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Lumbosacral orthoses reduce trunk muscle activity in a postural control task

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Abstract

Biomechanical modeling estimated that trunk muscle activity during various tasks could be reduced by 1–14% without the loss of spine stability when a lumbosacral orthosis (LSO) is worn [Cholewicki, J., 2004. The effects of lumbosacral orthoses on spine stability: what changes in EMG can be expected? Journal of Orthopedic Research 22, 1150–1155]. The present study experimentally tested these theoretical predictions in an unstable sitting task. This task required subjects to balance on a seat supported by a plastic hemisphere ($\varnothing = 30$ cm) and placed on a force plate that tracked the center of pressure (CoP). The average CoP velocity quantified subjects' performance. Healthy subjects (12 males, 11 females) balanced for 20 s in 3 trials performed with and without the LSO in random order. EMG was recorded bilaterally from rectus abdominis (RA), external oblique (EO), thoracic (TES) and lumbar erector spinae (LES), and expressed as the % of maximum voluntary activation (%MVA). There was no difference in the balance performance with and without the LSO ($p = 0.13$). However, EMG averaged across the trials was significantly lower in the LSO, as compared to the No LSO condition, for TES (5.8 ± 3.2 vs. 6.4 ± 3.7%MVA, $p = 0.02$) and LES (3.7 ± 1.5 vs. 5.9 ± 3.9%MVA, $p = 0.01$). No significant differences were present in the abdominal muscle activity. These results agree with earlier spine modeling simulations, which predicted the greatest reduction in muscle activity due to LSO to occur in TES and LES. It was hypothesized that such a reduction in muscle co-contraction could benefit patients with low back pain, who exhibit elevated muscular activity during postural tasks such as walking, standing and sitting.

Keywords: Abdominal belts; Back support; EMG; Balance; Neuromuscular control

1. Introduction

One popular notion about the mechanisms and function of abdominal belts and lumbosacral orthoses (LSOs) is that they provide support to the spine. Indeed, people with low back pain (LBP) perceive added support from wearing an LSO, which increases their confidence in undertaking various physical activities (Ahlgren and Hansen, 1978; Alaranta and Hurri, 1988; Jellema et al., 2002; Million et al., 1981). However, any attempts to objectively quantify such benefits have failed to date. No systematic improvement in strength or muscle endurance has been reported when belts were worn by healthy subjects (Ciriello and Snook, 1995; Lavender et al., 1998; Majkowski et al., 1998; Reyna et al., 1995; Smith et al., 1996). Similarly, no systematic reduction in the activity of erector spinae muscles has been found when tasks executed with and without the belt were compared (Ivancic et al., 2002; Marras et al., 2000; McGill et al., 1990). Nor are there any reports of a significant decrease in spine compression forces that could be attributed directly to the action of belts (Ivancic et al., 2002; Marras et al., 2000; Rohlmann et al., 1999). A few studies have found some reduction in muscle activity or spine compression forces, however, the task kinematics also changed when belts were worn (Granata et al., 1997; Woldstad and Sherman, 1998). Thus in these studies, the documented changes in muscle activity could

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be solely due to the different trunk kinematics. Ultimately, systematic reviews of the literature reject the hypothesis that abdominal belts or LSOs support the spine in a way that significantly reduces muscle forces and spinal loads (Calmels and Fayolle-Minon, 1996; McGill, 1993, 1998; van Poppel et al., 2000).

The rejection of the spine unloading hypotheses via LSOs does not preclude their possible function in spine stability. Abdominal belts and LSOs increase trunk stiffness and enhance spine stability by making the entire trunk more robust to perturbations (Cholewicki et al., 1999; Ivancic et al., 2002; Lavender et al., 2000; McGill et al., 1994). Increased trunk stiffness most likely comes from the passive interaction between the belt and abdomen, as no significant changes in trunk muscle activity could be demonstrated (Cholewicki et al., 1999). The magnitude of additional trunk stiffness around the spine's neutral posture is small and cannot result in appreciative moments that would aid the trunk musculature during the execution of strenuous tasks (Cholewicki et al., 2003). However, in postural control tasks, where the trunk muscle activity does not usually exceed 3% of maximum voluntary activation (%MVA) (Cholewicki et al., 1997), the added stiffness from an LSO could contribute significantly to spine stability. In these tasks, the spine is most vulnerable to buckling, because very little muscular effort is required (Cholewicki and McGill, 1996).

Cholewicki (2004) estimated that trunk muscle activity could theoretically be reduced between 1 and 14%MVA during various tasks, without loss of spine stability, when an LSO is worn. In a detailed biomechanical model, passive spine stiffness was increased by an amount equivalent to the stiffness provided by an LSO (Cholewicki, 2004). This augmentation of stiffness led to a higher spine stability index, as calculated by the model. Next, the trunk muscle forces, which were estimated from EMG, were reduced uniformly to the point where the spine stability index reached its original value. The differences in muscle forces obtained from these two simulations represented the muscular effort that could theoretically be relieved with the use of an LSO (Cholewicki, 2004). Based on these data, the expected reduction in trunk muscle activity is small, which explains why it could not be detected in previous studies examining the effects of abdominal belts and LSOs in tasks requiring considerable muscular exertions.

Can this enhancement of spine stability be perceived by the central nervous system (CNS) as the "support" and can any motor control adaptations to this support be measured? The answers to these questions have important implications, because a small reduction in muscle co-contraction could prevent muscle fatigue in patients with LBP, who exhibit elevated muscular activity during postural tasks such as walking, standing, and sitting (van Dieën et al., 2003). It is known that static contractions sustained above 5%MVA can lead to muscle fatigue and pain (Björkstén and Jonsson, 1977; Caldwell and Smith, 1966; Jonsson, 1978). Therefore guided by the theoretical estimates, the present study was designed to address the spine stabilizing function of LSOs in postural tasks. The purpose of this study was to test experimentally the theoretical predictions of muscle activity reduction due to LSO in an unstable sitting task. The seated postural task was selected, because it eliminates ankle and knee control strategies, leaving only the hip and lumbar spine to control balance. Thus, neuromuscular control of this postural task was reflected in the EMG activity of trunk muscles.

2. Methods

2.1. Procedures

Healthy subjects (12 males and 11 females, see Table 1) volunteered for this study, read and signed a consent form describing the protocol approved by Yale University Human Investigation Committee. After appropriate skin preparation, Ag–AgCl, disposable surface EMG electrodes were placed over the following muscles on both sides of the body: rectus abdominis (RA, 3 cm lateral to the umbilicus), external oblique (EO, medial to the mid auxiliary line at the level of the umbilicus), thoracic erector spinae (TES, 5 cm lateral to T9 spinous process), and lumbar erector spinae (LES, 3 cm lateral to L4 spinous process). Each pair of electrodes was spaced 3 cm center-to-center along the direction of the muscle fibers. A reference electrode was placed laterally over the 10th rib on the right side.

After verifying the quality of EMG signals on an oscilloscope, subjects performed maximum isometric exertions in trunk flexion, extension, and lateral bending on an examination table against the resistance provided manually by one of the investigators. These tasks were designed to elicit MVA levels from trunk muscles, for the purpose of EMG normalization. Next, subjects sat on a seat supported by a plastic hemisphere ($\varnothing$ = 30 cm) and placed on a force plate (Kistler Model 9286AA, Germany) that tracked the center of pressure (CoP). The seat was equipped with leg and foot supports to prevent any lower body movement (Fig. 1). A safety railing around the force plate provided security in case of loss of balance. Each subject was instructed to maintain his/her balance while sitting on the seat with arms crossed. Following a 1-min practice, subjects were asked to balance for three, 20-s trials with and without the LSO (QuikDraw PRO, Aspen Medical Products Inc., Irvine, CA).

Trials were randomized, so that half of the subjects performed the 3 trials with the LSO first and the other half performed the trials without the LSO (No LSO) first. The LSO tension was adjusted to reach 35 mm Hg (4.7 kPa) pressure between the LSO and abdomen laterally to the umbilicus. The pressure was measured with the Therapoint measurement system (Roho, Inc, Belleville, IL). Data collection and timing were initiated after the subject achieved a steady balancing state. Between each trial, a 30 s rest break was given, during which subjects were asked to hold on to the safety railing to prevent any additional practice.

2.2. Data processing and analysis

All EMG signals were band-pass filtered between 20 and 420 Hz, differentially amplified (input impedance = 100 GΩ, CMRR > 140 dB)

Table 1
Average age, weight and height of the subjects participating in the study

Gender	Age (years)	Body mass (kg)	Height (cm)
Males	27.9 (9.8)	77.0 (4.0)	179 (8)
Females	22.1 (2.1)	55.6 (4.3)	166 (7)

Standard deviations are in parentheses.

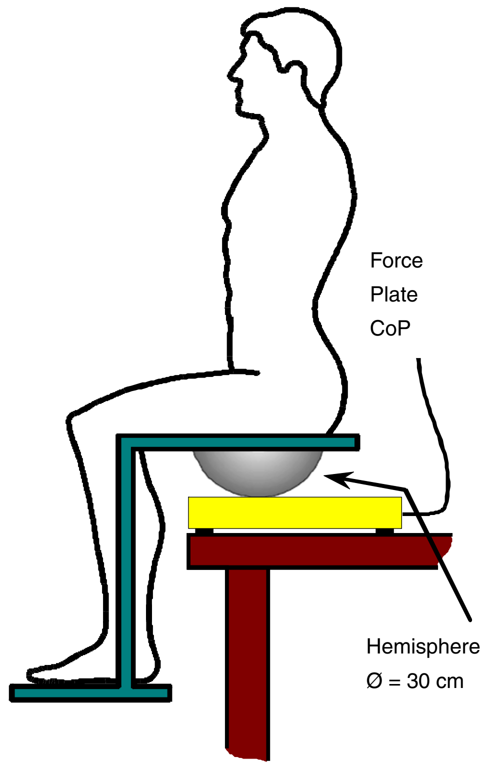

Fig. 1. A subject sitting on the unstable seat apparatus. A force plate recorded the movement of the center of pressure (CoP).

and A/D converted at a sample rate of 1600 Hz. Stored EMG data were digitally rectified, averaged over the entire 20-s trial, and normalized to provide %MVA. Because there were no significant differences in EMG activity between the left and right muscles, the EMG signals were averaged across both sides. The average CoP velocity (total CoP length/trial duration [mm/s]) in the lateral, antero-posterior, and resultant directions quantified balance performance (Cholewicki et al., 2000).

Initially, repeated measures MANOVA was applied to test for any effects of LSO on the 4 trunk muscle EMG activities. After a significant effect was returned ($p<0.05$), paired t-test comparisons were performed to identify which muscles had activity levels different between the LSO and No LSO conditions. Paired t-tests were also used to compare balance performance between these two experimental conditions.

3. Results

Following the practice, all subjects were able to complete all trials successfully. There was no significant difference in balance performance with and without the LSO ($p = 0.09$, 0.21, and 0.13 for CoP velocity in the antero-posterior, lateral, and resultant directions, respectively) (Fig. 2). However, EMG activity, averaged across the trials, was significantly lower in the LSO as compared to the No LSO condition for the TES ($p = 0.02$) and LES ($p = 0.01$) muscles (Fig. 3). These differences were 0.7%MVA and 2.2%MVA, respectively (Table 2). No significant differences between the two conditions were present in the RA ($p = 0.70$) or EO ($p = 0.62$) muscle activity (Table 2, Fig. 3).

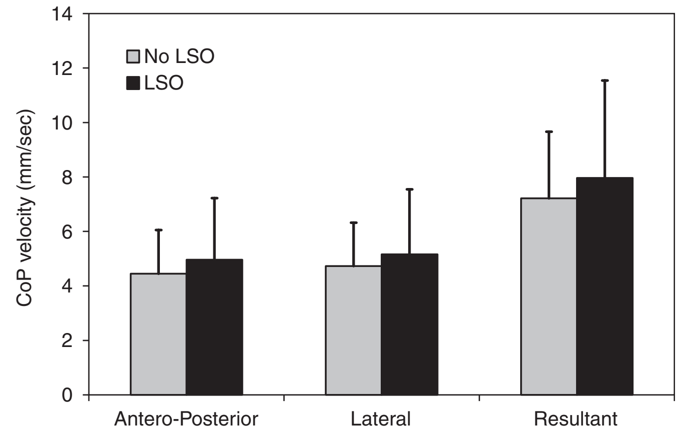

Fig. 2. Comparison of the center of pressure (CoP) velocity (CoP path/trial duration) between the LSO and No LSO conditions in the antero-posterior, lateral, and resultant directions ($p>0.05$). Error bars indicate standard deviations.

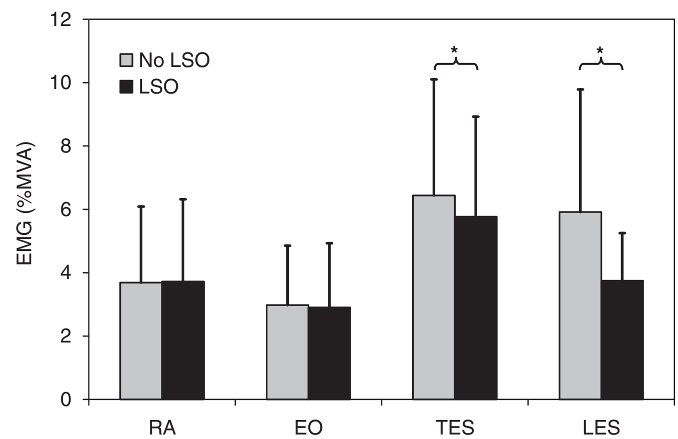

Fig. 3. Comparison of trunk muscle EMG activity between the LSO and No LSO conditions. Asterisks indicate significant differences ($p<0.05$). RA = rectus abdominis, EO = external oblique, TES = thoracic erector spinae, LES = lumbar erector spinae. Error bars indicate standard deviations.

4. Discussion

A significant reduction in the activity of back muscles due to LSO was documented during an unstable sitting task. These results agree with the spine and LSO modeling simulations, which predicted the greatest reduction in muscle activity to occur in the TES and LES across various tasks (Cholewicki, 2004). For unsupported sitting on a flat surface, a greater EMG decrease in the LSO condition was predicted across all muscles than was observed in our unstable sitting task (Table 2). Cholewicki (2004) stated that the theoretical values constituted the upper bound estimates and that much smaller effect of the LSO on trunk muscle activity should be expected in experimental settings. In order to obtain these upper bound estimates experimentally, full neuromuscular adaptation to the LSO would be necessary. It is likely that complete adaptation did not occur in the short acclimatization period of our

Table 2
Average EMG differences (standard deviations in parenthesis) between the LSO and No LSO conditions

	Rectus abdominis	External oblique	Thoracic erector spinae	Lumbar erector spinae
Current Study	0.0 (0.4)	0.1 (0.7)	0.7 (1.3)*	2.2 (3.4)*
Cholewicki (2004)	2.1	1.0	6.8	2.1

For comparison, the theoretically predicted EMG differences in unsupported sitting on a flat surface are presented (from Cholewicki, 2004). Units are %MVA.
*Indicates statistically significant differences ($p<0.05$).

experimental session. Furthermore, the LSO does not increase the stiffness at all intervertebral levels and degrees of freedom (Axelsson et al., 1992; Tuong et al., 1998) as was assumed in the Cholewicki (2004) study. Considering the above limitations, the comparison between the theoretical estimates and the current experimental results must be viewed as excellent.

It is very unlikely that our findings have been affected by systematic errors or bias. Trials were randomized, such that an equal number of subjects began their test with and without an LSO. The same subjects performed trials across both experimental conditions serving as their own controls. Furthermore, Jorgensen and Marras (2000) showed that wearing an abdominal belt did not affect EMG data recorded from the electrodes placed underneath the belt. If any changes of EMG signals were expected, they would be in the direction of increased signals with an LSO. The LSO presses the electrodes closer to the muscles and deeper into the skin, which should reduce impedance of the skin–electrode interface. Our findings revealed EMG changes in the opposite direction. Finally, although balance performance was not significantly different between the experimental conditions, there was slightly more CoP movement when the LSO was worn (Fig. 2). One could argue that more trunk movement, associated with increased CoP velocity, should produce more EMG activity. However, this was also not the case in our study.

Can the potential of LSOs to reduce the erector spinae activity by 1–2%MVA be perceived by the CNS as the "support" to the lumbar spine? The affirmative answer to this question is quite reasonable based on several previous studies. The most convincing experimental and modeling study showed that an addition of a 32 kg mass to the trunk required an increase in trunk muscle co-contraction of approximately 1–2%MVA above the 1–2%MVA level already necessary to maintain a stable upright position of the spine around the neutral posture (Cholewicki et al., 1997). Thus, such a small decrease in muscle activity might be perceived as a lessening of effort equivalent to removing a 32 kg mass off the trunk! Furthermore, static muscular contractions sustained over long periods of time, even at very low effort levels, can lead to fatigue and fatigue-related pain (Björkstén and Jonsson, 1977; Caldwell and Smith, 1966; Jonsson, 1978). From the Björkstén and Jonsson (1977) study, an approximate threshold of 5%MVA can be derived, below which muscle contraction could be sustained indefinitely. Muscular contractions above this threshold will eventually lead to muscle fatigue and pain (Fig. 4). Because postural, tonic trunk muscle activity must be maintained throughout most of the day, support from the LSO might be also perceived by the CNS. If this is the case, patients with LBP could benefit from wearing LSOs, which permit a slight reduction in trunk muscle co-contraction while maintaining spine stability. In turn, LSOs may prevent muscle fatigue and pain from compounding the existing pathology.

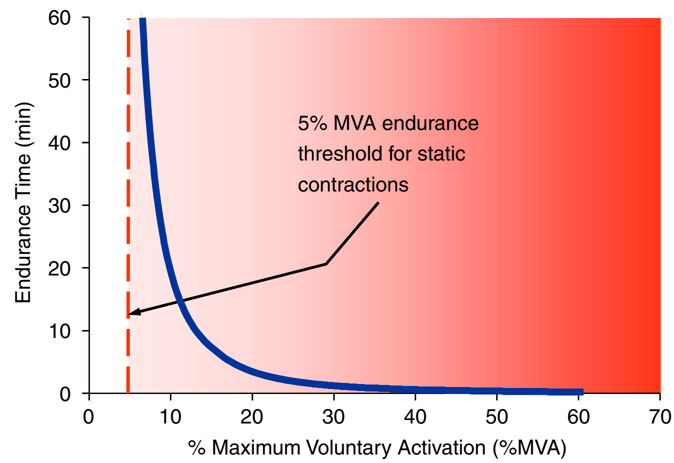

Fig. 4. Endurance times for sustained static muscular contractions. A threshold of 5% of the maximum voluntary activation (%MVA) can be derived, below which a muscle contraction can be sustained indefinitely. Adapted from (Björkstén and Jonsson, 1977).

While the supportive function of LSOs suggested in this study is reasonable for postural tasks, it is difficult to discuss this in the context of more strenuous activities, such as lifting. The 1–2%MVA reduction in the trunk muscle forces, when they are already at a 50% or 70% MVA level, does not appear functionally significant. It is possible that the benefit of LSOs is perceived during postural control when the LSO is initially worn and it becomes ineffective when more strenuous tasks are attempted. On the other hand, we did not measure activities of deep trunk muscles that were not accessible for the surface EMG recording. It is possible that these small muscles benefit to a proportionally greater extent from the LSO, in comparison to the large, superficial trunk muscles, even during more strenuous tasks. Generally, more EMG reduction would be expected from the muscles that are most active in a given task or posture (Cholewicki, 2004).

Finally, a question arises about the exact mechanism by which the CNS adapts to the increased trunk stiffness provided by an LSO. To address this question, balancing in a postural task must be viewed as a dynamic system, which consists of the body (including the spine) and the CNS as the controller. The CNS learns the dynamics of the system and then chooses the appropriate muscle recruitment strategy to execute the task while maintaining stability (Franklin et al., 2003). It appears that the CNS tunes stiffness to optimize system's performance (Franklin and Milner, 2003; Franklin et al., 2003; Reeves et al., 2006; Selen et al., 2006). It is possible that the CNS, sensing the added support from the LSO, reduces the level of tonic muscle activity to maintain optimal trunk stiffness to coincide with the learned dynamics. This scenario is consistent with our results, where a decrease of activity occurred in some of the superficial trunk muscles, but the balance performance in unstable sitting did not significantly change. Thus, the seated balancing task is an excellent experimental model illustrating the above dynamics, because the highly demanding balance control is superimposed on the need to maintain upright posture of the trunk and spine. Both tonic and phasic muscle activity takes place in such a task (Preuss et al., 2005) and it is likely that most of the EMG decrease due to the LSO came from the reduction in the tonic muscle activity.

If the CNS indeed adapts to the use of the LSO by reducing the active trunk stiffness, the long-term adaptation may become detrimental when the use of LSO is suddenly discontinued. In postural tasks, where muscle activity is already low, the risk of spine instability and injury will be magnified, if trunk stiffness remains reduced without the aid of an LSO. This mechanism could have been the cause of increased injury rates among the group of airline baggage handlers who stopped wearing belts in the Reddell et al. (1992) study. Therefore, the benefits and liabilities of LSO must be better understood before rational recommendations for their use in patients with LBP can be made.

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