DEPARTMENT OF HEALTH AND HUMAN SERVICES

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

Memorandum

Date: May 3, 2001

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

Subject: Final Report Submitted for Entry into NTIS for Grant 5 K01 OH000158-03.

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

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

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

Attachment
cc: Sherri Diana, EID, P03/C13

List of Publications

Orishimo KF, Granata KP: Quanitification of Trunk Muscle Co-Contraction from Biomechanical Stability. Journal of Biomechanics, in press, 2000

Granata KP, Orishimo KF, Sanford AH: Trunk Muscle Coactivation in preparation for Sudden Load. J Electromyo Kinesiology, in press, 2000

Granata KP, Sanford AH: Lumbar-Pelvic Coordination is Influenced by Lifting Task Parameters. Spine 25(11):1413-1418, 2000

Granata KP, Marras WSL: Cost Benefit of Muscle Co-Contraction in Protecting Against Spinal Instability. Spine 25(11):1398-1404, 2000

Title:	Trunk Stability and Spinal Load During MMH Lifting
Investigator:	Kevin P. Granata, Ph.D.
Affiliation:	University of Virginia
City & State:	Charlottesville, VA
Telephone:	(804) 982-0513
Award Number:	5 K01 OH000158-03
Start & End Date:	9/30/1997–9/29/2000
Total Project Cost:	$160,768
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Key Words:	

Abstract:
Occupationally related low back disorders (LBDs) are the leading cause of lost work days and the most costly occupational safety and health problem facing industry today. It is well known that LBD risk is associated with manual materials handling (MMH) and are influenced by MMH lifting parameters, specifically trunk posture and lifting task design. However, a major limitation in controlling the incidence of occupational LBDs is the inability to explain the injury mechanism to the lumbar spine (the most common site of injury). The scientific community has overlooked the influence of trunk and spinal stability as a cause of occupational LBD. When the trunk/spine is unstable, the tolerance to compressive forces on the spine is dramatically reduced. Therefore, an unstable spine may fail even if the applied load is a mere fraction of the recommended limits. The goals of this research were to evaluate whether the neuromuscular system attempts to control stability of the spine, and whether trunk posture during MMH lifting influences the stability of the spine. It was hypothesized that the risk of LBD in asymmetric and flexed postures is partially related to reduced spinal stability in these postures.

Result demonstrate MMH lifting parameters influence spinal stability. The neuromotor system aggressively controls spinal stability as demonstrated by the coactive response to static stability conditions. Conversely, in dynamic stability conditions the neuromotor system attempts to protect against spinal injury through active response characteristics. This suggests the stability of the spine is highly valued by the musculoskeletal system. MMH lifting parameters, specifically trunk posture can significantly affect spinal stability. Theoretical and empirical results agree that asymmetric postures require increased antagonistic coactivation to maintain stability. With increased trunk flexion, stability is improved, but the ability to maintain even minimal levels of stability may be compromised. In both conditions, i.e. asymmetric postures and trunk flexion, increased coactive recruitment of antagonistic muscles will aid stability, but will increase spinal load. Thus, stability may need to be sacrificed to prevent overload injury, or overload injury risk may be enhanced in the effort to maintain stability.

Epidemiologic data concludes MMH lifting posture significantly influences the risk of occupational LBDs. These postures have been related to increased spinal load as a result of increased antagonistic muscle coactivation. However, until now it was not understood why antagonistic co-contraction was increased in these postures. Our results illustrate that antagonistic co-contraction is necessary to maintain spinal stability and prevent

injury. It has been noted that many low-back injuries occur at low-loads, i.e. at compressive levels considered safe according to the NIOSH lifting guidelines. Stability readily explains this low-load risk. It also explains high-load risk, i.e. more stability is required in high spinal load conditions, but trunk flexion and asymmetry may reduce the ability of the musculoskeletal system from achieving that level of stability.

Publications

Orishimo KF, Granata KP: Quanitification of Trunk Muscle Co-Contraction from Biomechanical Stability. Journal of Biomechanics, in press, 2000

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FINAL REPORT :

Trunk Stability and Spinal Load During MMH Lifting

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Project Sponsor :	NIOSH
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Principle Investigator :	K.P. Granata, Ph.D.
Project mentor :	W.S. Marras, Ph.D. (The Ohio State University)

Acknowledgement : We wish to thank NIOSH of the Cneters for Disease Control for their support in this research effort. Assistance in data collection and analyses were provided by Mr A.Sanford, Mr K.Orishimo, and Dr.S.Wilson. We also wish to thank Dr. W.S. Marras for his guidance in this effort.

Key words: Low-Back Pain, Lifting, Stability, Muscle, Coactivation

ABSTRACT

Occupationally-related low back disorders (LBDs) are the leading cause of lost work days and the most costly occupational safety and health problem facing industry today. It is well known that LBD risk is associated with manual materials handling (MMH) and are influenced by MMH lifting parameters, specifically trunk posture and lifting task design. However, a major limitation in controlling the incidence of occupational LBDs is the inability to explain the injury mechanism to the lumbar spine (the most common site of injury). The scientific community has overlooked the influence of trunk and spinal stability as a cause of occupational LBD. When the trunk / spine is unstable, the tolerance to compressive forces on the spine is dramatically reduced. Therefore, an unstable spine may fail even if the applied load is a mere fraction of the recommended limits. The goals of this research were to evaluate whether the neuromuscular system attempts to control stability of the spine, and whether trunk posture during MMH lifting influences the stability of the spine. It was hypothesized that the risk of LBD in asymmetric and flexed postures is partially related to reduced spinal stability in these postures.

Results confirmed that the neuromotor system actively modifies muscle recruitment patterns in response to stability requirements. Results also demonstrate the spine is less stable in asymmetric postures. In trunk flexed postures, physiological constraints limit the ability to achieve spinal stability. Thus, the risk of LBDs may be related to spinal stability

SPECIFIC AIMS

The goal of this research was to identify occupational hazards for biomechanical LBD risk by quantifying factors that influence musculoskeletal stability. Three specific aims were identified in the original funding proposal.

1. Develop a biomechanical model of *in vivo* trunk / spinal stability

   Three spinal stability models were developed throughout the three-year project period. These models vary in complexity, biomechanical realism, and interpretability. These models were used to evaluate the stability of the spine in various lifting postures and conditions.

2. Assess the influence of MMH task parameters upon the relative stability of the trunk

   Our results demonstrate that muscle activation, specifically antagonistic co-contraction is recruited in response to stability requirements. Postures wherein spinal stability is reduced or wherein stability is physiologically difficult to maintain are the same postures cited as high-risk for occupationally-related low-back pain, i.e. asymmetric postures require greater muscle recruitment to safely maintain stability, stability in flexed postures may requires extreme compressive loads that may be related to risk of overload injury to the spine during MMH and lifting tasks. Thus, spinal stability is correlated with postures of high LBD risk and may contribute to the mechanics of occupational injury.

3. Quantify spinal loads and musculoskeletal behavior in response to unstable events

   Data illustrate that the musculoskeletal system responds to spinal stability. Conversely, the neuromuscular system does not change preparatory behavior in expectation of a sudden load or unstable condition. Thus, the neuromotor system greatly relies upon reflex response behavior to maintain dynamic stability during experimental sudden-load protocols. Fatigue and posture influence neuromotor response. Therefore, these risk-factors may limit the ability to safely maintain spinal stability during manual materials handling tasks.

STUDIES AND RESULTS

BIOMECHANICAL MODEL of SPINAL STABILITY

We developed a biomechanical model to evaluate the influence of antagonsitic co-contraction on spinal stability. Measured EMG data from trunk muscles recorded during MMH lifting exertions served as input to accurately describe coactivation patterns. The model demonstrated that recruitment of co-contraction, specifically antagonistic co-contraction contributed to improved spinal stability and potentially safer lifting performance (Granata & Marras 2000). However, antagonistic co-contraction is also known to increase spinal stability. A balance must be achieved between risk of overload injury and risk of spinal instability.

To evaluate stability using greater anatomic detail, an advanced biomechanical model of was developed including 7 vertebral segments (1 sacral, 5 lumbar, 1 thoracic) and 72 muscles (20 multifidus, 12 quadratus, 12 spinalis, 12 transversus, 12 longisimus, 2 internal obliques, 2 external obliques, and 2 recti abdominis). Each segment was represented in three-dimensions resulting in 18 degrees-of-freedom. The advantage of this model is that spinal curvature and euler buckling are more realistically represented. Model results illustrate that stability was reduced in flexed postures greater 20^0 to 30^0 of lumbar motion and in asymmetric postures (Granata 1998a, 1998b). This correlates with published LBD epidemiology. The disadvantage of this model is that there are many more muscles than degrees of freedom, so simplifying assumptions were required to describe muscle force distributions.

A third model of stability was developed to combine the geometric flexibility of the spine and Euler buckling principles with the simplicity and interpretability of inverted pendulum models. This model included a two segment spine to permit independent control of trunk angle and spinal curvature, as well as 12 muscle groups including bilateral rectus abdominis, external obliques, internal obliques, longisimus thoracic, and inter-segmental para-spinal muscles. The equilibrium and stability constraints provided 12 degrees-of freedom allowing prediction of force in the 12 muscles. Results demonstrate that the spine becomes less stable in asymmetric postures, agreeing with epidemiologic data. Stability is improved with trunk flexion, but may exceed the physiologic limits of muscle recruitment. Thus, maintaining spinal stability is increasingly difficult with sagittal flexion angle, suggesting potential risk factors for LBD. (Granata & Wilson 2001).

MMH PARAMETERS INFLUENCE TRUNK STABILITY

The biomechanical models illustrate that lifting task design, specifically trunk flexion and task asymmetry influence the biomechanical staiblity of the spine, and the risk of spinal injury from instability failure. Our previous research has demonstrated increased coactivation in these postures, presumably to augment spinal stability. To test whether antagonistic co-contraction is recruited in response to stability, we evaluated EMG recruitment from ten men and ten women during static lifting tasks. The lifting moment was maintained at a constant level while stability was changed. Results demonstrate the antagonistic muscle recruitment increased in proportion to stability requirements despite a constant lifting moment. This illustrates that the neuromuscular system responds to stability requirements of the spine. (Orishimo & Granata 2000).

The model results, in combination with the empirical results described above indicate trunk muscle activity is recruited to satisfy equilibrium and to maintain stability. Thus, in unstable postures, particularly asymmetric conditions, increased antagonsitic coactivation must be observed. To test this hypothesis, we recorded EMG activity from the eight major trunk muscles, including the bilateral erector spinae, internal obliques, external obliques, and rectus abdominis during static lifting exertions as a function of trunk posture including flexion angle and asymmetric posture. Results illustrated a correlation with predicted muscle activities necessary to maintain stability. These results clearly demonstrate that 1) the neuromotor system responds to stability requirements, and 2) the spinal stability is reduced in MMH lifting conditions historically noted as high risk postures (Granata & Wilson 2001). We conclude that the biomechanical cause of LBD risk may be partially related to spinal stability, and that stability must be considered in ergonomic attempts to control occupational LBDs.

MUSCULOSKELETAL RESPONSE TO UNSTABLE EVENTS

Recognizing that appropriate muscle activity must be recruited to maintain spinal stability, we hypothesized that in unstable conditions, increased preparatory trunk muscle coactivity will be demonstrated. We produced unstable events by implementing a sudden loading paradigm. A weight of 2.5% MVC suddenly added to a crate held by the test subjects. They were informed a sudden load would be applied, or no sudden load would be applied. We recorded the trunk muscle preparatory coactivity in response to the sudden loading conditions. Eleven males and 14 females were recruited into the study. Results demonstrated no change in preparatory coactivation when the subjects were anticipating a sudden load (Granata et al 2000). This indicates a reliance upon response or reflex muscle activity to maintain spinal stability. However, significant effects were observed as a function of postural asymmetry, fatigue, and gender. Thus, postural and personal factors may influence static stability recruitment. It is concluded that changes in static stability were associated with increased preparatory muscle activity (see section above, Orishimo & Granata 2000) but dynamic stability requirements rely on muscle response characteristics. These suggest further research must be performed to investigate the influence of muscle response dynamics on spinal stability. Ongoing analyses are examining the response characteristics as a function of the task asymmetry, fatigue state, and gender.

SIGNIFICANCE

Results demonstrate MMH lifting parameters influence spinal stability. The neuromotor system aggressively controls spinal stability as demonstrated by the coactive response to static stability conditions. Conversely, in dynamic stability conditions the neuromotor system attempts to protect against spinal injury through active response characteristics. This suggests the stability of the spine is highly valued by the musculoskeletal system. MMH lifting parameters, specifically trunk posture can significantly affect spinal stability. Theoretical and empirical results agree that asymmetric postures require increased antagonistic coactivation to maintain stability. With increased trunk flexion, stability is improved, but the ability to maintain even minimal levels of stability may be compromised. In both conditions, i.e. asymmetric postures and trunk flexion, increased coactive rectruitment of antagonistic muscles will aid stability, but will increase spinal load. Thus, stability may need to be sacrificed to prevent overload injury, or overload injury risk may be enhanced in the effort to maintain stability.

Epidemilogic data concludes MMH lifting posture significantly influences the risk of occupational LBDs. These postures have been related to increased spinal load as a result of increased antagonistic muscle coactivation. However, until now it was not understood why antagonistic co-contraction was increased in these postures. Our results illustrate that antagonistic co-contraction is necessary to maintain spinal stability and prevent injury. It has been noted that many low-back injuries occur at low-loads, i.e. at compressive levels considered safe according to the NIOSH lifting guidelines. Stability readily explains this low-load risk. It also explains high-load risk, i.e. more stability is required in high spinal load conditions, but trunk flexion and asymmetry may reduce the ability of the musculoskeletal system from achieving that level of stability.

Future efforts must focus on developing ergonomic evaluation tools to quantify stability in the workplace, and control measures that will enhance spinal stability. We have proposed further research to investigate assessment techniques to easily and quickly quantify musculoskeletal stability. We have also proposed further research to investigate neuromuscular factors that might be used to improve stability in specific individuals who demonstrated increased personal risk of stability injury.

It is our hope that this research will broaden the dimensional understanding of LBD risk. Currently, LBD risk (from a biomechanical perspective) is limited to spinal load. Our research demonstrated the added dimension of stability must also be considered to fully understand occupational LBD risk.

REFERENCES

The publications supported in part as a result of this research effort described above.

PEER REVIEWED MANUSCRIPTS

Granata KP, Sanford AH. *Spine* 2000; 25 (11), 1413-1418
Lumbar-Pelvic Coordination is Influenced by Lifting Task Parameters

Granata KP, Marras WS. *Spine* 2000; 25 (11), 1398-1404
Cost Benefit of Muscle Co-Contraction in Protecting Against Spinal Instability.

Orishimo KF, Granata KP. *Journal of Biomechanics*, 2000; in press
Quantification of Trunk Muscle Co-contraction from Biomechanical Stability

Granata KP, Orishimo KF, Sanford AH. *J Electromyo.Kinesiology*, 2000; in press
Trunk Muscle Coactivation in Preparation for Sudden Load

Granata KP, Wilson SE. *Clinical Biomechanics*, 2000; in review
Trunk Posture and Spinal Stability

CONFERENCE PROCEEDINGS

Granata KP, Sanford AH. *Proceedings of the American Society of Biomechanics* 1999
Lumbar-Pelvic Coordination is Influenced by Lifting Task Parameters

Granata KP. *Proceedings Human Factors and Ergonomics Society* 1998
LBD Risk Factors and the Structural Stability Tolerance of the Lumbar Spine.

Granata KP. *Proceeding of the North American Congress on Biomechanics* 1998
Structural Stability Tolerance of the Spine: Lumbar Lordosis in Lifting.

Granata KP, Marras WS. *Proceedings Human Factors and Ergonomics Society* 1998
Benefits of Trunk Muscle Co-Contraction in Protecting Against Low-Back Injury during MMH Lifting

Submitted to : Clinical Biomechanics

Trunk Posture and Spinal Stability

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Key words: Spine, Stability, Posture, Low-Back Pain

Running Title: Trunk Posture and Spinal Stability

ABSTRACT

Objective. The influence of trunk posture on musculoskeletal stability was investigated.

Design. A biomechanical model was developed to evaluate the influence of posture on spinal stability. Model validation was assessed by comparing predicted muscle recruitment patterns with measured EMG activity from the trunk muscles during static lifting exertions.

Method. An inverted double-pendulum model of the spine controlled by twelve muscle equivalents of the trunk was developed. Model input included trunk posture and lifted mass, output included muscle recruitment patterns necessary to achieve stability of the spine as well as the spinal load associated with the muscle recruitment patterns and applied external loads. EMG activity recorded from the trunk muscles of ten subjects were recorded during static exertions in various trunk flexion and asymmetric postures to compare with model output.

Results. Antagonsitic co-contraction was necessary to achieve spinal stability, particularly in upright postures. Stable spinal load was increased in asymmetric postures as a result of antagonistic muscle recruitment, suggesting greater neuromuscular control is necessary to maintain stability in asymmetric lifting postures. As trunk flexion angle increased, stability improved but spinal load was greater.

Conclusions. Results illustrate that muscle recruitment patterns are more accurately explained by spinal stability than equilibrium alone. Spinal stability is influenced by trunk posture.

RELEVANCE

Traditional assessment of low-back disorder risk have focussed on spinal loading. Results illustrate that postural risk factors for low-back pain may be partially attributable to stability considerations.

INTRODUCTION

Risk of occupationally-related low-back disorders (LBDs) may be correlated with the musculoskeletal stability of the spine. Biomechanical assessments of occupational LBDs have traditionally focused on spinal load with lifting guidelines that suggest spinal compression below 3400 N may considered safe for a majority of the working age population[1,2]. Conversely, epidemiologic reports demonstrate high incidence and risk of injury even at low spinal loads[3,4]. Injury associated with spinal instability may occur at compressive loads as low as 88 N[5] but neuromuscular control of spinal stability through appropriate recruitment of muscle activity permits safe support of extremely large spinal loads[6,7]. However, select postures may limit the ability of the neuromuscular system to maintain stability. Thus, the relation between lifting posture and risk of low-back injury may be related to spinal stability.

Trunk posture during occupational lifting is associated with the risk of suffering a LBD[8]. Retrospective studies have identified manual materials handling (MMH) as the most common cause of LBD accounting for 50% to 75% of all back injuries[9,10]. Lifting combined with twisting[11], lateral bending[3] and asymmetric postures[12] further increase the risk of LBD. Biomechanical measurements illustrate that antagonistic co-contraction of the trunk musculature is increased in these high-risk postures[13-15]. Recent efforts indicate antagonistic coactivity is recruited to maintain spinal stability[16,17]. Thus, high-risk lifting postures may be associated with reduced stability, requiring greater neuromotor activation to prevent unstable failure and injury of the spine. In other words, postural risk factors for LBDs may be partially explained by spinal stability. To understand the mechanisms of low-back injury, it is necessary to estimate stability as a function of lifting posture.

Spinal stability can be estimated from biomechanical models. These models fall into two categories of complexity, including inverted-pendulum analyses and indeterminate, multi-segment models. Several analyses have represented the spine as a two or three degree-of-freedom single inverted-pendulum[16,18,19]. These are an excellent means for investigating muscle recruitment because the number of muscles approximates the number of degrees of freedom of the model. However they are limited because they fail to represent anatomic detail of the spine including lumbar lordosis and flexibility of the spinal column. Other models include 5 to 7 vertebral segments with 90 to 180 muscle elements[20-23]. These provide anatomic detail and accurate characterization of lumbar curvature. Unfortunately, they cannot determine muscle recruitment distribution without significant simplifications and/or *a priori* assumptions regarding neuromotor control because the number of muscles far exceeds the number of degrees-of freedom of the system. Nonetheless, both of these approaches apply conceptually similar analyses to satisfy biomechanical equilibrum and stability.

Static equilibrium is satisfied when the applied forces and moments balance, i.e. potential energy of the system is an extremum. 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[24], i.e. the equilibrium potential energy extremum represents a minimum. Recognizing that active muscle stiffness is proportional to contractile force[25], trunk stiffness and stability are increased with antagonistic co-contraction. In fact, stability constraints may require increased coactive recruitment[17]. This provides a convenient mechanism by which the stability models may be validated, i.e. compare measured and predicted myoelectric coactivity. Once validated, these models can be used to compare spinal stability with postural factors associated with LBD risk.

The goal of this research was to evaluate the influence of trunk posture on spinal stability. It was hypothesized that spinal stability is reduced in asymmetric versus symmetric postures, i.e. reduced stability in high-risk conditions. Thus, risk of LBD may be related to musculoskeletal stability of the spine. To compensate for reduced stability, increased trunk muscle coactivation must be recruited in the asymmetric postures, thereby explaining previous myoelectric results. The increased coactivity necessary to maintain stability will contribute to increased spinal load and associated risk of overload injury. It was also hypothesized that stability is inversely correlated with trunk flexion. Trunk flexion is associated with increased muscle force requirements, thereby stiffening and stabilizing the spine. These hypotheses were tested by implementing a biomechanical model of spinal stability and computing the influence of stability constraints as a function of trunk postures. Validity of the model was assessed by comparing results with measured trunk muscle activity.

METHODS

Model

A three-dimensional, two-segment model was developed to determine spinal stability as a function of trunk posture and predict trends in antagonistic muscle activity (Figure 1). The two-segment geometry is unique, allowing independent control of trunk flexion angle and lumbar lordosis. Kinematic coordination between the two segments was established from measured values of lumbar lordosis and trunk angle[26]. Lordosis was described as the angle between the two segments and trunk angle as the anterior displacement of the superior aspect of the spine from vertical. Lordosis was prescribed as a fixed ratio of trunk flexion, with posterior angle of 15^0 in an upright posture declining linearly with flexion angle until the two segments were aligned, i.e. 0^0 lordosis in a flexed posture of 50^0. The model was exercised throughout a range of sagittal flexion angles from upright to 45^0 of forward flexion and at asymmetric postures from zero to 20^0 of trunk twist. Inter-vertebral resistance to motion or passive stiffness was ignored as research indicates these passive components contribute little to the stability of the trunk[16]

Twelve muscle equivalents were described including the right and left recti abdominis, external obliques, internal obliques, and paraspinal muscles incorporating one- and two-segment muscles, e.g. inter-transversus and longisimus thoracic equivalent muscles. Muscle origins, insertions and cross-sectional areas were established from published anatomy and modeling efforts[27,28] (Table XX). Vector lengths and directions of the muscles were computed from Euler rotations of the vertebral segments and associated muscle insertions. Three-dimensional moment generating capacity of each muscle about the vertebral base of the segments were determined from the vector product of the muscle-insertion vector and the unit-vector of muscle force.

Muscle force amplitudes were determined from equilibrium and stability constraints. Equilibrium requirements established six-degrees-of-freedom and stability provided six constraints, thereby allowing force estimation of up to twelve muscle groups. Equilibrium was satisfied by equating the sum of muscle moments with the external moments from a static load of 11.3 kg rigidly fixed 20 cm anterior to the superior surface of the modeled spine and a trunk mass equivalent to 55% of body weight was represented as a rigid mass fixed 5 cm anterior to the superior surface of the modeled spine[29]. Thus, external loads rotated with the trunk causing increased flexion and lateral moments and requiring greater restorative muscle forces with trunk flexion and asymmetry.

Stability requires the potential energy Hessian matrix of the system to be positive definite. In other words, a small perturbation in vertebral angle will change the external moment and will stretch the muscles in a mathematically predictable fashion. When stretched, stiffness from the muscles applies a restorative moment that can offset the change in external moment if the appropriate set of muscles are recruited. The potential energies of the system were determined from muscle stiffness and system geometry or posture (see Cholewicki[20] for a description of potential energy and stability calculation of the musculoskeletal spine). Muscle stiffness was established as a linear function of force with a constant of proportionality, $q = 5$ [21] and inversely proportional to the muscle equilibrium length,

$$k_m = q\frac{F_m}{L_m} \tag{1}$$

where k_m is the muscle stiffness of muscle m=1..12, F_m and L_m the muscle force and length respectively[23]. The system was considered stable when all eiganvalues of the Hessian matrix were greater than zero.

The model was designed to search for the set of muscle recruitment that simultaneously satisfied equilibrium and stability. Excessive co-contraction can be recruited to increase stability beyond minimum requirements but is costly in terms of spinal load and energy expenditure. Moreover, in some postures equilibrium and stability constraints may not be fully independent. Therefore, quadratic optimization was performed with an objective function to minimize the sum of muscle stress,

$$\min \sum_{m=1}^{12} F_m' \sigma F_m \qquad (2)$$

$$\text{st}:1)\ \sum_{m=1}^{12} (r \times \hat{f}_m) F_m = M_{Ext}, \qquad 2)\ \text{eig}\left(\frac{\delta^2}{\delta\theta_i\, \delta\theta_j} V\right) \geq 0, \qquad 3)\ F_m < \text{Gain} \cdot \text{Area}_m \qquad (3)$$

where F_m, and σ are the force amplitude and stress-squared matrix of muscles m=1…12. The analyses were subject to three constraints. First, the set of muscle forces must satisfy equilibrium where r, f_m, and M_{Ext} are respectively the muscle insertion moment-arm vector, unit-vector of muscle force and external moment at the vertebrae. Second the set of muscle forces and associated stiffness must satisfy stability wherein the eiganvalues of the Hessian matrix of potential energy, V are greater than zero. Third, set of muscle forces must be within physiological limits expressed in terms of muscle cross-sectional area, Area_m, and the force generating capacity per unit area, Gain = 50N/cm. Muscle force generating capacity was also modulated by muscle length as described as in published models of trunk biomechanics[30]. For presentation, predicted muscle forces were expressed as a percentage of the theoretical force generating capacity, i.e. percent of MVC. Model validity was evaluated by comparing predicted muscle force recruitment to measured values of myoelectric activity.

Spinal load was computed from the vector sum of muscle forces and external load at each vertebral level. Spinal load was computed from two separate conditions. First, an "equilibrium spinal load" was determined from a set of optimized muscle forces computed by satisfying equilibrium without consideration of stability, i.e. ignoring the second constraint. Second, a "stability spinal load" was determined from the set of optimized muscle forces satisfying both equilibrium and stability constraints. The analyses were performed at trunk postures including flexion and spinal twist, described heretofore as trunk asymmetry. Sensitivity analyses to lumbar lordosis was also investigated by changing the relation between trunk flexion and spinal lordosis by $\pm 10^0$. The influence of posture on spinal stability was evaluated by comparing "stability load" to "equilibrium load." This difference represented the relative effort beyond equilibrium necessary to maintain stability.

Experiment

Five healthy males and five healthy females, 24 to 40 years of age, with no prior history of low back pain voluntarily participated in this experiment. Mean (±std) subject height and weight was 172.0 ± 12.1 cm. and 74.2 ±17.7 kg respectively. All subjects provided informed consent approved by Human Investigations Committee of the university. Electromyographic (EMG) activities were recorded from the trunk muscles to compare with muscle recruitment predicted by the model. Subjects performed static lifting tasks at forward flexion angles of 0^0 (upright), 15^0, 30^0 and 45^0, and at asymmetric postures of 0^0 (sagittally symmetric), 10^0 left twist, and 20^0 left twist, and all combinations of these flexion and asymmetric postures. Subjects were required to hold a weight and 11.3 kg cradled in their arms with the arms crossed over their chest in each posture.

Static trunk postures were recorded using surface mounted electromagnetic tracking sensors (Ascension Technology Corp., Burlington Vt). Two sensors were placed over the subject's spinous processes at T10 and S1 and a third marker on the manubrium. Trunk flexion was computed from the mean angular displacement with respect to the upright posture, and trunk asymmetry from the rotation of the T10 and manubrium markers with respect to the S1 sensor about the trunk flexion vector-line. Trunk flexion and asymmetry angles were displayed as real-time feedback for the subject to observe. From this display the participants were able to achieve and maintain the prescribed postures.

EMG were collected at 1000 Hz using bipolar surface electrodes (Medicotest, Rolling Meadows, IL) from 4 bilateral sets of trunk muscles. These muscles included the right and left recti abdominis (RA), external obliques (EO), internal obliques (IO) and erector spinae (ES). Electrodes were placed according to Mirka[31], 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 45^0 to vertical; internal oblique 10 cm lateral to the midline within the lumbar triangle at a 45^0 orientation; and erector spinae 4 cm lateral to the L3 spinous process. Considering electrode locations, internal oblique activity was representative of extensor and lateral effort whereas external oblique myoactivity was considered flexor and lateral[30,32]. EMG signals were high-pass filtered at 30 Hz, low-pass filtered at 250 Hz prior to data collection, then rectified and integrated using a 5 Hz Hanning low-pass convolution filter in post collection analyses. After processing, signals from each muscle was normalized by the corresponding EMG values recorded during maximum isometric exertions including flexion, extension, right-lateral twisting and left lateral twisting. EMG data represented the average isometric value from the middle three seconds of the five second trial and were expressed as percent of MVC.

Statistical analyses were performed to determine the effects of the lifting posture and load on muscle activity. Repeated-measures ANOVA were performed for each muscle with independent variables of trunk flexion, asymmetry and external load. Analyses were performed using commercial statistical software (Statistica, 4.5, Statsoft, Inc., Tulsa OK) with a significance level of 0.05 for all tests. Trends in significant variables were investigated using post-hoc analyses with Bonferroni correction. Model validation was achieved by comparing trends in measured muscle activity with predicted trends in muscle recruitment computed from the model.

RESULTS

The biomechanical model predicted antagonistic coactivation must exist to maintain spinal stability during the extension exertions. Equilibrium conditions without consideration of spinal stability predicted no muscle force from the rectus abdominis muscles. This was a necessary result of the optimization routine that attempted to minimize the sum of muscle stresses thereby compelling antagonistic co-contraction to approach zero. Conversely, when stability constraints were applied in addition to equilibrium the RA was recruited at a mean level of 6.6% MVC. Measured EMG activity from the RA was significantly greater than zero with a bilateral mean value of 4% MVC across all postural conditions. The stabilizing RA force predicted by the model behaved similarly in both left and right muscles, declining to zero with greater flexion angles but increasing with asymmetry specifically in upright postures. Measured EMG data agreed with stability model predictions associated with asymmetry, increasing significantly with postural asymmetry. Empirical EMG data demonstrated increased activity with trunk flexion in contrast to model predictions. However, the change in EMG with trunk flexion was less than 1% MVC.

Stability constraints resulted in external oblique recruitment with an average force level of 13% MVC whereas the equilibrium model predicted mean bilateral activity less than 5%. Measured EO activity demonstrated a mean value 12% MVC in the left EO and 6% in the right EO as a result of the asymmetric loading. Myoactivity from the EO demonstrated a small but statistically significant increased with trunk flexion, i.e. 0.5% to 5% MVC change from upright to 45^0 flexed postures. Modeled EO recruitment declined with flexion angle allowing potentiation of extensor moment. Right EO increased with modeled trunk asymmetry while left EO recruitment declined with asymmetry to stabilize the lateral external moment according to the stability model. This asymmetric behavior was particularly notable in the upright postures wherein the right EO reached recruitment levels of 55% MVC. As predicted by the model the right EO increased significantly with postural asymmetry although the magnitude was less than predicted, peaking at 23% MVC in asymmetric postures. Modeled recruitment of the left EO was 24% MVC in sagittally symmetric upright postures and declined to zero in asymmetric and flexed postures. Measured EMG from the left EO was only 6% in sagittally symmetric upright postures but similar to the model demonstrated a trend toward reduced activation with asymmetry although this trend failed reach statistical significance.

The stability model predicted greater extensor muscle forces than required by equilibrium conditions alone, although the difference was notable only in near upright postures. This increased force was necessary to offset the stabilizing activity from the antagonistic muscles. To permit comparison with measured myoelectric activity recorded from the erector spinae (ES), the sum of the three paraspinal muscles are presented. Both the equilibrium model and the stability model predicted increased bilateral ES activity with trunk flexion. In flexed postures, the muscle force necessary to satisfy equilibrium was sufficient to achieve stability, so the predicted activity levels of the ES from the two models were identical in forward flexed positions with a mean level of 42%. However, in upright postures greater ES activity was recruited by the stability model than predicted from equilibrium alone, 28% versus 10% MVC. Measured ES EMG levels in the upright posture were 20% MVC and in flexed postures the mean activity was 37%. These agreed well with the stability model predictions. Neither equilibrium nor stability mechanics predicted a large change in ES recruitment with asymmetry, although in upright postures the stability model reduced right ES activity, -13% MVC, and increased left ES levels, 11%. Similarly, asymmetric lifting postures were associated with small trends that reduced right ES

EMG by very slightly, less than 2%, and increased left ES activity by 4.5%. However, these changes failed to reach levels of *a priori* statistical significance (p=.055 and p=.169 for the left and right muscles respectively).

Internal oblique muscles provided extensor moments and lateral stabilizing effects. Equilibrium model results, stability models results, and measured EMG demonstrated bilateral increase in IO activity with trunk flexion and asymmetry. In the flexed postures, predicted IO recruitment from the two models were identical, ranging from mean levels of 27% and 35% as a function of asymmetry. Measured IO activation in these same postures were 17% and 23% respectively. In the upright postures, the stability model predicted greater IO recruitment than the equilibrium analyses, particularly in upright asymmetric postures. The extrordinary recruitment of IO activity in these postures suggests reduced spinal stability in upright-asymmetric postures. Measured EMG also demonstrated increased recruitment with asymmetry, but the recorded myoelectric magnitudes were much smaller than predicted stability needs.

Spinal load was significantly increased when stability constraints were included in the model. Recognizing that added muscle recruitment was necessary to maintain spinal stability it should not be surprising that spinal load was markedly increased as well. Spinal compression increased with trunk flexion in proportion to the external moment associated with forward migration of the trunk mass and external load. The external load was small, resulting in a mean spinal compression of 720 N in the upright posture according to the equilibrium model. However, the compressive load increased to a mean value of 2150 N in forward flexed postures. In these flexed postures the high levels of muscle contraction necessary to satisfy equilibrium was sufficient to achieve spinal stability so the stability model predicted similar levels of spinal compression. Conversely, in upright postures additional muscle recruitment, specifically antagonistic co-contraction was necessary to maintain spinal stability thereby increasing spinal load. In the upright, sagittally symmetric posture, the stability model predicted a compressive spinal load 156% greater that the equilibrium model, with a mean level of 1815 N. Asymmetric lifting also generated increased spinal load. The stability load increased from a mean level of 1775 N in symmetric lifting tasks (averaged across all flexion angles) to a compression of 2155 N during simulated lifting tasks with 20^0 of trunk asymmetry. Thus, asymmetric postures were associated with reduced spinal stability, requiring increased muscle recruitment and spinal load to restore the stability of the system.

Lordosis of the spine was evaluated at levels of 5^0, 15^0, and 25^0. The change in lordosis did not influence modeled stable compression, i.e. the mean change in compressive load was 40N.

DISCUSSION

It was proposed that LBD risk associated with asymmetric and forward flexed lifting exertions may be related in part to reduced spinal stability in these postures. To test these hypotheses a model of spinal stability was developed and exercised with various spinal angles.. Spinal stability required greater muscle recruitment than was necessary to satisfy equilibrium alone. This resulted in greater spinal load when considering stability constraints, more than 150% greater than equilibrium load. The spinal load associated with stability increased with asymmetry, i.e. twisting postures, suggesting greater muscle recruitment was necessary to stabilize the spine in asymmetric conditions. Thus, several factors have been identified that help to explain the risk of LBD associated with asymmetric lifting. These include spinal load, muscle recruitment capacity and spinal stability.

To maintain stability, muscle recruitment was necessary that dramatically increased spinal load during the asymmetric lifting exertions. Spinal load has been cited as a principle biomechanical factor contributing to LBD risk[33-35]. It has been noted elsewhere that antagonistic coactivity is increased in asymmetric lifting tasks[14] that causes greater spinal compression in these postures[36,37]. Modified muscle recruitment patterns are necessary to offset lateral components of lifting moment in asymmetric loading and establish equilibrium[32,38,39]. The current results demonstrate these equilibrium recruitment patterns during 20^0 asymmetric postures were associated with as much as 21% increase in spinal compression over sagittally symmetric exertions. However, a biomechanical explanation for the increased antagonsitic coactivation has until now never been described. Antagonsitic co-contraction contributes up to 31% of the spinal compression in sagittal lifting exertions, i.e. 45% additional load when antagonistic co-contraction is considered[40]. Antagonistic co-contraction has recently been identified as a necessary component in the maintenance of stability[16]. Current results suggest this stabilizing recruitment dramatically increases spinal compression, and the change in compression with asymmetry may exceed 30%. Although the lifting loads and associated spinal compression levels were small in these analyses, at increased exertion levels or under dynamic conditions stability requirements may require spinal compression loads that approach injury tolerance levels.

Cost in terms of spinal load and muscle co-contraction energy requirements may entice the neuromuscular system to rely less on preparatory muscle recruitment and more on reaction mechanics to maintain spinal stability. To achieve static stability, antagonistic recruitment up to 45% MVC greater than equilibrium levels were required. Measured EMG data demonstrated significant antagonistic activity where none was required to satisfy equilibrium. Thus, the motor control system clearly responds to spinal stability requirements. This agrees with our previous efforts demonstrating that antagonistic coactivation changes with stability requirements even when equilibrium conditions remain constant[17]. However, the levels of antagonistic muscle activity did not approach the extrordinary levels predicted by the stability model. This difference may be explained by neuromotor response mechanics. It is well recognized that postural disturbances to the trunk elicit well coordinated response activation in the muscles[41-43]. Feedback control is a principle component of stability in engineering systems and musculoskeletal behavior alike. Unfortunately, existing stability models, including the model described in the current study, fail to account for the dynamic feedback response of the system and thereby over-estimate the required magnitude of recruitment coactivation. Analyses by Granata et al[44] and others by Cholewicki et al[45] agree that neuromuscular response dynamics play a significant role in stabilizing the spine. Thus, future efforts must incorporate reflex and response mechanics in biomechanical assessment of spinal stability. Nonetheless, the difference in predicted and measured muscle recruitment suggests greater reliance upon the dynamic neuromuscular response in asymmetric postures. Failure to recruit an appropriate and timely activation response will risk instability injury. Published measurements illustrate that a muscle responses are mediated by various factors including posture and fatigue[46]. Hence, risk of instability injury can be increased in asymmetric lifting exertions and may contribute to the risk of LBD in these postures.

Results suggest spinal stability may be improved with trunk flexion but risk of injury from spinal overload injury is enhanced with flexion. Muscle stiffness provides the primary mechanism of stability in analyses of the spine[29]. Research demonstrates active muscle stiffness increases with contractile force[25], therefore increased contractile force is necessary to achieve stability[23]. Trunk moments are low in upright postures requiring little contractile force from the posterior muscles but moments are large in flexed postures requiring increased extensor muscle force[47]. Hence, equilbirum conditions dictate that stiffness and stability are improved in high-moment low-load conditions such as trunk flexion. This is observable in the results wherein the compressive load from stability analyses and spinal load from the equilibrium model converge with trunk flexion. Unfortunately, spinal load is dramatically increased in these postures so the risk of injury is not improved. However, the model was limited to a three-dimensional representation of inter-vertebral motion, i.e. three dimensions of joint rotation. Trunk flexion also imposes large translation forces on the spine, requiring more advanced analyses involving 6-DOF per joint for the evaluation of stability[29]. In vitro studies suggest flexion loading may reduce stability of the osteoligamentous spine[48]. Moreover, sudden loading studies indicate the feedback gain of the trunk muscles may be influences by pre-load and associated trunk flexion, thereby modulating the stability of the spine in flexed postures[41]. Further research is necessary to document whether neuromuscular control of dynamic spinal stability in a system with 6-DOF per joint is influenced by trunk flexion.

Despite its relative simplicity the model performed well, predicting antagonistic co-contraction, correctly predicting general trends in extensor muscle activation and often recruiting the appropriate percent MVC of muscle activation. The two-segment inverted pendulum representation of the spine was unique, permitting analyses to independently control lordosis and trunk flexion and to predict activity in the major trunk muscles because the 6-DOF equilibrium and 6 stability constraints matched the number of muscles. Previous models have included 5 to 7 vertebral segments and up to 180 muscle elements[20-23] thereby providing anatomic detail and valuable biomechanical insight. However, the number of muscles in these models far exceeded the degrees of freedom requiring considerable neuromotor grouping assumptions and vertebral motion estimates. Simple models such as the analyses described here are limited in their anatomic detail but generate accurate results and require fewer *a priori* assumptions. Discrepancies between predicted and measured muscle recruitment were noted that illustrate model performance limitations. Specifically, the model predicted reduced flexor co-contraction with trunk flexion whereas the empirical measurements demonstrated a small but statistically significant increase in flexor activity. This is in contrast to results published by Cholewicki et al[16] wherein similar reduction in flexor activity was predicted from their stability model but contrary to our EMG measurements they reported a significant trend toward reduced anterior muscle activity with trunk flexion. An important assumption in stability models are the relation between muscle force and stiffness. Our model employed a constant of proportionality of q=5 based upon the theoretical results of Gardner-Morse et al[29]. Conversely, Cholewicki et al[16] employed a proportionality constant of q=30 in their single inverted-pendulum model of spinal stability. This value will influence the magnitude of antagonistic co-contraction and associated spinal load necessary to achieve stability but will not affect trends, i.e. stability will continue to be influenced by postural asymmetry and trunk flexion. The relation between active trunk stiffness and muscle load has never been empirically documented but may be a valuable measure of trunk stability and LBD risk.

CONCLUSION

This research has demonstrated that lifting posture can influence musculoskeletal stability of the spine. Analyses were upon a biomechanical model that predicted recruitment patterns in twelve muscle equivalents necessary to stabilize a two-segment, 6-DOF spine. The model generally agreed with measured muscle activity suggesting model validity. Results demonstrate antagonistic co-contraction is necessary to achieve stability and greater neuromuscular control is necessary to maintain stability in asymmetric lifting postures. As trunk flexion angle increased, stability improved but spinal load was greater. Risk of occupational LBDs associated with lifting posture may be partially influenced by musculoskeletal stability of the spine.

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LIST OF TABLE AND FIGURES

Table 1. ANOVA results and statistical significance (p-value) of integrated EMG of trunk muscles during static lifting exertions.

	RES	LES	RRA	LRA	REO	LEO	RIO	LIO
Gender	0.376	0.094	0.037	0.070	0.694	0.009	0.221	0.171
Flexion Angle	0.000	0.000	0.000	0.005	0.000	0.000	0.049	0.168
Asymmetry	0.169	0.055	0.001	0.000	0.215	0.000	0.020	0.001
Gender x Angle	0.000	0.004	0.091	0.160	0.032	0.000	0.053	0.068
Gender x Asymtry	0.002	0.341	0.599	0.270	0.795	0.150	0.137	0.835
Angle x Asmtry	0.968	0.866	0.340	0.267	0.652	0.808	0.647	0.394

RES : right erector spinae
RRA: right rectus abdominis
REO : right external oblique
RIO : right internal oblique

LES : left erector spinae
LRA: left rectus abdominis
LEO : left external oblique
LIO : left internal oblique

Table 2. Mean values of integrated EMG of trunk muscles during static lifting exertions.

Flexion Angle (deg)	Asymmetric Angle (deg)	RES (%MVC)	LES (%MVC)	RRA (%MVC)	LRA (%MVC)	REO (%MVC)	LEO (%MVC)	RIO (%MVC)	LIO (%MVC)
0^0	0^0	19.1%	21.7%	3.2%	3.5%	9.2%	6.1%	9.1%	9.8%
	10^0	21.3%	19.3%	3.3%	3.7%	10.7%	4.5%	10.8%	8.5%
	20^0	23.5%	20.6%	4.6%	4.1%	18.3%	5.3%	13.7%	12.1%
15^0	0^0	27.1%	29.4%	3.4%	3.5%	10.4%	6.0%	11.6%	12.6%
	10^0	28.2%	29.0%	3.8%	3.7%	14.3%	4.8%	12.7%	13.8%
	20^0	31.1%	28.0%	4.6%	4.2%	17.9%	5.5%	15.7%	15.3%
30^0	0^0	32.8%	35.4%	3.6%	4.0%	12.0%	4.6%	13.9%	15.2%
	10^0	34.2%	35.5%	4.2%	4.1%	16.1%	4.9%	14.8%	17.2%
	20^0	34.4%	32.0%	5.5%	5.1%	22.6%	5.9%	17.5%	19.9%
45^0	0^0	37.1%	40.0%	3.9%	4.7%	14.3%	7.1%	15.9%	17.7%
	10^0	37.1%	35.5%	4.5%	4.3%	17.5%	4.9%	14.2%	19.3%
	20^0	37.9%	34.6%	5.5%	5.2%	23.1%	5.8%	21.0%	23.9%

Figure 1. Predicted muscle recruitment from equilibrium model without consideration of spinal stability

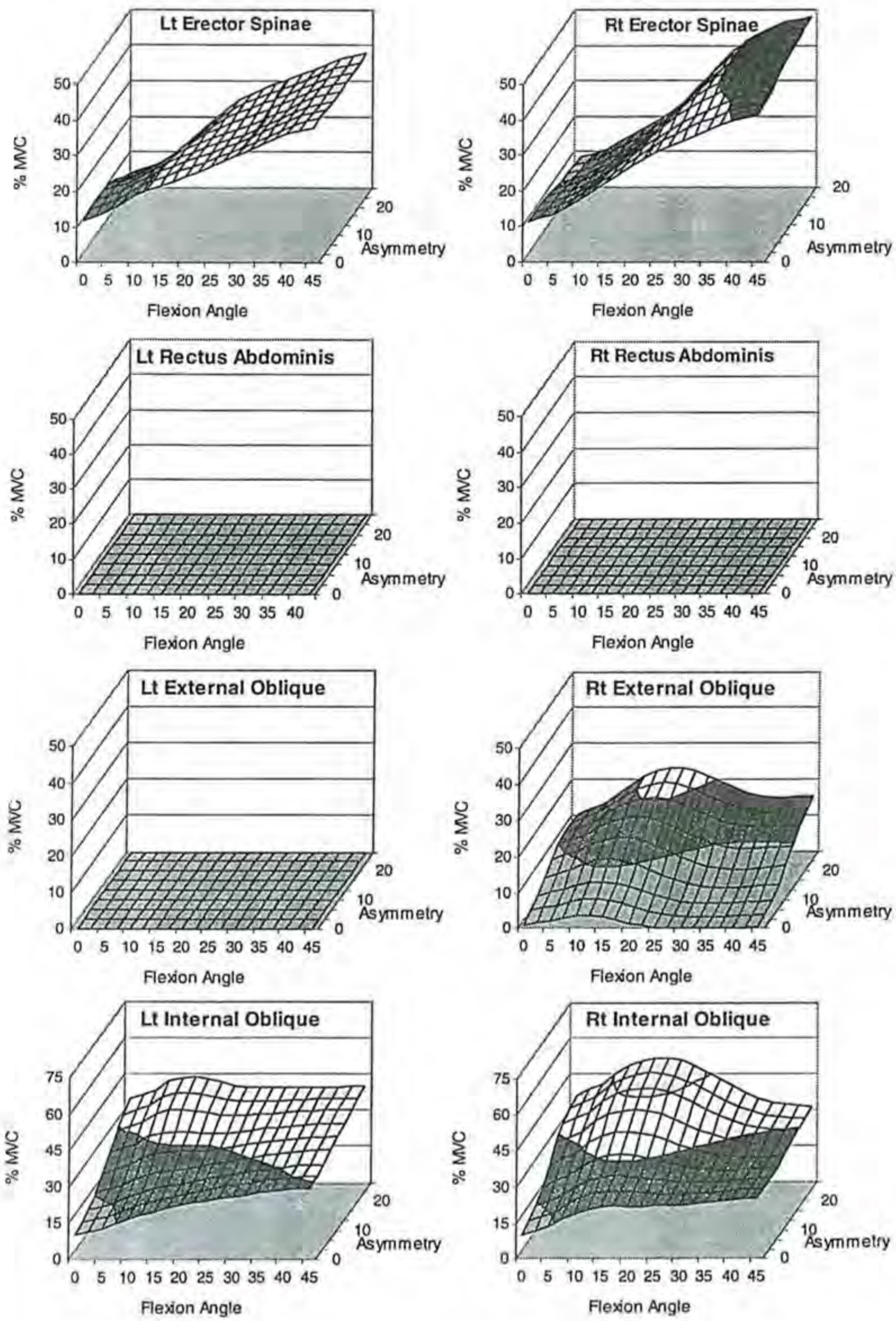


Figure 2. Predicted muscle recruitment from model satisfying both equilibrium and stability

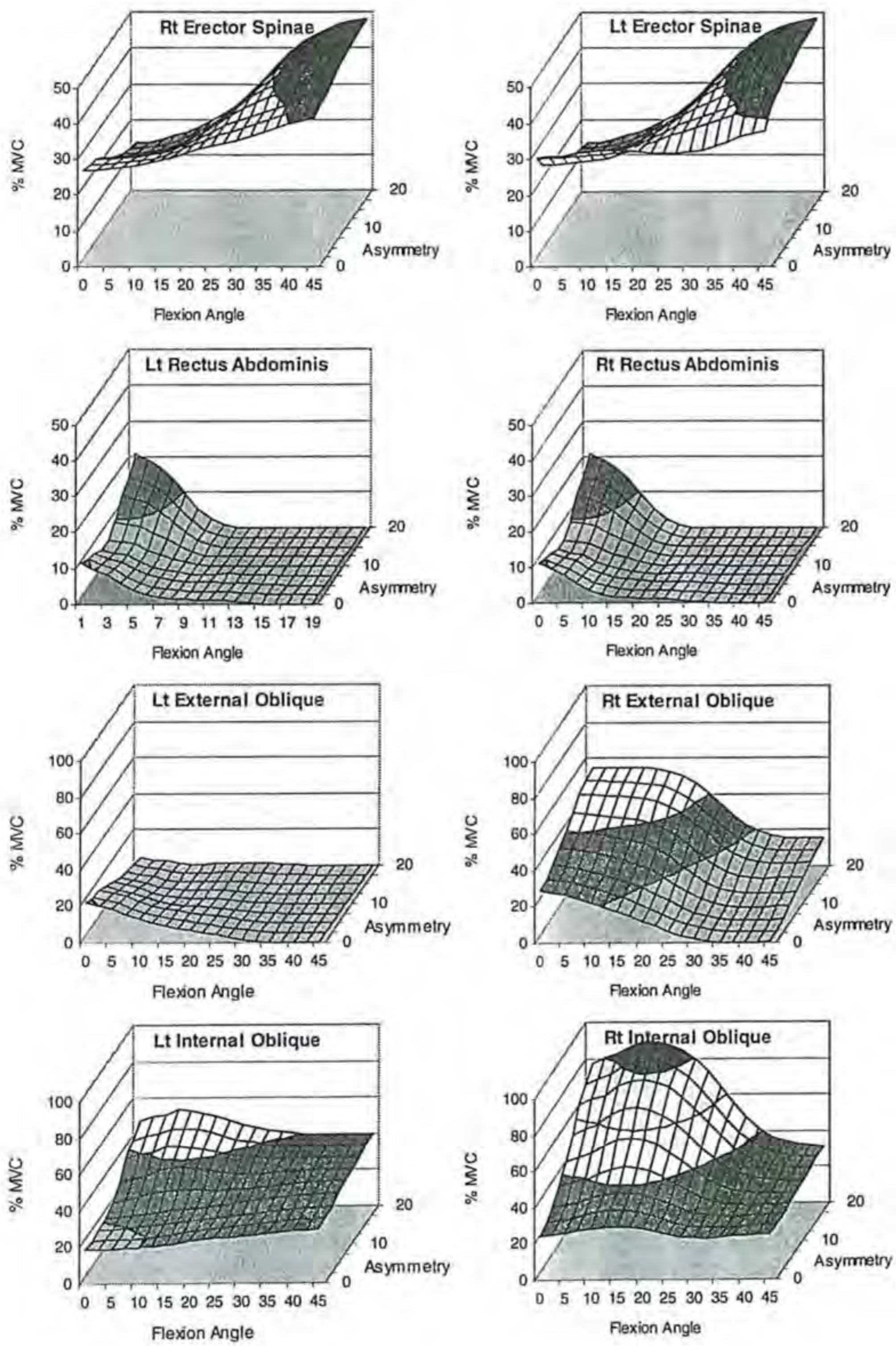
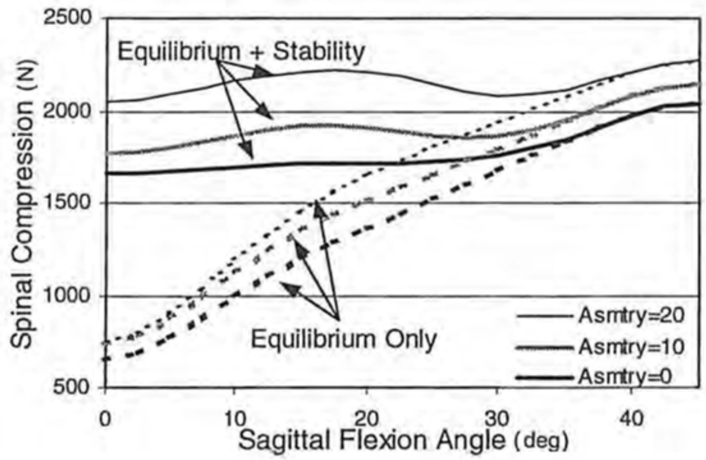


Figure 3. Stability constraints resulted in greater spinal compression than equilibirum analyses without consideration of stability. Greater neuromuscular control is necessary to maintain stability in asymmetric lifting postures resulting in increased stability spinal load.



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Trunk Muscle Coactivation in Preparation for Sudden Load

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ABSTRACT

Biomechanical stability of the lumbar spine is an important factor in the etiology and control of low-back disorders. A principle component of biomechanical stability is the musculoskeletal stiffening generated by preparatory muscle coactivation. The goal of this investigation was to quantify preparatory behavior, evaluating trunk muscle activity immediately prior to sudden trunk flexion loading during static extension tasks compared to activity observed when subjects were informed no sudden load would occur. Coactive excitation was also examined as a function of fatigue and gender. Results demonstrated increased extensor muscle and flexor muscle coactivation following static fatiguing exertions, potentially compensating for reduced trunk stiffness. Female subjects produced greater flexor antagonism than in the males. No difference in the preparatory coactive muscle recruitment patterns were observed when subjects were expecting a sudden flexion load compared to recruitment patterns observed in similar static postures when subjects were informed no sudden load would be applied. This indicates the neuromuscular system relies greatly on response characteristics for the maintenance of stability in dynamic loading conditions. Results provide insight into the control of spinal stability and identify a need for further research into neuromotor control of biomechanical stability of the spine.

tissue increases with myoelectric excitation and contractile force[19,20]. Mathematical models suggest muscle activation can be recruited to stiffen and stabilize the spine[8,21-24]. Reduced kinematic displacement and trunk stiffness following sudden flexion loading has been attributed to increased preparatory coactivation and associated intra-abdominal pressure[25,26] Recent measurements suggest antagonistic coactivation is actively recruited in response to stability requirements independent of trunk moment[27]. Thus, in an unstable environment, increased recruitment of antagonistic co-contraction is expected so as to improve stability and reduce injury risk. When a sudden load is anticipated it is expected that increased co-contraction must be recruited to stabilize the spine in a pre-emptive manner.

Sudden load paradigms are designed to investigated the neuromuscular preparation and response to biomechanical trunk perturbations. When a sudden flexion load is unexpectedly applied to the trunk a response in the form of antagonistic co-activation has been reported at levels up to 140% of the equivalent static value[28,29]. The response is influenced by asymmetry, fatigue, and the subject's history of low-back pain[30,31]. Unfortunately, preparatory behavior has been rarely reported. Measurements by Lavender et al[32] revealed some subjects increased preparatory antagonistic co-activation while others demonstrated no preparatory myoelectric activity. Thomas et al[25] also observed inconsistent behaviors when subjects were preparing for sudden loads. They reported that no preparatory myoactivation was observed when the timing of impact was unknown. When subjects were permitted to observe the falling mass activities in both extensor and flexor muscles ramped up in time to meet the kinetic impact. It was curious that increased coactivation was not observed in both blinded and unblinded conditions as the subjects were aware that a sudden load would occur in both conditions, but lacked timing information in the former.

[53]. To compensate for reduced active muscle stiffness, it is hypothesized that females may perform lifting tasks with greater coactivation to augment trunk stiffness and stability. The influence of gender on preparatory myoelectric activity have not been reported.

The goal of this investigation was to quantify trunk muscle electromyographic (EMG) activity in preparation for a sudden flexion load. It was hypothesized that increased coactivation would be observed when subjects were preparing for an impending flexion moment impact compared to equivalent conditions wherein subjects were informed no sudden load was to be applied. Furthermore, this investigation evaluated the influence of fatigue and gender on the preparatory behavior. It was hypothesized that increased preparatory coactivation must be observed in a fatigued state and greater coactivation may be demonstrated by female subjects to maintain biomechanical stability. Improved understanding of neuromuscular preparatory behavior may contribute to enhanced assessment of spinal stability and control of LBD risk.

was set at 0%, i.e. no added weight, and 20% of the subject's maximum voluntary exertion (MVE) strength in trunk extension. The MVE value was established from the maximum of three isometric exertions performed by pulling against a handle and cable attached to the floor via a load cell, with the cable length adjusted to create a 45^0 flexed posture. Sagittally symmetric sudden loading tasks were examined wherein the subject's feet were aligned with a transverse line on the floor and asymmetric tasks wherein the subject held the crate in the forward position while twisting such that the feet were aligned according to markings 45^0 to the left.

All six combinations of sudden load, pre-load and asymmetry were performed in an unfatigued state with a minimum of two minutes rest between exertions [54]. All conditions were repeated in a fatigued state. During the fatigued conditions no rest between exertions was provided so as to aid in the maintenance of the fatigued state. Fatigue was established by encouraging subjects to hold 20% of their MVE load with their trunk flexed at 45^0 degrees for as long as possible. When the subject could no longer hold the crate, the experimental trials were conducted. To assure the subject remained fatigued throughout all of the conditions, this process was repeated every two minutes, or after approximately four trials.

A brief secondary experiment was performed with nine volunteers. The posture in the primary protocol required the subjects to maintain a static flexed posture of 45^0. This was prescribed to simulate lifting postures observed in industry, to assure the crate held by the subjects could not rest on their thighs, and to apply a pre-load from the trunk mass flexion moment. All sudden load impulses resulted from a 2.5% MVC weight dropping 0.5 m. To investigate preparatory behavior in conditions with minimal pre-load, the secondary study subject's spinous processes at T10 and S1 and a third marker on the manubrium. Trunk flexion angle was displayed as real-time feedback for the subject to observe. From this display the participants were able to control static trunk flexion angle.

Statistical Analysis:

Statistical analyses were performed to determine the effects of the lifting parameters on the preparatory EMG. It is recognized that muscle activity and coactivity are markedly influenced by trunk flexion angle, asymmetric posture and static flexion moment[58]. We were specifically interested in the influence of sudden loading condition, fatigue, and gender. Thus independent mixed-measures MANOVA were performed for the two studies, i.e. flexed and upright postures were analyzed separately. For the secondary study, only sagittally symmetric postures with no pre-load have been reported as the influence of pre-load and asymmetry were identical in the two studies. Repeated measures variables included asymmetry, sudden load condition and fatigue while gender served as a between-subjects variable. Analyses were performed using commercial statistical software (Statistica, 4.5, Statsoft, Inc., Tulsa OK) using a significance level of 0.05 for all tests. Trends in significant variables were investigated using post-hoc analyses with Bonferroni correction.

increased coactivation was observed. Increased mean integrated magnitude of the EMG signal was recorded from the bilateral internal obliques and the right erector spinae (Figure 1), but only in the forward flexed postures (Tables 1 and 2). This increased extensor activation was likely recruited to compensate for the fatigue induced loss in force production from the erector spinae groups. Increased activity during the fatigued conditions was also noted in the flexor muscles including the right external obliques and left rectus abdominis. Both of these flexor muscles revealed a significant fatigue-by-gender interaction, with post-hoc analyses demonstrating increased flexor coactivation with fatigue was significant only for the female subjects. Further pair-wise analyses indicated the male subjects demonstrated greater reduction in median power frequency of the erector spinae EMG signal than the females, suggestion potential differences in the level of fatigue, and providing potential insight into the fatigue-by-gender interaction from the integrated EMG signal. It is nonetheless noteworthy that a change in flexor muscle activity was observed when the fatiguing task focussed on the extensor musculature.

Statistically significant changes in preparatory activity in anticipation of the sudden load were not observed contrary to the initial hypothesis (Table 1). When subjects were expecting the sudden load EMG increased by a mean value of 4% relative to the condition of no sudden load, which failed to achieve statistical significance. The mean power frequency of the bilateral rectus abdominis was significantly increased in preparation for the sudden load. However, the amplitude of the rectus abdominis signal was on the order of 2% MVC, thereby prohibiting any valid conclusions from the spectral data of this muscle. Recognizing that intrinsic active muscle stiffness is proportional to the contractile force, in conditions of high pre-loading the equilibrium level of muscle stiffness may be sufficient to maintain stability

DISCUSSION

Biomechanical stability describes the potential of the musculoskeletal system to maintain equilibrium in the presence of kinematic or kinetic disturbances. When the equilibrium posture is a state of minimum potential energy, the system will return to this minimum energy level if perturbed[59]. This is considered mechanically stable. One method of achieving this is to establish increased stiffness. The stiffness component of active muscle is well recognized[60] and contributes to voluntary control of active joint stiffness and motor control[61,62]. Hogan[63] observed that antagonistic muscle co-contraction serves to increase the system stiffness of the equivalent joint. Thus, preparatory co-activation can be recruited to stiffen the trunk[25,26]. Theoretical analyses suggest this coactive stiffening will effectively augment spinal stability[8,22,23]. Without added stiffness from the active muscles, the spine is highly unstable and susceptible to buckling and injury[4,5]. Empirical measurements demonstrate antagonistic coactivation is recruited in response to stability requirements of the trunk[27]. Thus, it was predicted that increased preparatory coactivation would be observed when expecting a sudden loading disturbance. Results from the current study failed to support this hypothesis.

There were no significant changes in muscle activation levels when subjects were expecting a sudden load. Previous studies conclude preparatory strategy may vary markedly from one subject to another[32]. Thomas et al[25] reported the area under the EMG curves during onset of activity was increased when subjects were permitted to observe the falling mass and predict the timing of the sudden load impact. However, EMG onset rate is slower when subjects are provided with improved timing queues[28]. Whether increased preparatory EMG area was attributable to increased myoactivation or longer duration of onset was not reported.

Recognizing that active muscle stiffness is proportional to contractile force[67,68], it is reasonable that trunk stiffness was greater in conditions of high pre-load [66] and antagonistic coactivation can be reduced while maintaining biomechanical stability in flexed postures [8,24]. It was thought that maybe the magnitude of the sudden load versus the pre-load in the current study failed to exceed the instability threshold, thereby requiring no added coactivation in preparation for the impact. To test this possibility a brief secondary study was performed wherein subjects were exposed to the sudden loading conditions in the upright posture, with no pre-load, and an increased sudden load impact energy, i.e. 5% MVC dropped from 0.5 m. As in the flexed postures, no increase in preparatory activity was observed. Thus, our results agree with others, concluding that the pre-load state may not contribute to preparatory and response behavior.

How was stability maintained in the absence of increased preparatory stiffening? The neuromuscular system can make use of both feed-forward and feedback components to maintain stability. Preparatory stiffening addresses primarily the feed-forward behavior, but no change was observed when stability conditions were modified. This leads us to conclude that the myoelectric response or feedback gain served was the primary stability mechanism in these experimental conditions. Although we have observed elsewhere that coactivation is a significant factor when stability is static in nature[27], the current results indicate dynamic stability relies greatly on neuromuscular feedback. This is supported by others who have proposed that neuromuscular response rate and magnitude may contribute to LBD risk[30,31,69,70]. Further research is necessary to model and quantify the control-feedback stability of the spine.

muscle stiffness was limited due to reduced pre-load, gender difference were observed in the external obliques, internal obliques, and rectus abdominis. Thus, results support the hypothesis that female subjects in the current study may require increased coactive recruitment to maintain stability. Unfortunately this increases the potential for fatigue related stability factors. Control of low-back injury may require gender specific preventative measures and more intensive research efforts focussing on gender specific biomechanical factors in musculoskeletal injury.

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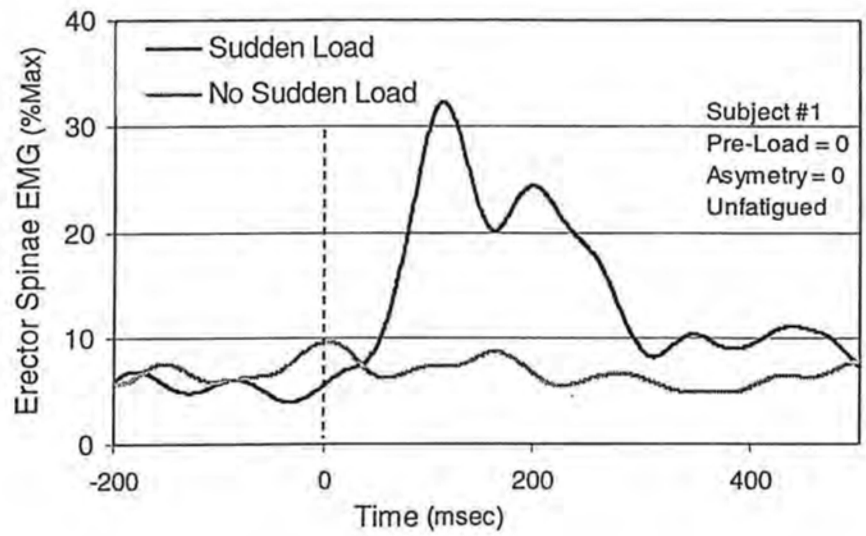
Table 2. ANOVA results (p-values) from the mean integrated preparatory EMG activity during trials performed in the upright posture with no pre-load.

	MUSCLE							
	RE0	RIO	RRA	RES	LEO	LIO	LRA	LES
Gender	**0.029**	**0.043**	**0.036**	0.224	**0.001**	0.093	**0.050**	0.579
Fatigue	0.553	0.433	0.589	0.913	0.718	0.182	0.822	0.058
Sudden Load	0.481	0.264	0.137	0.568	0.243	0.212	0.478	0.449
Gend x Fatig	0.361	0.516	0.478	0.340	0.482	0.307	0.904	0.675
Gend x Sudden	0.378	0.601	0.144	0.906	0.773	0.442	0.321	0.393
Fatig x Sudden	0.289	0.316	0.355	0.427	0.592	0.105	0.399	0.339
Gend x Fatig x Sudden	0.421	0.486	0.304	0.299	0.892	0.200	0.764	0.581

Bold indicates statistically significant main effect

Figure 2. Typical EMG preparatory and response activity from the erector spinae during a sagittally symmetric trials. The black line represents a sudden loading trial and the gray line as equivalent trial with no applied sudden load. The vertical dashed line at time=0 represents the time of the sudden load impact. Note the myoelectric response of the trunk muscles during static extension exertions suggesting the impact was sufficient to warrant a response beginning approximately 55 msec after the impact and peaking 115 msec after impact. No markedly different preparatory recruitment was observed in the two trials.



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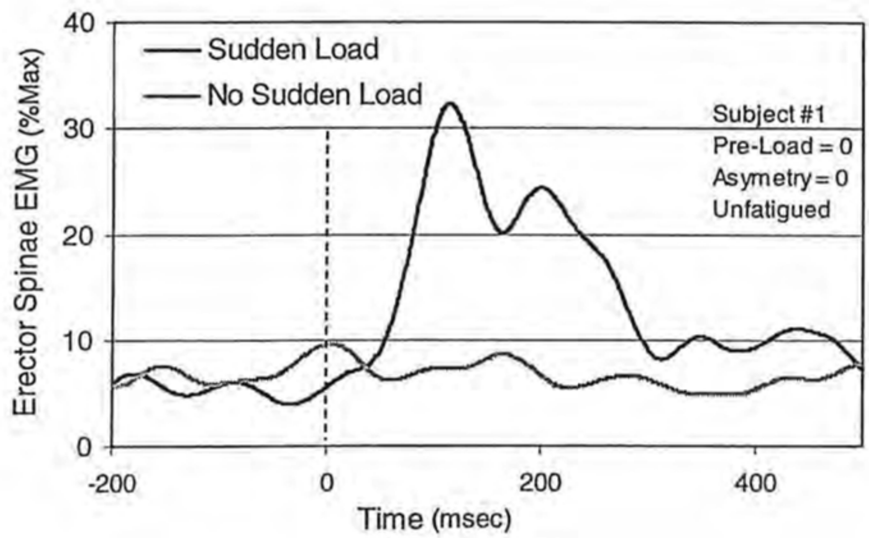

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Quantification of Trunk Muscle Co-Contraction from Biomechanical Stability

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Running Title: Co-Contraction and Stability

ABSTRACT

Purpose: Biomechanical models suggest antagonistic co-contraction may be related to stability constraints during lifting exertions. A model and experiment were designed to study flexor muscle co-contraction during trunk extension exertions, examining the independent influences of trunk moment and spinal stability. The goal was to assess the neuromuscular response to changes in spinal stability.

Methods: A two-dimensional model of the trunk muscle co-contraction was developed from biomechanical equilibrium and spinal stability constraints. Results 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.

Results: 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. Measured EMG activity in the trunk flexors increased with height of the external load as predicted by the model. Antagonistic activity also increased independently with trunk moment. Gender difference in spinal stability were noted.

Conclusions: 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. Results empirically demonstrate that the neuromuscular system responds to changes in biomechanical trunk stability.

Unfortunately, this does not prove that antagonistic activity is motivated by stability. Without empirical evidence it is difficult to validate the stability theory of antagonistic co-contraction.

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 J. et al 1998). However, others have shown antagonistic co-contraction is highly correlated with trunk moment (Mirka G.A., and Marras W.S. 1993) so it remains unknown whether co-contraction is related to trunk moment, spinal stability or both. Therefore, it is necessary to quantify the relationship between biomechanical stability and antagonistic co-activity, evaluating the influences of stability and with trunk moment independently.

The goal of this study was to assess the neuromuscular response to changes in spinal stability. 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. This co-activation is necessary to provide the increased stability required as potential energy increases. Secondary analyses were performed to examine the influence of gender on the co-active response to stability. Empirical measurement of trunk muscle EMG were compared to predictions from a biomechanical model of trunk stability and equilibrium.

(Thompson J.M.T., and Hunt G.W. 1984). Thus the equilibrium or moment equation in the upright posture becomes:

$$\frac{\delta V}{\delta \theta} = \sum M = F_e \cdot r_e - F_f \cdot r_f - F_{Ext} \cdot r_{Ext} \qquad (2)$$

where the elastic rest length, l_0, was represented as the equilibrium posture. In this posture stability is satisfied when

$$\frac{\delta^2 V}{\delta \theta^2} = -(W \cdot h_W + F_{Ext} \cdot r_{Ext}) + k_f \cdot r_f{}^2 + k_e \cdot r_e{}^2 = S \qquad (3)$$

where S is a general stability parameter greater than zero. Research has demonstrated that active muscle stiffness is proportional to contractile force (Hunter I.W., and Kearney R.E. 1982; Weiss P.I. et al 1988),

$$k_f = q \cdot F_f / h, \qquad k_e = q \cdot F_e / h, \qquad (4)$$

where q is the muscle stiffness coefficient described by Bergmark (Bergmark A. 1989). Simultaneous solution of the equilibrium, stability and muscle stiffness relations results in an expression for the contractile force of the antagonists flexor muscles during an extension task.

$$F_f = (W \cdot h_W + F_{Ext} \cdot h_{Ext}) / c_1 + (S - F_{Ext} \cdot c_2) / c_1 \qquad (5)$$

Coefficients c_1 and c_2 are constants involving the muscle moment arms, muscle lengths and stiffness coefficient. This model 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. Antagonistic muscle activity was measured to validate this proposed behavior.

Experiment

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.

Analyses

Flexor and extensor average EMG activities were plotted against the height of the external load. A significant positive slope suggests the flexor muscle EMG increases with height, validating the primary hypothesis. Dependent variables of flexor EMG averages were examined using a mixed-measured ANOVA with independent variables of height, external load and gender. Post-hoc and power analyses were performed to examine differences between levels of each independent variable at significance of α=.05.

statistical power analyses (Table 2) revealed the flexor muscle EMG in the two highest conditions were greater ($p<.05$) than the lowest conditions with statistical power greater than 0.8, i.e. $\beta = .2$. With the barbell at these heights, the center-of-mass height of the system was above that of the trunk under normal circumstances. This increase from the normal operating center-of-mass height decreased the stability of the system. 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, Figure 4). Weight significantly increased the flexor EMG levels at the four greatest heights ($p<0.001$). This interaction between weight and height is associated with the in-vivo value of the muscle stiffness parameter, q, relative to the barbell height, equation 5.

A significant ($p<.001$) gender difference in flexor stabilizing activity was demonstrated at the highest elevations, 60 and 80 cm. Although females recruited more flexor co-activation than males, gender failed to achieve significance as a main effect ($p = 0.065$). However, a significant statistical interaction ($p<.001$) demonstrate the difference between males and females increased with barbell height. Further research is necessary to understand this gender effect.

maintaining spinal stability (Bergmark A. 1989). Thus, based upon theoretical considerations, antagonistic activation of the trunk flexors during extension exertions has been attributed to the recruitment of biomechanical stability.

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. Historically, activity in the trunk extensors and flexors has been associated with equilibrium moment (Chaffin D.B. 1969). In our experimental protocol, external flexion moment was nearly identical when holding the barbell at high elevations and low elevations alike. Thus, based upon equilibrium analyses no change in EMG activation would have been be expected. However, stability demonstrated the need for increased trunk stiffness at increased load heights. Recruiting trunk stiffness through increased extensor activation alone would violate equilibrium constraints; so it was necessary to augment trunk stiffness through a concomitant increase in flexor and extensor activation. Results suggest the neuromuscular system responds to changes in biomechanical stability. This was achieved by modifying stability without changes in equilibrium conditions or trunk moment. Because the applied moment was constant at all heights, any changes in flexor co-activation must have been due to the changing stability requirements of the system.

Given the magnitude and height of the external load, flexor stiffness required for stability as well as the extensor force necessary for equilibrium can be predicted. Results agree with the theoretical analyses and measurements presented by Cholewicki et al (Cholewicki J. et al 1998). In their study there was no attempt to empirically discriminate between the effects of trunk moment and stability. Nonetheless, results presented by the authors demonstrate increased levels al 1995) but are difficult to validate because of their complexity. Nonetheless, principles of muscle recruitment as a function of stability versus equilibrium are readily understood and easily validated using the simple biomechanical analyses described above. Furthermore, the simple model of stability accurately predicted the activation trends observed in the measured data.

Stability is a controllable neuromuscular function. A fundamental assumption of this model was that the stability parameter, S, remained constant throughout the experimental testing session of each subject. Static mechanical stability requires the stability parameter, S, in equation 3 to be greater than zero but does not constrain the positive magnitude. It is clear from equation 5 that increased co-activation can be used to voluntarily modulate stability, i.e. the depth of the potential energy well at any given equilibrium condition. Unfortunately, the cost of increased stability is caloric energy expenditure and fatigue necessary to recruit co-contraction above the minimum equilibrium values. Consequently, most theoretical analyses assume "critical stability", i.e. the minimum value of stability represented by equating S to zero. It is likely that stability criteria are balanced against energy cost (Cholewicki J., and McGill S.M. 1996) wherein a small stability value greater than zero is maintained as a safety margin while simultaneously limiting excessive energy expenditure. The ability to voluntarily modulate S may have introduced variability into the measurements and results. Nonetheless, statistical ANOVA and power analyses confirm that the neuromuscular system responds to changes in biomechanical stability of the trunk through control of antagonistic co-contraction. Future research must examine the biomechanical and psycho-physical factors influencing the stability safety margin, S.

Independent analyses demonstrate that both height and weight caused significantly increased flexor co-activation. Since increasing either of these two factors raises the potential 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 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, greater effort and greater potential for fatigue in the female subjects for these specific lifting tasks. Males continue to represent the majority of employees performing heavy manual materials labor and therefore report significantly more workplace low-back injuries than females (Hillman M. et al 1996; MacDonald M.J. et al 1998). However, 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 injury (Feuerstein M. et al 1997; Jones B.H. et al 1993; Krause C. et al 1997; Macfarlane G.J. et al 1997). To improve gender inclusion in the workplace it is necessary to understand the cause of this gender bias. Stability requirements in workplace design may influence this risk.

Four factors may have contributed to gender difference in stability in our study. First, anatomical parameters such as trunk weight, center-of-mass and muscle moment arms are different in males and females, N.B. the female subjects were significantly ($p<.01$) smaller and lighter than the male subjects. However, reduced body weight and height would predict reduced average flexor antagonism according to the model equation 6. Conversely, reduced muscle moment arm versus muscle length, r_e/h, would predict increased co-contraction. Second, recent evidence (Granata K.P. et al 2000) indicates women may demonstrate reduced active muscle stiffness, e.g. q in equation 6, requiring increased flexor co-contraction to maintain stability.

Third, lumbar lordosis may have been significantly different in the male and female subjects. This can be roughly simulated by modifying the ratio of r_e / h, predicting a change in stability and flexor recruitment. Finally, the experimental protocol did not account for potential gender differences in the force-versus-EMG relationship. If female subjects demonstrated a stronger relation or steeper regression slope between force and EMG then muscle forces would appear as greater EMG activity in the women, particularly notable at the increased force levels associated with the higher barbell elevations. We could find no reported evidence suggesting inherent gender difference in the force-EMG relation. The probability of a random assignment of ten female subjects such that the force-EMG relation was significantly greater than in a group of ten 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 LBD risk and biomechanical 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

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<u>LIST OF TABLES AND FIGURES:</u>

Table 2

	Effect Size (mV)	Flexor EMG Activity (mV)
Height	6.0	
0 cm		9.4 ± 3.3
20 cm		10.1 ± 3.8
40 cm		14.1 ± 6.8
60 cm		18.9 ± 11.3
80 cm		19.8 ± 12.6
Weight	6.5	
4.5 kg		10.8 ± 4.6
9.0 kg		18.1 ± 11.5
Gender	6.5	
Male		12.5 ± 6.5
Female		16.5 ± 11.4

* Effect size calculated for a power of 0.80
Vertical lines indicate statistical similarity in post hoc analysis

Figure 1

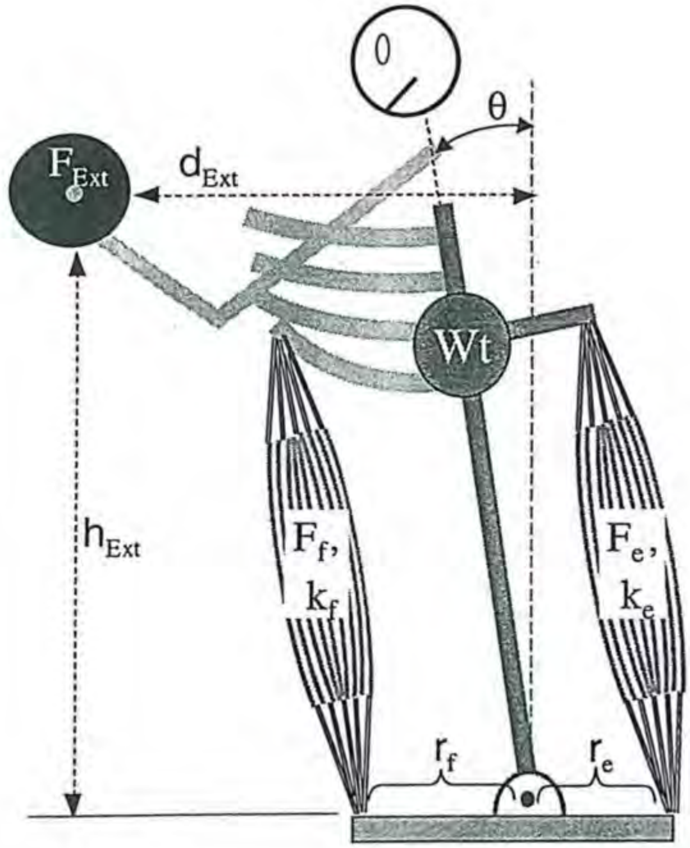
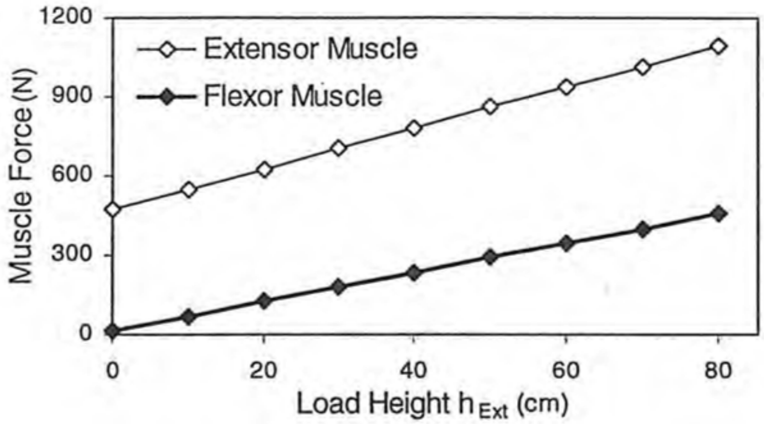


Figure 3



Extensor Muscle
Flexor Muscle
Muscle Force (N)
1200
900
600
300
0
0
20
40
60
80
Load Height h Ext (cm)

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Cost–Benefit of Muscle Cocontraction in Protecting Against Spinal Instability

Kevin P. Granata, PhD,* and William S. Marras, PhD†

Study Design. Lifting dynamics and electromyographic activity were evaluated using a biomechanical model of spinal equilibrium and stability to assess cost–benefit effects of antagonistic muscle cocontraction on the risk of stability failure.

Objectives. To evaluate whether increased biomechanical stability associated with antagonistic cocontraction was capable of stabilizing the related increase in spinal load.

Summary of Background Data. Antagonistic cocontraction contributes to improved spinal stability and increased spinal compression. For cocontraction to be considered beneficial, stability must increase more than spinal load. Otherwise, it may be possible for cocontraction to generate spinal loads that cannot be stabilized.

Methods. A biomechanical model was developed to compute spinal load and stability from measured electromyography and motion dynamics. As 10 healthy men performed sagittal lifting tasks, trunk motion, reaction loads, and electromyographic activities of eight trunk muscles were recorded. Spinal load and stability were evaluated as a function of cocontraction and trunk flexion angle. Stability was quantified in terms of the maximum spinal load the system could stabilize.

Results. Cocontraction was associated with a 12% to 18% increase in spinal compression and a 34% to 64% increase in stability. Spinal load and stability increased with trunk flexion.

Conclusions. Despite increases in spinal load that had to be stabilized, the margin between stability and spinal compression increased significantly with cocontraction. Antagonistic cocontraction was found to be most beneficial at low trunk moments typically observed in upright postures. Similarly, empirically measured antagonistic cocontraction was recruited less in high-moment conditions and more in low-moment conditions. [Key words: back, cocontraction, electromyography, low, model, spine, stability] **Spine 2000;25:1398–1404**

The role of trunk muscle cocontraction in lifting mechanics and spinal injury is poorly understood. Empirical measures have demonstrated significant muscle activity in the trunk flexor muscles during extension or lifting tasks.[49] Cocontraction may add protection against low back disorders (LBDs) by improving spinal stability.[6,12,21,40,47] However, cocontraction also contributes to spinal load[17] which has been cited as a risk factor for low back disorders.[20,36,37]

Cocontraction contributes to increased biomechanical stability.[6,12] Low back injury, low back pain, or both are thought to occur when spinal load exceeds tissue tolerance.[20,33,37] Vertebral tissue failure may be resisted at compressive loads up to 12,000 N,[3] with national standards advising against spinal compression in excess of 6400 N.[36] However, failure of the unsupported spinal column can occur as a result of mechanical instability at compressive loads less than 100 N.[7,9] Stability failure therefore may occur at spinal loads considered safe from a tissue tolerance standpoint. By recruiting antagonistic cocontraction of the trunk muscles, spinal stability can be improved[6,12] allowing the structure to withstand extreme compressive loads safely.[13] Recognizing the relation between cocontraction and stability[6,12] as well as the proposed relation between stability and LBD,[5,38,39] it may be hypothesized that antagonistic cocontraction can reduce the risk of low back injury by increasing spinal stability.

Spinal load also increases with antagonistic cocontraction during lifting exertions. Measurements demonstrate that trunk flexors cocontract simultaneously with the extensors during lifting tasks.[30,45,49] This cocontraction significantly influences spinal load,[23,28,44] accounting for 26% to 45% of the total compressive load.[17] Cocontraction is increased in high-risk lifting tasks such as in dynamic, asymmetric,[30,31] lateral,[27] twisting exertions.[28] Therefore, spinal load and the associated risk of overload injury also is increased in high-risk lifting tasks.[14,15,18,27,28]

The increased spinal load associated with antagonistic cocontraction challenges the stability of the spinal structure (*i.e.*, added load requires a greater stabilizing effort). For cocontraction to be considered beneficial, biomechanical stability must increase more than spinal load. Otherwise, it may be possible for cocontraction to generate spinal loads that cannot be stabilized. It remains to be demonstrated whether increased stability at the cost of increased spinal load is beneficial.

The objective of this research was to examine the influence of trunk muscle coactivity on stability of the spine relative to applied spinal load. Stability was quantified in terms of the maximum spinal compression that could be stabilized (*i.e.*, maximum stable load), as determined from *in vivo* measures of muscle activity. The *stability margin* was defined as the difference between the maximum stable load and the applied spinal load. It was

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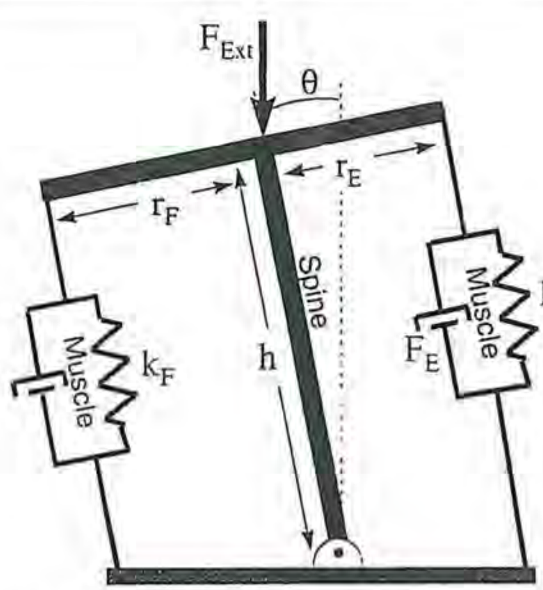


Figure 1. Simple model of spinal stability including trunk flexor and extensor muscle equivalents.

hypothesized that the stability margin would increase with antagonistic muscle cocontraction in the trunk flexors.

Background

The spine can be modeled as an inverted pendulum,[2,6] with a vertical external force, F_{Ext}, applied to the top, requiring muscular force to maintain equilibrium and muscular stiffness to maintain stability (Figure 1). Static equilibrium is achieved when the moment caused by the external force is balanced by the sum of moments caused by the extensor and flexor muscle forces, F_E and F_F, (1),

$$\sum M = F_E r_E - F_F r_F - F_{Ext} h\theta = 0 \qquad (1)$$

where angle θ and distances r_E, r_F, and h are described in Figure 1 and small angle approximation have been applied to simplify this discussion.

A system in equilibrium is said to be stable if it returns to equilibrium when perturbed.[16] Hence, any change in external moment resulting from small-angle perturbations must be offset by a change in internal, muscle-generated moments. The change in external moments are related to system geometry. Changes in muscle-generated moments are related to stiffness-based forces, in which muscles behave as nonlinear mechanical springs, and muscle stiffness, k, is linearly related to the equilibrium muscle force[1,24] and inversely related to length,[2,13] as expressed the following equation:

$$k = q\frac{F}{L} \qquad (2)$$

System behavior can be described from the change in external, flexor, and extensor moments:

$$\frac{dM_E}{d\theta} + \frac{dM_F}{d\theta} + \frac{dM_{Ext}}{d\theta} = q\frac{F_E}{L_E}\frac{dL_E}{d\theta}r_E + q\frac{F_F}{L_F}\frac{dL_F}{d\theta}r_F - F_{Ext}h \geq 0 \qquad (3)$$

where $dL/d\theta$ is the change in muscle length and q is a stiffness proportionality constant reported in the range of 5 to 30.[5,6,13]

It is common[2,5,12,13] to solve for the muscle stiffness coefficient, q, that minimally satisfies the stability condition (*i.e.*, critical stability). It also is possible to solve for the maximum external force that can be stabilized in the current equilibrium state[7,9] for a constant value of q. With the spinal load recognized as the vector sum of muscle and external forces, the maximum stable (spinal) load, F_Z^M, can be determined. This maximum stable load is

$$F_Z M \approx F_F q\, r_F\left(\frac{r_E + r_F}{h^2}\right) \qquad (4)$$

where the spatial derivatives of L_E and L_F are approximately r_E and r_F, respectively, and $L_E \approx L_F \approx h$ in near upright postures. Equation 4 shows that stability of the biomechanical system is proportional to the antagonistic cocontraction force, F_F, during an extension exertion. Antagonistic flexor force during an extension exertion will increase the maximum stable load more than the compressive load only if

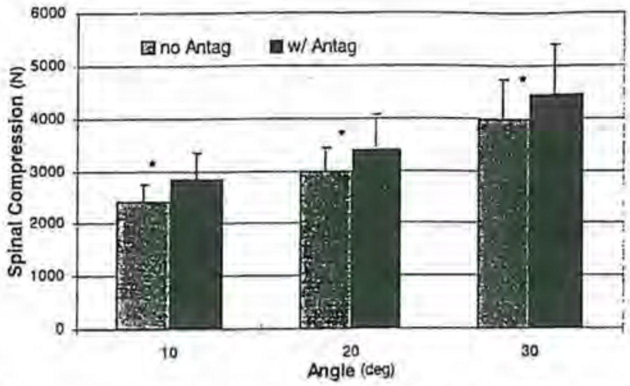
$$\frac{F_Z^M}{F_Z} \approx q\frac{r_E r_F}{h^2} > 1, \qquad (5)$$

In other words, antagonistic cocontraction is beneficial only when the ratio described in Equation 5 is greater than 1. Clearly, the cost–benefit of added antagonistic activity depends on the stiffness proportionality constant, q, and the kinematics of the lifting exertion. As more muscles and biomechanical realism are added, the model becomes increasingly complex. However, the concept of using maximum stable load to quantify the relation between stability and spinal load remains applicable even in more complex biomechanical models. A three-dimensional, electromyography (EMG)-assisted model of spinal load and stability was developed to examine the stability margin with greater biomechanical realism.

Methods

Model. An EMG-assisted model was developed to investigate the influence of muscle cocontraction during dynamic lifting tasks. The model included analyses of dynamic equilibrium and global stability. The equilibrium component generated spinal loads, muscle forces, and muscle kinematics from measured trunk motion, external trunk loads, and conditioned EMG signals, and has been reported extensively.[14,15,27,28]

Briefly, the spine was modeled as a three-degree-of-freedom inverted pendulum with muscle insertions along the iliac crest, vertebral transverse processes, and rib cage. Modeled muscles included the right and left erector spinae, internal obliques, external obliques, and rectus adbomini. Muscle kinematics were determined by vector rotations of the insertion points on the basis of trunk motions measured during the dynamic lifting tasks.[26] Muscle forces and associated moments were determined by satisfying dynamic equilibrium conditions and simul-



* indicates statistical significance between the two co-active conditions for each angle

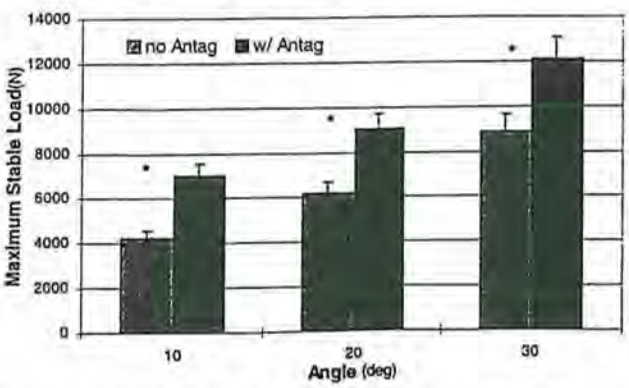
Figure 3. Spinal load increased significantly with trunk flexion angle and antagonistic cocontraction.

scaled by the erector spinae activity, the added variability reduced the statistical significance to $P < 0.06$ (Table 1). Relative coactivity in the external obliques also decreased with trunk flexion.

Spinal load increased by 12% to 18% when antagonistic muscle coactivity was included in the model (Figure 3). Spinal load also increased significantly with increased trunk flexion angle. Increased flexion moments with increased trunk angle required greater activity from the extensor muscles to achieve equilibrium, resulting in greater spinal load. Trunk flexion angle and coactivity both influenced spinal compression significantly (Table 1).

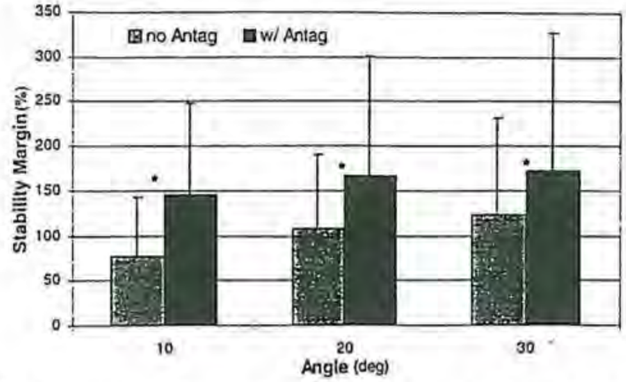
Spinal stability in terms of critical or maximum stable load increased by 36% to 64% as a result of antagonistic cocontraction (Figure 4). Stability also increased significantly with increased flexion angle (*i.e.*, structural stability improved with a posture of greater trunk flexion) (Table 1). Increased extensor muscle force associated with antagonistic cocontraction and trunk flexion resulted in greater muscle stiffness. The gains in stiffness generated improved biomechanical stability of the spine.

Although stability and spinal load both increased with antagonistic cocontraction, analyses of the stability mar-



* indicates statistical significance between the two co-active conditions for each angle

Figure 4. Spinal stability, defined as the maximum compressive load that could be stabilized, increased with trunk flexion angle and antagonistic cocontraction.



* indicates statistical significance between the two co-active conditions for each angle

Figure 5. Stability margin (*i.e.*, the difference between maximum stable load and spinal compression (F^{C_z}–F_z) was expressed as a percentage of the spinal compression. Stability margin increased significantly with antagonistic muscle cocontraction.

gin demonstrated that the increase in stability was significantly greater than the concomitant increase in spinal load (Table 1). Hence, the overall effect of cocontraction served to reduce risk in terms of spinal load *versus* stability (Figure 5). The stability margin increased with increased trunk flexion angle, but at a significance of $P < 0.07$ it failed to reach the *a priori* level of statistical significance.

■ Discussion

Improved spinal stability may be achieved by recruiting antagonistic cocontraction.[6,12] Cocontraction also contributes to increased spinal load,[17,28,44] which challenges the stability of the spinal structure (*i.e.*, added load requires a greater stabilizing effort). For cocontraction to be considered beneficial, the maximum stable load must increase more than the applied load. Otherwise, spinal load may exceed stability tolerance when cocontraction is recruited. It was hypothesized that the stability margin, defined as the difference between the maximum stable load and the applied spinal load, typically increases with antagonistic muscle cocontraction in the trunk flexors.

Spinal compression increased 12% to 18% with antagonistic activity in the flexor muscles of the trunk. Similar results have been reported in the literature at comparable lifting velocities.[10,12,17] Thelen et al[44] predicted a load increase of 220 to 575 N as a result of cocontraction during static flexion–extension exertions, which is consistent with the current results demonstrating a mean compression increase of 440 N.

Although many have suggested the stabilizing role of antagonistic cocontraction, only two published reports have quantified the relation. Using a two-muscle, one-degree-of-freedom model Cholewicki et al[6] successfully predicted the antagonistic activation necessary to maintain stability. A theoretical assessment by Gardner-Morse and Stokes[12] predicted a 50% reduction in the critical stiffness coefficient, q, when flexor cocontraction of muscle coactivity. Future implementation of more advanced models to portray biomechanical stability may permit greater understanding of the low back injury mechanisms. Future work also should examine the relation between cocontraction and stability during asymmetric lifting tasks wherein LBD risk[25,29,41] and muscle coactivity[15,30–32] are increased.

These results help to describe the biomechanical value of antagonistic cocontraction. Although the results were generated from a comparatively simple model of spinal motion and muscle cocontraction, the trends agree with those shown by others who have similarly demonstrated improved stability with cocontraction[6,12] and trunk increased flexion angle.[5] Using the biomechanical requirements of stability in addition to the traditional criterion of dynamic equilibrium, an improved understanding of motor control and trunk muscle cocontraction can be achieved.

Conclusions

There is a trade-off between the risk of injury associated with tissue overload and the risk of spinal instability. Trunk muscle cocontraction can be recruited to balance these risks. Model results suggest antagonistic cocontraction can be advantageous at low trunk moments by contributing to improved spinal stability. Similarly, empirical results demonstrated increased antagonistic coactivity when external moment was low (*i.e.*, in upright postures). Conversely, antagonistic coactivity was reduced when trunk moment was high (*i.e.*, in flexed postures). This helped to reduce the risk of spinal tissue overload injury when the stability was high as a result of increased muscle force and associated stiffness.

Acknowledgment

The authors thank K. Davis and B. Kirking for their assistance in these analyses.

Key Points

- Improved spinal stability may be achieved by recruiting antagonistic contraction.
- Spinal compression increased 12% to 18% with antagonistic activity in the flexor muscles of the trunk.
- Antagonistic cocontraction was found to be most beneficial at low trunk moments typically observed in upright postures.

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Lumbar–Pelvic Coordination Is Influenced by Lifting Task Parameters

Kevin P. Granata, PhD, and Adam H. Sanford, MS

Study Design. Low back kinematics, including relative lumbar and pelvic motions, were quantified during controlled lifting tasks.

Objectives. To evaluate the influence of load and lifting velocity on lumbar–pelvic (LP) coordination.

Summary of Background Data. Sagittal trunk extension is achieved through the coordinated motion of the pelvis and lumbar spine. There are no data to indicate whether lifting task design influences lumbar–pelvic coordination.

Methods. Lumbar and pelvic motions were recorded from 18 healthy subjects while performing isokinetic lifting tasks of 0.1 kg and 10 kg. Coordinated motions of the pelvis (sacral spine) and low-thoracic spine were evaluated using eigenvector analyses and a ratio of lumbar and pelvic angles (L/P).

Results. Eigenvector models of the lumbar–pelvic coordination accurately represented empirical coordination profiles. Weight significantly influenced lumbar–pelvic coordination. Trunk extension velocity demonstrated a small but statistically significant influence on lumbar–pelvic coordination. Weight and trunk flexion angle significantly influenced lumbar/pelvic angle ratios.

Conclusions. Trunk extension was achieved through simultaneous but nonlinear contributions from both the pelvis and lumbar spine throughout the range of motion. The lumbar spine accounted for 70% of the total, with increased pelvic contributions in flexed postures. Task weight increased the lumbar contribution to total trunk motion. When performing clinical evaluations of spinal kinematics, it is necessary to recognize that unloaded motions may not fully represent loaded behavior of spinal coordination. [Key words: low back, spine, pelvis, lifting, coordination] **Spine 2000;25:1413–1418**

Sagittal trunk motion is achieved through the coordinated rotation of the pelvis and flexion/extension in the lumbar spine. Subjects with no history of low back pain (LBP) demonstrate repeatable and consistent patterns of lumbar–pelvic (LP) coordination, but the nature of the motion is debated.[9,13,31,32,35] There is also disagreement about the magnitude of lumbar and pelvic contributions to total trunk motion, with lumbar/pelvic ratios reported from 0.4–1.97.[18,35] Reported discrepancies may be related to the influence of lifting task design and the influence of lifting parameters such as weight and velocity on LP coordination.

Lifting task parameters may influence dynamic LP coordination. Increased lumbar contribution to LP coordination influences dynamic spinal posture throughout the motion. Posture influences spinal load, nutrient transfer to the discs and associated risks of low back pain.[1,36] Although biomechanical analyses of lifting tasks typically represent trunk muscles as force (tension) generating elements,[4,14,28,37] recent studies have demonstrated the need to recognize the elastic behavior of trunk muscles to maintain spinal stability.[2,5,6,10,11,15] The elastic characteristics of active muscle tissues suggest that the weight of the load lifted and the dynamics of trunk motion can influence muscle length, similar to the way that the weight and dynamics of a mass will affect the length of a supporting spring. Recognizing that dynamic muscle length influences lumbar curvature and that lumbar curvature is a principle component of LP coordination, lifting task parameters may affect LP coordination.

Measures of LP coordination rarely control for dynamic and kinetic parameters. Mayer et al[26] reported lumbar–pelvic parameters during static exertions, Paquet et al[34] asked subjects to perform cyclic flexion/extension at "natural and comfortable" and "slower" cadences, Porter and Wilkinson[35] recorded motions during a single flexion cycle lasting 10 seconds, and Esola et al[8] had subjects move at a "self-selected velocity." Whereas most studies have examined unloaded exertions, Nelson et al[31] asked subjects to lift a 9.5-kg weight. Only two studies have considered the influence of load on spinal posture. Gracovetsky et al[12,13] reported increased "estimated intersegmental mobility" between vertebrae with increased load, but changes in lumbar lordosis failed to reach significance. Conversely, Mitnitski et al[30] reported that load increases lumbar lordosis at the onset of the extension motion. The coordination between lumbar spine motion and pelvic rotation is difficult to ascertain from these measurements. Hence, few studies have investigated the influence of weight on LP behavior, and they fail to achieve consensus. None have investigated the influence of velocity on LP coordination.

Differences in LP coordination between LBP patients and asymptomatic controls may be confounded and obscured by the manner in which tasks are performed. There is evidence to suggest that LP coordination may provide an effective means for discriminating between LBP and asymptomatic populations.[8,27,33,35] Research has demonstrated, however, that LBP patients move more slowly than asymptomatic control subjects.[25,29]

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Table 1. Statistical Results

	Weight (kg)		Velocity (°/sec)			Angle Range (°)		
	0.1	10	15	30	60	5–30	30–60	60–90
L/P ratio	2.13 ± 1.74	2.45 ± 2.39*	2.31 ± 1.97	2.22 ± 2.06	2.06 ± 1.30	2.37 ± 2.66	2.35 ± 1.79	1.84 ± 1.04*
Eigen coef 1	180.02 ± 45.95	187.02 ± 46.68*	184.90 ± 45.48	179.90 ± 49.01	185.80 ± 42.21			
Eigen coef 2	0.52 ± 12.52	1.52 ± 17.05	1.80 ± 16.08	2.08 ± 16.76	−1.24 ± 11.46			
Eigen coef 3	−0.68 ± 3.92	1.06 ± 5.51	−2.56 ± 4.69*	1.05 ± 4.81	2.30 ± 3.36			
R^2	0.996 ± 0.014	0.997 ± 0.011	0.995 ± 0.013	0.996 ± 0.018	0.998 ± 0.003			
RMS error (% mean)	1.73 ± 1.21	1.89 ± 1.26	2.20 ± 1.28*	1.70 ± 1.37	1.52 ± 0.92			

* Statistically significant ($P < 0.05$) differences within the category.
Coef = coefficient; RMS = root-mean-square.

vidual subject and trial. Thus, the shape of individual LP curves could be quantified and compared by evaluating the multiplicative coefficients. A linear model of the LP coordination also was implemented by least means square solution of the measured LP data to a straight line. To compare with previously published LP measurements, the mean ratios of lumbar and pelvic angles, L/P, were computed for trunk flexion angles of 0 to 30°, 30 to 60°, and 60 to 90°.[8,26,27,31,35,38] Recognizing that the exertions were extension tasks, the 60 to 90° range describes the early phase of the lift, the 30 to 60° range the midrange of the lift, and the 0 to 30° range describes the terminal phase of the motion.

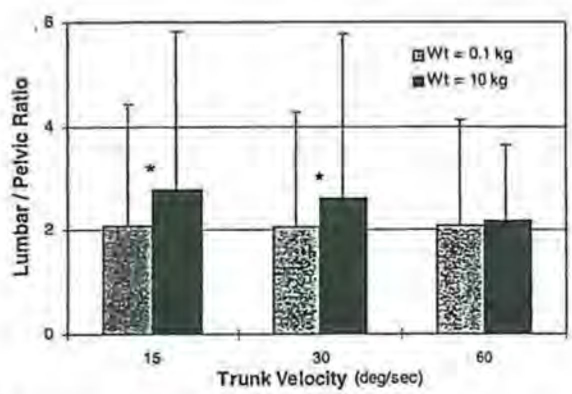
Independent variables included lifting velocity and weight of the lifted load. Dependent variables included coefficients from the linear and eigenvector models describing the LP coordination profiles and the mean L/P ratios at each trunk flexion range. Both eigenvector and linear model estimates were compared with empirical LP data using correlation analyses (R^2) to assess accuracy of estimated shape and root-mean-square (RMS) error to assess magnitude accuracy. Repeated measures (within-subject) analysis of variance (ANOVA) were performed to assess statistical significance. An added advantage of the within-subject design was that it helped to reduce potential intersubject error from sensor placement and skin motion artifact. Significance was noted for all effects with $\alpha < 0.05$.

Results

The mean ratio of lumbar to pelvic angle was 2.38 ± 2.68, demonstrating that the lumbar spine typically accounted for 70% of the total trunk motion in these subjects. Coordination between the pelvis and lumbar spine changed throughout the lifting motion as demonstrated by statistically significant reduction in L/P ratio with increasing trunk flexion (Table 1, Figure 2). This behavior also is observed in a plot of the lumbar angle *versus* the pelvic angle (Figure 3). Note that the slope of the L *versus* P profile decreases from left to right, suggesting that the L/P ratio decreases with trunk angle.

Eigenvector estimates of the LP profiles accurately represented the empirical data, with a mean correlation coefficient of $R^2 = 0.996 \pm 0.013$. Average RMS error, representing the magnitude of differences between the estimated and empirical data, were 1.8 ± 1.2% of the mean. Evaluation of the eigenvector coefficients confirmed the correlation analyses. The first eigenvector and associated trial-specific weighting coefficient described more than 95% of the LP coordination variability. This eigenvector described the relative contribution to total trunk motion from lumbar and pelvic motions. Addition of the second and third eigenvectors increased the accuracy to more than 99.5%. The second eigenvector modified the extent to which the pelvic motion leads the lumbar motion. The third eigenvector tended to increase the initial acceleration and endpoint deceleration of the lumbar spine compared with the motion of the pelvis. The accuracy from the eigenvector model generally was not influenced by lifting conditions (Table 1) except for a statistically significant increase of 0.43% RMS error during 15 deg/sec motions.

The eigenvector model performed significantly better than linear models when representing the dynamic relation between lumbar and pelvic angles (Table 2). Improved performance from the eigenvector model suggests that the coordination between lumbar and pelvic motions was nonlinear. These results are supported by statistically significant changes in the L/P ratio as a function of trunk angle range (Table 1), demonstrating the



* indicates statistically significant differences between weight conditions within each angle.

Figure 2. The mean ratio of lumbar *versus* pelvic flexion within the trunk angle ranges of 0 to 30° flexion, 30 to 60° flexion, and 60 to 90° flexion illustrate a significant increase in lumbar contribution with added weight at velocities of 15 degree/sec and 30 degree/sec.

1.70 in these ranges of motion. Lumbar/pelvic ratios were significantly reduced at greater trunk flexion angles, as also reported by Esola et al[8] and McClure et al.[27] The L/P magnitudes reported Esola et al[8] and McClure et al,[27] however, were approximately half of the values in the present study. This difference is likely because of the definition of pelvic angle. Whereas the present data and Porter and Wilkinson[35] define pelvic angle relative to a ground reference, Esola et al[8] and McClure et al[27] reported the angle between the thigh and pelvis. Their subjects were free to flex their knees during the lifting task, and the pelvic–leg angle is likely influenced by leg motion, thereby reducing the LP ratio. Thus, despite protocol differences, the results of the present study agree in trend, and in some cases in magnitude, with values reported in the literature.

Lumbar and pelvic motions occurred simultaneously. Farfan[9] suggested that trunk motion is generated by sequential contributions from the lumbar and pelvic systems. If this were true, then the plot of the lumbar *versus* pelvic motion would show a region of pure vertical alignment, *i.e.*, lumbar motion without pelvic rotation. Similarly, pure pelvic motion without concomitant lumbar extension would be illustrated by an horizontal region in the LP profile. Neither of these conditions were observed (Figure 3), indicating simultaneous motion. It is interesting to note, however, that the coordination was dominated by pelvic motion in extreme flexion angles. Measurements of LP coordination at flexion angles greater than 90° demonstrated similar behavior, *i.e.*, pelvic contribution to trunk motion increased with angle but failed to demonstrate purely sequential coordination.[8,27,35,38] Nelson et al[31] concurred that the motions are simultaneous, but noted that extension tasks were characterized by a greater level of sequential coordination, whereas flexion tasks were more simultaneous.

Results suggest pelvic motion leads the lumbar motion during the extension task, as noted by the statistically reduced L/P ratio in flexed postures (Table 1) and the flattening of the LP profile in flexed postures (Figure 3). This timing effect was unaffected by weight, *i.e.*, the second and third eigenvector coefficients were unchanged with weight, and there was no significant interaction between weight and trunk angle when evaluating L/P ratios. By measuring the height of the lifted weight throughout a trunk extension task, Davis et al[7] concluded that lumbar motion was delayed with increased load when using a "bent knee" lifting technique, but found no statistically significant effect in lumbar motion delay when using the straight leg or "stoop" technique. Although Davis et al[7] did not measure lumbar–pelvic coordination directly, the results support the findings of the present study and suggest the need for future research to investigate the influence of "bent knee" postures on lumbar–pelvic timing.

The influence of trunk angle on L/P ratio was statistically significant (Table 1), confirmed by the fact that nonlinear representations of LP coordination, *i.e.*, eigenvector models, were superior to linear estimates (Table 2). These results suggest a transfer of motion from the lumbar spine to the pelvis with increasing flexion. Several reports have observed this behavior, but only Esola et al[8] and McClure et al[27] have statistically validated the effect of angle on LP coordination. Farfan[9] and Gracovetsky et al[12] concluded that ligamentous tissues resist lumbar flexion even at relatively small flexion angles. Ligamentous resistance to spinal flexion suggests pelvic motion must dominate when in flexed postures, illustrated by the results of the present study. Although measurements were limited to 90° of flexion, reports have indicated that the transfer of motion from lumbar flexion to pelvic rotation continues to increase as subjects approach their maximum flexion angle.[8,27,31,35,38]

Trunk extension velocity statistically modified LP coordination, but the size of the effect was very small. The L/P ratio during the slow, 15 degree/sec motions was significantly greater than the faster exertions, but only under weighted conditions. Velocity influenced the third eigenvector coefficient, increasing the simultaneity of the lumbar and pelvic coordination during the lift. The third eigenvector accounted for less than 0.4% of the total variability, however, suggesting that velocity effects on LP coordination can be largely neglected.

Weight lifted during trunk extension tasks influenced LP coordination. The lumbar spine contributed more to the total trunk motion during the 10-kg lifting tasks than during the unloaded (0.1 kg) lifts. During the 10-kg lifting tasks, L/P ratios increased significantly in each of the trunk flexion ranges. Similarly, the first eigenvector coefficient increased significantly with weight (Table 1), suggesting increased lumbar motion per unit of pelvic rotation.

Although these results may be the first to directly measure the influence of weight on LP coordination, the effect might be expected from previous evaluations of lumbar flexion. Comparing the three-dimensional relation between sacral and low-thoracic angles, Marras et al[19,24] concluded lifting task parameters as well as age and gender influence lumbar spine behavior. Gracovetsky et al[13] demonstrated that weight affects intersegmental mobility during trunk extension tasks, and Mitnitski et al[30] reported that lumbar lordosis was significantly affected by lifted weight, *i.e.*, increased lumbar kyphosis at peak trunk flexion. Thus, the results of the present study agree with published research and illustrate that dynamic lifting parameters can influence the motion behavior of the lumbar spine as well as lumbar–pelvic coordination.

Clinical assessments of low back pain often include a kinematic evaluation, some as simple as measuring maximum trunk flexion, others involving more complex quantification of multidimensional dynamics.[23] Differences in L/P ratios between asymptomatic controls and LBP patients have been reported during both loaded and unloaded tasks, suggesting a difference exists regardless of the loading.[8,27,31,35] Low back pain patients, however, tend to move more slowly than asymptomatic con-

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Structural Injury Tolerance of the Spine: Stability and Lumbar Lordosis in Lifting

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INTRODUCTION

Stability of the lumbar spine was biomechanically modeled to examine the influence of low-back posture, i.e. lordosis, and trunk flexion on the tolerance to structural failure and injury

REVIEW AND THEORY

Traditional advice for lifting tasks suggests the maintenance of a lordotic curvature during the exertion, i.e. the "squat" lifting technique. Analyses demonstrate this reduces ligament and intervertebral shear loads on the spine (Potvin et al, 1991). Conversely, lifts performed with a flattened or kyphotic spine, i.e. stoop lift, are associated with reduce fatigue, greater strength, higher performance levels, and reduces spinal compression (Garg et al, 1985; Hagen et al, 1993; Kumar et al, 1992). An intermediate posture is indicated for optimal transfer of nutrients and mechanical load through the intervertebral disc (Adams et al, 1985). These studies have consider material tolerance and performance issues. However, stability tolerance associated with spinal curvature remains unknown.

Analyses of spinal stability suggest the spine may buckle and fail at compressive loads significantly lower than the material tolerance of the spine (Crisco et al, 1992). Although NIOSH (1981) recommends a compressive limit of 6400 N, measurements demonstrate the spine can suffer from structural failure, i.e. unstable buckling, at compressive loads less than 100 N. It is the task of the trunk musculature to maintain spinal stability, thereby allowing the spine to withstand loads approaching the material tolerance limits.

Clearly the shape of the lumbar spine will influence the stability of the structure (Gardner-Morse et al, 1995). Hence, the risk of structural failure and spinal injury is related to the lordotic curvature as a function of the applied load and muscle stiffening capacity. Although the biomechanical load and material tolerance of the spine has been examined as a function of lifting technique, there are no published analyses documenting the influence of lifting method on spinal stability considering the biomechanical effects of stability during lifting.

In the current effort, two hypotheses were examined [illegible] during lifting exertions is influenced by the lordotic curvature. 2.) The curvature of the spine necessary to maintain stability changes with trunk flexion angle. The purpose of this research was to biomechanically model spinal stability and evaluate these two hypotheses.

METHODS

A 15-degree of freedom biomechanical model of spinal stability was developed to represent the three-dimensional behavior of the lumbar spine. The model included a pelvic/sacral base, six vertebral elements, and eleven muscle equivalents. Stability analyses were performed by solving the generalized eiganvalue problem resulting from the Hession matrix of potential energy. The methods are similar to those described elsewhere (Cholewicki et al, 1996; Gardner-Morse et al, 1995). The eiganvalue analyses yield a constant representing the muscle stiffness rate, Q (Gardner-Morse et al, 1995), necessary as a minimum requirement for stability. Increased values of Q suggest reduced spinal stability, requiring greater stabilizing effort from the trunk musculature.

Stability analyses were performed for lumbar lordotic curves from 0^0 (straight spine) to 20^0. Lordosis was defined in the upright posture. Trunk flexion was modeled as an incremental flexion of each vertebrae from the upright lordotic posture according to the distribution described in McGill and Norman (McGill et al, 1986). At each lordosis/angle combination, a value of Q was determined to describe the relative stability of the static spinal posture.

RESULTS

Review of the literature suggests physiologic value of muscle stiffness rate, Q, ranges from 250 to 1000 Nm/deg per Newton of muscle force. Except for extreme conditions, e.g. lordosis=0^0, flexion>15^0, modeled values of Q agreed with the physiological levels.

Biomechanical stability (associated with reduced Q) increased with lordosis, asymptotically approaching a constant Q value for each flexion angle (Figure 1). The stabilizing effort also increased with trunk flexion angle. At greater trunk

QUANTIFICATION OF LUMBAR-PELVIC COORDINATION IN HEALTHY ADULTS

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INTRODUCTION

Sagittal trunk motion is accomplished through a coordination of pelvic rotation and lumbar spine flexion/extension. As trunk range of motion in healthy subjects is not fundamentally different to persons with low-back pain (LBP), other means of discriminating the two populations must be used. Research suggests that lumbar-pelvic (LP) coordination could be a potential diagnostic metric for LBP. Though LP coordination has been studied at length, there is little consensus regarding the gross contributions of the lumbar spine and pelvis to total trunk motion; literature values for lumbar to pelvic (L/P) ratios range from 0.4 to 1.97. Such a spread of literature values would pose difficulties for the comparison of healthy and LBP populations. Previous studies have not controlled lifting parameters such as weight or trunk extension velocity as possible influences of LP coordination. A study of the effect of lifting parameters upon LP coordination was performed to provide a complete description of lifting behavior and to assess possible sources of error in previous quantifications of LP coordination.

PROCEDURES

18 subjects (13 M, 5 F) with no prior history of LBP volunteered to participate in the study. Subjects performed lifting exertions at 15, 30 and 60 deg./s trunk extension velocity with 0 and 10 kg loads carried in the hands. To control trunk extension velocity, subjects were required to follow a target region in a real time display that was pre-programmed to travel at the desired trial velocity. Spinal motions were recorded from six degree of freedom electromagnetic sensors placed over the spinous processes of the S1 and T10 vertebrae. The shape and magnitude of the LP coordination was quantified using principal component analysis (PCA). For direct comparison to literature values, L/P ratios were calculated over three 30° windows as reported in previous studies. A repeated measures ANOVA was performed on all findings, and significance was noted for all effects with $\alpha < 0.05$.

RESULTS AND DISCUSSION

PCA results show that the weighting coefficient of the most powerful eigenvector was significantly affected by task weight. This eigenvector describes the basic lifting behavior of all subjects, so significance with weight describes a gross effect upon the lumbar and pelvic contributions to total trunk motion.

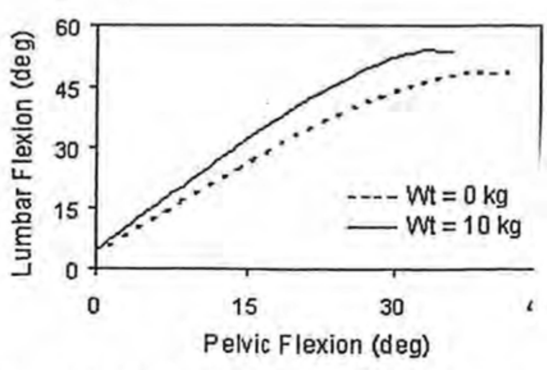

Fig. 1: Weight effect on LP Coordination

LBD RISK FACTORS AND THE STRUCTURAL STABILITY TOLERANCE OF THE LUMBAR SPINE

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A model of low-back injury is presented suggesting structural instability is a risk factor in occupation injury. The purpose of this research was to document the potential for spinal buckling as a function of asymmetric and sagittal trunk angle during lifting. A biomechanical model was developed to compute the Euler stability and determine the structural tolerance of the lumbar spine in work-related postures. When applied load and associated spinal compression exceeds the structural tolerance, i.e. buckling load, the spinal column fails. Analyses demonstrate the structural tolerance is often less than material tolerance estimates, particularly in flexed and asymmetric postures. Furthermore, the structural tolerance is reduced in sagittally flexed and asymmetric lifting postures. Hence, the relation between stability and trunk posture correlated with low-back disorder (LBD) risk factors. Results suggest musculoskeletal instability may help explain the relation between LBD risk and lifting posture.

INTRODUCTION

Occupationally-related low back disorders (LBDs) are the leading cause of lost work days and the most costly occupational safety and health problem facing industry today. It is well known that LBD risk is associated with manual materials handling (MMH) lifting tasks particularly when combined with lateral, bending and asymmetric postures (Punnet et al 1991; Marras et al 1995). However, a major limitation in controlling the incidence of LBDs is the inability to explain the injury mechanism to the lumbar spine.

Traditional assessments of spinal load often fail to explain injury. Cumulative injury is assumed to result when spinal load exceeds the material tolerance of the spine. However, high rates of injury are often associated with spinal loads significantly below biomechanical tolerance estimates and at levels considered safe according to NIOSH recommendations (Herrin et al 1986). Despite incidence rates indicating a significant LBD problem in the automotive industry, Punnett et al (1991) found less than three percent of the sampled jobs imposed static compressive forces greater than the 1981 NIOSH action limit. Granata et al (1996) demonstrated workers who had survived more than two years without low-back injury generated significantly greater spinal loads than their inexperienced counterparts. Moreover, analyses of controlled lifting tasks suggest injury tolerance cannot be exceeded even during maximum exertions (Hutton and Adams 1982). Although spinal load may correlate with LBD risk, as a single factor it often fails to explain injuries.

Structural failure and material failure represent two different injury mechanisms. Spinal tolerance is typically associated with the material failure, e.g. end-plate fracture, disc rupture, vertebral body injury. However, a multi-segment spine will become unstable and buckle, i.e. structural failure, at loads safely within the limits of material tolerance. An unsupported multi-segment spine will buckle at compressive loads as low as 100 N (Crisco and Panjabi 1992a; Crisco et al 1992b) whereas endplate fracture may be resisted up to 12,000 N (Chaffin and Page 1994). Hence, musculoskeletal instability and associated spinal buckling may be a significant contributor to LBD risk.

The purpose of this research was to document the potential for spinal buckling as a function of asymmetric and sagittal trunk angle during lifting. A model was developed to compute the stability and buckling load at which the lumbar spine will fail. Analyses demonstrate spinal stability is correlated with LBD risk factors, specifically task asymmetry and trunk angle. Moreover, the compressive load at buckling failure can be significantly less than the recommended limits associated with material tolerances. Results

BENEFITS OF TRUNK MUSCLE CO-CONTRACTION IN PROTECTING AGAINST LOW-BACK INJURY DURING MANUAL MATERIALS LIFTING

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There is a trade-off between the risk of low-back injury associated with tissue overload versus the risk of spinal instability. The benefit of antagonistic co-contraction in helping to prevent low-back pain associated with instability injury during manual materials handling tasks was examined. Ten healthy males performed sagittal lifting tasks while trunk motion, reaction loads, and EMG activity were recorded. A biomechanical model was developed to compute spinal load and spinal stability during the tasks. Antagonistic co-contraction was found to be beneficial in terms of the stability versus spinal load, i.e. compression increased 12 - 18% while stability increased 36 - 64%. Stability was a minimum at low trunk moments, e.g. upright postures. Conversely, as trunk moment increased the risk of stability failure was reduced but the risk of spinal tissue overload injury was increased. To compensate, subjects recruited antagonistic co-contraction less in high moment conditions and more in low moment conditions.

INTRODUCTION

The role of trunk muscle co-contraction in lifting mechanics and spinal injury is poorly understood. Significant muscle activity in the trunk flexor muscles during extension or lifting tasks have been demonstrated (Zetterberg C. et al 1987). Biomechanical analyses suggest this antagonistic activity may contribute to increased stability of the spine(Cholewicki J. et al 1998; Gardner-Morse M., and Stokes I.A. 1998), which may add protection against low-back disorders (LBDs) (Hodges P.W., and Richardson C.A. 1996; Wilder D.G. et al 1988). Antagonistic co-contraction also contributes significantly to spinal load(Granata K.P., and Marras W.S. 1995b) which has been cited as a risk factor for low-back disorders(NIOSH 1981; Norman R.W. et al 1998). Increased load applied to the spine through muscular co-contraction requires greater spinal stability for support. It is unknown whether the increased stability associated with antagonistic co-contraction is sufficient to meet the increased demands of spinal load. The goal of this research was to examine how the relation between spinal compression and spinal stability is influenced by antagonistic trunk muscle co-contraction during dynamic lifting tasks.

Injury and /or low-back pain is thought to occur when spinal load exceeds tissue tolerance(Norman R.W. et al 1998). Vertebral tissue failure may be resisted at compressive loads up to 12,000 N(Chaffin D.B., and Page G.B. 1994), with national standards advising against spinal compression in excess of 6400 N(NIOSH 1981). However, failure of the unsupported spinal column can occur as a result of mechanical instability at compressive loads less than 100 N(Crisco J.J. et al 1992). Thus, stability failure may occur at spinal loads considered safe from a tissue tolerance standpoint. Fortunately, the musculoskeletal system can voluntarily control spinal stability by recruiting antagonistic co-contraction of the trunk muscles(Gardner-Morse M., and Stokes I.A. 1998)·(Cholewicki J. et al 1998; Gardner-Morse M. et al 1995). Hence, antagonistic co-contraction may reduce the LBD risk by increasing spinal stability.

Antagonistic co-contraction also increases spinal load during lifting exertions. Measurements demonstrate co-contraction significantly influences spinal load(Granata K.P., and Marras W.S. 1995b; Marras W.S., and Granata K.P. 1996; Thelen D.G. et al 1995). Increased spinal load from muscle co-contraction are particularly evident in high-risk lifting tasks such as in dynamic, asymmetric(Marras W.S., and Mirka G.A. 1990) lateral(Marras W.S., and Granata K.P. 1995) and twisting exertions(Marras

W.S., and Granata K.P. 1996). Thus, the associated risk of overload injury may be increased in high-risk lifting tasks.

Increased spinal load associated with antagonistic co-contraction challenges the stability of the spinal structure, i.e. added load requires a greater stabilizing effort. For co-contraction to be considered beneficial, biomechanical stability must increase more than spinal load. Otherwise, it may be possible for co-contraction to generate loads that cannot be stabilized. The objective of this research was to examine the influence of trunk muscle coactivity on stability of the spine relative to the applied spinal load. Is the trade-off between spinal compression and spinal stability biomechanically beneficial? Thus, the research seeks to discern whether antagonistic co-contraction is biomechanically beneficial.

METHODS

An EMG-assisted model of biomechanical stability was implemented to investigate the cost-benefit of muscle co-contraction during dynamic lifting tasks. The model employed three components including a geometric model of musculoskeletal trunk dynamics, a model of musculoskeletal equilibrium, and a model of global biomechanical stability.

The geometric module represented the spine as a 3-degrees-of-freedom inverted pendulum with transverse sacral and thoracic planes for muscle attachments. Modeled muscles included the right and left erector spinae, internal obliques, external obliques, and rectus adbomini. Spine and muscle kinematics were determined by vector rotations of the insertion points based upon trunk motions measured from an electrogoniometer(Marras W.S. et al 1992) during dynamic lifting tasks. The musculoskeletal equilibrium model accepted input EMG data, three-dimensional external trunk load data, and muscle dynamics from the geometric module. Muscle forces and associated moments were determined by satisfying dynamic equilibrium conditions and simultaneously distributing the external trunk moments in relation to the conditioned EMG signals. Spinal load at the lumbo-sacral junction was computed from the vector sum of muscle and external forces. Details of the geometric and equilibrium components of the model have been reported extensively(Granata K.P., and Marras W.S. 1993; Granata K.P., and Marras W.S. 1995a; Marras W.S., and Granata K.P. 1995; Marras W.S., and Granata K.P. 1996).

The model of biomechanical stability determined critical stability, i.e. maximum load that could be stabilized on the current equilibrium condition. Muscle stiffness was determined from the quotient of muscle force accepted from the equilibrium module and muscle length. Elastic muscle force and associated trunk moments were determined from the product of stiffness and the spatial derivative of muscle length, $\Delta F = k \, \Delta L$ for each of the modeled muscles, three-dimensions of moment, and three-dimensions of perturbation. The changes in external moments were determined from partial derivatives of the moment arm vector. The solution was a 3-by-3 matrix describing the three-dimensions of critical stability associated with the three-dimensions of perturbation.

EMG, trunk motion, and external force and moment data were collected from ten healthy males with no history of low-back pain, age 21 to 35 years with mean (std) weight and height of 72.7 (6.6) kg and 176.7 (4.2) cm.. Subjects were asked to perform a sagittally symmetric lift with a 22.7 kg box from a platform 52 cm. from the floor and 51 cm. anterior to the ankles and place it on a second platform 107 cm from the floor and 25 cm. anterior to the ankles. Exertions were performed at freely selected lift rates while three-dimensional trunk motion data were recorded from an electrogoniometer(Marras W.S. et al 1992). Data collection methods for kinematic, kinetic and EMG data have been described elsewhere(Fathallah F.A. et al 1997; Granata K.P. et al 1995; Mirka G.A., and Marras W.S. 1993).

Analyses were performed to examine the relationship between trunk muscle coactivity and the increase in biomechanical stability versus spinal load. The influence of coactivation was demonstrated by running the model once using the full set of EMG data and comparing the results with analyses wherein the antagonistic coactivity were eliminated from the model. Independent variables included trunk flexion angle and coactivation level. Dependent variables included spinal load and maximum stable load. Analysis of variance (ANOVA) procedures were performed to assess statistical significance.

RESULTS

Antagonistic EMG activities and coactivities were significantly greater than zero in all conditions. Antagonistic activity from the rectus abdominis EMG activity and external obliques coactivity decreased significantly with trunk flexion angle (Table 1).

beneficial the maximum load the system can stabilize must increase more than the applied load. Otherwise spinal load may exceed stability tolerance when co-contraction is recruited. The goal of this research was to examine the relation between spinal compression and spinal stability as a function of antagonistic trunk muscle co-contraction.

Spinal stability increased significantly with trunk flexion angle and antagonistic co-contraction. Stability increased by 36% to 64% when co-contraction was included in the model. This represents a mean increase in the compression tolerance of 2925N. Stability also improved with flexion angle. Both trunk flexion and antagonistic co-contraction work in a similar manner to stabilize the system. Increased flexion moments from co-contraction and trunk flexion require increased extensor muscle force. Recognizing that active muscle stiffness is proportional to contractile force(Hoffer J.A., and Andreassen S. 1981), the increased flexion moment and associated extensor muscle force stiffened and stabilized the biomechanical structure.

Although antagonistic activity increased spinal compression up to 18% in the free-speed, sagittal lifting task, stability tolerance increased as much as 64%. Mean stability margin, expressed as a percent of the applied spinal load, was 103% without antagonism and 161% when co-contraction was included. If injury risk is associated with the relation between tolerance and load, then co-contraction reduced the risk of stability failure despite the fact that load was increased. It is interesting to note that without co-contraction the biomechanical system was unstable in the near upright postures, i.e. stability margin less than 100%. This indicates flexor antagonism is necessary to maintain stability in the upright postures. In an upright posture the trunk mass and lifted load present a low flexion moment and the force in the extensor muscles are insufficient to recruit the necessary mechanical stiffness for stability. Consequently, flexor muscles must be activated to increase the demand and stiffness in the trunk extensors to achieve stability. Accordingly EMG of the rectus abdominis increased significantly in more upright postures. Results indicate that muscle co-contraction is more beneficial ($p<.07$) in upright, low-moment postures. Hence, antagonistic co-contraction may be recruited to complement biomechanical need, i.e. increased activity in the upright posture to stabilize the system, reduced activity in flexed postures to reduce spinal load.

The benefit of antagonistic co-contraction in terms of stability must be balanced against the risks associated with increased spinal load. In vitro experiments have demonstrated increased inter-vertebral stability with co-contraction(Quint U. et al 1998; Wilke H.J. et al 1995). The measurements indicate that inter-vertebral stability may be improved by pure spinal load, in addition to the muscle stiffness behavior. However, there is little evidence that antagonistic activity will improve the instantaneous tissue tolerance of the spine. Hence, the requirement for biomechanical stability must be balanced against the risk of tissue overload. Reduced coactivity in the trunk flexors with flexion angle suggests the motor control system attempts to achieve a balance between stability and spinal load.

These analyses were limited by simplified anatomic and biomechanical representations of the spine and muscle geometry. The lumbar spine was modeled as a single inverted pendulum with three degrees-of-freedom of motion, and eight muscle equivalents were implemented to represent the muscular anatomy of the trunk. However, these simplifications require fewer neuromotor assumptions, allow muscle forces to be distributed empirically(Granata K.P., and Marras W.S. 1996), and have been shown to produce accurate results (Cholewicki J. et al 1998; Granata K.P., and Marras W.S. 1993; Granata K.P., and Marras W.S. 1995a; Marras W.S., and Granata K.P. 1995; Marras W.S., and Granata K.P. 1996). Nonetheless, future implementation of more advanced models of biomechanical stability may permit greater understanding of the low-back injury mechanisms.

These results help to describe the biomechanical value of antagonistic co-contraction. Antagonistic co-contraction was found to improve spinal stability despite concomitant increases in spinal load. However, as trunk moment increases, the risk of stability failure was reduced and the risk of spinal tissue overload injury was increased. Correspondingly, antagonistic co-contraction was recruited less in high moment conditions, i.e. during trunk flexion, and more in low moment conditions, i.e. in upright postures. Using the biomechanical requirements of stability in addition to traditional criterion of dynamic equilibrium, an improved understanding of motor control and trunk muscle co-contraction can be achieved.