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The Effects of a Back-Belt on Lumbar Disc Deformation During Stoop Type Lifting

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Abstract-Low-back pain and injury are responsible for a major portion of lost workdays and injury compensation claims. The use of back support belts has been forwarded as a counter measure towards reducing low-back injuries in the industrial setting. PURPOSE: The purpose of this study was to determine if a back support belt relieves stresses encountered by the lumbar spine during stoop type lifting and potentially reduce the risk of injury. METHODS: Twelve male participants (49.7±3.7 years) performed two sessions of stooped type lifting with a loaded milk crate (11.5 kg), at 4 repetitions per minute, for 15 minutes in accordance with the NIOSH lifting equation. One lifting session was performed without a support belt, while the other with a support belt. Three sets of fluoroscopic images were collected with the participants positioned at the initiation (flexed trunk), mid-range, and completion of the lift (erect standing). The first series of images were collected under a no-load condition, while the second (no support belt) and third series (support belt) of images were collected with the participants lifting the 11.5 kg milk crate. Images were imported into AutoCAD where lumbar disc deformation and joint angles were measured by calculating changes in position of adjacent vertebra (L3-4 and L4-5). A reduction of disc deformation was deemed indicative of reduced stress. RESULTS: Analysis of variance revealed that compressive and shear disc deformation were reduced while in the erect trunk posture for the support belt condition ($p< 0.05$). No significant reduction in disc deformation was detected while in flexed trunk postures for the support belt condition ($p> 0.05$). CONCLUSIONS: During stoop type lifting, support belts provide a measurable amount of stress reduction of the lumbar spine when the trunk is in the erect posture, with little effect during flexed trunk positions.

Keywords- *back belts, lumbar stress*

I. Introduction

Occupational back disorders have plagued man for centuries (1) and recent years have shown little departure from this trend. It is estimated that 60-70% of the work force will experience at least one serious incidence of sciatica or back strain during their lifetime (2, 3). The US Bureau of Statistics reported 182,270 cases involving injuries to the back in year 2011 (4). The US National Center for Health Statistics reports that 18% of all work place injuries are spine or back related (5). Mitchell et al. (6) correlated these injury occurrences to average 28.6 lost work days per 100 workers per year. The financial burden associated with work place back disorders has been estimated to cost U.S. industry in excess of $50 billion dollars a year (7).

The use of back belts has been forwarded as a counter measure towards reducing low-back injuries in the industrial setting. In light of the personal and financial burden associated with low back pain, further research investigating the relation between back belts, external loads, and stress encountered by the lumbar spine during lifting tasks is warranted.

The purpose of this study was to determine if a commonly used back belt can relieve stresses encountered by the lumbar spine during stoop type lifting.

II. Methods

Participants

Twelve male participants 40 to 55 years of age participated in this study. The participants were recruited from a heavy industrial facility and were free of back injury or pain at the time of data collection. The participants averaged approximately 20 years of employment and were primarily assigned to physically demanding labor positions associated with a heavy industrial site. Subject height, mass, and age were: 177.4 (±6.4) centimeters, 87.0 (±10.7) kilograms, and 49.7 (±3.7) years. The rationale for recruiting these participants was based on the desire to collect a subject pool which bore some resemblance of a cross-section of the labor force. Prior to the execution of the study, all participants were verbally informed of the details of the study and required to read and sign an informed consent document approved by an Institutional Review Board for the use of Human Subjects.

Procedures

The dependent variables measured were: compressive and anterior shear disc deformation (L3-L4, and L4-L5), and the associated joint angles. The methodology utilized to measure the variables was fluoroscopic imaging. Although fluoroscopic imaging does not measure soft tissue characteristics, it does allow measurement of changes in position between adjacent vertebrae (8). Changes in position of adjacent vertebrae are directly related to disc deformation and the associated stresses encountered. Lateral fluoroscopic images of participants under three different conditions were used to determine their effects on the aforementioned dependent variables. The conditions were: from a stooped position with spine flexed to standing erect under no load, from a stooped position with spine flexed to standing erect under load, and from a stooped position with spine flexed to standing erect under load with a back belt.

The load lifted (11.5 kg) was based on the Revised NIOSH lifting equation (9) and was so selected to address NIOSH's criticism of previous research efforts where loads were inconsistent with NIOSH lifting recommendations (9). The load was placed in a milk crate such that when lifted from the floor, the load was suspended just below waist level (Figure 1). In this position the arms did not interfere with the lateral fluoroscopic images. Additionally, this lifting procedure is commonly undertaken during manual handling tasks.

In order to achieve the minimum volume of mass lifted to induce spinal shrinkage (consistent with the previous research efforts), a lifting frequency of 4 lifts per minute was selected along with a 15-minute stimulus period. The mass lifted was 690 kg for the stimulus period (11.5 kg load, 15 minute stimulus period, and 4 lifts per minute). This loading duration is consistent with the methodology and findings of Tyrrell, Reilly, and Troup (10) and was intended to assure that the lumbar discs reached hydrostatic equilibrium due to the load and loading pattern. The participants were monitored to assure a controlled repeatable movement that was based on the body mechanics unique to each subject.

The dependent variables measured during the no load condition served as the baseline values. To assure that loads experienced during the course of the day (prior to testing) did not confound baseline measures, each subject was instructed to assume the Fowler's position for six minutes. The Fowler's position is typically recommended for the relief of back pain, the subject is supine with knees and hips flexed (both at 90 degrees) and the legs supported. This position has been demonstrated to return stature lost during loading (spinal shrinkage) to preloading conditions (10). Further standardization prior to the baseline fluoroscopic images included the participants standing for 20 minutes with their body weight evenly distributed on both feet (11, 12). This additional period of standing assured that the discs returned to a hydrostatic equilibrium that was due to body weight alone.

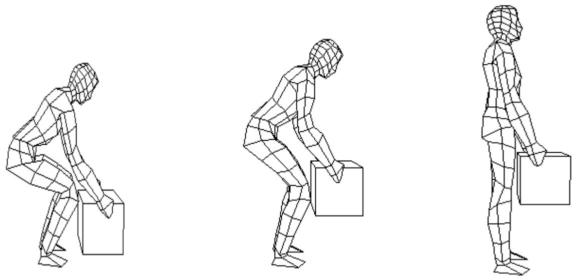

Figure 1. Stoop lifting positions: initiation, mid-range, and completion.

Upon completion of the standardization period, lateral fluoroscopic images were taken of the participants going from a stooped position with spine flexed to erect standing (under no load). This series of fluoroscopic images provided the baseline from which changes in the dependent variables were compared.

The first stimulus period consisted of the participants lifting the 11.5 kg load for 15 minutes at a frequency of 4 repetitions per minute. The participants performed the stoop lift, lifting the load from the floor to knuckle height (no back belt).

Following the stimulus period the participants were positioned for a series of fluoroscopic images. The participants were positioned uniformly with the position assumed for the initial series of fluoroscopic images. Once the participants were properly aligned, they again lifted the 11.5 kg load to knuckle height (going from a stooped position with spine flexed to erect standing) and maintained that posture while the lateral fluoroscopic images were collected (no back belt).

Following the second series of fluoroscopic images, the subjects were instructed to assume the Fowler's position for another six minutes followed by 20 minutes of standing with their body weight evenly distributed on both feet. Again, this procedure was intended to re-establish hydrostatic disc equilibrium due to body weight alone and mitigate any prior effects of spinal loading.

During the second treatment each subject was wearing a support belt. The support belt selected for this study was the Ergodyne Proflex 2000 SF®. This support belt is constructed of a light weight, elastic material. Velcro is used to secure the tightness of fit. The SF suffix referred to in the belt's name stands for "sticky fingers" which are rubber stays on the rear of the belt. These rubber stays secure the position of the belt on the trunk both laterally and longitudinally.

With the belt securely fitted to the subject, the lifting task prescribed for the first condition was repeated; the 11.5 kg load was lifted for 15 minutes at a frequency of 4 repetitions per minute. The subjects performed the stoop lift, lifting the load from the floor to knuckle height. Following this treatment the subjects were positioned for lateral fluoroscopic images. The subjects were positioned uniformly with that assumed for the two prior series of fluoroscopic images. Once the subjects were properly aligned, they again lifted the 11.5 kg load to knuckle height (going from a stooped position with spine flexed to erect standing) and maintained that posture while the lateral fluoroscopic images were collected.

Subjects crossed-over with regard to the condition of wearing the support belt. The subjects were randomly assigned to the order of stimulus period in which they were wearing a support belt. Half of the subjects were wearing a support belt during the first loading period (or stimulus period) followed by not wearing a support belt during the second stimulus period. The other half of the subjects were not wearing a support belt during the first stimulus period followed by wearing a support belt during the second stimulus period.

The fluoroscopic images were taken by a certified technician. The images were collected with an Infimed 2000 fluoroscopic imaging system. Three fluoroscopic images were collected for each of the three conditions. For each condition, the first image was collected at the initiation of the movement (stooped position with spine flexed), the second image was collected at mid-range of the movement, and third image was collected at the completion of the movement in the erect standing position (see Figure 1). The radiation exposure rate was 200 mA at an intensity of 60-85 kV, and the time of .19 seconds/image. Therefore, for the three conditions there was a total of 342 mA s of radiation exposure (200 mA/second x .19 seconds/image x 3 images/condition x 3 conditions). The total radiation exposure was less than 60% of a standard lumbar examination.

Careful attention was given to the participant's sagittal positioning and distance relative to the collection plate and beam emitter between conditions. This minimized artificial changes in the dependent measures due to out-of-plane body movement and image distortion due to beam dispersion (13). Additionally, the same technician was used throughout the data collection to minimize error. Repeated images for each position were not collected in order to avoid additional radiation exposure to the participants (14).

The maximum distance between the beam emitter and the fluoroscreen was 80 cm. Therefore participants were positioned in a manner such that the lumbar spine was centered at the mid-point between the emitter and the fluoroscreen (i.e. approximately 40 cm). The beam was centered at the forth lumbar vertebrae, this minimized beam distortion at the L3-L4 and L4-L5 junctures. A calibration grid (1/8"x1/8") was placed at the same field depth as the participant's lumbar spine. The true size of the grid allowed for the calculation of actual kinematic measures collected from the fluoroscopic images. The fluoroscopic images were imported into AutoCAD for data analysis.

A 1/8" x 1/8" (3.175 x 3.175 mm) calibration grid provided the means for characterizing the distortion within the fluoroscopic field. Comparison of the grid size in the fluoroscopic field where measurements were to be recorded varied by less than 0.10 mm. Since distortion of the fluoroscopic image was comparable to that observed in the Kanayama study (15), it was deemed negligible in this study as well.

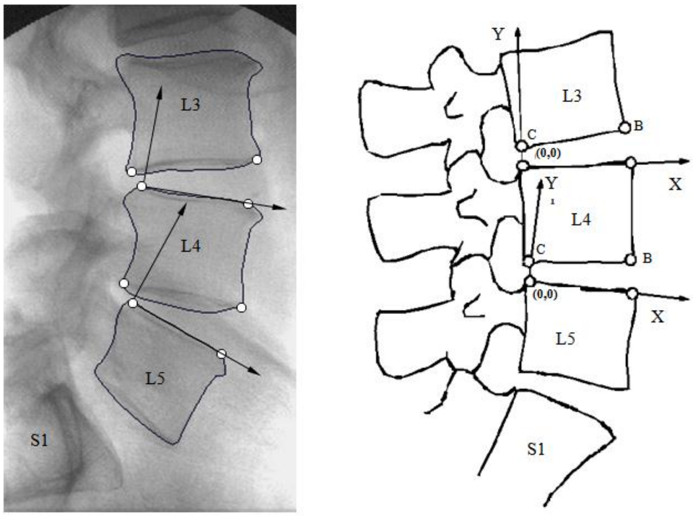

Figure 2. Local coordinate system to define lumbar disc deformation.

Disc deformation was characterized in a manner consistent with Kanayama et al. (15). A local coordinate system (see Figure 2) was established to define disc deformation for both discs L3-L4 and L4-L5. In the local coordinate system for L4-L5, the posterior-superior corner of L5 served as the origin. The X-axis extends out along the superior border of the fifth lumbar vertebrae and the Y-axis is perpendicular to it. The displacement (Δx and Δy) of the inferior corners (anterior and posterior: points B and C) of L4 served as the measure of L4-L5 disc deformation. X and Y displacements defined shear and compressive disc deformation, respectively. In the local coordinate system for L3-L4, the posterior-superior corner of L4 served as the origin. The X-axis extends out along the superior border of the forth lumbar vertebrae and the Y-axis is perpendicular to it. The displacement (Δx and Δy) of the inferior corners (anterior and posterior: points B and C) of L3 served as the measure of L3-L4 disc deformation. X and Y displacements defined shear and compressive disc deformation, respectively.

The angle formed by the intersection of the lines extending across the caudal border of the forth lumbar vertebrae and the cranial border of the fifth lumbar vertebrae defines the L4-L5 joint angle. Similarly, the angle formed by the intersection of the lines extending across the caudal border of the third lumbar vertebrae and the cranial border of the forth lumbar vertebrae defines the L3-L4 joint angle. The importance of measuring these angles relates to their impact on the degree of lumbar lordosis (i.e. greater L3-L4 and L4-L5 joint angles correlate with accentuated lordosis).

The local coordinate systems were established through the use of AutoCAD release 12 (Autodesk, Inc.). Silhouettes of the vertebrae L3, L4, and L5 were sketched. The local coordinate system for L3-L4 was affixed to the superior border of the L4 silhouette. The local coordinate system for L4-L5 was affixed to the superior border of the L5 silhouette. These silhouettes were maintained in layers, where they could be retrieved and superimposed onto other images. This procedure is essentially the same as that described by Dvorak, Panjabi, Chang, Threiler, and Grob (16), except that the silhouettes were generated and superimposed with AutoCAD instead of by hand.

All images were analyzed by the same author (MD). Twenty images were randomly selected for re-analysis in order to quantify intra-observer variance or repeatability. The correlation between the repeated measures was .999, and the mean and SD of the intra-observer difference (17) were 0.00 +/- 0.12 mm. All of the intra-observer differences were within ± 2 SD of the mean difference.

A personal computer with SuperANOVA software package (Abacus Concepts, Inc. Berkeley, Ca.) was utilized for data management and statistical analysis. Standard descriptive statistics (mean and standard deviation) for age, height, and weight were calculated.

At the completion of the lift (the erect standing position), a paired t-test was utilized to determine differences between conditions (no support belt and 11.5 kg load, support belt and a 11.5 kg load) for the dependent variables of compressive and anterior shear disc deformation (L3-4, L4-L5). It should be noted here that disc compressive and shear deformation are a measure of change in disc shape from the no support belt and no load condition, while in the erect standing position.

At the beginning and mid-range of the lift, a 1x3 analysis of variance (ANOVA) design (completely within) with repeated measures was utilized to determine differences between conditions (no support belt and no load, no support belt and 11.5 kg load, and support belt and a 11.5 kg load), for the dependent variables of compressive and anterior shear disc deformation (L3-4, L4-L5). It should be noted here that disc compressive and shear deformation are a measure of change in disc shape from the no support belt and no load condition, while in the erect standing position. Therefore, disc compressive and shear deformation are measurable quantities during the no support belt and no load condition while the spine is flexed in the stooped position (initiation of the lift) and in the mid-range position.

For the joint angles L3-L4 and L4-L5, a 1x3 analysis of variance (ANOVA) design (completely within) with repeated measures was utilized to determine differences between conditions (no support belt and no load, no support belt and 11.5 kg load, and support belt and a 11.5 kg load) at completion of the lift in the erect standing position. It should be noted here that the joint angles were compared directly to the resting condition angular measures.

An evaluation of power for this study is based on an alpha = .05, number of subjects (n = 12), 3 treatments, and the largest difference among means (or effect size) = 1.25. Power is estimated to be approximately 0.70 (18).

III. RESULTS

Support belt efficacy was determined by collecting lateral fluoroscopic images of the lumbar spine under three different conditions. These three conditions were: from a stooped position with spine flexed to standing erect under no load without a support belt, from a stooped position with spine flexed to standing erect under load without a support belt, and from a stooped position with spine flexed to erect standing under load with a support belt. For each condition, the first image was collected at the initiation of the movement (stooped position with spine flexed), the second image was collected at mid-range of the movement, and third image was collected at the completion of the movement (standing erect).

TABLE I. LUMBAR DISC DEFORMATION

No Load No Belt

	Point B		Point C		
L3-4	Δx	Δy	Δx	Δy	Angle
Flexed	2.7 (1.0)	-3.6 (1.0)	1.9 (.8)	3.6 (1.6)	1.4 (5.3)
Midrange	1.8 (0.9)	-2.5 (1.4)	1.1 (0.7)	2.3 (1.4)	5.1 (6.2)
Erect	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)	12.6 (3.7)
L4-5					
Flexed	2.8 (1.6)	-3.7 (1.6)	1.6 (1.1)	3.5 (1.9)	4.2 (5.4)
Midrange	2.2 (1.3)	-2.9 (1.5)	1.1 (0.8)	2.6 (1.7)	7.1 (6.3)
Erect	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)	16.0 (5.1)

Load No Belt

	Point B		Point C		
L3-4	Δx	Δy	Δx	Δy	Angle
Flexed	2.8 (1.1)	-4.2 (1.2)	2.0 (0.8)	2.8 (1.5)	1.7 (5.2)
Midrange	2.2 (0.8)	-3.3 (1.3)	1.4 (0.7)	1.9 (1.1)	4.5 (4.9)
Erect	0.3 (0.2)	-0.7 (0.5)	0.4 (0.2)	-0.8 (0.3)	12.8 (4.0)
L4-5					
Flexed	3.2 (1.9)	-4.3 (1.9)	2.0 (1.3)	2.8 (1.4)	4.4 (5.6)
Midrange	2.5 (1.5)	-3.8 (1.8)	1.4 (0.9)	2.2 (1.3)	6.2 (6.1)
Erect	0.4 (0.3)	-0.8 (0.4)	0.4 (0.2)	-0.8 (0.5)	15.9 (4.9)

Load Belt

	Point B		Point C		
L3-4	Δx	Δy	Δx	Δy	Angle
Flexed	2.7 (1.0)	-3.9 (1.0)	1.9 (0.8)	3.2 (1.3)	1.6 (5.0)
Midrange	1.8 (0.8)	-3.2 (1.3)	1.1 (0.6)	2.4 (1.3)	4.6 (4.6)
Erect	0.2 (0.2)	-0.4 (0.3)	0.2 (0.2)	-0.4 (0.3)	12.7 (3.8)
L4-5					
Flexed	3.1 (1.7)	-4.0 (1.8)	1.7 (1.0)	3.3 (1.8)	4.8 (3.7)
Midrange	2.5 (1.4)	-3.5 (1.8)	1.3 (1.0)	2.4 (1.3)	7.1 (4.4)
Erect	0.3 (0.2)	-0.4 (0.3)	0.3 (0.2)	-0.3 (0.3)	15.8 (5.0)

Note: Lumbar disc compressive and shear deformation are a measure of change in position (Δx, Δy for points B and C) from the no support belt and no load condition, while in the erect standing position. Δx and Δy are in millimeters, angular values are in degrees, mean (SD).

Initiation of the Movement

No significant differences were detected between the belted and non-belted conditions for shear (Δx) or compressive (Δy) disc deformation at either point B or C for both the L3-4 and L4-5 functional units ($p>.05$).

Mid-range of the Movement

No significant differences were detected between the belted and non-belted conditions for shear (Δx) or compressive (Δy) disc deformation at either point B or C for both the L3-4 and L4-5 functional units ($p > .05$).

At the initiation of the movement (stooped position with spine flexed) and at mid-range of the movement, the belt was found ineffective in terms of reducing compressive or shear disc deformation at the L3-4 and L4-5 junctures. In the absence of any significant reduction of disc deformation during belted conditions, it is concluded that the support belt is not successful in reducing the amount of stress on the lumbar spine.

Completion of the Movement

Significant differences were detected between the belted and non-belted conditions for shear (Δx) (point C) and compressive (Δy) disc deformation (point B and C) at both the L3-4 and L4-5 functional units ($p < .05$). No significant differences were detected between the belted and non-belted conditions for the L3-4 or L4-5 joint angles ($p > .05$).

At the completion of the movement (standing erect), the support belt was found to reduce disc deformation. Shear deformation was significantly reduced on the posterior aspect of the functional unit while compressive deformation was reduced on the anterior and posterior aspects of L3-4 and L4-5 junctures. A reduction of disc deformation is deemed indicative of reduced stress in the disc, and therefore it is concluded that the support belt is effective in reducing lumbar spine stresses while in this posture.

IV. DISCUSSION

Previous investigators examining the efficacy of back support belts focused on monitoring the dependent variables of: EMG, inner-disc pressure, inner-abdominal pressure, gross trunk motion, intersegmental mobility, stature loss, and maximum acceptable weights (12, 14, 19, 20, 21, 22, 23, 24, and 25). These researchers hoped that by monitoring changes in these variables due to the introduction of a support belt, they could determine if the belt was indeed unloading the spine and, if so, by what mechanisms this unloading was occurring. The National Institute for Occupational Health and Safety critiqued these and other related research efforts and concluded, "there are insufficient data indicating that typical industrial back belts significantly reduce the biomechanical loading of the trunk during manual lifting" (26). Further, the professional journals of the American Medical Association and the Canadian Medical Association both report that there is a lack of evidence supporting the use of back belts for the reduction of workplace back injuries (27, 28)

This study utilized fluoroscopic imaging and measured disc deformation in a manner consistent with Kanayama et al. (15), focusing on changes in disc deformation as a function of support belt usage. This approach circumvented the need to determine what physiologic and/or mechanical mechanisms were responsible for the support belt's effectiveness in reducing spinal loading (or lack thereof). The theory being, that if the belt was in some manner unloading the spine it would be manifested via a reduction in disc deformation. Since there is a relationship between disc deformation and stress, a reduction in disc deformation would then imply a reduction in stress. Hence, this methodology provided the ability to measure compressive and shear loading of the intervertebral disc.

The displacement of points B and C during the initiation of the lift (stooped position with spine flexed, no load, no belt) are listed in Table 1. Displacements at the L4-L5 juncture were larger than those observed at the L3-L4 juncture (with the exception of point C). These results compare favorable with Kanayama et al. (15) where displacement of both points was greatest at the L4-L5 juncture (with the exception of point C Δx). The magnitudes of displacements listed in Kanayama et al. (15) are slightly higher than those reported here. In this study the initial position for the lift was a stooped position with the trunk flexed. The subjects were allowed to flex their trunk with a combination of both pelvic rotation and spinal flexion. In Kanayama et al. (15) the subjects flexed their trunks while having their pelvis fixed such that trunk flexion was accommodated via spinal flexion only. It is likely that greater spinal flexion was achieved due to fixation of the pelvis and thus likely explains the larger maximum displacements. It is also possible that the subjects in this study achieved maximum spinal flexion during the initiation of the lift, as maximum spinal flexion is usually achieved at approximately 50-60 degrees of trunk flexion (29). If this is the case, the small margin of difference between the results reported here and Kanayama et al. (15) are likely due to inter-individual variance or age-related differences.

Table 1 also lists the L3-L4 and L4-L5 joint angles achieved by the subjects when they were in the flexed trunk position at the initiation of the lift (no load, no belt). Most studies investigating the lumbar range of motion reported the results as a total range of motion, flexion combined with extension. Pearcy, Portek and Shepherd (30) reported their results by separating the lumbar range of motion into flexion range and extension range. The authors of that study reported the flexion range of motion for the L3-L4 and L4-L5 as 12(±1) and 13(±4) degrees, respectively. The angles in this study compare favorably with Pearcy, Portek and Shepherd (30) both in magnitude and descending order (cranial to caudal) of increasing range. The slightly lower angular values noted in this study are likely due to the age-related decrease in lumbar range of motion (31). Pearcy, Portek and Shepherd's (30) subject pool averaged 29.5 years of age, as compared to an average age of 50 years for this subject group.

The direction and displacement of points B and C defines the criteria by which disc deformation was used to assess the effectiveness of the support belt. Point B refers to deformation encountered on the anterior aspect of the functional unit, while point C refers to deformation encountered on the posterior aspect of the functional unit. Delta Y refers to compressive deformation and delta X refers to shear deformation. Changes in the position of point C are postulated as the most critical, as tissue failures or impingement occurring along the posterior surface of the functional unit are most likely to be associated with pain due to their location relative to the spinal cord and nerve branches (32).

The belted conditions of this study were balanced with respect to the two stimulus periods. In order to determine if there was an effect due to stimulus order, paired t-tests were performed comparing the disc deformation measured during the first stimulus period and the second stimulus period. No significant differences were found between the two stimulus periods due to order effect. Therefore, any order effect due to stimulus period was deemed negligible.

During the initiation and mid-range of the lift, the loads encountered by the lumbar spine are at their peak. Body segment weight moments as well as the 11.5 kg load moment are at their greatest due to large lever arms. The erector spinae muscle group must exert large forces over a small lever arm to counter the torques associated with the body segment's weight and 11.5 kg load. As the erector spinae generates these high forces, the lumbar spine experiences large compressive stresses. The inclination of the trunk during the initiation and mid-range of the lift contributes large shear forces to the lumbar region as well. The posterior elements of the lumbar vertebrae (including the posterior portion of the intervertebral discs) bare the brunt of these forces and are typically the location of pain development. It is believed that the lumbar discs are most prone to failing when the trunk is in a flexed or rotated position while under load (32). The rationale for this is based on the knowledge of annular fiber disruption or failure criteria. It is believed that annular fibers fail after 5 degrees of vertebral rotation relative to a given juncture level (32). The annular fibers are essentially stretched apart. Flexion of the spine while under load places the annular fibers under the same sort of loading pattern. The posterior portion of the disc is the location of greatest stress concentration and ultimately is the site of annular fiber failure.

This study attempted to determine if by some physiological or mechanical mechanism the belt was facilitating the unloading of the lumbar spine during a given lifting activity. The data suggest that the support belt used was not effective in unloading the lumbar spine at the initiation or mid-range of the lift when lumbar stresses are at their greatest and annular fiber failure most likely.

During the completion of the lift, the loads encountered by the lumbar spine are at a minimum. Body segment weight moments as well as the 11.5 kg load moment are at their lowest due to decreased lever arm lengths. The erector spinae muscle group need only exert small forces over a small lever arm to counter the torques associated with the body segment's weight and 11.5 kg load. As the erector spinae force is reduced, the lumbar spine experiences lower compressive stresses. In the erect standing position, the inclination of the trunk contributes only small shear forces to the lumbar region. In this position the support belt seems to be effective in reducing the stresses encountered by the lumbar spine. It is curious as to why the belt seems to unload the lumbar spine in the erect position but not in the earlier mid-range or flexed portions of the lift. The physiologic or mechanical mechanism afforded by the support belt in the erect position is not of a significant nature during the flexed trunk portions of the lift. Therefore, the mechanism must either be absent during the flexed trunk portions of the lift, overwhelmed by larger forces in the lumbar region associated with the greater weight moments via extended lever arms and/or greater shear forces due to trunk inclination, or be masked by some other mechanism.

Table 1 provides a comparison of joint angles between conditions for the L3-L4 and L4-L5 junctures during the completion of the lift (i.e. standing erect). No significant differences ($p>.05$) in mean joint angle measures were found between conditions for the L3-L4 and L4-L5 junctures. The results of this comparison suggest that the support belt is not effective in terms of minimizing significant increases of the L3-L4 and L4-L5 joint angles at the completion of the lift. Thus, the belt demonstrated no impact on the amount of lumbar lordosis in the erect position during lifting tasks of this nature.

Lumbar lordosis is an accumulation of the joint angles L1-L2 through L4-L5, with the L5-S1 and sacral horizontal angles also having a measurable impact. The relationship between extreme lordosis and low back pain has been well established (32). The relationship between L5-S1, the sacral horizontal angle and low back pain has also been acknowledged (8). In this study the L5-S1 and sacral horizontal angles were not measured. The age related degeneration of the L5-S1 joint made it difficult to measure the L5-S1 or sacral horizontal angles in any reliable fashion (i.e. the joint was fussed in most subjects). The L3-L4 and L4-L5 joint angles were measured as an indicator of the amount of lumbar lordosis present. As previously stated, no differences between conditions were detected indicating that the support belt had no effect on the amount of lumbar lordosis in the erect posture for lifting tasks of this nature. This result is in direct contrast to suggestions that corsets be worn to reduce lumbar lordosis in patients with low back pain (32). However, in this study comparisons were not made between unloaded belted and unloaded non-belted conditions which may explain this discrepancy. Additionally, none of the subjects in this study were currently patients with low back pain.

A study by Bourne and Reilly (12) attempted to determine the effect of a "standard" weight-lifting belt on spinal shrinkage during circuit weight training. Spinal shrinkage is a measure of stature loss as a result of spinal loading. Stature loss could be due to compression of intervertebral discs and/or changes in the kyphotic or lordotic curves of the spine. Examination of the L3-L4 and L4-L5 joint angles in this study suggest that insignificant changes occurred in the amount of lumbar lordosis due to wearing a belt or the load lifted. Therefore, it could be postulated that the spinal shrinkage observed in earlier studies (10, 11, 12, 33, 34, 35, 36, 37, and 38) was a function of disc deformation only and not changes in kyphotic or lordotic curves of the spine.

The lumbar support belt used in this study provided a measure of stress reduction in the lumbar spine. In this study subjects flexed their trunk with a combination of spinal flexion and pelvic rotation. The positive effects of the belt were either not present or were undetectable when the trunk was flexed. However, in the erect trunk position (spine not flexed), the lumbar stress reduction manifested itself. Other studies have detected positive benefits from a support belt while the trunk was in an erect position (or the trunk was flexed but the spine was not flexed). Nachemson, Schultz, and Andersson (39) found reduced inner disc pressures (IDP) as a result of wearing a support belt for isometric extension resistant tasks while the trunk was in the erect position. Lander, Simonton, and Giacobbe (40) measured a number of variables and used a model to calculate L5-S1 forces during a squat exercise. Assuming the squat was performed in a proficient manner, trunk flexion occurs primarily via pelvic rotation with little or no spinal flexion. The author's calculations suggested that the two belted conditions had "significantly smaller forces than the non-belted conditions." Bourne and Reilly (12) attempted to determine the effect of a "standard" weight-lifting belt on spinal shrinkage during circuit weight training. Spinal shrinkage is measured in the erect standing position. The stature loss for the belted group was less than that of the non-belted group. McGill, Seguin, and Bennet (41) studied the effect of belt wearing and breath holding on passive stiffness of the upper torso. Stiffness measures were taken with the subjects standing in the erect position with bending moments applied to the trunk. The authors concluded that belts and breath holding appear to increase trunk stiffness in the frontal and transverse planes with little effect on sagittal plane torso stiffness.

The results of this study combined with those of similar findings might suggest that using a support belt in any scenario where the spine is not flexed might be beneficial in terms of unloading the spine. For example, sitting in a chair requires the hip to be flexed with minimal spinal flexion. It is has been hypothesized that prolonged sitting causes a posterior migration of nuclear material within the disc (42), and a reduction of stature (35). Wearing a support belt might mitigate or minimize the stresses on the lumbar spine due to prolonged sitting. As mentioned above, squatting with proper technique insures trunk flexion occurs as a result of pelvic rotation with little or no spinal flexion. Under these conditions, the stresses encountered by the lumbar spine may be reduced through use of a support belt. Carrying a load over a distance where the trunk is erect is another example when a support belt might facilitate spinal unloading.

The question as to why the support belt appears to function only when the lumbar spine is not flexed (i.e. normal lordosis) must be asked. First, it should be noted that even when the spine was flexed, there was a tendency towards reduced disc deformation for the belted conditions at the initiation of the lift and at mid-range of the lift. This tendency was present at the L3-L4 and L4-L5 junctures (see Table 1). However, this tendency was not considered statistically significant. It is possible that had a greater load been utilized during the lift this tendency towards reduced disc deformation during belted conditions may have exhibited a statistical significance. Likewise, a more rigid belt may have lead to a significant reduction in disc deformation.

Another possible explanation may have to do with intra abdominal pressure (IAP) and it's inter-relationship with breath holding and belt wearing. McGill, Norman, and Sharratt (23) demonstrated that breath holding as well as belt wearing significantly increased IAP during squat lifts. Breath holding was not controlled during this study for two reasons. First, it reduces the external validity of the results. Second, it has been postulated that there is an elevated cardiovascular risk associated with breath holding and lifting (43). The subjects in this study averaged 50 years of age. It was decided that a protocol with instructions for breath holding might be potentially dangerous. However, the subjects were allowed to hold their breath if it was natural for them to do so, as "casual observation of all sorts of lifting indicates that people hold their breath during exertion" (23). It is possible that lack of controlling breath holding may have in some manner confounded the effects of wearing a support belt.

Previous research suggests that during lifting, IAP is greater when the trunk is flexed (with or without spine flexed) as compared to when the trunk is erect (20). The reason for this is due to the volume of the abdominal cavity being reduced during trunk flexed conditions and thus a compensatory increase in IAP must result. Breath holding and belt wearing are known to increase IAP and thus would affect the amount of IAP increase due to trunk flexion. If this is the case, it is possible that the increase in IAP due to trunk flexion and breath holding were of sufficient magnitude to mask any potential reductions in disc deformation which might have been afforded by increases in IAP due to the support belt. This of course presumes that increased IAP facilitates unloading of the lumbar spine in some manner.

During erect standing conditions, the volume of the abdominal cavity re-expands and a compensatory reduction in IAP occurs. Under these circumstances, the relative contribution of IAP due to belt wearing is now larger then during the flexed trunk conditions. This might explain why a significant reduction in disc deformation was observed at the completion of the lift, in the erect standing position during the belted condition.

The hypothesis forwarded above is based on the theory that IAP is unloading the spine in some manner. The proponents and opponents of this theory have debated this topic at length. The issue is still unresolved. Proponents of the theory suggest that thoracic and abdominal cavities act as "rigid wall cylinders", potentially resisting compressive loads that would otherwise be placed on the lumbar spine (44). Others suggest that the spinal compressive loads are reduced by a trunk extensor moment which is generated by increased IAP (45, 46). The magnitude of this extensor moment is a product of diaphragm area, the moment arm connecting the diaphragm to the lumbar region, and IAP. The extensor moment associated with increased IAP is thought to reduce the muscle activity of the trunk extensors, and hence reduce spinal compression. Finally, Gracovetsky and Farfan (47) proposed that increases in IAP exert a posterior hydraulic action on extensor tissue. Tension is generated within the tissue, thus producing an extensor moment. Opponents of IAP's role in unloading the spine argue these points. McGill and Norman (48) provided a synopsis of the current state of thought with respect to the role of IAP in reducing spinal loading, "the generation of IAP during load-handling is well documented, the role of IAP is not."

McGill and Norman (48) forwarded the postulate that IAP was likely related to a mechanism by which the lumbar spine is stabilized with little or no effect on reducing compressive loads. Likewise, they suggested that the activated abdominals which increase IAP "create a rigid cylinder of the trunk, resulting in a stiffer structure." If this is the case, the authors are unknowingly stating that IAP must be actively involved in reducing shear stresses or providing shear stress relief of the lumbar spine. The stiffer a structure is, the greater it's resistance to changes in shape. If trunk stiffness is providing resistance to changes in shape in an anterior-posterior direction, then increased trunk stiffness is resisting loads that would otherwise be placed on the lumbar spine.

McGill and Norman's (48) contribution related to lumbar support belts is one which expands this rationale to encompass how IAP may provide a mechanism for stabilizing the lumbar spine. It is stated that lumbar support belts increase IAP by 21% (48). Thus, IAP increases due to belt wearing are forwarded as a means of further stiffening the lumbar spine. Additionally, the authors suggest that the abdomen might benefit from the structure afforded by a support belt by minimizing anterior-posterior shear, much the same as the rib cage in the thoracic spinal region.

This research effort focused on the premises forwarded by McGill and Norman (48). Supporting information not elucidated by the aforementioned authors is forwarded here.

Assume the spine and trunk behave as a column. Column buckling theory suggests that the buckling limit of a column can be elevated by increasing the mass moment of inertia (I) or by providing lateral support to the column (49). A support belt satisfies both these criteria. First, the mass of the support belt (albeit small relative to the trunk) increases the area moment of inertia of the trunk. The increase in IAP as a result of wearing the support belt acts directly against the posterior wall of the abdominal cavity. The posterior wall of the abdominal cavity is coincident with the anterior perspective of the lumbar functional units. The increase in IAP is directly applied to the functional units and thus supports the spinal column and improves stability. This notion could best be visualized in the following manner: whatever pressure is exerted against the support belt is the minimum pressure exerted against the lumbar spine. Further, Spinal erector muscle action is essentially zero at the initiation of lifting activities when the trunk is in a flexed position (32). When these passive tissues are pre-stressed via IAP, it is possible that they become active load bearing members of the column (or lumbar spine). As such, they enlarge the mass distribution, which in turn increases the area moment of inertia and the critical buckling limit. Increasing the mass distribution increases the radius of gyration. An increase in the radius of gyration reduces the slenderness ratio (a ratio of column length to radius of gyration); the square of the slenderness ratio is inversely proportional to the critical stress at buckling (49). Any support to the lumbar spine via increased IAP pre-stressing passive tissue would be of great importance in terms of minimizing the potential of the spine buckling. Studies by Miyamoto and colleagues (50, 51) examined the effects of back belts on trunk muscles and trunk cross sectional shape utilizing fast MRI and computer tomography. The results indicated that there is a change in the trunk architecture when back belts are in use, lending support to the spine-column buckling hypothesis presented above.

McGill and Norman (48) discuss a similar effect: "the co-contracting musculature of the lumbar spine can perform the role of stabilizing guy wires to each lumbar vertebrae bracing against buckling." If McGill and Norman's hypothesis is valid, then the contention related to IAP and lumbar stabilization presented in the previous paragraphs must also hold some validity.

Additional verification for this notion of improved stability due to belt wearing is observed on a number of occasions in the respective literature. McGill, Seguin, and Bennet (41) studied the effect of belt wearing and breath holding on the passive stiffness of the upper torso. The authors concluded that belts and breath holding appear to increase trunk stiffness.

Lander, Hundley, and Simpton's (52) research effort focusing on the squat determined that IAP was significantly increased from 25-40% for belt-wearing conditions over the non-belt conditions. Ground reaction force data during the up phase (or ascent phase of the lift) of the belted condition was of significantly less duration than the non-belted condition. This same phenomena was exhibited in an earlier study by Lander, Simonton, and Giacobbe (40). Woodhouse, Heinen, Shall, and Bragg (53) measured the effects of lumbosacral supports on isokinetic lifting parameters of the squat. The data demonstrated a trend towards greater peak force and greater average muscular power while wearing a support belt. The data of these studies implies that the subjects were able to perform the lifting tasks much more rapidly during belted conditions. Further, Zink et al. (54) demonstrated that the use of back belts increased the velocity of bar movement while performing the squat exercise.

It is suggested here that the ability to perform lifting tasks in a smaller time interval is the result of improved spinal stability, the result of increased IAP afforded by a lumbar support belt. A similar phenomenon is observed when comparing lifting activities performed on a lifting apparatus (machine lifting) versus free weights. Subjects can invariably lift more weight at a greater rate on a lifting machine than for the identical movement with free weights. The reason for this differential is that the weight or resistance associated with a lifting apparatus is pre-stabilized by the equipment itself. The need for the neuromuscular system to stabilize the weight is alleviated. The neuromuscular system need only direct it's efforts towards pushing as hard as possible in the direction of the pre-stabilized resistance provided by the lifting apparatus. When handling free weights, the neuromuscular system is challenged to both move and balance the load being lifted. It is hypothesized that in a similar manner, a lumbar support belt stabilizes the lumbar spine (via increased IAP), which in turn allows the neuromuscular system to focus on moving the weight, with a reduced burden of balancing and stabilizing the lumbar spine. This hypothesis is best left to motor control experts to explore further.

The results of this study suggest that a support belt may be effective in terms of reducing compressive and shear stress in the lumbar region when the trunk is in an erect posture, with no measureable benefit while in flexed trunk positions. However, the lumbar spine is most vulnerable to injury during extreme flexion as well as rotation, with an even greater susceptibility to injury during lifting tasks requiring these postures (Calliet, 1988). If the support belt is not effective in reducing lumbar stresses while in these most critical postures, the efficacy of support belt usage must be questioned. The data collected during this study do not support the contention that support belts are reducing compressive or shear stress in the lumbar region while in the most critical positions (when compressive shear forces are at their greatest). The data in this study suggest that support belts are effective in non-critical postures in terms of reducing compressive and shear stress in the lumbar spine.

It must also be kept in mind that the results of this study are based on static lifting postures. Lifting activities are more often of a dynamic nature. Researchers such as Stewart McGill have suggested that low back injuries may be related to a momentary motor control failure, which leads to tissue damage or nerve impingement. If this is the case, it would be unlikely to identify such an event during a static lifting event. Therefore, the impact of a support belt on dynamic motor control movements cannot be addressed based on the results of this study. Additionally, the interaction of the inertial characteristics of the trunk during a dynamic lifting activity cannot be addressed by this study.

The question as to whether the number of workplace back injuries could be reduced by wearing support belt remains unanswered based on the results of this study. It appears likely that in some postures one's risk of back injury maybe reduced while wearing a support belt. Until researchers can definitively resolve the controversy over support belts, medical professionals, ergonomists, engineers, epidemiologists, and biomechanists must carefully weigh theoretical data and empirical observations in order to arrive at a reasonable suggestion concerning belt usage.

V. CONCLUSIONS

Within the limits of this study, it is concluded that:

1. Support belts reduce stress on the lumbar spine when the trunk is in the erect posture.

2. Support belts are not effective in reducing stress on the lumbar spine when the trunk is in flexed postures.

3. Lumbar lordosis is not favorably impacted while wearing a lumbar support belt in the erect trunk position.

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