during the symmetric loadings; however, there was a reduction in peak response from two of the anterior muscles. With asymmetric unexpected loadings, the peak EMG values increased in the posterior contra-lateral muscles and in the contra-lateral External Oblique. Peak activity in the ipsilateral Erector Spinae was reduced with the lifting belt.

In sum, these results suggest the benefits of the lifting belt may be limited to unexpected loadings that are sagittally symmetric. Even though the flexion moment benefit continues to persist with the asymmetric unexpected loads, the additional contra-lateral muscle recruitment associated with the belt may increase the risk of muscle overexertion injury, thereby offsetting the potential benefit of reduced muscle recruitment ipsilaterally. Given that unexpected loads are unpredictable by definition, and will likely involve some degree of asymmetry, the data reported here suggest that a lifting belt may be of little help.

Publications

Thomas JS, Lavender SA, Corcos DM, Anderson GBJ: Trunk Kinematics and Trunk Muscle Activity during aA Rapidly Applied Load. Journal of Electromyography and Kinesiology, accepted, 1998

Final Performance Report

DO LIFTING BELTS PROTECT THE BACK DURING SUDDEN LOADING?

NIOSH Research Grant 5 R03 OH03339-02

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June 23, 1998

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List of Abbreviations

<u>General</u>

LBD(s)	Low back disorder(s)
EMG	Electromyography
NEMG	Normalized electromyographic signal
MVC	Maximum voluntary contraction

<u>Muscle Names</u>

ERSL	Left Erector Spinae
ERSR	Right Erector Spinae
EXOL	Left External Oblique
EXOR	Right External Oblique
LATL	Left Latissimus Dorsi
LATR	Right Latissimus Dorsi
LGTL	Left Longissimus Thoraces
LGTR	Right Longissimus Thoraces
RABL	Left Rectus Abdominus
RABR	Right Rectus Abdominus

<u>Units:</u>

N	Newtons
Nm	Newton-meter
%MVC	Percentage of maximum voluntary contraction

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SIGNIFICANT FINDINGS

This study tested the following hypotheses with regards to the role of lifting belts play during sudden unexpected loading conditions:

Hypothesis

The use of a lifting belt stiffens the torso during sagittally symmetric and asymmetric sudden perturbations as evidenced by a reduction in the trunk motion in the sagittal, frontal, and transverse planes.

Findings

The lifting belt did not result in any change in the trunk motion in the first study where the subject, standing in a apparatus that prevented pelvic rotation and lower extremity motion, was loaded via a chest harness. Thus, limiting the degrees of freedom with respect to compensatory motion to the lumbar and lower thoracic spine.

In the second study, where the free-standing subjects were holding a box in the hands to which the sudden loading was applied, a significant reduction in the forward bending of the torso was observed during the sagittally symmetric loadings. Further, there was a significant reduction in the torso flexion to hip flexion ratio with the lifting belt tensioned indicating that the increased spine stiffness resulted greater motion at the hips.

Hypothesis

The lifting belt reduces the peak electromyographic activity of the major trunk muscles supporting the torso during sagittally symmetric and asymmetric sudden perturbations.

Findings

In the first study the Erector Spinae muscles showed significantly less activity in response to the sudden loadings with the lifting belt. And while the decrement was approximately 9 percent relative to the conditions without the lifting belt, the absolute change in the normalized peak electromyographic signal was only about 3 percent of the signals maximum range.

The previous finding was not replicated in the second study in which the subjects were free-standing. In fact, the two contra-lateral posterior muscles showed increased activity with the belt during the asymmetric unexpected loadings in the male subjects. Posterior muscle activity was unchanged for all subjects during the symmetric loadings and unchanged for the female subjects during the unexpected loadings when the lifting belt was used. However, the lifting belt was associated with a reduction in the peak electromyographic activity in right External Oblique and Rectus Abdominus muscles irrespective of the load orientation, and in the right Erector Spinae (ipsilateral side) during the asymmetric loadings.

Hypothesis

The use of a lifting belt results in a greater sharing of the peak bending moments acting on the spine with the articulations of the lower extremities, thereby, reducing the magnitude of the impulse loaded delivered to the torso.

Findings

The spine moments were computed based on kinematic and ground reaction force (kinetic) data obtained during the second study. In males, the peak spine flexion moments were reduced during the unexpected sudden loadings when the lifting belt was used by approximately 9 percent. The flexion moments at the right hip in the males also decreased by a similar amount. The moments at the left hip and both knees were unchanged due to the lifting belt. There were no changes in any of the moments measured in the female subjects on account of the belt.

USEFULNESS OF FINDINGS

The overall goal of this research was to provide information as to the biomechanical impact of wearing a lifting belt in a sudden unexpected loading situation. Sudden unexpected loads have been shown by epidemiological studies to be events that trigger the onset of low-back pain. The information provided in the following report will assist those who are responsible for the health and safety of employees that perform manual material handling tasks (lifting, etc.). Essentially, this report shows that there was no clear cut benefit to the lifting belt. The first study in which the subject was in a constrained posture demonstrated a small but significant reduction in the Erector Spinae electromyographic signal with the lifting belt. In the second study with free unconstrained subjects, there was a reduction in total postural disturbance with the belt but there was no clear benefits from the electromyographic signals. Thus, this report suggests that a lifting belt is not a panacea for the prevention of back injuries due to sudden loading. And that the best approach to preventing back injuries which result from unexpected loading is to focus on prevention the loading event.

ABSTRACT

The biomechanical role lifting belts play in the prevention of low-back injury has been the subject of considerable debate. Recent studies have shown that lifting belts restrict trunk motion in the frontal and transverse planes during passive motion tests and when lifting. It is was theorized that through this restriction of trunk motion, or in other words the stiffening of the torso, that lifting belts may protect the back from injury, particularly when the body is subjected to sudden unexpected loads. Epidemiological studies have indicated that sudden unexpected events, whether they consist of a sudden load imposed upon the body or a rapid trunk motion during a slip, are frequently related to the onset of low-back disorder (LBD's). Biomechanically, these events create large internal loadings on the spine and it's supporting structures as the muscles attempt to null the perturbation. It was hypothesized that the stiffening effects of the lifting belt may protect workers' backs from the extreme loads encountered during unanticipated loading events. Thus, the objective of the two studies was to determine whether lifting belts protect individuals exposed to sudden loading of the torso. The approach taken in both studies was simulate sudden loading by rapidly applying a load either directly to the torso (experiment 1) or via a container held in the hands (experiment 2) In the "unexpected" trials within each study the subjects were blindfolded and auditory cues were masked so that the temporal onset of the loading could not be determined. In half the trials a lifting was belt was tensioned versus the remaining half of the trials where the same belt was extremely loose. Half of the trials were symmetric about the torso's mid-sagittal plane and in half it was asymmetric (45 degrees). In each study 8 trunk muscles were sampled with surface electromyography (EMG) prior to and during the sudden loading. In the first experiment the subject's pelvis was fixed to a reference frame structure and the loads were applied directly to the torso. This allowed for the isolation of the trunk response to sudden loading independent of other body segments. In this study 20 subjects, 10 male and 10 female, experienced 24 sudden loads (3 trials of each combination of the belt, expectancy, and asymmetry conditions). When the unexpected trials were examined the benefits of the lifting belt were only apparent during the asymmetric loading conditions. On average, the normalized left and right erector spinae EMG decreased from 31 to 28 percent of the maximum voluntary contraction level (MVC), respectively. Kinematic changes measured with a Lumbar Motion Monitor (LMM), with the exception of a slight decrease in lateral bending ($p<.05$), were not existent when the belt was tensioned. No changes were observed in the EMG or the trunk kinematics prior to the unexpected loading with the lifting belt tensioned.

In the second unexpected loading study free-standing subjects were loaded by rapidly applying a force to a container held in the hands. The 10 male and 8 female subjects participated in 12 loading trials that investigated all combination of the three independent variables: belt use, preload, and load symmetry. The applied force was normalized to each subjects isokinetic trunk strength. Kinetic and kinematic data were obtained using two forceplates combined with an electro-magnetic motion measurement system (The Motion Monitor™). These data allowed for the determination of postural changes and the computation of moments using a bottom-up model. EMG data were obtained from eight trunk muscles following the onset of the sudden unexpected load. In this study the belt reduced forward bending of the spine during symmetric unexpected loadings. In females the belt slightly reduced the lateral bending of the spine. In males the belt reduced the flexion moment acting on the spine. Little change was found in the peak EMG signals from the posterior muscles during the symmetric loadings, however, there was a reduction in peak response from two of the anterior muscles. With asymmetric unexpected loadings the peak EMG values increased in the posterior contra-lateral muscles and in the contra-lateral External Oblique. Peak activity in the ipsilateral Erector Spinae was reduced with the lifting belt. In sum, these results suggest the benefits of the lifting belt may be limited to unexpected loadings that are sagittally symmetric. Even though the flexion moment benefit continues to persist with the asymmetric unexpected loads, the additional contra-lateral muscle recruitment associated with the belt may increase the risk of muscle overexertion injury, thereby offsetting the potential benefit of reduced muscle recruitment ipsilaterally. Given that unexpected loads are unpredictable by definition, and will likely involve some degree of asymmetry, the data reported here suggest that a lifting belt may be of little help.

BODY OF REPORT

BACKGROUND AND SIGNIFICANCE

2.1 Introduction

Occupational low back pain represents an enormous cost to society both financially and in terms of morbidity. Epidemiological studies indicate that up to 80 percent of the population can be expected to suffer from low back pain sometime in their lives (Andersson, 1991). In a majority of cases this pain is considered idiopathic, thereby making treatment difficult and often expensive. Several occupational factors have been explored as to their contribution to low back disorders (LBD's). Frequently combinations of lifting, bending, twisting, and general material handling tasks are described as the precursors of back injuries (Andersson, 1991; Bigos et al., 1986; Frymoyer et al., 1983, Punnett et al., 1991). Furthermore, low back pain represents the leading cause of activity limitation in individuals under 45 years of age (Andersson, 1991). It is this age group who performs most of the occupational manual material handling tasks.

Snook et al. (1978) evaluated the effectiveness of three potentially viable preventative approaches taken toward controlling occupational LBD's: job design, training, and employee selection. These authors concluded that job design was the most promising of the three alternative approaches and attempts to prevent the incidence of low-back disorders (LBD's) through ergonomic redesign have enjoyed good success in some manufacturing environments. Unfortunately, in many work environments jobs are not easily redesigned due to constantly changing work demands. For example, those employed in the construction industry, delivery type jobs, or in nursing occupations perform frequent lifting but rarely under repeated circumstances. Nurses report the use of mechanical assistive devices (ie: hoists) to be awkward. Owen and Garg (1991) report that where assistive devices lowered the biomechanical stress on the handler, the patient comfort ratings were less than satisfactory. Thus, in the nursing environment where the work layout and location is constantly changing, job design principles are exceptionally difficult to put into place. Similar problems are encountered in construction and delivery occupations.

Historically, ergonomic efforts in environments where quick "fixes" are not available have relied heavily on training employees in the proper techniques for lifting. Snook et al. (1978) concluded that such training programs were ineffective at reducing low back injury claims. More recently, Daltroy et al. (1997) reported that a controlled trial of "back-school" type program, designed to teach safe lifting techniques, was ineffective at controlling LBDs in their study of 4000 postal employees. Thus, one of the approaches taken to control LBD's in jobs comprised of these less structured tasks has been to use lifting belts.

Personal protective devices are not usually recommended when developing ergonomic control measures. Such devices rely on employee compliance in order to be effective. However, more and more organizations are now requiring employees in occupations with substantial material handling demands to wear lifting belts. Unfortunately there is limited epidemiological data supporting the use of lifting belts (NIOSH, 1994).

Walsh and Schwartz (1990) found that a group which received a combination of lumbosacral orthosis and back school had significantly fewer lost days than a group receiving only the back school, or a control group which did not experience either treatment. Strength, productivity and LBD incident rate were unaffected by the back support or the back school. However, when the treatment group was sub-divided according to back injury history, the reduced time away from work was only found in the group that had a history of back injury.

Reddell et al. (1992) recently studied the back injury rates over an eight month period in 642 airline baggage handlers. The employees were divided into four treatment groups: belt only, belt plus training, training, and a control group with no intervention. Over the course of the study there were no significant differences between any of these four groups with regard to incident rates, lost time, or restricted time. But it should also be noted that 58 percent of the subjects in the groups administered lifting belts discontinued using the belt prior to the completion of the eight month study. Accompanying surveys indicated that the belts were too hot, rubbed, pinched, and bruised the lower ribs. Moreover, the data indicated that those who were originally issued belts but discontinued their use reported more LBD's than the control group that received no belt at all. This suggests the possibility of physical deconditioning with extended use of the belt. Mitchell et al. (1994) reported that the low-back injuries sustained by workers wearing a lifting belt were more costly and required more intensive treatment than the injuries experienced by workers who did not wear a lifting belt. More recently, Kraus et al (1996), reported that lifting belts reduced injury rates by about a third in a study of their implementation within a large home-improvement retail chain. Best results were obtained for employees who performed jobs characterized as having a low intensity level of lifting, or in younger (<35 years) or older (55 years or more) employees who performed high intensity lifting. The employees who showed the greatest benefit of wearing a lifting belt had been with the company between one and two years.

There are two theorized biomechanical mechanisms by which lifting belts assist the spine and it's supporting tissues during lifting activities. First, lifting belts are believed to increase the intra-abdominal pressure (IAP). Second, lifting belts are believed to alter the body's kinematic and kinetic response to material handling situations. Such changes in lifting style would potentially reduce the external moments acting on the spine during lifting and or reduce the occurrence of hazardous trunk motions, for example, twisting. Each of these mechanisms will be reviewed in greater detail.

Intra-Abdominal Pressure

The pressure within the abdomen potentially creates an extension moment on the spine (Bartelink, 1957; Morris et al., 1961), thereby reducing the tension in the back muscles required to complete an exertion or lifting task. The reduced tension in the back muscles in turn reduces the mechanical loads transmitted to the spine. Early studies have documented that IAP increases with increased external load (Davis, 1956; Morris et al., 1961; Andersson, 1977), although, less so during dynamic as opposed to static trunk exertions (Cresswell, 1993, Marras et al., 1985). Nachemson et al. (1986) reported that the increased IAP generated in a valsalva maneuver led to increased paraspinal muscle activity and increased intra-discal pressure, thereby questioning the true role IAP plays in spinal loading.

Several investigators have shown lifting belts or orthoses to increase the IAP measured during lifting tasks (Harman et al., 1989; Lander et al., 1990, 1992; Liggett 1989; McGill et al., 1990) while others have not (Kumar and Godfrey, 1986; Nachemson et al., 1983). Trained weight lifters reportedly increased their IAP with lifting belts only when loads magnitudes reached 90 percent of maximal strength capacity (Harman et al., 1989; Lander et al., 1990). Liggett (1989) found that IAP increased with lifting belts when weights lifted from the floor to knuckle level were 25, 50, and 75 percent of the lifters one trial maximum lift. However, with maximal lift values ranging from 184 to 245 kg, the 25 percent values were far beyond most industrial loading tasks. McGill et al.'s (1990) subjects were instructed to select a load which was heavy yet could be safely lifted. This instruction resulted in loads that ranged from 73 to 91 kg. The resulting IAP were 21 percent greater with the lifting belt. McGill noted that the erector spinae electromyographic (EMG) activity decreased with breath holding, thereby supporting the notion that IAP does create an extension moment. However, the erector spinae EMG activity was unaffected by lifting belt use.

It should be emphasized that the lifting belt studies above focussed on athletes or individuals working at very near their maximal capacity. The situation is very different in most occupational settings. In most cases the loads are smaller, the lifting is much more repetitive, and the lifting tasks frequently involve twisting and lateral bending. Kumar and Godfrey (1986) measured IAP in 20 subjects performing manual material handling tasks similar to those observed in industry. These authors fitted each subject with 6 different abdominal supports. The results showed that as these subjects performed a variety of 7 kg and 9 kg lifts, both symmetric and asymmetric with regard to the body's mid-sagittal plane, the IAP was unaffected by the type of abdominal support or whether a support was used at all. Similarly, no clear trend was apparent in the IAP's reported by Nachemson et al. (1983) as they tested three different orthoses in response to flexion, extension, lateral bending, and torsional isometric loads of 15 to 20 kg. In summary, it appears that any IAP benefit gained from lifting belts is not apparent when low to medium loads are handled.

Lifting Kinetics and Kinematics

As mentioned above, several epidemiological studies have associated deviated trunk postures with increased risk of LBD. Of particular concern are tasks that require extreme forward bending, lateral bending and twisting (Andersson, 1991; Frymoyer et al., 1983; Marras et al., 1993; Punnett, 1991). Some investigators have evaluated whether lifting belts affect the trunk kinematics. Reduced trunk motion would effectively result in a changed lifting style. Previous research has shown that the peak moments predicted at L5/S1 with a biomechanical model have been shown to be dependent upon lifting style and lifting speed (Buseck et al., 1988; Bush-Joseph, 1988; De Looze et al., 1993; Dolan et al., 1994; Gagnon and Gagnon, 1992; Schipplein, 1991).

Lander (1990), used cinema-graphic data to quantify the trunk kinematics and showed that there were no differences in the absolute or relative joint angles in the sagittal plane for the knee or the torso when their subjects lifted with and without lifting belts. McGill et al. (1994) investigated the change in the passive bending resistance of the torso with the lifting belt. These authors used biofeedback via EMG to insure the muscles were not recruited to resist the bending moments applied to the torso. They found that the bending resistance of the torso with the belt was unchanged in response to bending moments in the sagittal plane. However, these authors did observe reduced frontal plane and transverse plane motion with the lifting belt. Thus, they concluded that the passive resistance to lateral bending and twisting motions was increased with the lifting belt.

Lavender et al. (1995) quantified the changes in trunk kinematics due to a lifting belt as subjects performed a lifting task with varying degrees of asymmetry. Trunk motions were measured with the lumbar motion monitor as subjects lifted a box and placed it on a shelf. Subjects participated in two lifting sessions: one session with the lifting belt and one session without the lifting belt. In half of the lifts foot movement was restricted, whereas in the other half of the lifts foot movement was encouraged. These authors reported that the twisting and lateral bending motions were dependent upon the both the belt and restrictions in foot movement during the asymmetric lifts. Their results showed that the frontal and transverse plane motions were reduced in asymmetric lifting when foot motion is encouraged, but that the transverse plane motions were even further reduced when a lifting belt was worn. The trunk's angular velocity and acceleration in each of these planes was also significantly reduced with the lifting belt.

This latter study was limited in that the internal muscle forces were not evaluated. It is possible that the limited motion could have been accompanied by larger internal muscle forces due to the motion resistance. These studies do suggest however, that the lifting belts potentially stiffen the trunk with regard to preventing motion in the frontal and transverse planes. Thus, should the torso be subjected to a sudden perturbation in an oblique plane, the lifting belt should act as a stiffening aid and reduce the internal muscle force required to stabilize the torso.

Sudden Loading

Manning et al. (1984) determined that 66 percent of the back injuries recorded in an industrial setting were preceded by some type of underfoot accident. Often these were slips without falls (Manning et al., 1984). This suggests an injury scenario whereby the neuromuscular system over-reacts to an unanticipated event and in the process damages tissue containing nociceptors. A similar scenario can be expected to occur when a sudden load is imposed upon the body. For example, a sudden load applied to the hands will require the rapid generation of muscular forces in the torso in order to maintain the body's stability, where stability is defined as the maintenance of the body's balance and posture. Such a scenario describes 12.3 percent of accidental injuries evaluated by Mitchell et al. (1983). Similarly, Magora (1973) reported an epidemiological link between the frequency of sudden maximal efforts, especially when unexpected, and the occurrence of occupational low back pain.

The consequences of sudden unexpected loads or sudden changes in postural stability have been investigated as sudden loads were applied to the hands (Carlson et al, 1981; Marras et al., 1987), as loads were applied to the torso directly (Omino and Hayashi, 1992) and during impending falls (Romick-Allen and Schultz, 1988). All of these events lead to increased loading of the spine and it's supporting tissues. In sum, when the system is unexpectedly loaded a startle response is generated wherein the system "over-reacts" (Greenwood and Hopkins, 1976). This response further tensions the muscles and accentuates the mechanical loading of the spine.

An individual's expectancies regarding the temporal occurrence of a loading event will significantly affect the magnitude of the startle response. Studies evaluating sudden loading of the torso have reported strong relationships between warning time and muscle response (Lavender et al., 1989; Marras et al., 1987). For example, as warning time was increased from 0 to 400 ms, the severity of the impulse load delivered to the spine decreased (Lavender et al., 1989). It was theorized that the internal loadings were reduced through the formation of temporal expectancies as to when the sudden loading would occur. Similarly, expectancies as to the magnitude of the loading were theorized to develop. When subjects were led to expect a 25 pound lift, Khalil et al., (1990) found that the EMG response of the biceps to an actual 5 pound lift was equivalent to that of a 25 pound lift. Butler et al. (1993) found a jerking motion and a significant increase in the peak L5-S1 moment, as detected via kinetic and kinematic analyses, when the magnitude of the lifted load was much less than anticipated. Pattersson et al. (1986) reported an oscillatory lifting pattern under such conditions while the body attempted to adjust it's response based on the new information regarding the load's magnitude.

While the epidemiological studies into sudden loading have not been precise enough to identify the directional component of the perturbation responsible for the resulting LBD's, several biomechanical factors suggest that it may be the loads in the oblique planes which create the greatest LBD risk. Asymmetric spine loading leads to an increase in the restorative moments generated by the contra-lateral musculature (Ladin et al., 1989, 1991; Lavender et al., 1992 Marras and Mirka, 1992; Seroussi and Pope, 1987). Lavender et al. (1989) showed that when sudden loads were applied which would create forward bending and right lateral bending moments on the spine, collectively the peak responses of the contralateral (left) Latissimus Dorsi and Erector Spinae increased by 37 percent. Conversely, the ipsilateral posterior muscles (right side) showed a 55 percent decrease in their peak combined response. These results indicate the extreme disparity between the left and the right sides of the body under such conditions. This type of loading would be expected to increase the lateral shear forces acting on intervertebral disks and the potential for injury. Moreover, because fewer muscles are anatomically arranged to resist the complex bending moments generated during asymmetric loading there is greater recruitment of antagonistic muscles in order to stabilize the torso. This increased co-contraction will lead to increased compression on the lumbar spine.

The internal muscle responses prior to voluntary or anticipated involuntary motions have been observed using EMG. Several investigators have published data showing EMG activity in the Gastrocnemius prior to expected destabilizing perturbations (Bouisset and Zattara, 1981; Branch et al., 1991; Cordo and Nashner, 1982). Similarly, when subjects were dropped from heights between 20 and 120 cm an anticipatory activation of the Soleus muscle prior to landing was observed (Greenwood and Hopkins, 1976). The onset of the muscle activity in this study occurred at a consistent time prior to contact with the ground. Likewise, muscle activity under sudden loading conditions has been shown to begin earlier when adequate warning time is available (Lavender et al., 1989). These studies suggest that the body's natural response to sudden unanticipated events is to stiffen the joints. This increased stiffness reduces the amplitude and the time delay of the musculoskeletal response (Winters et al., 1988). Thus, tissue strain is reduced and the disturbance to postural stability is minimized.

Normally increased stiffness of the musculoskeletal response is achieved through the coactivation of antagonistic muscles (Humphery and Reed, 1983). In the torso several muscle groups could contribute to the stiffness in the lumbar intervertebral joints. The nature of the muscular coactivation has been shown to be a function of several factors, for example, the rate of force development across a joint (Gordon and Chez, 1984), whether there is spatial uncertainty regarding the required response, whether a compensatory force correction is required (De Luca and Mambrito, 1987), the magnitude of limb deceleration (Meinck et al., 1984), and the behavioral control strategy to unanticipated events (Flanders and Cordo, 1987). In the torso the coactivation would increase both the compressive and shear forces placed on the spine.

Lavender et al. (1993) reported that the four subjects tested in a sudden loading paradigm developed ways to stiffen their torsos primarily through co-contraction of the anterior muscles. After five testing sessions all of the subjects showed significant changes in their response to the sudden loading. All of the subjects reduced their forward trunk flexion during the sudden loading, and reduced the peak compressive loads acting on their spines as determined via an EMG driven biomechanical model. It was theorized by these investigators that the changes in the coactivation during the preparatory period resulted in the increased stiffness observed during the loading. This increased stiffness reduced the destabilizing effects of the loading, thereby reducing the internal forces acting on the spine. In a second study Lavender and Marras (1995) showed that when a warning signal was available to enhance the subject's temporal expectations the subjects' preparatory strategies shifted to allow an even stiffer trunk response to the sudden loading. Specifically, all subjects increased their posterior muscle responses prior to the sudden loading by between 18 and 95 percent. In conclusion, the subjects attempted to use warning information to further stiffen their torsos during the perturbation.

<u>Summary</u>

Epidemiological studies have demonstrated a link between sudden loading and the onset of low-back disorders (Manning et al., 1984). Biomechanical studies have shown dramatic increases in the muscle and joint loads in response to the sudden loading. The literature suggest there are two components to sudden loading which make it particularly hazardous. First, there is the dynamic component of the loading. The effects of this component have been observed by comparing EMG data when anticipated loads were rapidly applied versus the identical loads held in a static posture (Marras et al., 1987). Second, there is the expectation component. In many cases where back injuries have occurred the loading was unanticipated. This latter component suggest that when individuals are forewarned as to the potential for sudden loading, the trunk is prepared in such a way that it provides a stiffened response to the sudden loading. Thus, while it appears that the key to the prevention of these injuries may be in the body's preparation, often this is not possible.

Recent research into the effects of lifting belts on trunk kinematics has indicated that the belts do limit the trunk motion in the frontal and transverse planes. Limiting the torso displacement in these directions during a sudden perturbation should reduce the additional bending and torsional moments generated by the upper body mass. Thus, less muscle force would be required to restore the trunk to its pre-perturbation posture if a lifting belt stiffens the torso under these conditions. As a result, lifting belts <u>potentially</u> protect workers who are subjected to unexpected loading events or perturbations.

SPECIFIC AIMS

The goal of the research performed here was to provide information that can reduce the likelihood of low-back pain being triggered when the body is subjected to unexpected suddenly applied loads. Immediately after such loadings occur, as the body recovers from the sudden perturbation, large internal forces are generated thereby creating an enormous potential for injury. While previous research has shown that when sudden loads can be anticipated the magnitude of the muscle response is substantially reduced, unfortunately, in many situations there is little or no warning as to when a sudden loads will occur. The stiffening of the torso seen in recent research using lifting belts suggest that a lifting belt may prove beneficial to those in occupations in which sudden unexpected loading of the torso occurs frequently. Thus, this research addressed the issue of whether lifting belts protect the torso through reducing the kinematic response (increased stiffness) and or through reduced activation of the trunk muscles following a sudden perturbation.

The following specific hypotheses were tested as part of this study:

1. A lifting belt stiffens the torso during sagittally symmetric and asymmetric sudden perturbations as evidenced by a reduction in the trunk motion in the sagittal, frontal, and transverse planes.

2. A lifting belt reduces the peak electromyographic activity of the major trunk muscles supporting the torso during sagittally symmetric and asymmetric sudden unexpected loading.

3. During unexpected sudden loading the effectiveness of a lifting belt in minimizing the muscular loading of the torso is enhanced relative to expected sudden loading conditions.

4. The use of a lifting belt results in a greater sharing of the peak bending moments acting on the spine with the articulations of the lower extremities, thereby, reducing the magnitude of the impulse load delivered to the torso.

EXPERIMENT 1: The effect of lifting belts on trunk muscle activation during sudden unexpected loading.

METHODS

Experimental Design

A mixed model design was used, in which the independent variable gender was a between subjects factor and, the independent variables of direction of the applied load (symmetric, asymmetric), and the tension of the lifting belt (very loose, tight) were the within subject factors. The session consisted of 3 trials at each combination of the experimental conditions with a minute rest between trials. The sequence of the 24 trials was randomized.

The dependent measures included the trunk kinematics, measured with the Lumbar Motion Monitor (LMM) (Chattanooga Corp), and the normalized surface EMG from the left and right: longissimus thoraces (LGTL and LGTR), erector spinae (ERSL and ERSR), external oblique (EXOL and EXOR), and rectus abdominus (RABL and RABR).

Subjects

Twenty subjects (10 male and 10 female), 20 to 33 years of age participated in this study. The subjects were screened for a history of low back pain (LBP). Only individuals with no history of low back pain in the last year were allowed to participate in the study.

Apparatus

The subjects stood in a reference frame constructed of steel tubing. In the center of the reference frame was a smaller structure which allowed the experimenter to secure the subject in an upright position so that motion below the lumbar spine was restricted. The sudden load was delivered to the subject 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 meter. The weight of the bag was normalized to a value that was 5% of each individuals maximum isometric trunk extensor strength. The applied load was delivered both in the mid-sagittal plane, and in an oblique plane rotated 45 degrees to the right of the mid-sagittal plane, for the symmetric and asymmetric conditions respectively (Figure 1.1). A load cell attached to one of the pulleys was utilized as an event marker.

Disposable surface EMG electrodes (Nikomed Corp) were used for this study. The inter-electrode distance was 2 cm. The EMG signals were pre-amplified (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 msec. The integrated

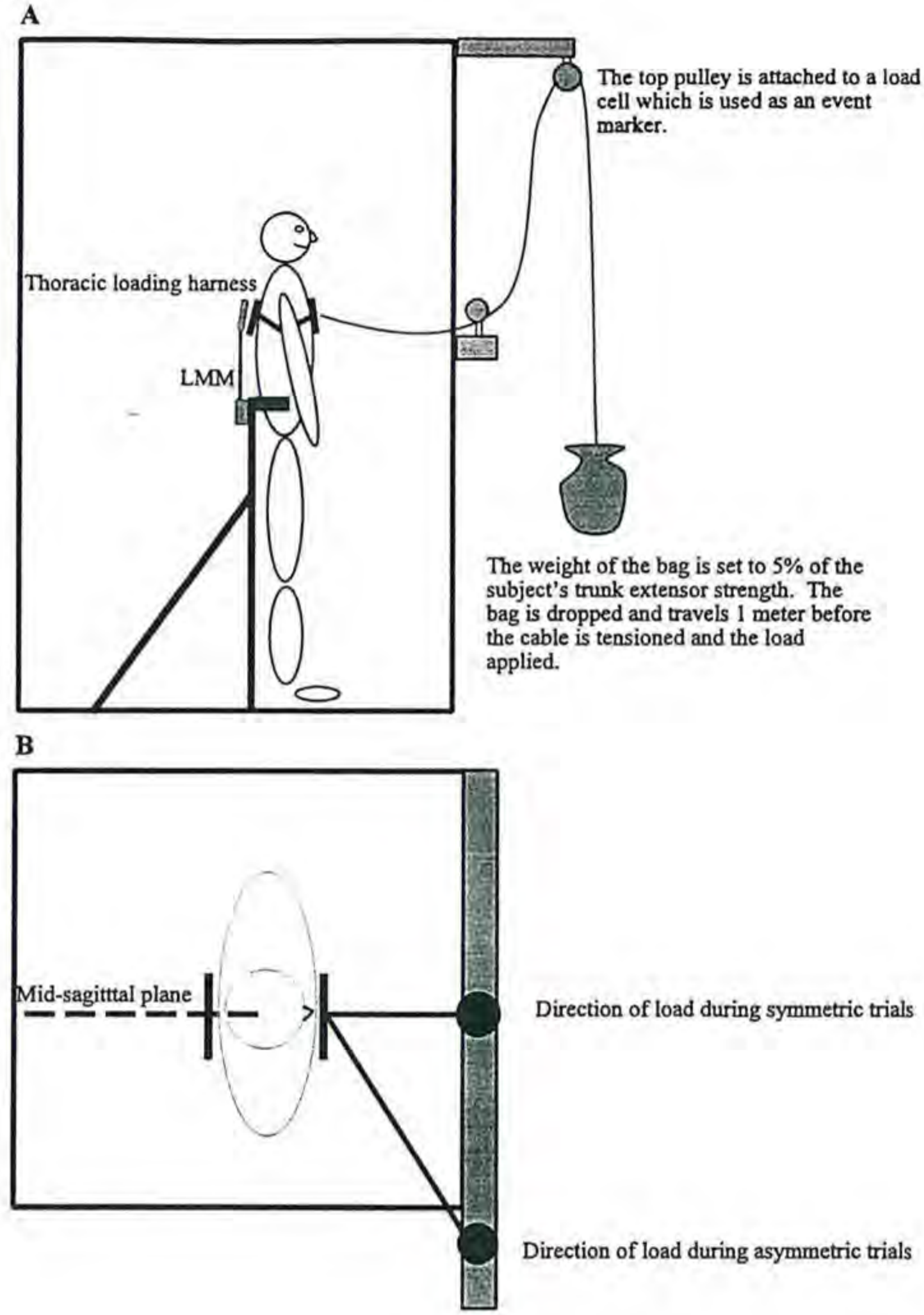

Figure 1.1: Apparatus for Experiment 1. Subject's pelvis was fixed and load was applied through a thoracic harness.

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 tri-axial electro-goniometer), 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 et al. (1992)

Procedure

Upon entering the laboratory, subjects signed an informed consent which described the protocol of the study. The subjects' height, weight, age, leg length, lumbar spine length (L5 to T1), and waist circumference were measured.

Surface EMG electrodes were placed on the skin overlying the muscle bellies of the eight trunk muscles. The electrodes were attached at the level of T10 approximately 4 cm from the midline for the LGTL and LGTR, and at the level of L3, approximately 4 cm from the midline for the ERSL and ERSR. Electrodes for the RABR and RABL were attached at the level of the umbilicus 2 cm from the midline. Placement for the EXOL and EXOR electrodes was at the level of the umbilicus, approximately halfway between the iliac crest and the anterior superior iliac spine. This is usually 2 cm medially and 2 cm laterally from these respective bony landmarks and rotated 45 degrees from the vertical. The common ground electrode was attached between the sixth and seventh rib in the mid-axillary line. The skin at these sites was cleaned with alcohol and lightly abraded. Baseline or resting EMG values were recorded with the subject standing in a relaxed posture.

Maximum isometric muscle forces, for the purpose of EMG normalization, were measured by having the subjects perform resisted isometric trunk flexion, extension and rotation. The subject stood in the reference frame with the pelvis firmly secured, a harness was attached to the thoracic region, that in turn was connected to a dynamometer via steel cable. The subject 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 minute intervals until the force measured from each muscle group no longer increased and the two greatest trials were within 10 percent 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.

The subject remained in the reference frame with his or her pelvis secured as described above. The method of attaching the LMM to a subject was modified in this experiment. The base of the unit was secured to the stand in which the subject's pelvis was secured. The LMM was adjusted so that the base was aligned with the subject's lumbosacral junction. The thoracic attachment of the LMM was attached with a modified loading harness system, and secured so that the top of the LMM was aligned with the third thoracic vertebrae. The modified attachment of the LMM allowed us to restrict the contributions of the pelvis and lower extremities during the sudden load while not limiting trunk motion. Furthermore, the modification of the thoracic attachment enabled the experimenters to deliver the sudden load through the thoracic harness at the level of the tenth thoracic vertebrae. Padding was placed around the EMG electrodes to prevent compression from the thoracic harness and from the lifting belt. The subject was then instructed to stand straight with the eyes facing forward and a measure of relative neutral posture was recorded.

The load was applied either symmetric to, or asymmetric to the torso's mid-sagittal plane. During symmetric loading, the cable was attached to the mid-point of the thoracic loading harness and run through two pulleys in the mid-sagittal plane. During asymmetric loading, the cable was attached to the mid-point of the thoracic harness and was run through two pulleys set 45 degrees to the right of the mid-sagittal 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. The lifting belt was worn loosely for the no belt trials (enough space to slide a hand between the belt and the abdomen), or fully tensioned for the belt trials. The subject wore a blindfold to block any visual cues and a noise generator was used block all auditory cues of the impending load. Trunk kinematics and EMG data were collected for 1 second prior to the release of the load and for 2 seconds after the suddenly applied load.

<u>Data Treatment</u>

The integrated EMG (IEMG) data were normalized for each subject with respect to the EMG data collected during maximal trunk exertions, and to the resting EMG levels according to the equation (1).

$$\mathbf{NEMG(i,j)= (IEMG(i,j)- REST(i))/ (MAX(i)- REST(i))} \qquad \mathbf{(1)}$$

Where:

i=	muscles 1 through 8
j=	experimental conditions 1 through 4
NEMG(i,j)=	the normalized EMG for muscle i in condition j
IEMG(i,j)=	the current Integrated EMG value for muscle i in condition j
REST(i)=	the minimum resting IEMG value for muscle i for the relaxed standing measure
MAX(i)=	the maximum IEMG value from muscle i during the maximal isometric exertion

The pre-load area of normalized EMG, was used as a measure of the muscle activity prior to the onset of an impending load, was determined by equation (2).

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

Where:

i=	muscles 1 through 8
j =	experimental conditions 1 through 4
NEMG(i,j)=	the normalized EMG for muscle i in condition j
m=	sample coinciding with the onset of activity in muscle i
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 utilized in this investigation. Trunk position data were obtained from the LMM. The position data were smoothed with a three point moving average, and the second central point difference method was used to calculate the velocity. The same procedure was repeated on the velocity data to determine acceleration.

Four-way multivariate analyses 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 4-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 4-way mixed model analysis of variance (ANOVA) procedures on individual muscles or kinematic variables.

RESULTS: EXPERIMENT 1

Analysis of the lifting belt on peak normalized EMG

There was a significant interaction of the belt condition and the direction of the applied load on the peak normalized EMG of the left and right erector spinae ($p<0.01$, $p<0.009$) (Figure 1.2). Analysis of the simple effects of this interaction revealed that when the applied load was symmetric to the mid-sagittal plane, the peak NEMG of the left erector spinae muscle (ERSL) were, on average 36% MVC with the lifting belt and only 34% MVC without the belt ($p<0.055$). In contrast, during asymmetric loading the peak NEMG for the left erector spinae (contra-lateral muscle) were on average 33% MVC when the lifting belt was tensioned and 36% MVC when the belt was loose ($p<0.036$) (Figure 1.2a). The right (ipsilateral) erector spinae (ERSR) peak responses were 35 % MVC with the lifting belt tensioned and 34% MVC when the belt was loosened during symmetric loading ($p>0.1$) (figure 1.2b). During asymmetric loading

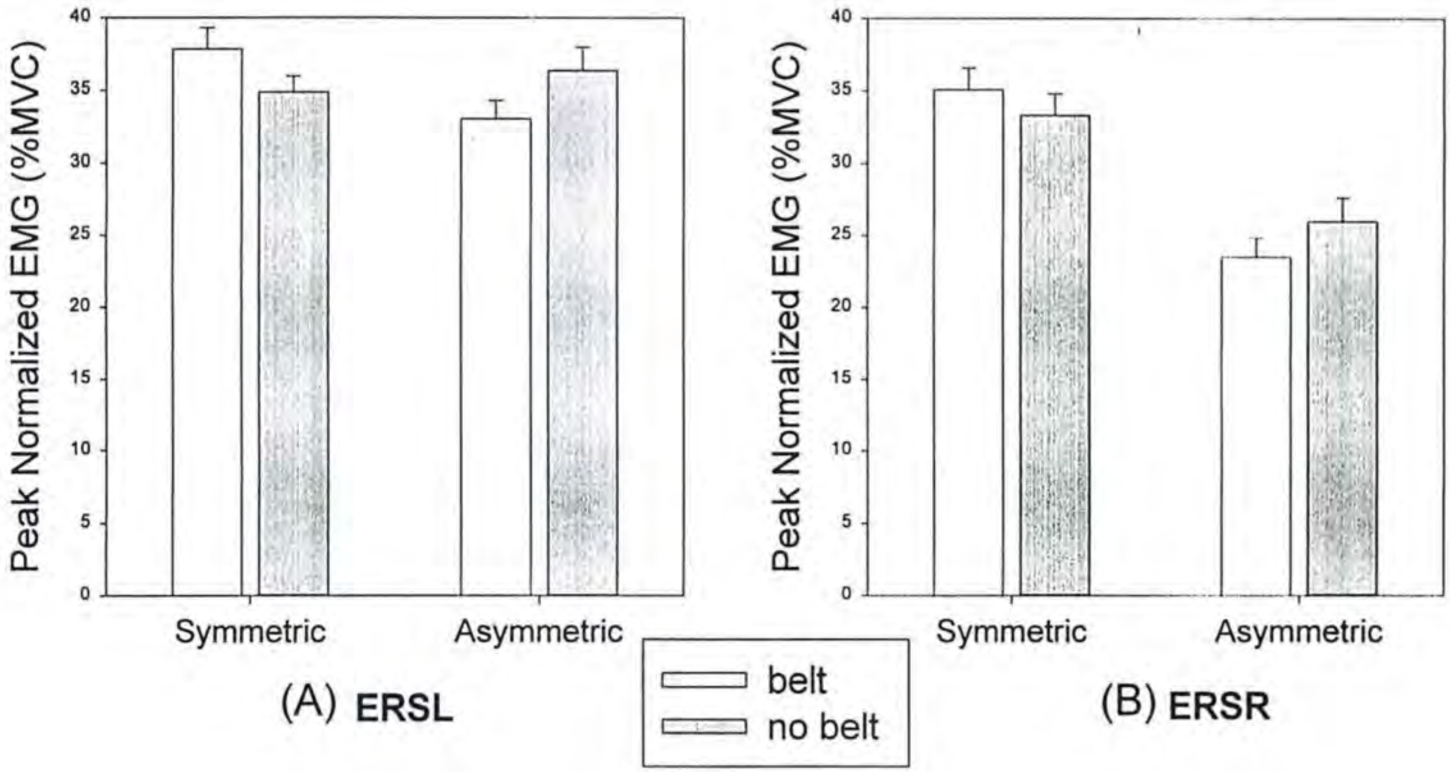

Figure 1.2. The peak normalized EMG response of the left (a) and right (b) Erector Spinae muscles (ERSL and ERSR) to the unexpected sudden loading as a function of the loading direction and belt condition.

conditions, the right erector spinae (ERSR) peak responses were 23 % MVC with the lifting belt tensioned and 26% MVC when the belt was loosened ($p>0.1$) (figure 1.2b). There were no significant interactions of gender or trial with the belt conditions for peak normalized EMG of the eight trunk muscles tested.

Analysis of the lifting belt on pre-load area of normalized EMG

The pre-load area of normalized EMG, which is a measure of muscle activity prior to the onset of the impending load, was not significantly affected by the lifting belt, direction of the applied load, trial or gender.

Analysis of the lifting belt on trunk kinematics

Multivariate analysis of the kinematic data revealed a main effect of the lifting belt for frontal plane trunk motion ($p<0.0001$). Trunk motion in the sagittal or transverse planes were not affected by the lifting belt. Neither gender or trials influenced trunk motion in any of the three cardinal planes ($p>.05$). While there was a statistically significant difference in the peak displacement and velocity of the trunk in the frontal plane ($p<0.0001$, $p<0.001$), the difference was too small to have any clinical significance. The reduction was, on average, 0.5 degrees and 2.0 degrees/second respectively when the lifting belt was worn.

Other findings

The peak normalized EMG of the right longissimus thoraces (LGTR) (ipsilateral to the asymmetrically applied load), and the right erector spinae (ERSR) were significantly reduced during asymmetric loading. The peak normalized EMG responses of left and right rectus abdominus (RABL & RABR) were increased during asymmetric loading ($p<0.001$, $p<0.026$) compared to symmetric loading. The peak normalized EMG of the left and right external oblique (EXOL & EXOR) were also increased during asymmetric loading conditions ($p<0.0001$, $p<0.0004$) compared to symmetric loading.

DISCUSSION

Sudden unexpected loading has been identified as a significant risk factor for developing costly low back injuries due to the increased stresses placed on the spinal structures from increased trunk muscle contractions (Bigos et al., 1986; Magora et al., 1973; Troup et al., 1981). Given the work of McGill et al. (1994) and Lavender et al. (1995), who report 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 which suggest that wearing a lifting belt reduced the peak NEMG of the left erector spinae muscles on average by 3% MVC during asymmetric loading (Figure 1.2b). In contrast, during symmetric loading, the peak NEMG of the left erector spinae activity increased on average by 2% MVC when the lifting belt was worn (Figure 1.2a). 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 et al., 1996). Perhaps during sudden loading the subject uses increased effort to resist the perturbation while the belt is worn, though there were no significant kinematic changes. If peak muscle activity is increased and trunk displacement is not reduced then the lifting belt may potentially increase the risk of injury during sudden loads applied in the mid-sagittal plane.

In this investigation, the effect of lifting belts on the peak response of the longissimus thoraces, 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 thoraces muscles. The activity of these muscles were measured because of a concern that the bending stress from the applied load would be displaced to the thoracic column and increase the demand on the longissimus. However, the lifting belt had no effect on the peak NEMG of the longissimus thoraces muscle. The peak NEMG of the four abdominal muscles were also not affected by the lifting belt which is consistent with McGill et al. (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 pre-load area of normalized EMG in this paradigm of unexpected sudden loading. This finding suggests that the subjects did not have any sense of increased trunk stability due to the lifting belt since they did not alter their pre-load preparation strategy. However, no specific questions were asked to determine the subjects perception of trunk stability.

Several studies on lifting belts have included female subjects (McGill et al., 1994; Reyna et al., 1995; Smith et al., 1996; Sullivan et al., 1996). Sullivan et al. (1996) reported that only their male subjects had increased isometric force production due to lifting belts, while Smith et al (1996) reported that on average the 69 women participating the their study could lift one kg more from the floor to waist level when the lifting belts were worn. The latter group of authors concluded, however, that the magnitude of this change in lifting strength was not sufficient to advocate the use of lifting belts as a way of increasing lifting capacity. McGill et al. (1994) found that female subjects had less trunk stiffness due to the belt during forward flexion and left side bending, thereby suggesting less of a protective effect during sudden loading conditions. But we found the no interaction of gender and the lifting belt conditions indicating that the lifting belts did not have a greater effect on the men or women in this study. This may be due in part to the experimental design. The subjects were constrained so that motion was only available 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.

In this experimental paradigm asymmetric loading resulted in an increased peak muscle response of all four trunk flexors, and a decreased peak muscle response of the ipsilateral trunk extensors (right longissimus thoraces and erector spinae). The NEMG from the contralateral trunk extensors did not significantly change between the symmetric and asymmetric conditions. These results are in contrast to other investigators who have reported an increase in peak muscle response of the contralateral trunk extensors to asymmetric loading (Lavender et al., 1989), and are most likely due to methodological differences in load application. In our investigation, asymmetric loading conditions produced a much smaller bending moment compared to the symmetric loading conditions. As such, symmetric loading would require a larger peak muscle response to counteract the applied load. During asymmetric loading, a large twisting moment was created that would necessitate increased peak response of the external oblique muscles to counteract the rotation moment at the torso.

EXPERIMENT 2: A biomechanical evaluation of a lifting belt used during sudden unexpected loading: A study of the body's electromyographic, kinematic and kinetic responses.

METHODS

Experimental Design

Four independent variables were included in this study. These were: the tension of the lifting belt (tensioned or not-tensioned), the symmetry of the applied load with respect to the torso's mid-sagittal plane (symmetric or 45 degrees asymmetric), the initial weight of the box (pre-loaded or not preloaded), and the subjects gender. The experiment consisted of 8 trials, one for each combinations of the four variables just mentioned presented in a randomized sequence.

The dependent variables in this study can be classified into four groups: electromyographic (EMG), dynamometric, kinematic, and kinetic. The electromyographic measures were comprised of the peak normalized EMG values following the sudden load application obtained bilaterally from the following muscle groups:

- Latissimus Dorsi
- Erector Spinae
- External Oblique
- Rectus Abdominis

Dynamometer readings measured the actual force applied during a given trial. Kinematic data were comprised of the angular orientation changes of 12 body segments following the application of the load. The kinetic variables analyzed here were the three dimensional external moments at L5/S1, and the flexion-extension moments at each knee and hip.

Subjects

The subjects were recruited from Rush Medical Center, University of Illinois and other surrounding educational facilities. Ten female and eleven male subjects participated in the experiment. The subjects were screened for the history of back pain. Only subjects with no back pain during the last year were permitted to participate. Subject description is provided in table 2.1.

Apparatus

Subjects stood on two solid wooden blocks (23½"×16"×14") placed over two force plates. The force platforms, from Bertec Corporation, provided the ground reaction force vector, moment vector and center of pressure under each foot. Surrounding the two wooden blocks was wooden platform 14 inches high. This platform served to mount the loading apparatus constructed using PVC tubing (figure 2.1). The PVC loading apparatus was used to position the two pulleys used in the application of the sudden load. A kevlar line (no-stretch) tied to the bottom of the

	Age (years)		Height (inches)		Weight (lbs)	
	Males	Females	Males	Females	Males	Females
Mean	30.82	26.80	69.44	64.55	163.45	129.45
Std. Deviation	7.03	5.81	2.84	2.99	23.03	15.55
Range	23 - 47	22 - 40	64 – 74.8	60 – 68.5	135 - 207	105 - 158

Table 2.1 Summary of subject anthropometry.

plastic box held by the subject was connected to a force gauge and weight via the two pulleys. Two equally weighted bags of lead shot were used for the sudden load and the pre-load. The weight of each bag was normalized to each subject's maximum voluntary extension force and weighed 7.5% of this value. The sudden load was applied by allowing one bag of lead shot to fall one meter prior to the line becoming taught at which point 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 event marker within the data stream. The pre-load bag used on the pre-load trials was also suspended from the line coming from the bottom of the box. The wooden blocks on top of the force plate and the elaborate design of the loading apparatus were required to minimize the amount of metal in the calibrated space used by the electro-magnetic motion measurement system.

The Motion Monitor™ by Innovative Sports Training (I.S.T.) was used to obtain kinematic data. The Motion Monitor uses an electromagnetic tracking system in which up to fourteen sensors are tracked within the generated magnetic field to provide six degree of freedom position and orientation data. The transmitter was placed 111 cm above the floor and 62 cm behind the two force platforms. Twelve sensors were used in the current study. These were positioned on the head, first thoracic vertebrae (T1), first lumbar vertebrae (L1), top of sacrum (S1), left and right upper arm, forearm, thigh and shank. Sensors are connected via cable to the data collection computer.
The sensors were used within a calibrated region that measured 137 cm by 91 cm by 183 cm. The calibration of the magnetic field, or "mapping", was necessary because of the distortion created by metallic structures (cabinets, force plates, metal beams in the floor and ceiling, etc) in the laboratory. Software incorporated in the Motion Monitor system was used for the mapping process. Points within the measurement space were modeled using a 3rd order polynomial function. Mean dynamic error averaged .7 cm (sd=.3 cm). The largest errors tended to be on the periphery of the measurement space.

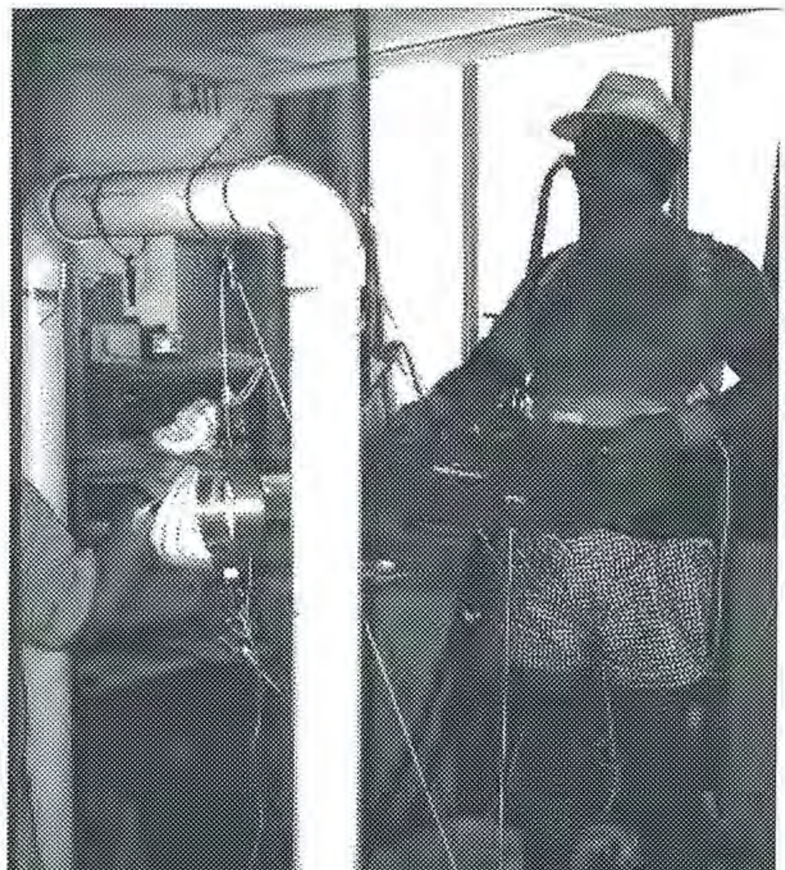

Figure 2.1. The experiemental apparatus for the second experiment. The blindfolded subject is about to receive the sagittally symmetric sudden load in a loose lifting belt trial without preload.

Eight active bipolar surface electrodes (Medical Devices, Inc.) were placed on trunk muscles to obtain the electromyographic signals. Each electrode had a gain of 10 and had an inter-electrode spacing of 1 cm. A pre-amplifier (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 EMG signal. The signal was again amplified (gain= 3), rectified and integrated (time constant= 50 msec) before being passed to the two 16 channel National Semiconductor A/D boards used to obtain the analog EMG, forceplate, and dynamometer data. The A/D cards reported ±2048 units with 4.9 mV resolution per unit. A 75 MHz serial computer was used to run data acquisition software, store, analyze and export all data.sets. The EMG signals from electrodes were monitored throughout the experiment using an oscilloscope and on a computer screen using a virtual instrument constructed with LabView™ software.

A Lido Back machine was used to measure each subject's trunk extension strength and obtain maximum voluntary contractions from the eight muscles sampled. The subjects were strapped into the machine which restricted upper and lower body movement allowing only extension and flexion with resistance. The machine recorded the maximum amount of force exerted by the subject. This extension strength was used to normalize the weight applied during the sudden loading to each subject's capacity.

In the study the lifting belt was constructed of webbed material covered by an elastic band 17 cm wide that stretched anteriorly and attached with Velcro. The two lifting belt conditions consisted of either this elastic being tensioned or the elastic being completely slack.

Temporal conditions were controlled by a blindfold that was used to block visual cues. Audible cues, signaling the onset of the load, were masked with a noise generator. A pre-load of weight equal to sudden load weight was used in pre-loaded trials.

<u>Procedure</u>

Subjects was given consent forms to read and sign. Demographic data on weight, height and age was collected. All subjects were asked to change into a pair of shorts. Females subjects were provided with a T-shirt that was tied up so that their waist was exposed. Surface electrodes were placed on the skin in line with the muscle's line of action using surgical tape at the following bilateral sites:

a) Latissimus Dorsi: T7 level over the belly of the muscle. at the T10 level approximately 2 cm lateral form the midline.
b) Erector Spinae: at the L3/L4 level approximately 5 cm lateral form the midline.

c) External Oblique: along the line between the superior iliac spine and the umbilicus approximately 5 cm medial from the superior iliac spine.

d) Rectus Abdominis: at the level of the umbilicus 2 cm lateral from the midline.

Maximum voluntary contractions (MVC) were performed for the purpose of normalization of EMG. The subject was put in the Lido Back machine and asked to flex, extend, twist right, and twist left at his/her maximum ability. The machine was set to it's lowest isokinetic velocity setting (5 deg/sec). The EMG was sampled as the trunk was flexed or extended approximately 15 degrees. The maximum extension trunk strength was recorded and used to normalize the weight and pre-load used in sudden loading trials. The baseline EMG was taken in the in measurement space within the magnetic field as the subject stood in a relaxed standing posture.

The sensors from the motion monitor were placed securely using Velcro raps, self adhesive surgical raps and a cap on the following twelve "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 the 1st lumbar vertebrae, over the spinous process for the first thoracic vertebrae, 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 using an adjustable cap. An additional sensor was used to locate the segment ends for the software by placing it over the left and right shoulders, elbows, wrists, hips, knees, and ankles.

Closed cell foam gads (plastisote) 7 mm thick were placed surrounding the electrodes using surgical tape. These pads were used to prevent the lifting belt from exerting direct pressure on the electrodes. The lifting belt of correct size was placed loosely around the waist. The belt was only tensioned during the trials that require lifting belt, otherwise belt was kept loose enough that a hand could be easily slipped between the belt and subject's abdomen.

Subjects were asked to stand straight on the force plates with forearms bent ninety degrees, with no abduction of the shoulder and forearms in a neutral orientation. A reading with the Motion Monitor was then obtained with the subject in this "calibration" position.

Prior to each trial subject was blindfolded and instructed about the belt tension and symmetry conditions. The box was then handed to the subject and the condition of pre-load was checked. The weight was raised one meter and dropped freely after a variable interval of 3 to 8 seconds. Data were collected 3 seconds before and 1 second after the loading.

Data Treatment

The integrated EMG were normalized with regards to each subject's maximum voluntary contraction (MVC) obtained from the exertions performed while in the Lido™ machine. The normalization calculation was as follows:

$$\mathbf{NEMG(i,j)= (IEMG(i,j)- REST(i))/ (MAX(i)- REST(i))} \qquad \mathbf{(3)}$$

Where:

i=	muscles 1 through 8
j=	experimental conditions 1 through 8
NEMG(i,j)=	the normalized EMG for muscle i in condition j
IEMG(i,j)=	the current Integrated EMG value for muscle i in condition j
REST(i)=	the minimum resting IEMG value for muscle i for the relaxed standing measure
MAX(i)=	the maximum IEMG value from muscle i during the maximal isometric exertion

Peak values from the data following the load onset were analyzed using a repeated measures statistical analyses. This analysis included three within subjects factors (belt condition, preload condition, and loading symmetry) plus the between subjects gender factor.

Kinetic and kinematic data were used to compute the external moments on the spine using a linked segment model. The three-dimensional model is comprised of 7 rigid bodies: Two feet, two shanks, two thighs, and the pelvis. Joint centers are obtained from marker locations for the knees and ankles by using half the associated breadth measurement. The hip joint center is positioned medially at 25 percent of the hip breath measure. L5/S1 was assumed to be in the mid-sagittal plane 9.04 cm from the L5 spinous process along a vector directed anteriorly and angled 6 degrees below the horizontal (Chaffin and Andersson, 1991).

A fast fourier transform was used to transform the time varying position data to the frequency domain where all the frequencies above four Hz were set to zero. The data were then transformed back to the time domain where the now low-pass filtered position data were used to compute the angular positions of the body segments using the projection angle method. A 5-point numerical differentiation which uses the best fit fourth-order polynomial is used to obtain the angular velocity (Lanczos, 1988). The same 5-point numerical differentiation technique is 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 about 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 about the longitudinal axis of each segment were considered negligible. Body segment inertia properties were approximated using the radius of gyration data published by Webb Associates (1978). The center of mass and the mass distributions of each body segment were approximated by data obtained from Dempster and Gaughran (1967) and from Zatsiorsky and Seluyanov (1983). 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. (1979).

By knowing the ground reaction force and assuming the shank as a rigid body, the equations of dynamic equilibrium were used in conjunction with the computed angular accelerations to obtain the external forces and moments at the knee joint. The forces and moments at the hip was obtained by treating the thigh as a rigid body and solving the equations of dynamic equilibrium knowing the moments and forces at the knee joint. Similarly, the forces and moments at L5/S1 was computed assuming the pelvis as a rigid body and computing the equilibrium moments using the moment and forces at the hip joints. The L5/S1 moments were plotted over time and the peak value of each directional component was extracted for analysis.

Statistical Analysis

Repeated measures ANOVA's were conducted for each dependent measure. These included the EMG responses, the postural changes, and the moments at L5/S1 predicted by the model described above. Significant effects ($p<.05$) and trends ($p<.07$) are reported in the results. Additional ANOVA's 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 one error but instead present the outcome of each statistical test performed so that the reader could evaluate it's significance.

RESULTS

Electromyographic Analysis

The top part of table 2.2 shows which muscles were significantly affected by the use of the lifting belt either by itself or in combination with other factors (interaction). The lifting belt affected the peak activation levels of both Erector Spinae muscles, however, this was also dependent upon the symmetry of the loading, and for the *Left Erector Spinae* this effect was also dependent upon the gender of the subject. Figure 2.2 shows that tensioning the belt yielded greater peak activity in the *Left Erector Spinae* in the symmetric conditions for both genders, although this increase was non-significant in post-hoc tests ($p>.10$). As would be expected, during asymmetric conditions this contra-lateral muscle showed increase activation, but the belt further increased the peak response in males ($F=5.93$; $df=1,9$; $p=.038$). The response of the

Table 2.2: The results from the repeated measures analysis of variance for each dependent variable. Only the effects which involved the belt and were significant for at least on of the dependent variables are shown. F-Test values (F), degrees of freedom (df) and the probability associated with each effect (p) are presented.

Dependent Variable		EFFECT						
		B	B X S	B X P	B X G	B X S X P	B X S X G	B X P X G
	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				

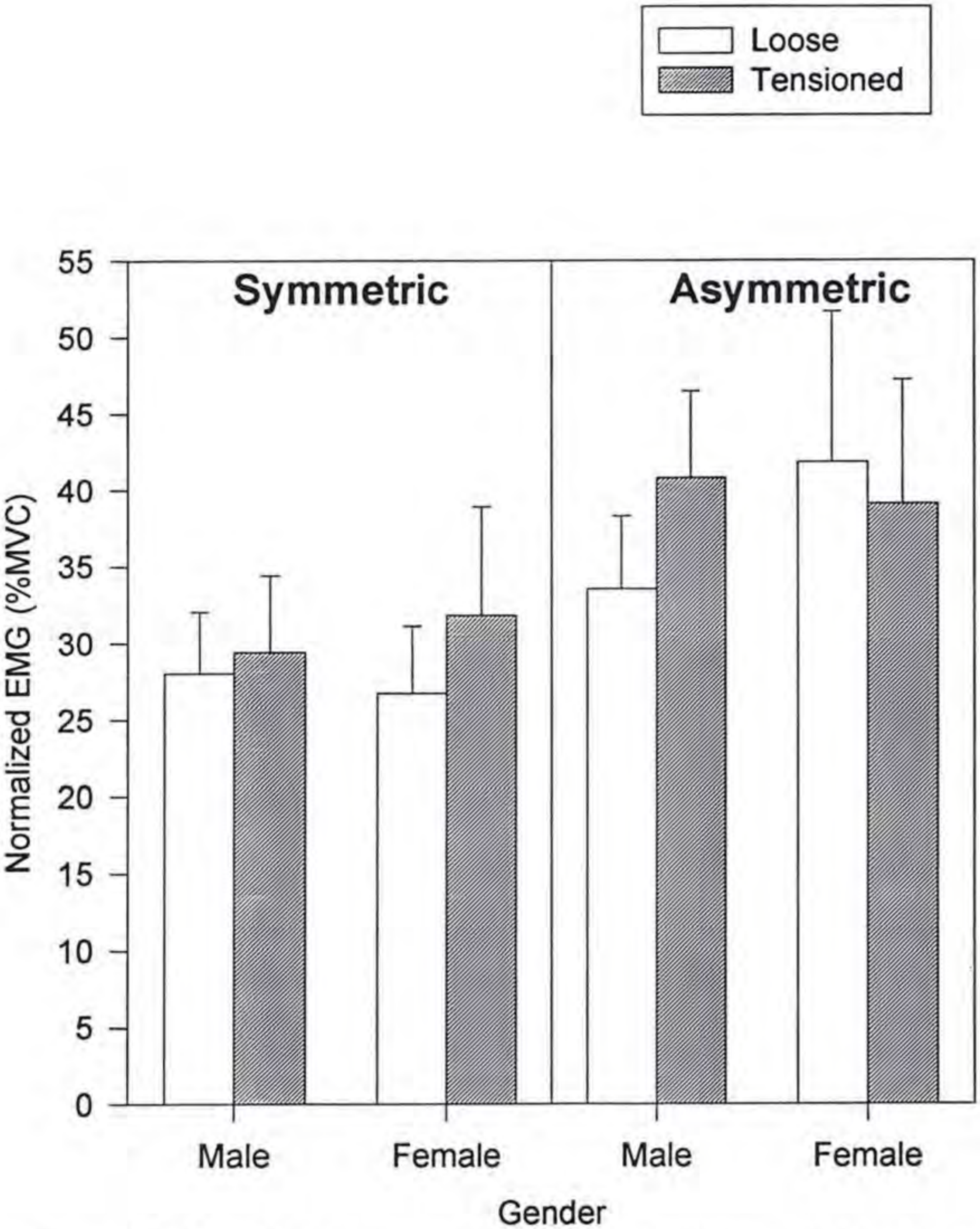

Figure 2.2. Mean ERSL response during the sudden unexpected loadings as a function of the subjects' gender, the symmetry of the loading, and the lifting belt tension.

Left Erector Spinae in females was essentially unaffected by the belt tension during the asymmetric loadings. Meanwhile, the ipsilateral *Right Erector Spinae's* peak response (figure 2.3) was reduced during the asymmetric loading, but more so when the lifting belt was tensioned (F=5.14; df=1,14; p=.038). The increase shown in the *Right Erector Spinae's* response shown in figure 2.3 during the symmetric loadings was not statistically significant.

With regard to the Latissimus Dorsi muscles, only the *Left Latissimus Dorsi (LATL)* muscle was significantly affected by the lifting belt (table 2.2). Figure 2.4 shows that the peak response of this muscle was also dependent upon the gender of the subject. While it appears that females showed lower LATL NEMG signals with the belt tensioned, this change was not significant. Males, however, showed significantly greater peak LATL response with the belt tensioned (F=4.63; df=1,9; p=.060). Independent of gender, the response of the LATL muscle to the lifting belt was dependent upon the symmetry of the loading and the preload conditions (Figure 2.5). And while the response of this contra-lateral muscle to the asymmetric loadings was greater, the only significant change associated with tensioning the lifting belt detected in post-hoc tests was found to be a reduction in the peak EMG during the symmetric non-preloaded conditions (F=8.16; df=1,16; p=.011).

Three of the four anterior muscles sampled were affected to some degree by the lifting belt. Only the *Left Rectus Abdominus*' response was not affected. The peak activities in the *Right Rectus Abdominus*, and the *Right External Oblique*, albeit relatively small, were reduced by 41 and 19 percent, respectively (Figure 2.6).

The peak response of the *Left External Oblique (EXOL)* to the lifting belt was also dependent upon the symmetry of the loading and the gender of the subject (table 2.2). During the symmetric loadings the belt had little impact on the EXOL response. Asymmetric loadings, while leading to larger a larger response in this contra-lateral muscle, showed gender specific responses to the lifting belt (Figure 2.7). Post-hoc tests suggest that females had a lower peak EXOL response with the belt tensioned during the asymmetric loadings (F=3.92; df=1,7; p=.088). The increased EXOL response by the males with the tensioned belt was non-significant in the post-hoc statistical analysis (p=.22).

<u>Kinematic Analysis</u>

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=.001). On average the decrease with the lifting belt tensioned during the symmetric loadings was 1.6 degrees, or 17 percent of the sagittal plane motion measured in the thoracic and lumbar spine (Figure 2.8). It is interesting to note that while the hip flexion did not change with the lifting belt, the average magnitude of this motion in both hips was approximately 20 degrees. The total flexion motion of the hips and spine decreased

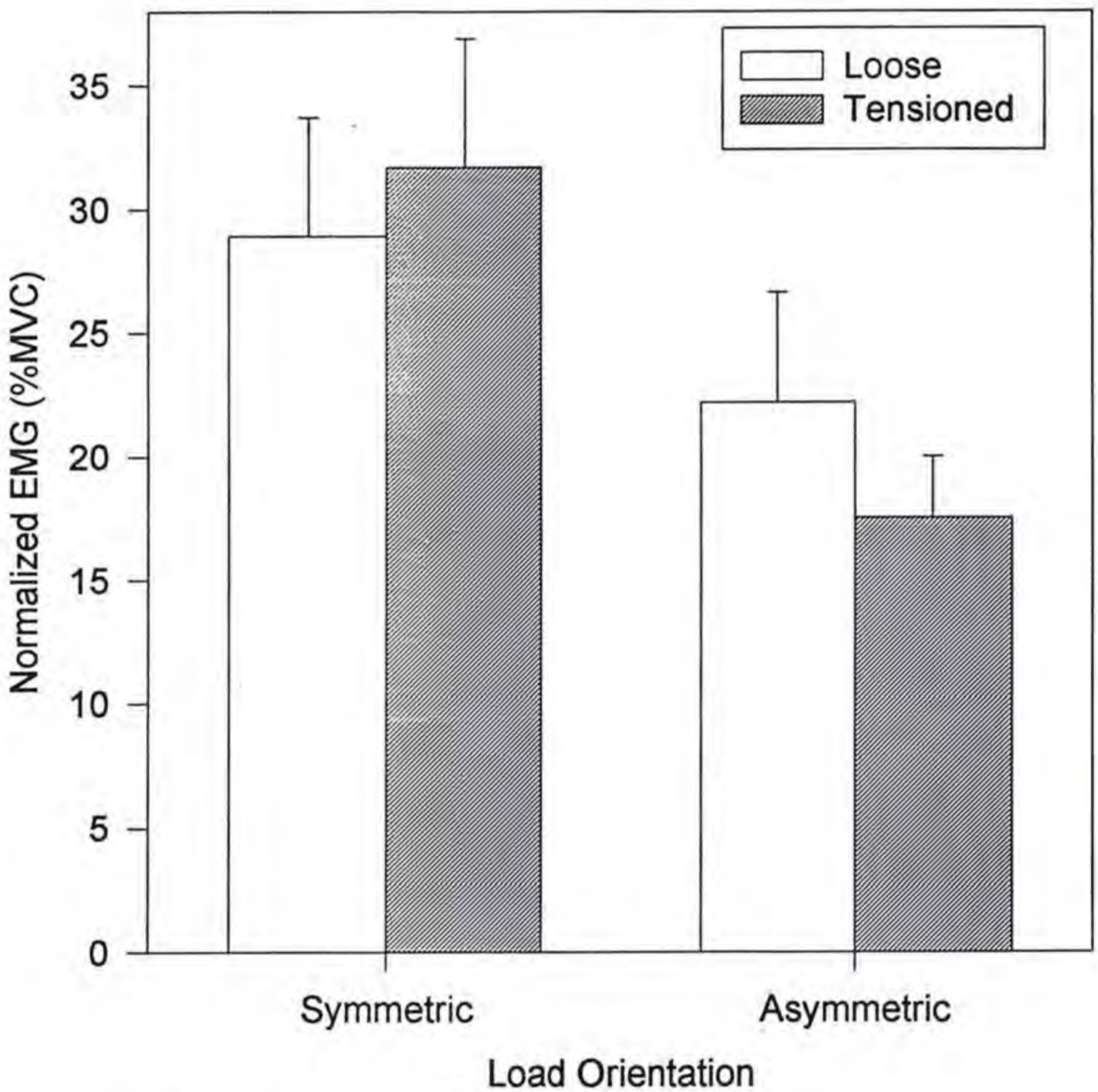

Figure 2.3. Mean ERSR response during the sudden unexpected loadings as a function of the load symmetry and the lifting belt tension.

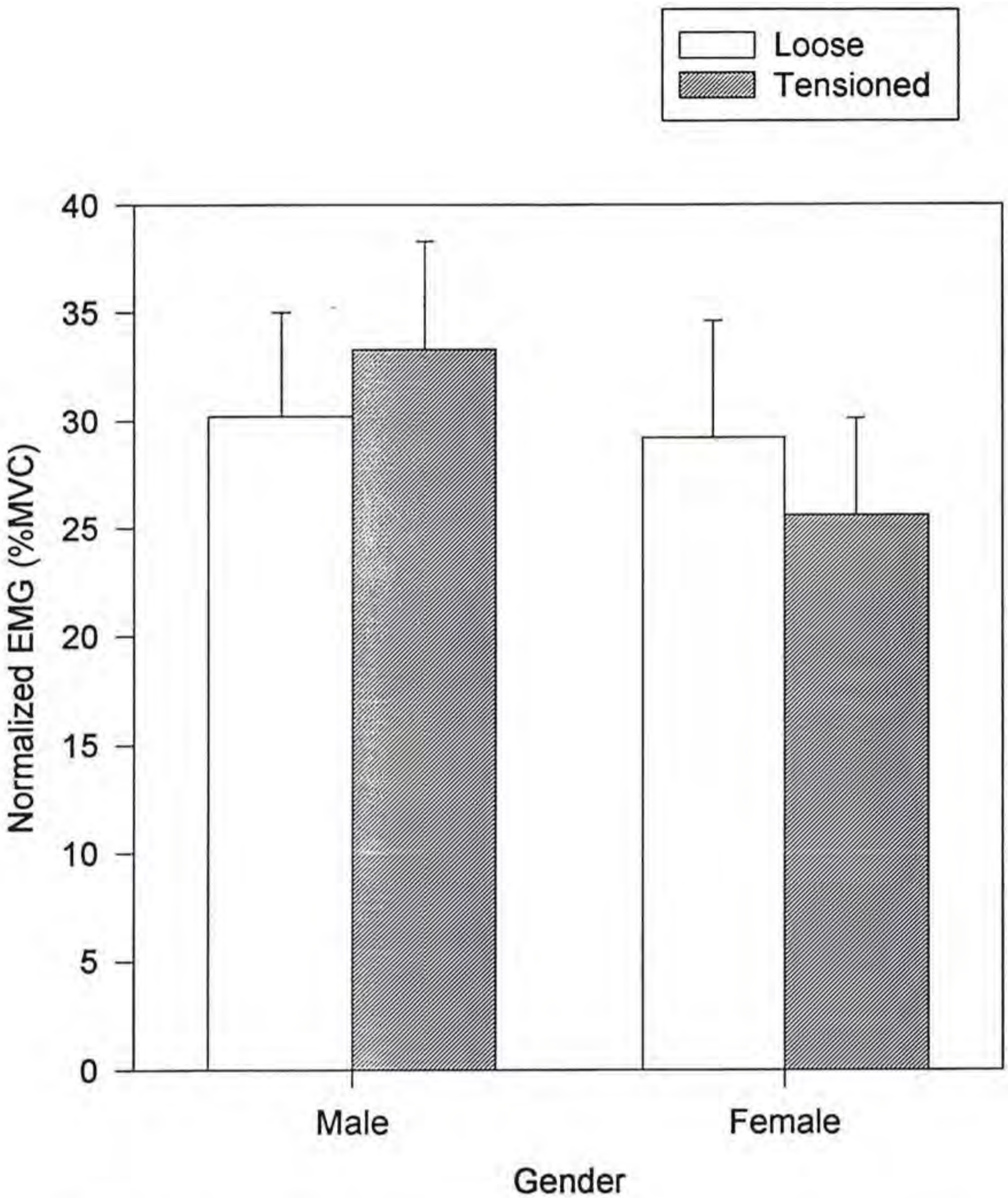

Figure 2.4. Mean LATL response during the sudden unexpected loadings as a function of the subjects' gender and the lifting belt tension.

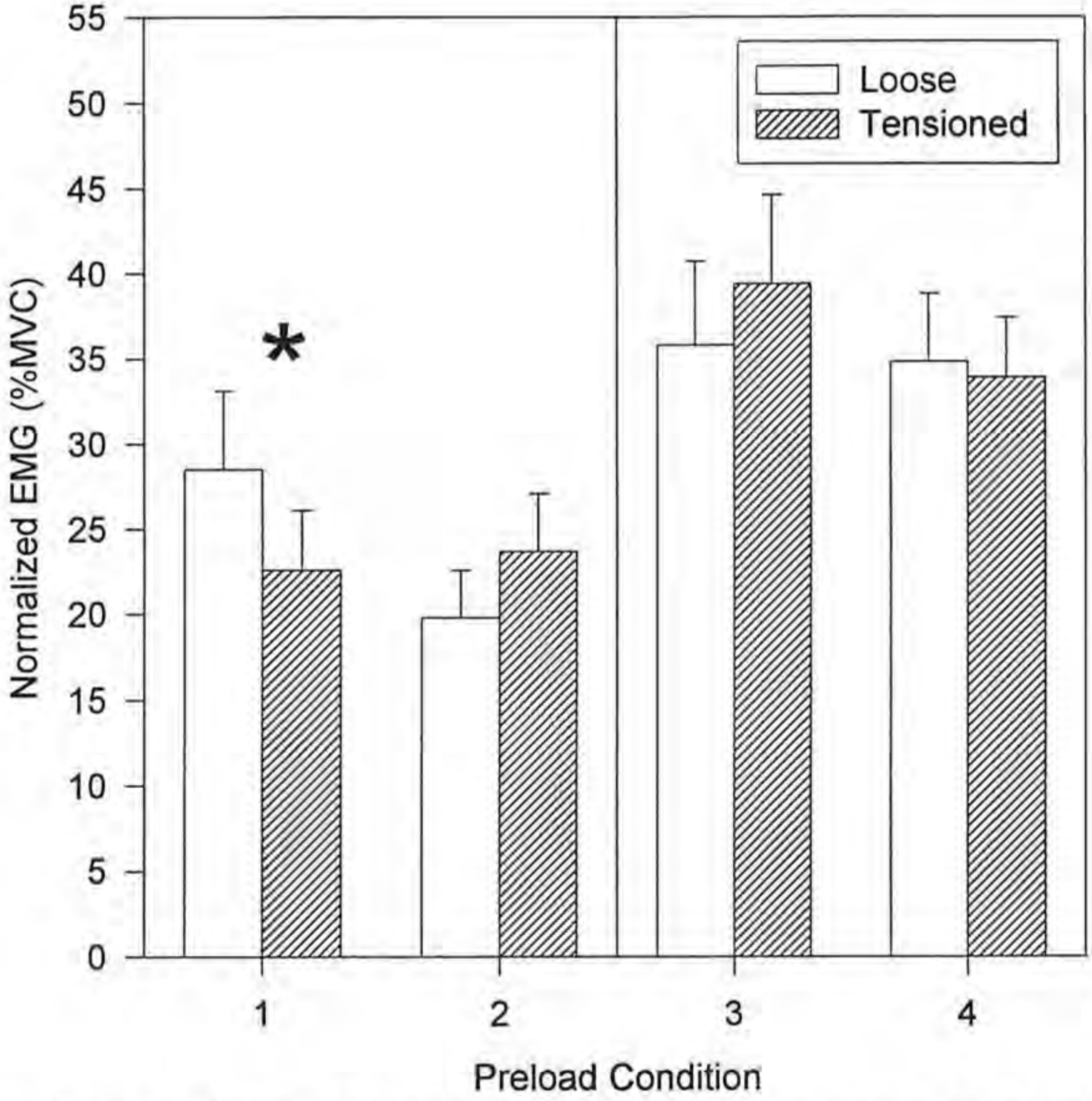

Figure 2.5. Mean LATL response during the sudden unexpected loadings as a function of the symmetry of the loading, the preload condition, and the lifting belt tension. *p<.05

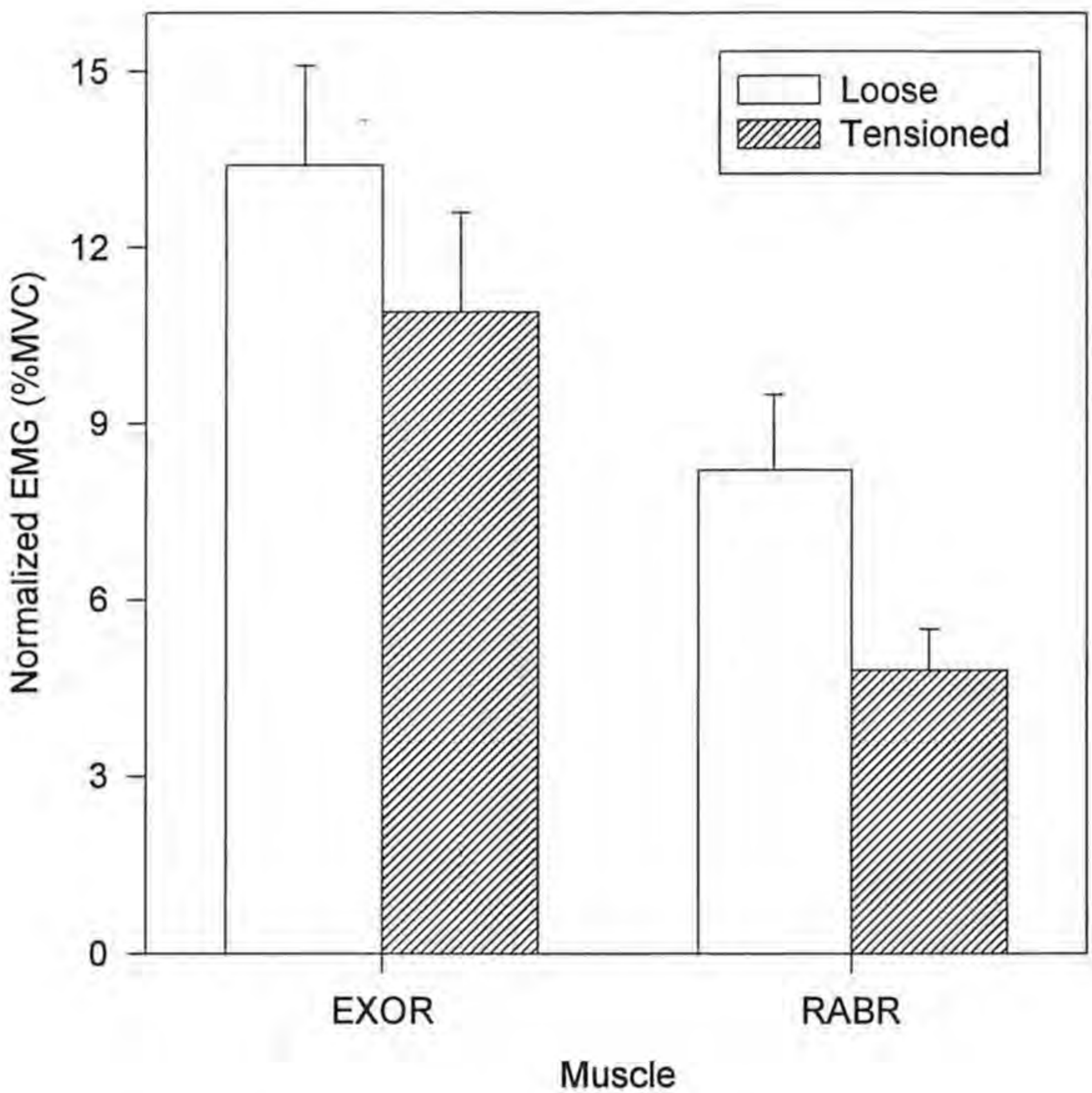

Figure 2.6. Mean EXOR and RABR responses during the sudden unexpected loadings as a function of the lifting belt tension.

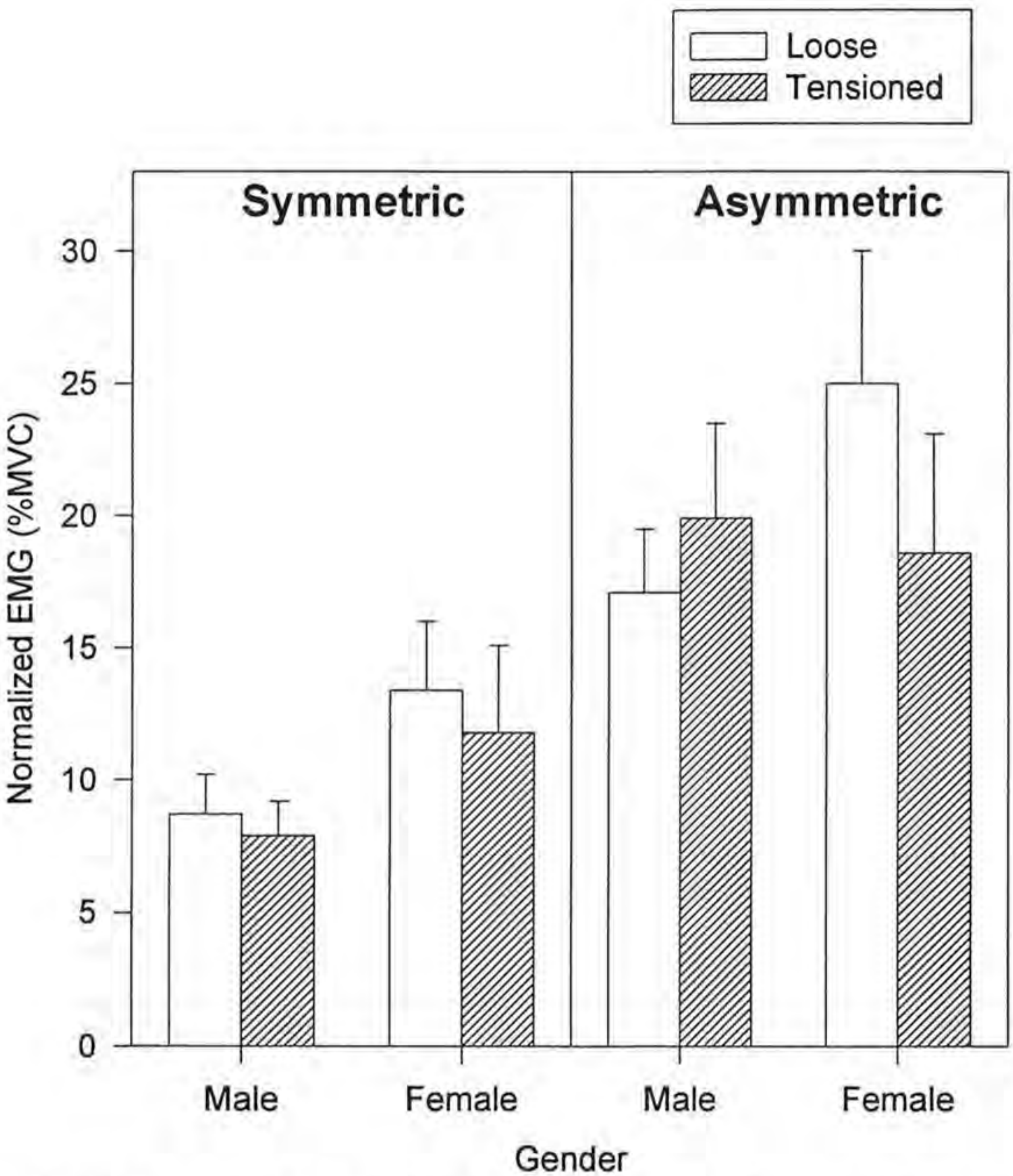

Figure 2.7. Mean EXOL response during the sudden unexpected loadings as a function of the subjects' gender, the symmetry of the loading, and the lifting belt tension.

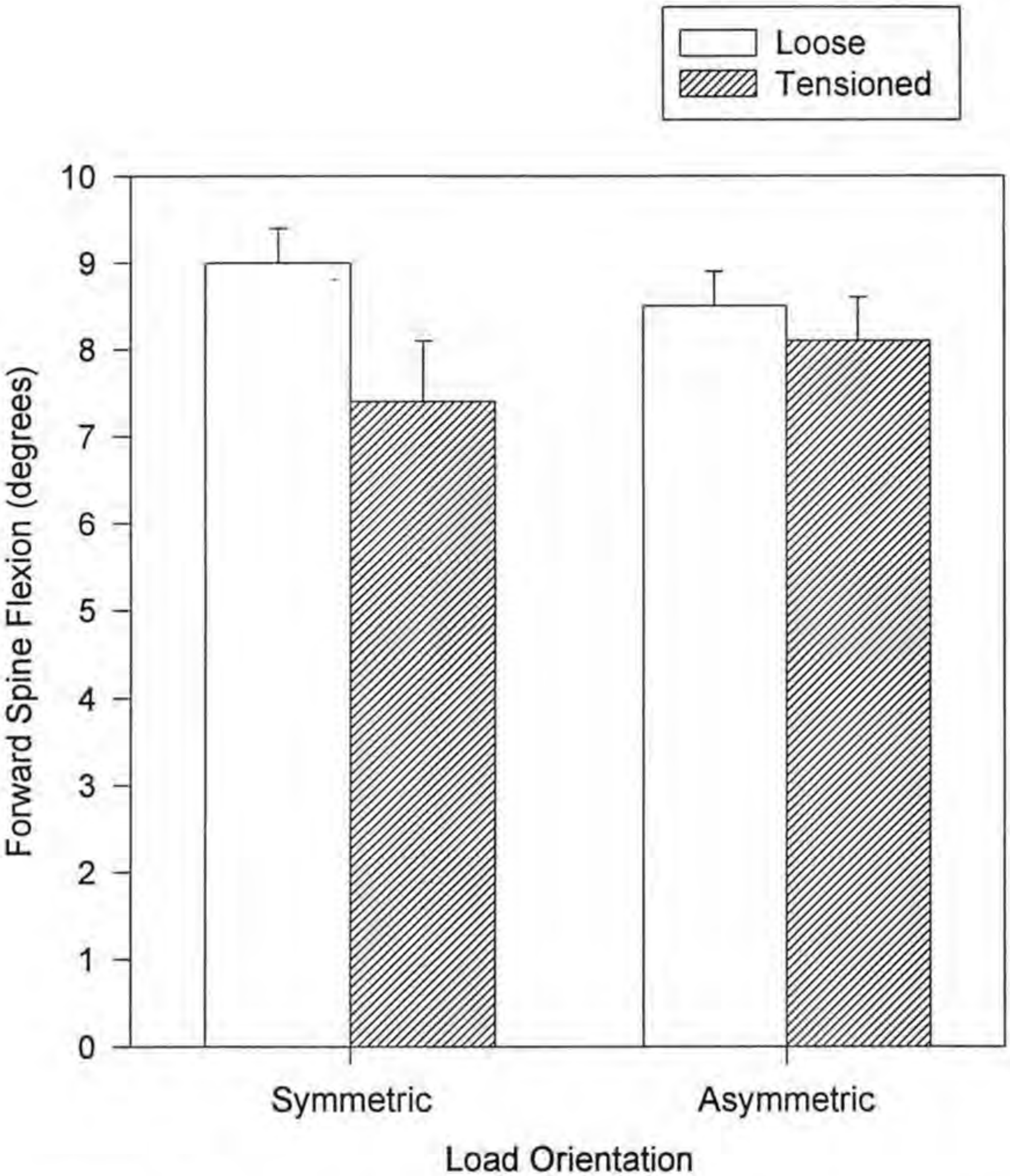

Figure 2.8. Mean trunk flexion during the sudden unexpected loadings as a function of the load symmetry and the lifting belt tension.

with preload during symmetric and asymmetric loads. The lifting belt reduced the total spine and hip flexion during the symmetric non-preloaded trials (figure 2.9a). An analysis of the ratio of forward spine flexion to bilateral hip flexion showed that while the lifting belt significantly decreased the spine motion relative to the hip motion (F=5.72; df=1, 16; p=.029), the effect was also dependent upon the preload, and symmetry conditions (F=4.83; 1,16; p=.043). In essence, this interaction highlighted that the spine to hip flexion ratio was reduced during the sagittally symmetric preloaded trials with the belt tensioned (figure 2.9b). This identifies the reduction in total hip and spine motion which occurred during the preloaded trials with the belt tensioned, was due to less motion in the spine as opposed to in the hips. Whereas the preloaded trials, with the belt loose, showed motions at the spine and hips reduced by approximately equal proportions.

The lateral flexion of the spine was significantly reduced by tensioning the lifting belt (F=12.8; df=1,16; p=.003), although the belt effect was also dependent upon the preload condition and the subjects' gender (F=10.7; df=1,16; p=.005). Further analysis of the data displayed in figure 2.10 revealed that the lateral bending was reduced in the preload conditions in both genders when the belt was tensioned (F=8.39; df=1,16; p=.011). During the conditions without the preload the belt only reduced the lateral bending in the female subjects. The twisting motions in the spine were unaffected by the lifting belt, even during the asymmetric loadings.

Other kinematic changes associated with the lifting belt were observed in the knees. Overall, tensioning the lifting belt significantly reduced flexion in the right knee by about 1 degree (F=5.91; df=1,16; p=.027). However, the motion in the right knee was also dependent upon the symmetry of the loading and the subject's gender (Figure 2.11). Knee flexion in males was reduced by a little over 2 degrees during asymmetric loading. Females, on the other hand, showed no change during asymmetric loading, but nearly a 2 degree reduction in knee motion during symmetric loadings with the belt tensioned. The left knee showed only a weak interactive trend (p=.08) suggesting that belt use was associated less motion during only during symmetric loadings without preload.

<u>Kinetic Analysis</u>

The forces measured by the strain gauge placed in series with the load averaged 387 N (S.D. = 67 N) for males and 322 N (S.D.= 57.8 N) for females. Statistical analysis of the force data indicated a significant belt by preload interaction (F=8.30; df=1,16; p=.011). Figure 2.12 shows that 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=.010) from 323 N (sd= 40 N) to 338 N (sd=44 N) as the belt was tensioned.

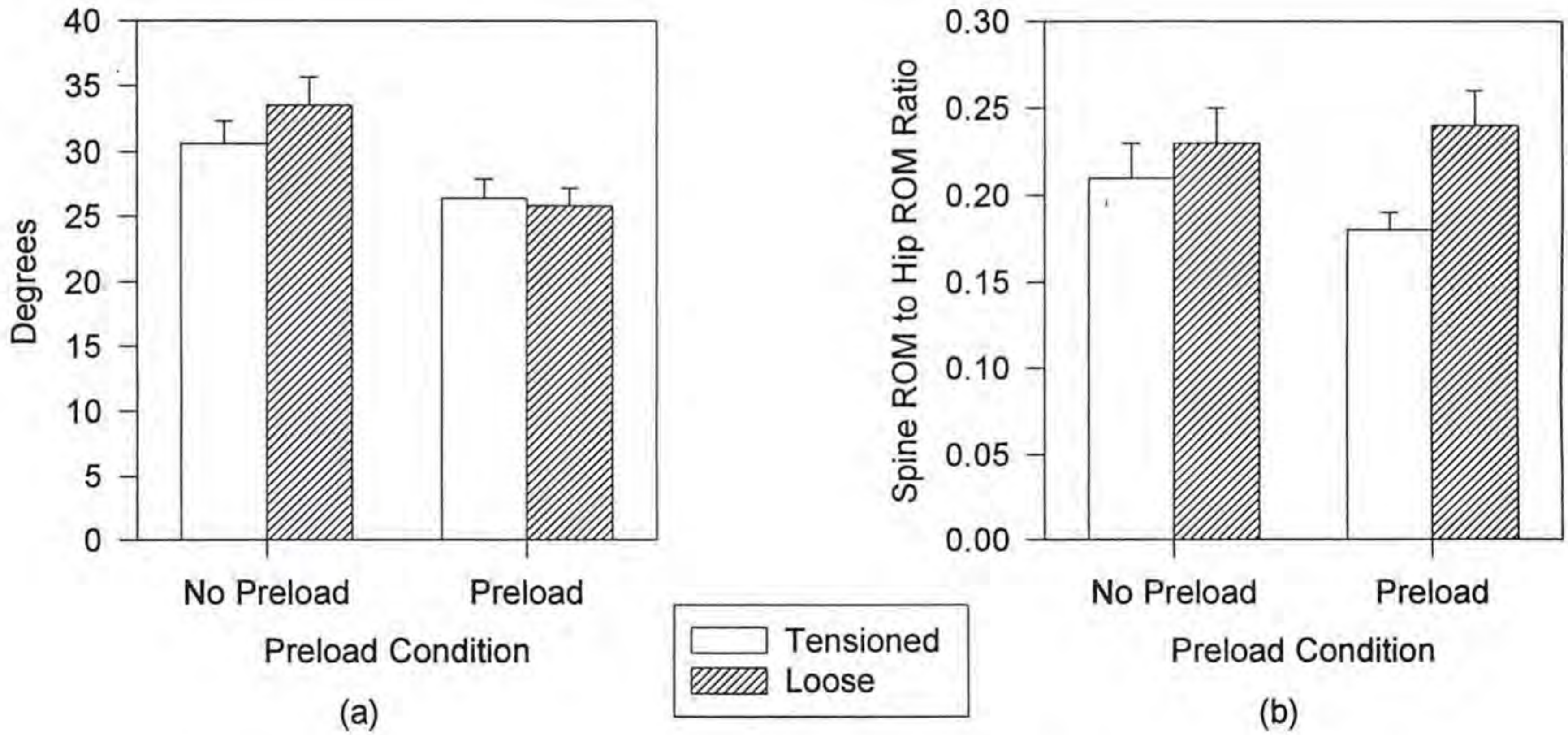

Figure 2.9. The total trunk flexion due to spine flexion and hip flexion (a) during the symmetric loadings. The ratio of spine motion to hip motion in the symmetric loadings (b).

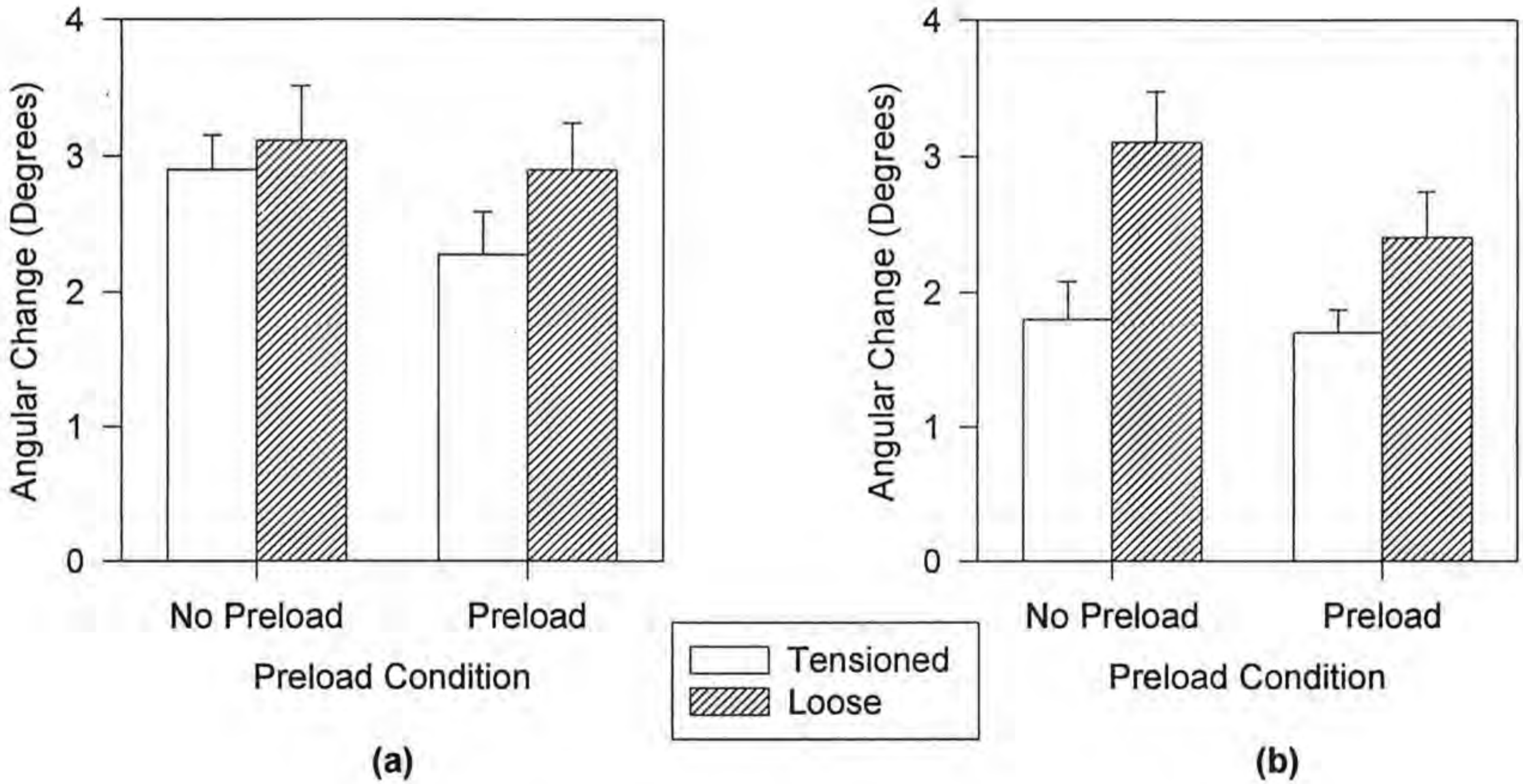

Figure 2.10. The mean lateral bending measured following the onset of the sudden load in males (a) and females (b) as a function of the belt tension and preload conditions.

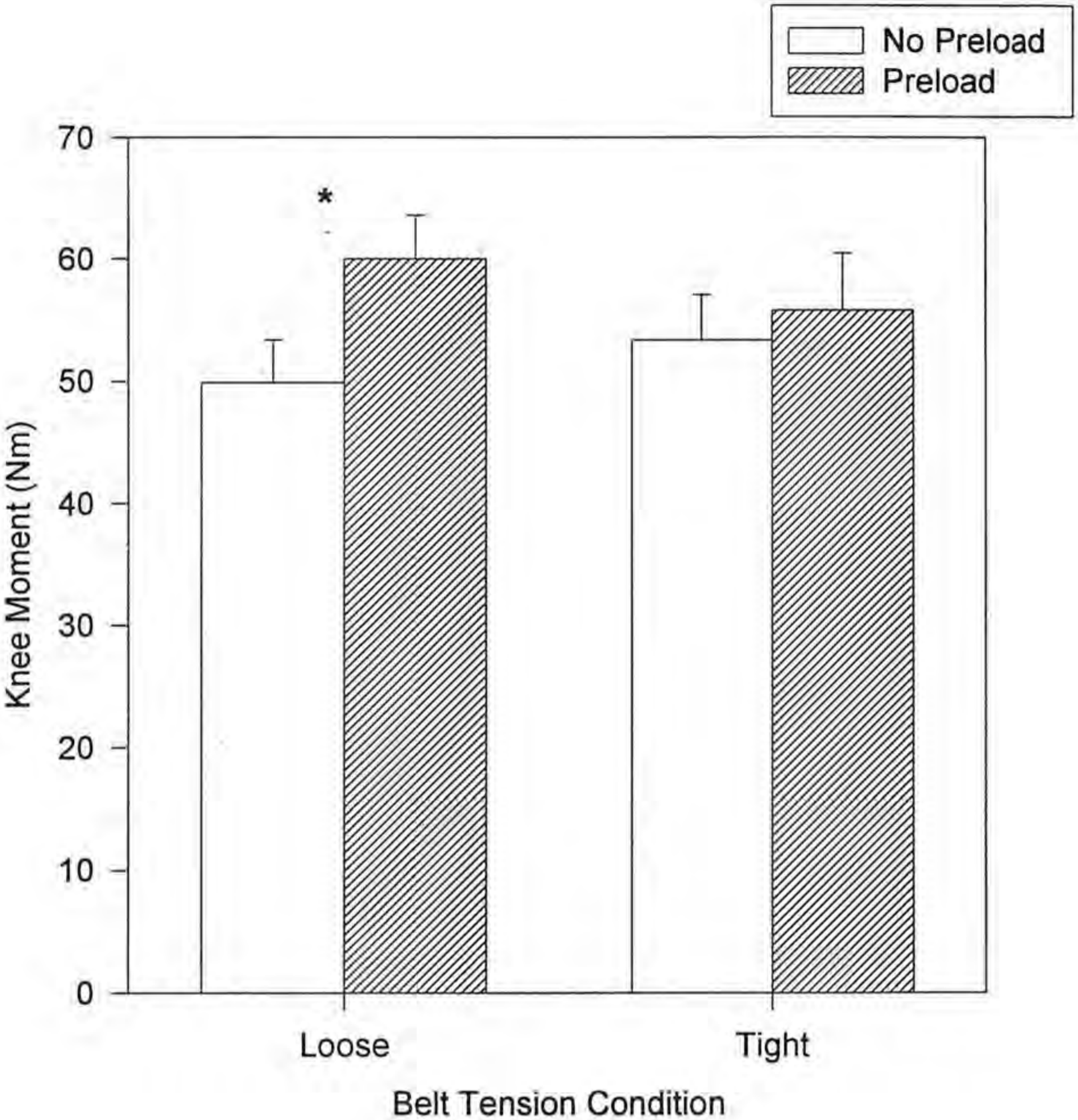

Figure 2.11. The maximum flexion moment in the right knee was affected by the preload condition when the belt was loose (* p<.05) but was unchanged when the belt was tensioned.

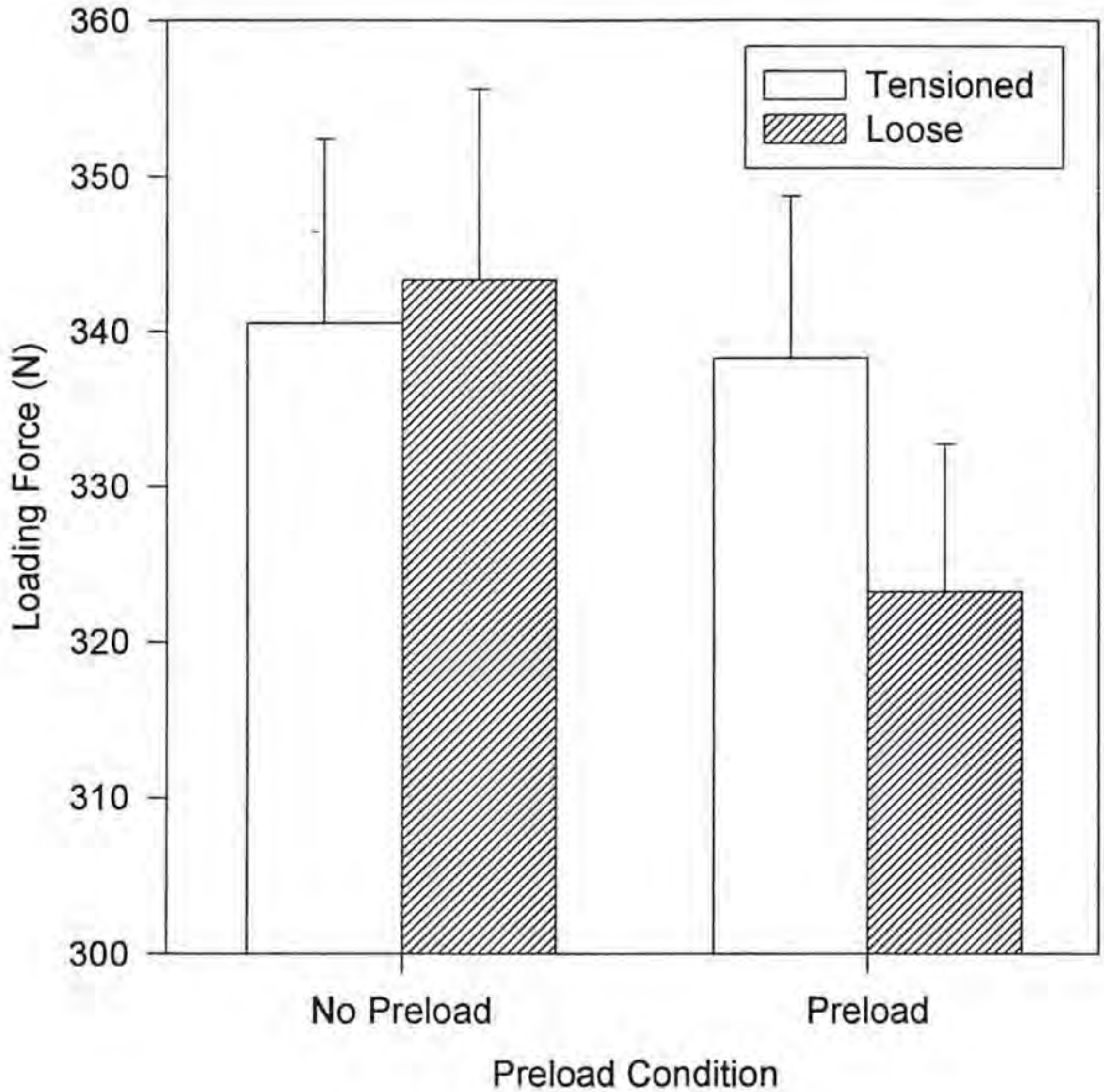

Figure 2.12. The mean impluse loading force measured by the strain gauge as a function of the preload condition and the lifting belt tension.

External moments were calculated using the kinematic and the ground reaction force data in a 3D 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 only occurred in the males ($F=7.89$; $df=1,16$; $p=.013$; $F=4.58$; $df=1,16$; $p=.049$). Figure 2.13 shows the spine moments in the males decreased from a mean of 200 Nm (sd=49 Nm) to 181 Nm (sd=43 Nm), a 9 percent change. The change in the male right hip flexion moment decreased 12 Nm, also a 9 percent change. There were no significant changes in the lateral bending or twisting moments at the spine associated with the lifting belt.

DISCUSSION

The data presented above suggest that the effects of the lifting belt on spinal loading are more complicated than originally hypothesized. It's biomechanical impact while not large, is observable. Essentially, in this second study we have found that experiencing sudden unexpected loading with a lifting belt (tensioned) results in the following:

- Reduced forward bending of the spine during symmetric loadings.
- Reduced lateral bending of the spine.
- Reduced forward flexion moment in males
- Greater peak forces at the hands during the preloaded trials.
- Increased contra-lateral posterior muscle activity during asymmetric loading
- Decreased ipsilateral Erector Spinae muscle activity with asymmetric loads
- Decreased anterior muscle activity

However, assembling these effects into a coherent explanation is much more difficult. The change in muscle activations suggest that the belt impacts the underlying strategy employed in by the body when dealing with potentially destabilizing perturbations.

The overall hypothesis tested in this study was that the lifting belt would stiffen the torso, and in so doing, protect the torso. The support for the stiffening hypothesis comes from the reduced forward bending in the spine and from the change in the spine to hip motion ratio, especially during preloaded trials. This shows that the stiffness of the torso is dependent upon both the belt and the presence versus absence of a preload.

When the preload was present, the elbows flexed less (~30%) and the spine flexed less (~20%). Perhaps as a consequence of the reduced motion, the left and right Latissimus Dorsi muscles, as well as the right External Oblique muscle, had significantly lower peak EMG values when the person was preloaded. In many ways the preloaded conditions may be more relevant to every day material handling situations where the sudden load is the result of an object that is being held or carried starting to slip. Here the system is essentially preloaded. Overall, the system was less compliant with the preload, suggesting that the pretensioning of the muscles reduced

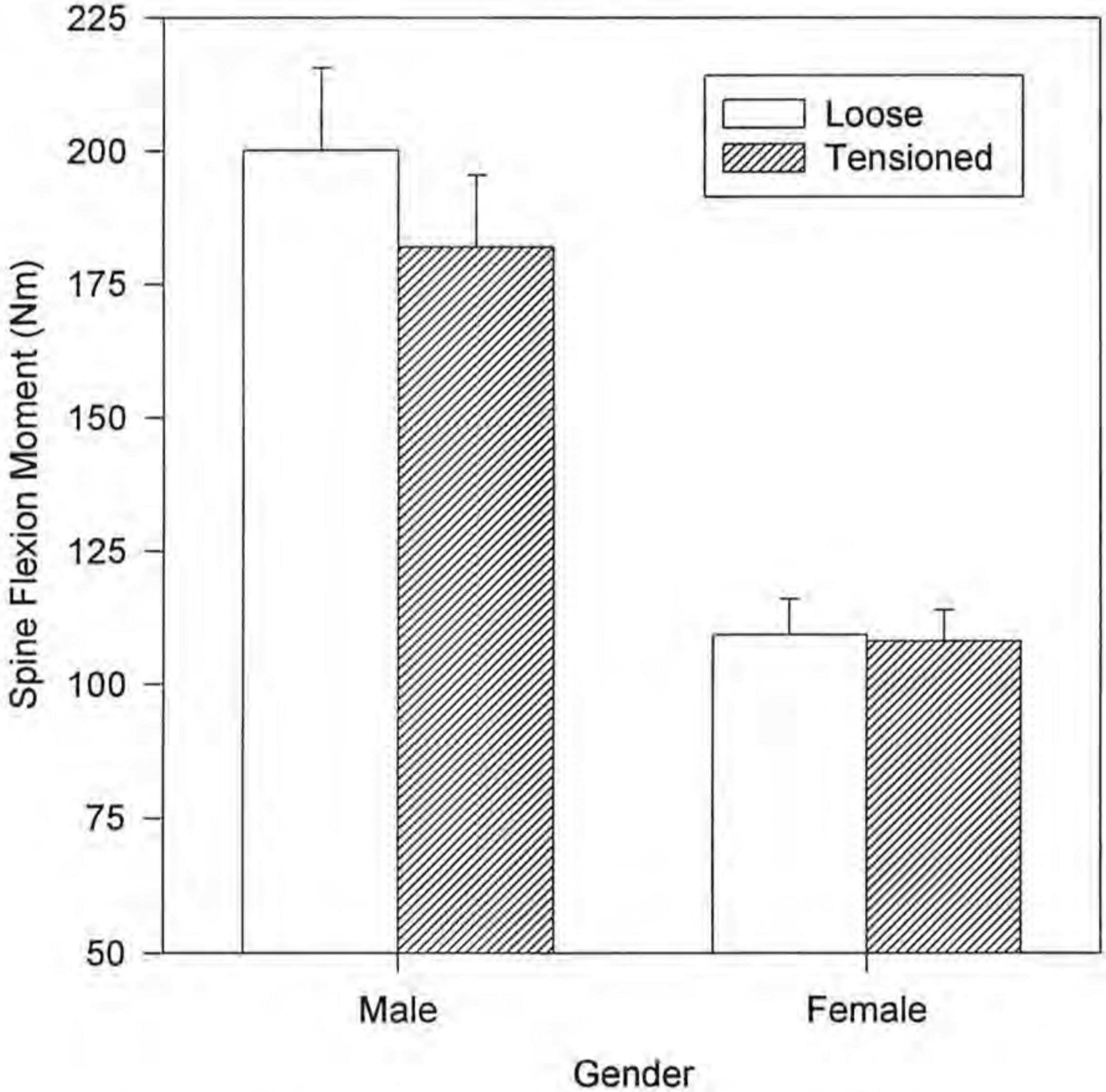

Figure 2.13. Mean forward flexion moment on the spine during the sudden unexpected loadings as a function of the subjects' gender and the belt tension.

the destabilizing effects of the sudden load, thereby lessening the peak muscle responses. Similar results have been observed when people can accurately predict the temporal onset of a sudden load (Lavender and Marras, 1995). Normally, the reduction in compliance should have been accompanied by an increase in force measured by the strain gauge during the preloaded trials. This did not occur. There was a borderline significant trend ($p=.077$) towards a three way interaction effect between the belt, preload, and gender factors with regard to the initial trunk flexion posture during the symmetric trials. In essence, females assumed a slightly more flexed posture than males during the preloaded conditions, especially without the belt. With the belt differences were minimal. There was however, considerable variability in this response.

The forward bending motion of the trunk was clearly affected by the belt. However, the significant changes occurred with the sagittally symmetric loadings. The finding of reduced forward bending is consistent with data from Magnusson et al. (1996) who reported reduced forward bending in a small sample of subjects performing a lifting task. Previous work from our laboratory, which quantified tri-axial trunk motions when lifting with and without a belt, found that the belt had no impact on sagittal plane spine motions but did reduce lateral bending and twisting motions. Other have found similar trunk motion results when studying the passive stiffness changes in the torso with and without a belt (McGill et al., 1994). Thus, the reduction in lateral bending we observed in the current study with the lifting belt tensioned was consistent with these previously published findings. It is interesting to note that the symmetry of the loading did not affect the maximum amount of lateral bending. This suggest that: (a) the overall angular change in the frontal plane posture relatively small, and (b) the symmetric loads resulted in more than just sagittal plane motion. It is likely that the rapid loading resulted in non-planer motions or spinal instability. Preload resulted in less motion, however, the tensioned lifting belt combined with the preload resulted in the least motion across both genders.

It should not be surprising that no changes in the twisting motion or twisting moments were detected even in the asymmetric loadings. The asymmetric applied load would initially create only a small twisting moment. Only after the trunk and pelvis flexed forward, would the forces acting on the hands result in a shift from lateral bending and flexion moments to axial rotation moments, and hence, 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.

Because we measured the hip motion and the spine motion we were able to quantify the change in total trunk motion, in addition to spine motion when the lifting belt was used. Rotation of the pelvis on the femurs results in a forward bending moment on the spine that must be resisted by the trunk musculature, even though there may be no motion in the spine. In the absence of the preload, the belt reduced the total motion by approximately 3 degrees during the sagittally symmetric loadings. About half of this reduction in motion occurred in the lumbar spine and about half in the pelvis. This resulted in the relative motion between the spine and the pelvis staying constant across belt conditions. With preload, however, the total spine and hip flexion was reduced, regardless of the lifting belt condition. However, the belt reduced the ratio between the spine and hip motion. This finding suggests that the stiffened spine flexion due to the belt and the trunk extensors results in relatively greater motion at the hips.

It is theorized that belt, in damping spine flexion, reduced the acceleration enough to decease the flexion moment calculated at L5/S1. It is interesting that this effect only occurred in males. There are two issues that could have contributed to this finding. First, anatomical differences between the male and female pelvis could have affected the effectiveness of the belt in controlling spine motion. However, if this were true then a belt by gender interaction effect in the spine forward flexion motion should have been found. Second, the moments were nearly twice the magnitude in the males as opposed to the females. This is largely due to the normalization procedure whereby the magnitude of the sudden load was based on the trunk extension strength. Plus, males flexed their spines about 28 percent further than females ($p<.01$) in response to the loadings, thereby adding more moment due to trunk orientation and motion. Thus, the moments were already relatively small for the females, thereby allowing less room for variation.

The greater peak muscle response in the LATL, ERSL, EXOL is consistent with the reduced forward bending and lateral bending motions following the loading, but appears inconsistent with the reduction in the flexion moment observed in the male subjects. Overall, the drop in EMG activity observed in the right external oblique, rectus abdominus, and right erector spinae (asymmetric trials only) with the belt tension could be viewed as a reduction in the co-contraction response. Thus, belt may have been perceived by subjects as being protective, which in turn resulted in reduced use of antagonistic muscles normally recruited for stabilization purposes (Ladin et al., 1989; Lavender et al., 1992). Greater agonistic muscle force, combined with less antagonistic force would result in a greater deceleration of the trunk following the onset of the loading, which reduced the motion, which in turn, lowered the peak moment.

In sum, the results from this study do not provide an easy answer with regards to the benefits of the lifting belt. In part this is due to the interactions found between the belt and the other factors considered in the experimental design. At first glance, the data support the hypothesis that belt stiffens the torso's response to sudden loading. Clearly the effects are small and the individual differences are large. This suggest that a belt may protect some individuals but not others. Further research is needed to identify the nature of these individual differences. Until this is done recommendations regarding belt use or non-use cannot be made.

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LIST OF PRESENT AND POSSIBLE FUTURE PUBLICATIONS

In Process:

1. Thomas J.S., Lavender, S.A., Corcos, D.M., Andersson, G.B.J: Trunk kinematics and trunk muscle activity during a rapidly applied load. Journal of Electromyography and Kinesiology, accepted.

2. Thomas, J.S., Lavender, S.A., Corcos, D.M., Andersson, G.B.J.: The effect of lifting belts on trunk muscle activation during a suddenly applied load. Human Factors, submitted.

Planned:

1. The effect of a lifting belt on the trunk motion and moments during sagittally symmetric and asymmetric unexpected sudden loads.

2. The effect of a lifting belt on trunk muscle recruitments during expected and unexpected suddenly applied loads.