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The effects of lumbosacral orthoses on spine stability: What changes in EMG can be expected?

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Abstract

Antagonistic trunk muscle activity is normally required to stabilize the spine. A lumbosacral orthosis (LSO) might reduce the need for this antagonistic activity by providing passive stiffness to the trunk and increasing spine stability. The maximum reduction in trunk muscle EMG and in the resultant spine compression force due to the LSO was estimated using a biomechanical model. The lumbar spine stability was first quantified for the average trunk muscle EMG recorded from 11 male subjects performing various isometric trunk exertion tasks. Subsequently, the spine-stiffening effects of the LSO were implemented in the model and trunk muscle forces were reduced iteratively until the original level of spine stability without the LSO was achieved. The upper bound estimates of the reduction in trunk muscle EMG due to LSO ranged from 0.6% to 14.1% of the maximum voluntary activation depending on the task and the muscle. The resultant spine compression force averaged across all tasks decreased by only 355 N. A much larger variance of the experimental data precluded the detection of these effects at statistically significant levels. However, the small effects size does not necessarily exclude the possibility of functional benefits of slightly reducing muscle activity in patients with low back pain.

Keywords: Spine stability; Lumbar spine; Lumbar orthoses; Low back pain; Muscle activity

Introduction

The use of lumbosacral orthoses (LSO) in conservative and postoperative treatments of low back pain continues despite the lack of scientific explanation for their function. The LSOs possess limited ability to reduce intervertebral motions [3,30] and their spine unloading mechanism could not be demonstrated [29,32]. Other hypotheses, such as enhanced spine proprioception, are still uncertain [27]. Nevertheless, a high satisfaction rate among patients with low back pain reinforces the use of lumbar orthoses [1,2,19]. In a much similar way, abdominal belts are used in industry as either protective or assistive devices in manual load handling activities without convincing justification from the scientific literature [26].

The most common rationale for the use of LSOs or belts is based on the premise that they directly, or via the intra-abdominal pressure mechanism, provide some assistance to the trunk extensor moment, reducing the activation of erector spinae muscles and consequently the magnitude of the spinal compression forces. Therefore, numerous studies have examined the differences in trunk muscle EMG in an attempt to demonstrate this spine unloading function of braces. Although both positive and negative results can be found [21–23], collectively the scientific literature does not support the hypothesis that orthoses reduce muscle forces [26,32].

Based on similar assertions, spine compression forces during lifting tasks were also estimated. While lumbar belts modify trunk kinematics, which may result in lower spinal loads [33], the reduction in spine compression force has not been observed when the task kinematics were controlled [24,29]. Postural shrinkage, as another indicator of the cumulative mechanical stress placed on the spine during physical activity, has also been compared between the belt and no belt conditions with inconsistent results [23,28].

Previous studies that have investigated changes in trunk muscle EMG or spine compression forces have

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approached this problem from a mechanical equilibrium point of view. That is, the reduction in muscle activity would reflect the ability of the brace to support a portion of trunk moment that must be exerted to perform a given task. A brace would need to provide relatively large moments to be effective in practical terms. However, apart from the end-range of motion, the LSO resists very little moment near the neutral spine posture [6]. Not surprisingly then, significant reduction in trunk muscle EMG could not be demonstrated.

One consistent research finding is that abdominal belts and orthoses passively increase trunk stiffness [7,22, 25] and enhance stability of the lumbar spine [18]. Therefore, the function of LSOs could be viewed differently from the perspective of spinal stability [7,10]. A certain amount of trunk antagonistic muscle co-contraction is necessary to provide structural stiffness to the lumbar spine and to maintain its stability [13,14,16]. On average, as little as 2% of the maximum voluntary activation (% MVA) from all trunk muscles may be sufficient to stabilize the spine in a neutral, upright standing position [13]. A small amount of additional trunk stiffness provided by the LSO significantly increases stability of the spine [7,10] and could in effect reduce the overall demand on antagonistic muscle co-contraction. Therefore, the purpose of this study was to theoretically estimate the reduction in trunk muscle activation that could occur when additional spine stability is derived from the LSO and less antagonistic muscle co-contraction is necessary to stabilize the spine. It was hypothesized that this reduction would be small relative to the intra-individual variability in muscle activation, which would explain mostly non-significant results obtained from past EMG and spine loading studies. However, knowledge of the magnitude of this possible effect could be helpful in designing future studies examining the proposed mechanism of LSO function.

Methods

The model simulation procedures were designed to reflect the experimental study summarized below and described in detail elsewhere [12]. Briefly, 14 subjects without low back pain wore an LSO for a 3-week period and were tested at 0, 1, and 3 weeks. Among other tests, the subjects performed exertions in isometric trunk flexion, extension, lateral bending, and axial rotation at a specific target force (100 N for men and 70 N for women). Once the target force was reached, it was suddenly released causing large trunk perturbation. This quick force release procedure was used only for the calculation of trunk stiffness in the experiment [12] and was not related to the present study. The surface EMG was recorded for one second before and two seconds after the quick release, bilaterally, from rectus abdominus, external and internal oblique, latissimus dorsi, and thoracic and lumbar erector spinae muscles. However, only the EMG data recorded immediately before the force release were used as input to the biomechanical model in the present study. The model output was the estimate of spine compression forces and spinal stability in the form of a stability index (SI) [10]. Both of these values quantified the aggregate effects of trunk muscle activity prior to quick release. Computations of spine compression force and stability were also performed for trials in unsupported sitting and isometric lifting of 30% of body mass.

The simulation of the effects of LSO on trunk muscle EMG was performed in two steps. First, the additional spine stability provided by the LSO via passive trunk stiffening was estimated. Subsequently, the amount by which trunk muscle activation could be reduced to bring spine stability back to the level obtained without the LSO was computed.

Biomechanical model

A detailed description of the biomechanical model of the lumbar spine was published earlier [10]. The model consisted of five lumbar vertebrae between the rigid pelvis and ribcage and 90 muscle fascicles. Each intervertebral joint was represented by a lumped parameter, nonlinear disc, and ligament equivalent for stiffness about the three axes of rotation. Thus, the system had 18 degrees of freedom (6 joints × 3 rotational degrees of freedom for each).

Muscle forces were estimated based on 200 ms of EMG data taken from 12 trunk muscles immediately before the force release. The EMG data were rectified, averaged, and normalized to the maximum values obtained from maximum voluntary exertion trials. After accounting for the contributions of the passive tissues, the moments and forces necessary to balance the external loads and the upper body weight were distributed among all 90 muscle fascicles using an EMG-assisted optimization method [11]. The muscle forces and muscle stiffness were first estimated from EMG using a cross-bridge bond distribution moment model reflecting muscle contraction dynamics [9]. A quadratic optimization algorithm was subsequently applied to minimize the adjustments (muscle gains, g_i) of individual muscle forces estimated from EMG, while requiring that the moments about the three rotational axes of every intervertebral joint were balanced [8]:

$$\sum_{i=1}^{90} M_i(1-g_i)^2 = \min \qquad M_i = \sqrt{M_{x_i}^2 + M_{y_i}^2 + M_{z_i}^2} \tag{1}$$

subject to the following constraints:

$$\sum_{i=1}^{90} g_i M_{x_i} = M_x \quad \sum_{i=1}^{90} g_i M_{y_i} = M_y \quad \sum_{i=1}^{90} g_i M_{z_i} = M_z \quad g_i \geqslant 0 \tag{2}$$

where M_{x_i}, M_{y_i}, M_{z_i} represent the moments estimated from EMG, which the "i"th muscle produces about the three joint axes of rotation of a given joint, and M_x, M_y, M_z represent the total muscle moments necessary for equilibrium at that joint. After the gain adjustment, all muscle forces and external loads were then summed along the axis perpendicular to the L4–L5 intervertebral disc to obtain a joint compression force.

The stability of the lumbar spine was quantified with the stability index (SI) representing the mean curvature of the system's potential energy in the vicinity of static equilibrium [10]. Thereby, the ability of the system to return to its original position upon any arbitrary perturbation from its equilibrium was explored using the minimum potential energy principle. Formulated mathematically, the potential energy of a system must be at a local minimum to ensure stability. Upon mathematical perturbation applied to each of the 18 generalized coordinates (Euler angles), the change in potential energy of the spine system was computed as the sum of the elastic energy stored in both the linear springs (representing muscles and tendons) and the rotational springs (representing intervertebral discs, ligaments, and other passive tissues), minus the work performed by the external load. The 18th root of the determinant of the Hessian matrix, formed from the second partial derivatives of the potential energy with respect to each generalized coordinate, constituted the SI, [10]. The SI quantified with the average curvature of potential energy, is equivalent to the SI quantified with the lowest eigenvalue as a relative measure of spine stability [17]. In either case, the assumption must be made that the motor control system senses the overall spine stability and adjusts the level of trunk muscle co-contraction accordingly.

Simulation procedures

For the simulation of LSO effects, a generic EMG input was created first by averaging the EMG data for all males ($n = 11$). Only males were simulated in this study because the females had a lower isometric target force and the model's anatomy was based on a 50th percentile male. The L4–L5 joint compression force and spine stability

Table 1
The effects of LSO on lumbar range of motion (ROM) and trunk stiffness simulated in the biomechanical model

Motion	Modified coefficients "a" for the LSO case	ROM with no LSO (deg)	ROM with LSO (deg)	Increase in stiffness at 10° (N m/rad)	Increase in stiffness at 15° (N m/rad)
Axial rot.	6.8	44.4	25.7	33.9	58.5
Extension	7.6	39.5	22.9	41.0	73.3
Flexion	0.8	72.0	41.8	9.0	19.7
Lat. bend	4.8	35.0	20.3	45.2	99.3
Average	–	47.7	27.7	32.3	62.7

were calculated for each loading condition in the "No LSO" case. Next, the trunk stiffening effects of LSO were modeled with the increased stiffness of the rotational springs representing passive properties of the lumbar intervertebral joints. Because the model contained only rotational degrees of freedom, the actual placement of the springs representing the effects of LSO was irrelevant as long as the increased trunk stiffness was equivalent to the additional stiffness gained from the LSO. Based on our previous work with the same LSO (Aspen Medical Products, Inc., Long Beach, CA, USA), on average it restricted 42% of the lumbar range of motion (ROM) [6]. This ROM was measured at the limit of trunk motion that the subjects could reach with the same effort, which was verified by EMG [6]. Therefore, the coefficients "a" in the general exponential expression of passive intervertebral joint properties in the simulation model [10] were increased to reduce ROM by 42% (Eq. (3) and Table 1). In the model, the ROM was the limit of motion calculated under the moments of upper body weight:

$$M_p = a(e^{b\theta} - 1) \tag{3}$$

where M_p is a passive joint moment, a, b are coefficients given in [10], and θ is the intervertebral joint angle.

Consequently, the resultant trunk stiffness increased on average by 32 and 63 N m/rad calculated at 10° and 15° around the trunk neutral posture, respectively (Table 1). These values were very similar to previous estimates of LSO and belt effects under static conditions (34 and 46 N m/rad [6] and 18 and 49 N m/rad [25] calculated at 10 and 15 degrees). The calculations of spine stability were repeated with the trunk stiffening effects of LSO.

The simulations were then repeated iteratively, uniformly reducing the stiffness of all muscles at each step, until the original baseline stability index (no LSO condition) was obtained. Theoretically, the overall muscle forces could be reduced by the same amount without compromising spine stability, because muscle stiffness is proportional to muscle force [9]. Because the pattern of muscle response to the increase in trunk stiffness is unknown, the uniform reduction of all muscle forces was a reasonable solution. These reduced muscle forces had to be re-adjusted again with the EMG-assisted optimization procedure to satisfy the static equilibrium constraints (Eq. (1)). Finally, the gains obtained with and without the LSO for the superficial trunk muscles, corresponding to the EMG recording sites, were applied to the EMG data. Thus, the possible differences in the trunk muscle EMG activity due to the LSO and the resultant changes in spine compression forces were computed.

Results

The simulated reduction in trunk muscle EMG due to wearing the LSO ranged from 0.6 to 14.1% MVA of maximum voluntary exertion depending on the task and the muscle (Table 2). The largest decrease in EMG occurred in the isometric trunk extension and axial rotation (by 4.9 and 4.2% MVA, respectively). The smallest decrease in EMG was predicted in the sitting task (by 2.4% MVA). The simulation of LSO had the largest effect on the thoracic erector spinae and latissimus dorsi muscles (decrease between 4.4 and 7.2% MVA averaged across all tasks). The smallest effect was seen in the internal oblique muscles (decrease of 1.7% MVA). On average, the overall muscle EMG was predicted to diminish by 3.8% MVA due to the trunk stiffening effects of LSO (Table 2).

The spine compression force, which resulted mainly from the activity of all 90 muscle fascicles represented in the model, decreased on average by 26.6% (355 N) due to the LSO (Fig. 1). Interestingly, the smallest effect was predicted for the lifting task (194 N reduction), which produced the highest spine compression force among all of the tasks.

The corresponding experimental data indicated only a 0.03% MVA reduction in EMG and 5.8% (99 N) reduction in spine compression force at the end of the third week of wearing the LSO [12]. Neither effect was statistically significant [12].

Discussion

The theoretical prediction of the changes in trunk muscle activity due to additional spine stability provided by the LSO suggests a very modest overall reduction. Because of the assumption that all of the muscles would uniformly diminish their activity, their combined predicted effects on spine compression force resulted in a more pronounced reduction of 26.6% (355 N). However, these differences most likely constitute only the upper bound estimates. First, full adaptation to the enhanced spine stability was assumed. In reality, the motor control system may not fully appreciate additional spine stability provided by the LSO, it may take much longer to adapt to it, or the full adaptation might never occur. Second, the stiffening effects of the LSO may not affect all of the lumbar levels and/or degrees of freedom, as was assumed in our study. In fact, radiological studies suggest that orthoses are most effective in limiting the intervertebral rotations in the upper lumbar segments (L1–L3), and they may actually increase the intervertebral rotations at the lower lumbar levels [30]. The intervertebral translations are not reduced at all [3]. Such an incomplete restriction of motions within the

Table 2
The simulated effects of LSO on trunk muscle EMG expressed in % of the maximum voluntary activation

	RRA[a]	REO	RIO	RLD	RTE	RLE	LRA	LEO	LIO	LLD	LTE	LLE	Average
No LSO													
Axial rot.	2.1	2.9	10.6	10.8	18.5	6.5	4.3	10.2	0.6	5.1	4.9	4.7	6.8
Extension	2.4	1.7	2.0	21.4	17.0	3.5	4.2	1.7	2.4	13.6	15.7	6.4	7.7
Flexion	9.0	9.1	2.6	2.7	5.3	2.2	9.0	8.4	2.8	4.3	6.6	1.6	5.3
Lateral bend	2.3	4.1	0.9	4.6	5.6	3.2	4.7	7.8	7.6	6.2	12.3	8.0	5.6
Lift	0.7	2.0	1.6	10.2	25.5	36.2	1.6	1.6	1.6	9.7	27.1	37.7	13.0
Sit	1.2	1.1	1.2	1.5	7.6	2.4	3.0	1.0	1.2	1.9	8.5	2.0	2.7
Average	2.9	3.5	3.2	8.6	13.3	9.0	4.5	5.1	2.7	6.8	12.5	10.1	6.8
LSO													
Axial rot.	0.0	0.0	6.9	5.5	7.3	2.0	0.7	5.9	0.0	0.6	1.5	0.3	2.6
Extension	0.3	0.4	0.4	7.3	7.0	1.8	0.5	0.4	0.5	5.0	6.6	3.3	2.8
Flexion	4.5	2.6	0.8	0.3	0.7	0.0	4.6	2.2	0.9	0.6	0.9	0.0	1.5
Lateral bend	0.0	0.6	0.0	0.9	1.6	0.2	1.4	3.2	4.3	2.1	4.6	3.2	1.8
Lift	0.0	0.4	0.4	6.4	18.8	30.2	0.0	0.3	0.3	6.1	20.0	31.7	9.6
Sit	0.0	0.1	0.0	0.2	1.2	0.2	0.0	0.1	0.0	0.2	1.4	0.1	0.3
Average	0.8	0.7	1.4	3.4	6.1	5.7	1.2	2.0	1.0	2.4	5.8	6.4	3.1
Difference													
Axial rot.	−2.1	−2.9	−3.7	−5.4	−11.2	−4.5	−3.6	−4.3	−0.6	−4.5	−3.4	−4.4	−4.2
Extension	−2.1	−1.3	−1.6	−14.1	−10.0	−1.7	−3.7	−1.3	−1.9	−8.5	−9.1	−3.2	−4.9
Flexion	−4.5	−6.5	−1.8	−2.5	−4.6	−2.2	−4.4	−6.2	−1.9	−3.7	−5.7	−1.6	−3.8
Lateral bend	−2.3	−3.5	−0.9	−3.7	−4.0	−3.0	−3.4	−4.6	−3.3	−4.0	−7.7	−4.8	−3.8
Lift	−0.7	−1.5	−1.3	−3.8	−6.7	−6.0	−1.6	−1.3	−1.3	−3.6	−7.1	−6.0	−3.4
Sit	−1.2	−1.0	−1.2	−1.4	−6.4	−2.2	−3.0	−0.9	−1.2	−1.7	−7.1	−1.9	−2.4
Average	−2.1	−2.8	−1.8	−5.1	−7.2	−3.3	−3.3	−3.1	−1.7	−4.4	−6.7	−3.6	−3.8

[a] The first letter in a muscle symbol stands for right (R) or left (L) side; the next two letters identify the muscle: RA, rectus abdominus; EO, external oblique; IO, internal oblique; LD, latissimus dorsi; LE, lumbar erector spinae; TE, thoracic erector spinae. A negative difference indicates a reduction in EMG.

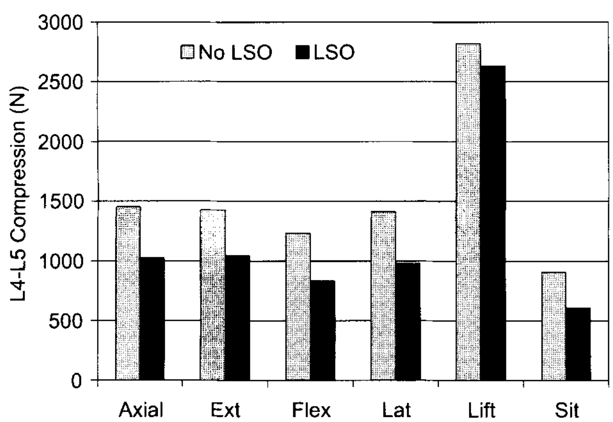

Fig. 1. The simulated effects of LSO on the spine compression force during various isometric trunk exertions: Axial = axial rotation, Ext = extension, Flex = flexion, Lat = lateral bending, Lift = lifting, Sit = unsupported sitting.

lumbar spine would lessen the actual amount of spine stability provided by the orthoses as compared to our estimates. Finally, the intervertebral joint stiffening effect from the spine compression forces due to muscle contraction [15] was not incorporated in the model. Such an effect would decrease spine stability more rapidly due to reduced muscle forces. Therefore, in experimental settings, much smaller effects of the LSO on trunk muscle activity should be expected than those estimated in the present study.

Even if all of the predicted differences were realized experimentally, the overall standard deviation of 8.0% MVA in the EMG results obtained from such an experiment [12] would preclude statistical significance of the estimated 3.8% MVA effects. Perhaps only the differences in the thoracic erector spinae and latissimus dorsi muscles in the isometric trunk extension trials (8.5–14.1% MVA Table 2) would emerge as significant. Interestingly, several studies reported perhaps an incidental decrease of activity in a few isolated muscles when some type of a lumbar orthosis was worn [7,22].

The lack of statistically significant reduction in spine compression force frequently reported in the literature [26,32] is also consistent with the suggested mechanism of LSO function. Resulting mostly from the variance in muscle activity, the overall standard deviation of the experimentally obtained spine compression force was 547 N [12], while a reduction of only 355 N in the spine compression force was predicted in this study. Approximately 39 subjects would be necessary to detect this "best case scenario" effect with a statistical power of 0.8.

Because the actual differences are expected to be much smaller than these estimates, the lack of experimentally documented effects of orthoses on spine loading should therefore not be surprising.

The obvious question arises whether such small mechanical effects of the orthoses could have any practical significance in the lumbar spine function. If we consider that only 1–2% MVA of co-contraction is required from all trunk muscles to maintain the spine in a stable upright position [13], the estimated changes in muscle activity become quite significant by comparison. For example, in the unsupported sitting task, our study predicted a change from 2.7 to 0.3% MVA in the overall trunk muscle activity due to the LSO—nearly an 89% reduction. In practical terms, such a reduction in muscle forces might be perceived as the equivalent of removing a 32 kg mass from the upper body. Cholewicki et al. [13] demonstrated theoretically and experimentally that an addition of 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 without any additional loads. Therefore, such small changes in trunk muscle activity may have relatively large practical implications for the lumbar spine function.

A small reduction in trunk muscle activity afforded by the LSO may be even more functionally significant in the population of patients suffering from low back pain. These patients exhibit an increased muscle co-contraction during their activities of daily living presumably to enhance the stability of their lumbar spines [31]. Because spine stability must be maintained at all times throughout the day during walking, standing, and even unsupported sitting, the elevated muscle co-contraction creates a significant physiological burden. From the early ergonomic studies, we know that static muscle contractions sustained above 5% MVA will lead to muscle fatigue and pain [4,5,20]. Contractions below 5% MVA could be maintained indefinitely [4,20]. A small reduction in trunk muscle activity may be sufficient to bring it below the 5% MVA threshold. Therefore, the LSO may provide significant symptomatic relief to some low back pain patients by reducing the necessary static trunk muscle co-contraction and preventing muscle fatigue and pain from compounding the existing causes of the pain.

The rationale for this study was based on the assumption that in a given task, the motor control system seeks a certain level of spine stability via muscle co-contraction, which could be reduced if the additional stability is provided by the LSO. Because the increase in passive trunk stiffness and spine stability is the most consistently reported finding in literature [7,18,22,25], this rationale seems well justified. However, other physiological and mechanical effects of orthoses might be possible, but were not considered in this study.

The EMG data used for the simulations came from the subjects who anticipated trunk perturbation and might have increased their level of antagonistic muscle co-contraction. However, such anticipation was present during the tests with and without the LSO and is not likely to cause systematic errors in the present study. There are also a number of anatomical, physiological, and mathematical assumptions specific to the biomechanical model itself. They are too numerous to be listed here, but the readers are referred to the original publication for an in-depth discussion of this model's limitations [10]. While the verification of this type of a model is impossible, its internal validity has been confirmed in several previous studies [10,18,31]. Furthermore, the numerical results of the present study need not to be taken literally. Instead, they were meant to guide the design of future research into the function of lumbosacral orthoses. In this context, our results suggest that the methodology should be adjusted and the measurement tools recalibrated to meet much lower effect expectations when testing the hypotheses raised in this discussion.

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