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Response of trunk muscle coactivation to changes in spinal stability

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Abstract

The goal of this effort was to assess the neuromuscular response to changes in spinal stability. Biomechanical models suggest that antagonistic co-contraction may be related to stability constraints during lifting exertions. A two-dimensional biomechanical model of spinal equilibrium and stability was developed to predict trunk muscle co-contraction as a function of lifting height and external load. The model predicted antagonistic co-contraction must increase with potential energy of the system even when the external moment was maintained at a constant value. Predicted trends were compared with measured electromyographic (EMG) data recorded during static trunk extension exertions wherein subjects held weighted barbells at specific horizontal and vertical locations relative to the lumbo-sacral spine junction. The task was designed to assure the applied moment was identical during each height condition, thereby changing potential energy without influencing moment. Measured EMG activity in the trunk flexors increased with height of the external load as predicted by the model. Gender difference in spinal stability were also noted. Results empirically demonstrate that the neuromuscular system responds to changes in spinal stability and provide insight into the recruitment of trunk muscle activity. © 2001 Elsevier Science Ltd. All rights reserved.

Keywords: Low-back; Spine; Stability; Muscle; Co-contraction; Model

1. Introduction

Neuromuscular control of spinal stability may play a significant role in the etiology and prevention of low-back pain. Lifting guidelines suggests spinal compressive loads below 3400 N may be considered safe for a majority of the working age population (Konz, 1982; Waters et al., 1993) but epidemiologic reports demonstrate high incidence and risk of injury even at low spinal loads (Granata et al., 1996; Punnet et al., 1991). The lumbar spine without muscle support becomes unstable and may suffer strain injuries at compressive loads as low as 88 N (Crisco and Panjabi, 1992). Clearly, a spinal stability limit of 88 N is substantially less than the spinal compression experienced during most lifting tasks. To maintain stability in realistic conditions muscle activation must be recruited to stiffen and stabilize the spine (Bergmark, 1989). Mechanical stiffness of active muscle tissue increases with myoelectric excitation and contractile force (Morgan, 1977). Consequently, antagonistic muscle co-contraction is known to increase the equivalent stiffness of a joint (Hogan, 1980). Thus, it has been assumed that antagonistic activity during lifting exertions is attributable to spinal stability requirements (Zetterberg et al., 1987). Unfortunately, this does not prove that antagonistic activity is motivated by stability.

To date there is no direct evidence to suggest that muscle recruitment changes in response to spinal stability requirements. Recent analyses predicted that antagonistic muscle excitation must exist to simultaneously maintain lifting moment and stability (Cholewicki et al., 1998). However, others have shown that antagonistic co-contraction is highly correlated with trunk moment (Mirka and Marras, 1993). It remains unknown whether co-contraction is related to trunk moment, spinal stability or both. Potential energy of the system influences stability requirements and can be controlled independent of trunk moment to assess the relation between stability and antagonistic co-activity. Recent analyses indicate potential gender differences in

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active muscle stiffness that may influence spinal stability and associated coactive muscle recruitment (Granata et al., 2001a).

The goal of this study was to assess the neuromuscular response to changes in spinal stability during lifting exertions. It was hypothesized that antagonistic co-activation in the trunk flexors must increase with potential energy of the system at constant levels of external trunk flexion moment. Furthermore, it was hypothesized that gender differences must be observed in the coactive response to spinal stability.

2. Methods

A two-dimensional, inverted-pendulum model was developed to predict trends in antagonistic muscle activity from equilibrium and stability constraints similar to the analyses by Cholewicki (Cholewicki et al., 1998). The lumbar column was represented as a massless, rigid rod. A point-mass represented the weight of the head, arms and trunk, W, was centered on the spine at an elevation h_W above the lumbo-sacral (L5-S1) joint (Fig. 1). Flexor and extensor muscle groups were represented as force and stiffness generating elements

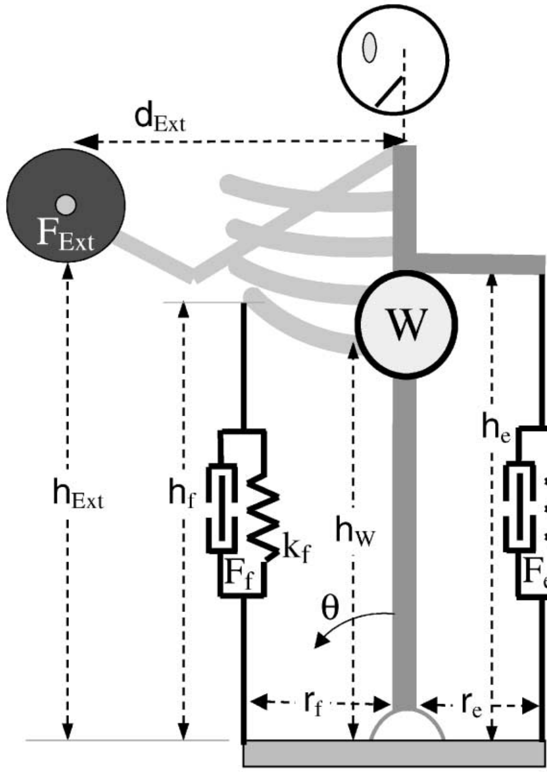

Fig. 1. Biomechanical model. Potential energy, static equilibrium and stability were determined from a two-dimensional representation of the trunk. Both the flexor and extensor muscles were capable of generated contractile force and stiffness.

with moment arms of r_f and r_e with upright posture lengths of h_f and h_e, respectively. A vertical external load, F_{Ext} carried in the hands acted at a constant horizontal distance, d_{Ext} anterior to L5-S1 and at a height of h_{Ext} above the lumbo-sacral junction. Potential energy of the system, V, was approximated as

$$\begin{aligned} V = {} & Wh_W \cos\theta + F_{Ext}[h_{Ext}\cos\theta - d_{Ext}\sin\theta] \\ & + F_f(h_f - r_f\sin\theta) + F_e(h_e + r_e\sin\theta) \\ & + 1/2\,k_f[(h_f - r_f\sin\theta) - l_0]^2 + \\ & 1/2\,k_e[(h_e + r_e\sin\theta) - l_0]^2, \end{aligned} \tag{1}$$

where W represented the weight of the trunk with center-of-mass at height h_W. F_f and F_e represented the contractile force while k_f and k_e were the stiffness in the flexor and extensor muscles, respectively. The expressions in parentheses were the muscle lengths at a flexion angle θ from the vertical equilibrium position with l_0 the equilibrium length of the elastic terms.

Static equilibrium was satisfied by the homogenous first derivative of potential energy with respect to trunk angle,

$$\frac{\delta V}{\delta\theta} = \sum M = F_e r_e - F_f r_f - F_{Ext} d_{Ext} = 0, \tag{2}$$

where the upright posture was assumed and the elastic rest length, l_0, was represented as the equilibrium muscle lengths. Stability measures the ability of the system to return to its original equilibrium energy state after a perturbation, thereby requiring the second derivative of potential energy to be greater than zero (Thompson and Hunt, 1984). Thus, stability in the upright posture is satisfied when

$$\frac{\delta^2 V}{\delta\theta^2} = -(W_t h_w + F_{Ext} h_{Ext}) + k_f r_f^2 + k_e r_e^2 = S, \tag{3}$$

where S is a general stability parameter greater than zero. Research has demonstrated that active muscle stiffness is linearly proportional to contractile force (Weiss et al., 1988),

$$k_f = q_f F_f / h_f, \quad k_e = q_e F_e / h_e, \tag{4}$$

where q_e and q_f represent the stiffness gradient with respect to muscle force described by Bergmark (Bergmark, 1989). Force in the antagonists or flexor muscles during a trunk extension task is determined through simultaneous solution of the equilibrium, stability and muscle stiffness relations.

$$F_f = \{F_{Ext}(h_{Ext} - d_{Ext} q_e r_e / h_e) + Wh_W + S\}/c_1. \tag{5}$$

Coefficient c_1 is a constant involving muscle moment arms, muscle lengths and stiffness gradient coefficients and is independent of muscle excitation levels.

This model (Eq. (5)) suggests that during extension exertions antagonistic activity of the flexors must increase with the height or potential energy of the external load to maintain stability. The model also predicts an interaction between external load, F_{Ext}, and external load height, h_{Ext}, i.e. the height of the external load will dictate the strength of the relationship between F_{Ext} and antagonistic coactivity. Low heights h_{Ext} cause the parenthetical term in Eq. (5) to approach zero and the influence of F_{Ext} on antagonistic activity is small. Conversely, at high heights h_{Ext} the parenthetical term becomes largely positive and the influence of F_{Ext} on flexor activity is large. Research indicates the stiffness gradient, q, in female subjects may be less than in males (Granata et al., 2001a). If true, then the model (Eq. (5)) predicts this gender difference in q requires increased flexor activation in women with a gender difference that increases with external load height.

To illustrate these recruitment behaviors a 50th percentile North American male was modeled (Cholewicki et al., 1998) and predicted activity displayed in Fig. 2, i.e. trunk weight $W = 52$ kg at an elevation $h_W = 20$ cm with flexor and extensor moment arm lengths $r_f = 8.5$ and $r_e = 6.0$ cm, respectively, and muscle stiffness gradient of $q_e = q_f = 5$ according to analyses by Gardner-Morse et al. (Gardner-Morse et al., 1995). For simplicity, values of $h_e = h_f = h_w$ and $S = 0$. However, the specific coefficient values affect only minor changes in relative slope and intercept whereas predicted trends are independent of reasonable coefficient values, e.g. antagonistic activity of the flexors must increase with the height of the external load regardless of the coefficient values employed by the model.

Antagonistic muscle activity during static lifting exertions was measured to validate the predicted muscle recruitment trends. Ten men and ten women with no prior history of low-back disorders and no history of cardiovascular conditions volunteered to participate (Table 1). All subjects signed an informed consent form approved by the University of Virginia, Human Investigations Committee. Subjects were required to hold a weighted barbell between two vertical guide-bars at randomly ordered heights of $h_{Ext} = 0, 20, 40, 60$ and 80 cm. above the sacrum (Fig. 3). Each subject performed the exertions at two weight levels, 4.5 and 9.0 kg. Two trials were performed at each height and weight condition. The guide-bars assured the horizontal moment arm distance was constant at $d_{Ext} = 30$ cm thereby generating external moments of 13.2 and 26.5 Nm during all height conditions.

EMG signals were collected from bipolar surface electrodes (Medicotest, Rolling Meadows, IL) over four bilateral sets of trunk muscles. Electrodes were placed according to Mirka (Mirka, 1991): for the rectus abdominis, 3 cm lateral and 2 cm superior to the umbilicus; external oblique 10 cm lateral to the umbilicus with an orientation of 45° to vertical; lateral

Table 1
Subject Demographics

	Male	Female
# of Subjects	10	10
Age (yr)	23.89 ± 1.45	26.32 ± 4.60
Height (m)[a]	1.80 ± 0.10	1.68 ± 0.05
Weight (kg)[a]	80.69 ± 7.17	67.75 ± 11.26

[a] Indicates significant differences between males and females with $p < 0.01$

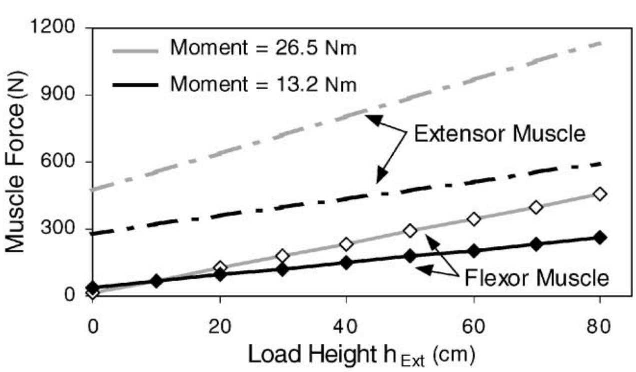

Fig. 2. Theoretical flexor co-contraction must increase with height of the external load above L5-S1. The elevation of the load influences potential energy of the net biomechanical system, requiring modified flexor activity to maintain spinal stability. Predicted extensor activity must increase with applied external flexion load and to offset flexor muscle moment.

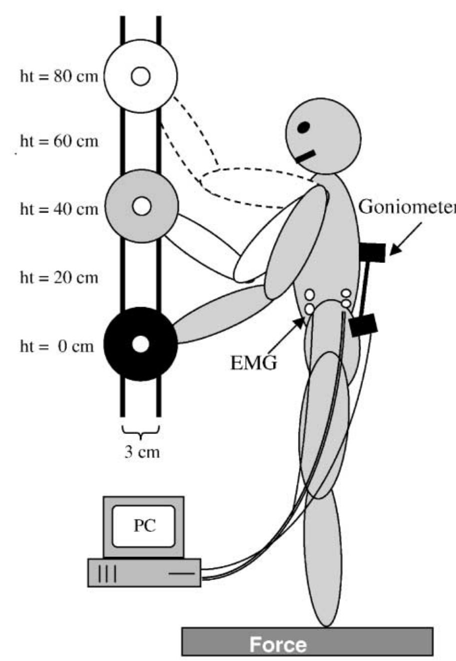

Fig. 3. Experimental configuration. Potential energy and external trunk moments were independently controlled. Subjects were required to hold a weighed barbell at specified heights while maintaining the effective moment arm distance at a constant value. EMG was recorded and spinal posture controlled during the static exertions. N.B. five height conditions were measured but only three are illustrated for graphic clarity.

extensors 8 cm lateral to the midline within the lumbar triangle at a 45° orientation; and erector spinae 4 cm lateral to the L3 spinous process. Data were sampled at 1000 Hz, band-pass filtered between 30 and 250 Hz, rectified, then smoothed using a 5 Hz Hanning low-pass filter. Mean integrated EMG amplitudes were determined from the middle 3 s of the 5 s static trials. Activity from the isometric mean values of the recti abdominis and external obliques were averaged to represent an equivalent trunk flexor muscle and activity from the erector spinae and lateral extensors were averaged to represent an equivalent trunk extensor muscle.

Spinal posture was monitored by an electrogoniometer (Penny and Giles®, Biometrics Ltd., UK) placed over the spinous processes of the T10 and S1 vertebral levels. A neutral value was recorded while subjects stood in their normal comfortable posture. Subjects were instructed to maintain this posture as displayed on a video data screen during each task. This assured the trunk was upright and lumbar lordosis was consistent between conditions.

Average EMG activity levels were compared with the height of the external load. A significant positive slope, i.e. a significant ANOVA main effect for height, suggests the flexor muscle EMG increases with height, validating the primary hypothesis. Dependent variables of flexor EMG averages were examined using a repeated-measures ANOVA with independent variables of height, external load and between-subjects variable of gender. To assure group differences in variability did not influence ANOVA procedures, between-groups assessment with independent variances was performed in support of the statistical results. Results agreed identically with ANOVA findings and therefore will not be discussed further. Post-hoc analyses were performed to examine differences between levels of significant independent variables at significance of $\alpha = 0.05$.

3. Results

The biomechanical model predicted that muscle force in the abdominal and paraspinal muscles must increase with height of the external load. Results in Fig. 2 represent model output for a 50th percentile North American male (Cholewicki et al., 1998) and muscle stiffness gradient of $q_e = q_f = 5$ (Gardner-Morse et al., 1995). Contractile force in the muscles must increase with height despite the fact that external trunk flexion moment was the same at each height (Fig. 2). This trend of increasing co-activation with height must be observed regardless of the anthropometry applied to the model and is expected in large males and small females alike.

Abdominal activity must increase to augment trunk stiffness necessary to stabilize the system as potential energy increases. The model also predicts that extensor muscles concomitantly increase with height to offset the antagonistic flexion moment from abdominal muscle activity. The difference in force levels between the extensor and flexor muscles represents the internal extensor moment necessary to maintain equilibrium. Statistical analyses of myoelectric data demonstrated EMG levels increased with height of external load (Table 2, Fig. 4). Thus, when the external moment was held constant at 13.2 Nm measured flexor muscle activity increased significantly ($p<0.001$) with height of the barbell above L5-S1, i.e. antagonism increased with potential energy at constant moment. Similar results were observed with the 26.5 Nm moment condition. Post-hoc and statistical power analyses revealed the flexor muscle EMG in the two highest conditions were greater ($p<0.05$) than the lowest

Table 2
Statistical results of ANOVA[a]

	Flexor EMG	Extensor EMG
h_{Ext}	$p<$**0.000**	$p<$**0.001**
F_{Ext}	$p<$**0.001**	$p<$**0.001**
Gender	$p<$**0.037**	$p<0.182$
$h_{Ext} \times F_{Ext}$	$p<$**0.001**	$p<$**0.001**
h_{Ext} × Gender	$p<$**0.001**	$p<0.261$
F_{Ext} × Gender	$p<0.259$	$p<0.681$
$h_{Ext} \times F_{Ext}$ × Gender	$p<0.255$	$p<0.103$

[a] Bold values highlight statistical significance at $p<0.05$.

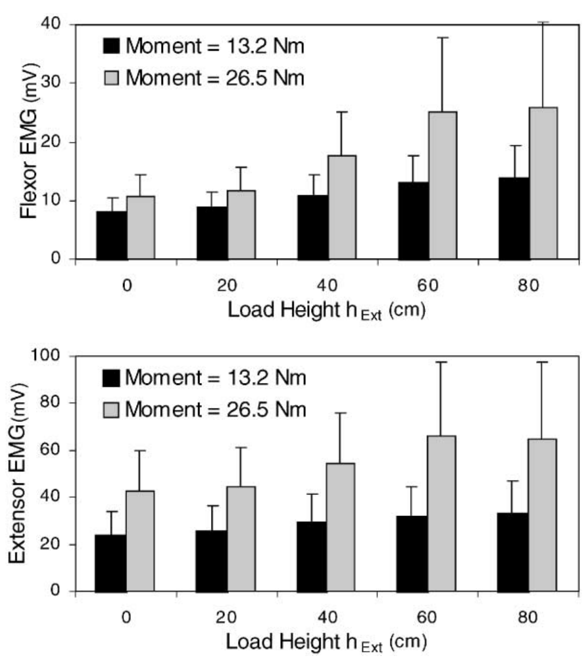

Fig. 4. EMG activity of the trunk flexors muscles and extensor muscles. Flexor co-activation was significantly influenced by height of the external load despite constant external flexion moment. Extensor activity necessarily increased with height to offset the flexor antagonism. Weight of the external load, i.e. external flexion moment also significantly influences flexor and extensor EMG activity.

conditions with statistical power greater than 0.8, i.e. $\beta = 0.2$. Thus, flexor co-activation was recruited to restore and maintain musculoskeletal stability of the trunk even though external trunk flexor moment remained unchanged with height.

Recognizing that potential energy and the need for increased biomechanical stability increases with both height and mass, it was not surprising that antagonistic activity also increased with added external load at each height (Table 2, Fig. 4). Weight significantly increased the flexor EMG levels at the three greatest heights ($p < 0.009$). This interaction between weight and height was predicted by the model.

A significant ($p < 0.037$) gender difference in flexor stabilizing activity was observed. Female subjects recruited 32% more flexor co-activation than males. A significant ($p < 0.001$) gender-by-height interaction supported the model prediction that the gender difference in flexor activity must increase with height. Post-hoc analyses of this interaction revealed that the gender difference was significant only at the two greatest barbell elevations. Further research is necessary to understand this gender effect.

4. Discussion

Neuromuscular control of stability is necessary to maintain equilibrium in the presence of potential kinetic or kinematic disturbances and to reduce the risk of musculoskeletal injury. Theoretical analyses indicate antagonistic co-contraction may be recruited to control trunk stiffness and spinal stability (Gardner-Morse and Stokes, 1998; Granata and Marras, 2000). However, predicting that antagonistic co-activity may contribute to stability does not prove that antagonistic co-activity is motivated by stability. Several studies have measured the recruitment response to simultaneous changes in equilibrium moment and stability (Cholewicki et al., 1998, 2000), but it was difficult to discern whether the co-contraction resulted from equilibrium or stability. To demonstrate that antagonistic co-activity is motivated by stability, it was necessary to observe the myoelectric response to changes in stability independent of equilibrium conditions.

Results from the current effort demonstrate antagonistic co-contraction in the flexor muscles of the trunk increased in response to greater need for biomechanical stability despite a constant trunk moment. In the experimental protocol, external flexion moment was nearly identical when holding the barbell at high elevations and low elevations alike. Based upon equilibrium analyses no change in EMG activation would have been be expected (Chaffin, 1969). However, stability required increased trunk stiffness at greater load heights. Recruiting trunk stiffness solely through increased extensor force would violate equilibrium constraints; so it was necessary to augment trunk stiffness through antagonistic co-contraction of the flexors and extensors. Because the external moment was constant at all heights, the changes in flexor co-activation must have been in response to changes in stability requirements.

Model predictions and experimental results agree with the analyses and measurements in the published literature (Cholewicki et al., 1998; Stokes and Gardner-Morse, 2000). In the study by Cholewicki et al. (Cholewicki et al., 1998) there was no attempt to empirically discriminate between the effects of trunk moment and stability. Nonetheless, results presented by the authors demonstrate increased levels of myoactivity in the upright posture when greater loads were borne on the shoulders (see Cholewicki et al., 1998, Figs. 4 and 6). In the upright posture with the load on the subjects' shoulders, trunk moment was approximately zero in both weight conditions while potential energy was proportional to the external load. Hence, their results support our hypothesis and results, i.e. trunk muscle co-contraction increases with potential energy of the net biomechanical system to maintain stability even when trunk moment remains unchanged.

Flexor co-activation was also associated with barbell weight. Extensor muscle recruitment was increased with weight in agreement with equilibrium models (Chaffin, 1969; Ladin et al., 1989), but the stability requirements dictate increased flexor antagonism as well. Since height and weight both contribute to potential energy and external stability, it is logical that both factors influence flexor co-activation. Eq. (5) illustrates that flexor muscle force must increase with greater F_{Ext} when the multiplying factor in parentheses is greater than zero. The model also predicted the influence of external load on flexor activity must be small at low heights h_{Ext}, i.e. term in parentheses of Eq. (5) is small. At large values of h_{Ext} a change in external load is amplified by the parenthetical relation. This prediction was validated by the measured data wherein a statistically significant interaction revealed increased antagonistic activity with the heavier load specifically at the highest elevations. If the experimental protocol had employed a larger external moment arm, d_{Ext}, or if the in vivo stiffness gradient were greater, one should expect the antagonistic activity to decrease with heavier external loads. This suggests a critical design factor when considering lifting task performance. Stability is improved with external load for lifts performed beyond the critical moment arm value, albeit at the cost of increased spinal load. Lifts performed at distances less than the critical moment arm will observe a loss in stability with increased external load, requiring increased antagonistic co-activation. This may provide insight into the safety of lifting task design.

A significant gender difference in flexor stabilizing activity was observed, particularly at the highest elevations, 60 and 80 cm. This may indicate the female subjects were disadvantaged by reduced external stability when performing similar tasks as the male subjects. This necessarily required greater muscle recruitment, effort and potential for fatigue in the female subjects for these specific lifting tasks. When normalized to the number of employees and work hours, epidemiologic research indicates females suffer more than twice the risk of musculoskeletal and low-back injuries than males (Feuerstein et al., 1997; Krause et al., 1997; Macfarlane et al., 1997). To improve gender inclusion in the workplace it is necessary to understand the cause of this gender bias. These results indicate stability may play a role in gender factors that contribute to injury risk and biomechanical analyses of stability may contribute to improved injury control.

Several factors may have contributed to gender difference in stability in our study. First, anthropometric differences suggest trunk weight, center-of-mass and muscle moment arms were different in the male and female subjects (Table 1). However, reduced body height and weight would predict reduced average flexor antagonism in the female participants. Conversely, reduced muscle moment arm versus muscle length, r_e/h_e, would predict increased co-contraction. Second, recent evidence (Granata et al., 2001b) indicates women may demonstrate reduced active muscle stiffness and a reduced stiffness-versus-force gradients requiring increased flexor co-contraction to maintain stability. Third, lumbar lordosis may have been significantly different in the male and female subjects and research suggests lordosis may influence spinal stability (Granata, 1998). However, caution is warranted when considering the gender difference in EMG data. The experimental protocol did not account for potential gender differences in muscle mass, skin impedance or maximum EMG potential. Conversely, reduced muscle mass in the extensor muscles must reduce flexor antagonism for equilibrium. The probability of a random assignment of 10 female subjects such that raw EMG levels were significantly greater than in a group of 10 male subjects is exceedingly small. Thus, attributing gender difference to EMG artifact is unlikely, but cannot be dismissed. Further research is necessary to understand potential gender differences in risk of low-back disorders and spinal stability. Although gender differences were observed, both males and females demonstrated the same statistically significant trend wherein flexor antagonistic co-activation increased with the elevation of the external load while trunk moment was held constant

Despite limitations in the model and experiment, trends in muscle excitation were accurately predicted. The model was limited to a two-dimensional analysis with a single equivalent extensor muscle and a single equivalent flexor muscle. Passive stiffness from muscles and vertebral motion segments were ignored as it contributes little to the net system behavior in the loaded upright posture (Cholewicki et al., 1998). The inverted pendulum representation of the spine prohibited assessment of spinal curvature and lordosis in these analyses. Instead, lordosis was controlled at a constant posture throughout the experimental conditions. The posture of the arms was different at each barbell height to assure identical trunk flexion moments at each height. It was assumed that arm posture and shoulder musculature did not impact the trunk flexor activity as no muscles are common to both region. Further investigations are warranted in this regard. Finally, it was assumed that the stability parameter, S in Eqs. (3) and (5) was independent of the barbell height. Static mechanical stability requires that the stability parameter, S must be greater than zero but does not constrain the positive magnitude. Large positive magnitudes of S provide improved stability but expend excessive physiologic energy through recruitment of co-contraction. Most theoretical analyses assume "critical stability", i.e. minimum static energy required to maintain stability by equating S to zero. In reality, it is likely that a small stability value greater than zero is maintained as a safety margin while simultaneously limiting excessive energy expenditure (Cholewicki and McGill, 1996). However, the ability to voluntarily modulate S through neuromuscular control of antagonistic co-contraction may have introduced variability into the measurements. Nonetheless, results confirm that the neuromuscular system responds to changes in biomechanical stability of the trunk through control of antagonistic co-contraction.

In conclusion, by applying a constant external moment and varying the height of the external load, it was possible to examine the influence of stability independent of trunk moment. Experimental data supported the proposed hypothesis demonstrating increased muscle activation when external loads were held at greater heights even when external moment was similar in each condition. Results illustrate that the neuromuscular system responds to changes in stability. The information gained from this effort provides insight into the neuromotor recruitment of muscle activity, spinal stability and lifting biomechanics.

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