ELSEVIER

Clinical Biomechanics 16 (2001) 650–659

CLINICAL BIOMECHANICS

www.elsevier.com/locate/clinbiomech

Trunk posture and spinal stability

K.P. Granata [a,*], S.E. Wilson [b]

[a] *Motion Analysis and Motor Performance Laboratory, Kluge Children's Rehabilitation Center, University of Virginia, 2270 Ivy Road, Charlottesville, VA 22903, USA*

[b] *Department of Mechanical Engineering, University of Kansas, Lawrence, KS, USA*

Received 19 February 2001; accepted 25 June 2001

Abstract

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

Design. A biomechanical model was developed to evaluate the influence of posture on spinal stability. Model performance 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 12 muscle equivalents of the trunk was implemented to determine spinal load and stability. Model input included trunk posture and lifted mass, output included muscle recruitment patterns necessary to achieve stability of the spine and spinal load. EMG activity recorded from the trunk muscles of 10 subjects were recorded during static exertions in various trunk flexion and asymmetric postures to compare with model output. Stable spinal load was examined as a function of trunk flexion and asymmetry during the lifting exertions.

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 stability than by equilibrium alone. Spinal stability is influenced by posture. Specifically, control of spinal stability is reduced in asymmetric postures associated with low-back disorder risk.

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.

Keywords: Spine; Stability; Posture; Low-back pain

1. Introduction

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

Trunk posture during occupational lifting is associated with the risk of suffering a LBD [8]. Retrospective studies conclude LBD risk during lifting is increased when combined with twisting [9], lateral bending [4] and asymmetric postures [10]. Biomechanical measurements illustrate that antagonistic co-contraction of the trunk musculature is increased in these high-risk postures [11–13]. Recent efforts indicate antagonistic co-activity is recruited to maintain spinal stability [14,15]. Thus, in-

* Corresponding author.
E-mail address: kpg8n@virginia.edu (K.P. Granata).

0268-0033/01/$ - see front matter
PII: S 0 2 6 8 - 0 0 3 3 (0 1) 0 0 0 6 4 - X

creased antagonistic co-contraction in these high-risk lifting postures may indicate greater neuromotor effort is necessary to prevent unstable failure and injury of the spine. In other words, postural risk factors for LBD 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 computational models of neuromuscular mechanics. Several analyses have represented the spine as a two- or three-dimensional inverted-pendulum [14,16,17]. These are an excellent means for investigating muscle recruitment because the number of muscles approximates the degrees-of-freedom (DoF) of the model but they fail to represent anatomic detail of the spine. Other models include 5–7 vertebral segments with 90–180 muscle elements [18–21]. These provide anatomic realism but cannot determine muscle recruitment distribution without significant simplifications and/or a priori assumptions because the number of muscles far exceeds the DoF of the system. A combined approach to evaluate spinal stability may provide improved insight.

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 postures traditionally recognized as high-risk conditions. To compensate for reduced stability, increased trunk muscle co-activation must be recruited in these postures, thereby explaining previous myoelectric results. The hypotheses were tested by implementing a biomechanical model of spinal stability and comparing results with measured trunk muscle activity.

2. Methods

2.1. 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 (Fig. 1). The two-segment geometry is unique, allowing independent control of trunk flexion and lumbar lordosis as well as requiring stabilizing support against global and local buckling behavior [21]. The model was exercised throughout a range of sagittal flexion angles from upright to 45° of forward flexion and at asymmetric postures from 0° to 20° 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 [14].

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 (Fig. 1).

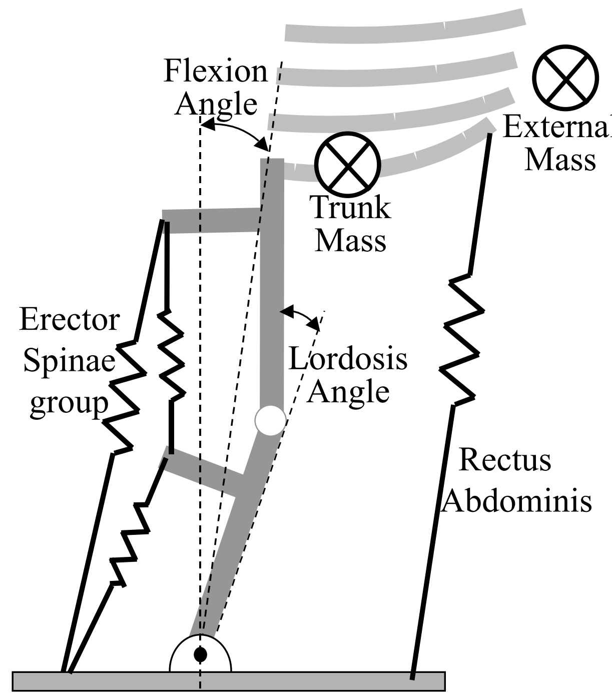

Fig. 1. Schematic representation of the stability model included a 3D inverted double-pendulum supported by 12 muscles (six bilateral muscle groups). Trunk mass and external load were applied at fixed vector locations relative to the superior surface of the upper vertebra. For clarity, the external oblique and internal oblique muscles have been omitted from the figure but were included in the computational analyses.

Muscle origins, insertions and cross-sectional areas were established from published anatomy and modeling efforts [22,23] (Table 1). Three-dimensional unit-moments of each muscle about the vertebral base of the segments were determined from the vector product of the muscle-insertions and the unit-vector of muscle force. Muscle force amplitudes were determined from equilibrium and stability constraints. Equilibrium requirements established six DoF and stability provided six constraints thereby allowing force estimation of up to 12 muscle groups. External load was represented as an 11.3 kg mass fixed 15 cm anterior to the superior surface of the modeled spine plus a trunk mass of 42 kg fixed 5 cm anterior to the superior surface of the modeled spine [24]. Thus, external loads rotated with the trunk causing increased sagittal and lateral moments with trunk posture and requiring greater restorative muscle forces in proportion to trunk flexion and asymmetry.

Stability requires the Hessian matrix of the system potential energy must be positive definite. The potential energies of the system were determined from muscle stiffness and system geometry or posture (see [18] 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

Table 1
Anatomic representation of muscle origins and insertions[a]

	Vertebral inferior surface			Vertebral superior surface		
	X Rt lateral (cm)	Y Anterior (cm)	Z Cranial (cm)	X Rt lateral (cm)	Y Anterior (cm)	Z Cranial (cm)
Vertebra L_0 (origin)				0.0	0.0	0.0
Vertebra L_{Lower}	Origin			0.0	0.0	10.5
Vertebra L_{Upper}	Superior surface of L_{Lower}			0.0	0.0	10.5
	Muscle origin			Muscle-insertion		
Erector spine #1	$4.5L_0$	$-6.0L_0$	$-4.2L_0$	$4.5L_{\mathrm{Lower}}$	$-6.0L_{\mathrm{Lower}}$	$6.3L_{\mathrm{Lower}}$
Erector spine #2	$4.5L_{\mathrm{Lower}}$	$-6.0L_{\mathrm{Lower}}$	$6.3L_{\mathrm{Lower}}$	$4.5L_{\mathrm{Upper}}$	$-6.0L_{\mathrm{Upper}}$	$6.3L_{\mathrm{Upper}}$
Erector spine #3	$4.5L_0$	$-6.0L_0$	$-4.2L_0$	$4.5L_{\mathrm{Upper}}$	$-6.0L_{\mathrm{Upper}}$	$6.3L_{\mathrm{Upper}}$
Rectus abdominis	$4.0L_0$	$8.0L_0$	$-3.1L_0$	$4.5L_{\mathrm{Upper}}$	$13.0L_{\mathrm{Upper}}$	$10.5L_{\mathrm{Upper}}$
External oblique	$6.0L_0$	$10.0L_0$	$-3.1L_0$	$13.0L_{\mathrm{Upper}}$	$6.0L_{\mathrm{Upper}}$	$10.5L_{\mathrm{Upper}}$
Internal oblique	$6.5L_0$	$-6.0L_0$	$0.0L_0$	$10.0L_{\mathrm{Upper}}$	$2.0L_{\mathrm{Upper}}$	$10.5L_{\mathrm{Upper}}$

L_0: Superior surface of sacrum = inferior surface of L_{Lower}; L_{Lower}: Lower vertebral segment; L_{Upper}: Upper vertebral segment; Values represent vector displacements from the inferior surface of the indicated vertebrae. All vectors rotate rigidly with the indicated vertebrae. All vectors with respect to L_0 do not move or rotate.

[a] All values are relative to the indicated vertebral base and rigidly rotate with the indicated vertebral segment. Only right side muscles are represented, left side muscles are sagittally symmetric (negative value for X).

proportionality, $q = 5$ and inversely proportional to the muscle equilibrium length [19,21].

The model solved 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

$$\min \sum_{m=1}^{12} F_m' \sigma^2 F_m, \tag{1}$$

where F_m, and σ are the force amplitude and stress-squared matrix of muscles $m = 1, \ldots, 12$. The analyses were subject to three constraints:

$$\sum_m (r \times \hat{f}_m) F_m = M_{\mathrm{Ext}}, \tag{2a}$$

$$\mathrm{eig}\left(\frac{\delta^2}{\delta\theta_i\,\delta\theta_j} V\right) \geqslant 0, \tag{2b}$$

$$F_m < \mathrm{Gain} \cdot \mathrm{Area}_m. \tag{2c}$$

First, the set of muscle forces must satisfy equilibrium where r, f_m, and M_{Ext} are, respectively, the muscle-insertion 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 eigenvalues of the Hessian matrix of potential energy, V are greater than zero. Third, the 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, $\mathrm{Gain} = 50\ \mathrm{N/cm^2}$. Muscle force generating capacity was also modulated by muscle length as described as in published literature regarding trunk biomechanics [25]. For presentation, predicted muscle forces were expressed as a percentage of the theoretical force generating capacity, i.e., percent of maximum voluntary contraction (MVC). Model performance was evaluated by comparing predicted muscle recruitment to measured values of myoelectric activity.

Spinal load was computed from the vector sum of muscle forces and external load at the lumbar-sacral inter-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 (Eq. (2b)). Second, a "stability spinal load" was determined from the set of optimized muscle forces satisfying both equilibrium and stability constraints. 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.

2.2. Experiment

Five healthy males and five healthy females, 24–39 years of age, with no prior history of low-back pain voluntarily participated in this experiment. Mean (SD) 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° (upright), 15°, 30° and 45°, and at asymmetric postures of 0° (sagittally symmetric), 10° left twist, and 20° left twist and all combinations of these flexion and asymmetric postures. Subjects were required to hold a weight of 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, Burlington VT, USA). 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 axial rotation of the T10 and manubrium markers with respect to the S1 sensor. Trunk flexion and asymmetry angles were displayed as real-time to allow participants to achieve and maintain the prescribed postures.

EMG were collected at 1000 Hz using bipolar surface electrodes (Medicotest, Rolling Meadows, IL, USA) from four bilateral sets of trunk muscles. Electrodes were placed according to Mirka [26] including right and left recti abdominis (RA), external obliques (EO), internal obliques (IO) and erector spinae (ES). Considering electrode locations, internal oblique activity was representative of extensor and lateral effort whereas external oblique myoactivity was considered flexor and lateral [25,27]. Note that three modeled paraspinal muscles per side were necessary to establish equilibrium and stability. It has been assumed the sum of these activities was represented by the surface EMG recording from the ES. EMG signals were band-pass filtered between 30 and 450 Hz in hardware prior to data collection. The data were rectified and integrated using a 5 Hz Hanning low-pass convolution software filter and normalized to values collected during isometric maximum flexion, extension, right-lateral twisting and left-lateral twisting exertions. 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 two independent variables including trunk flexion and asymmetry and one stratification factor, i.e., gender. Analyses were performed using commercial statistical software (Statistica, 4.5, Statsoft, Tulsa OK, USA) with a significance level of 0.05 for all tests.

3. 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 (Fig. 2). 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 was considered in addition to equilibrium the RA was recruited at a mean level of 6.6% MVC (Fig. 3). Measured EMG activity from the RA was significantly greater than zero with a bilateral mean value of 4% MVC across all postural conditions (Table 2). 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 (Table 3). 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% (Figs. 2 and 3). 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 (Table 2). Myoactivity from the EO demonstrated a small but statistically significant increased with trunk flexion, i.e., 0.5–5% MVC change from upright to 45° flexed postures (Table 3). Modeled EO recruitment declined with flexion angle allowing potentiation of extensor moment. Right EO increased while left EO declined with modeled trunk asymmetry to stabilize the lateral moment. This asymmetric behavior was particularly notable in the upright postures wherein modeled right EO reached recruitment levels of 55% MVC. As predicted by the model the right EO increased significantly with postural asymmetry (Table 3) 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 towards 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

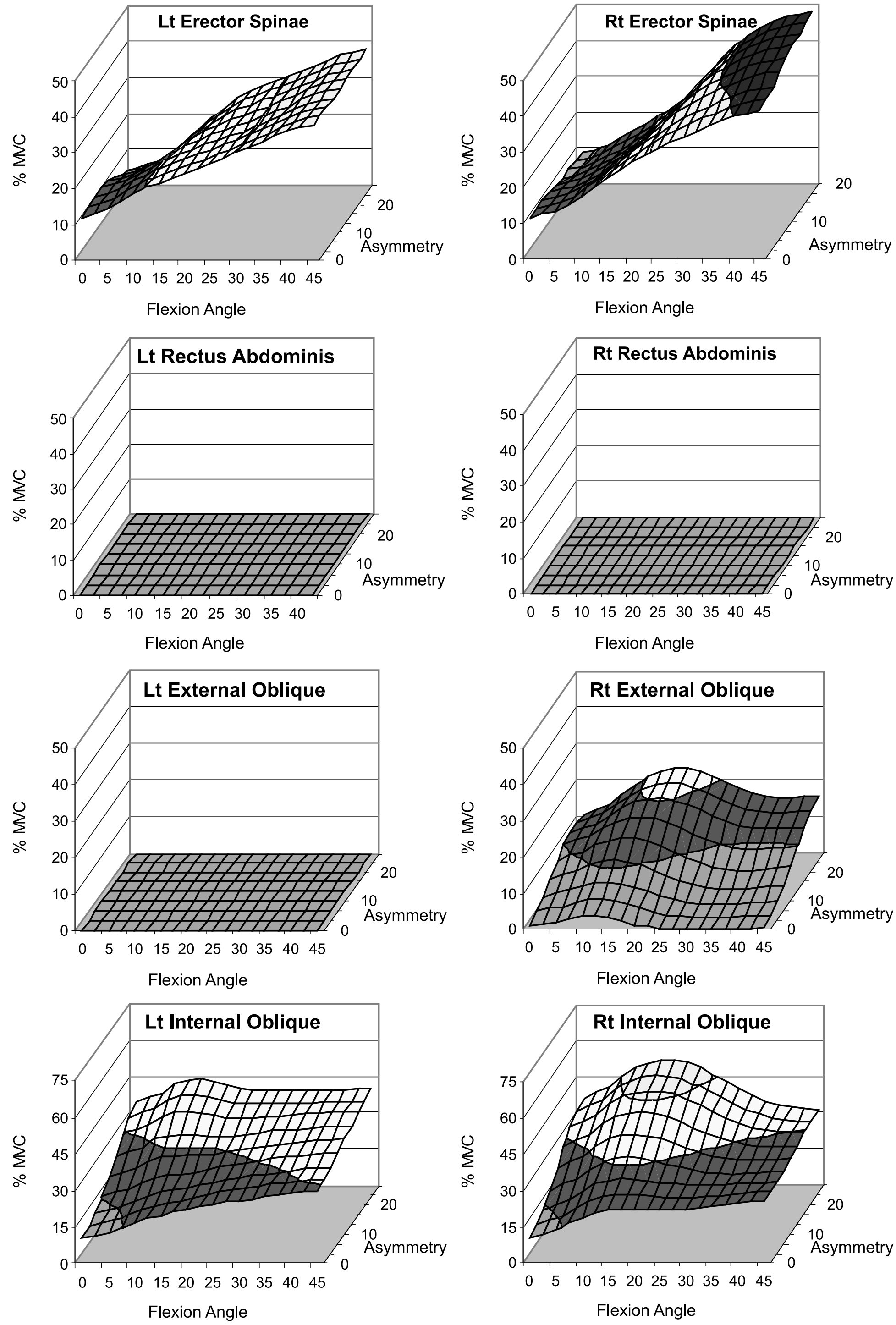

Fig. 2. Predicted muscle recruitment from equilibrium model without consideration of spinal stability.

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% vs 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

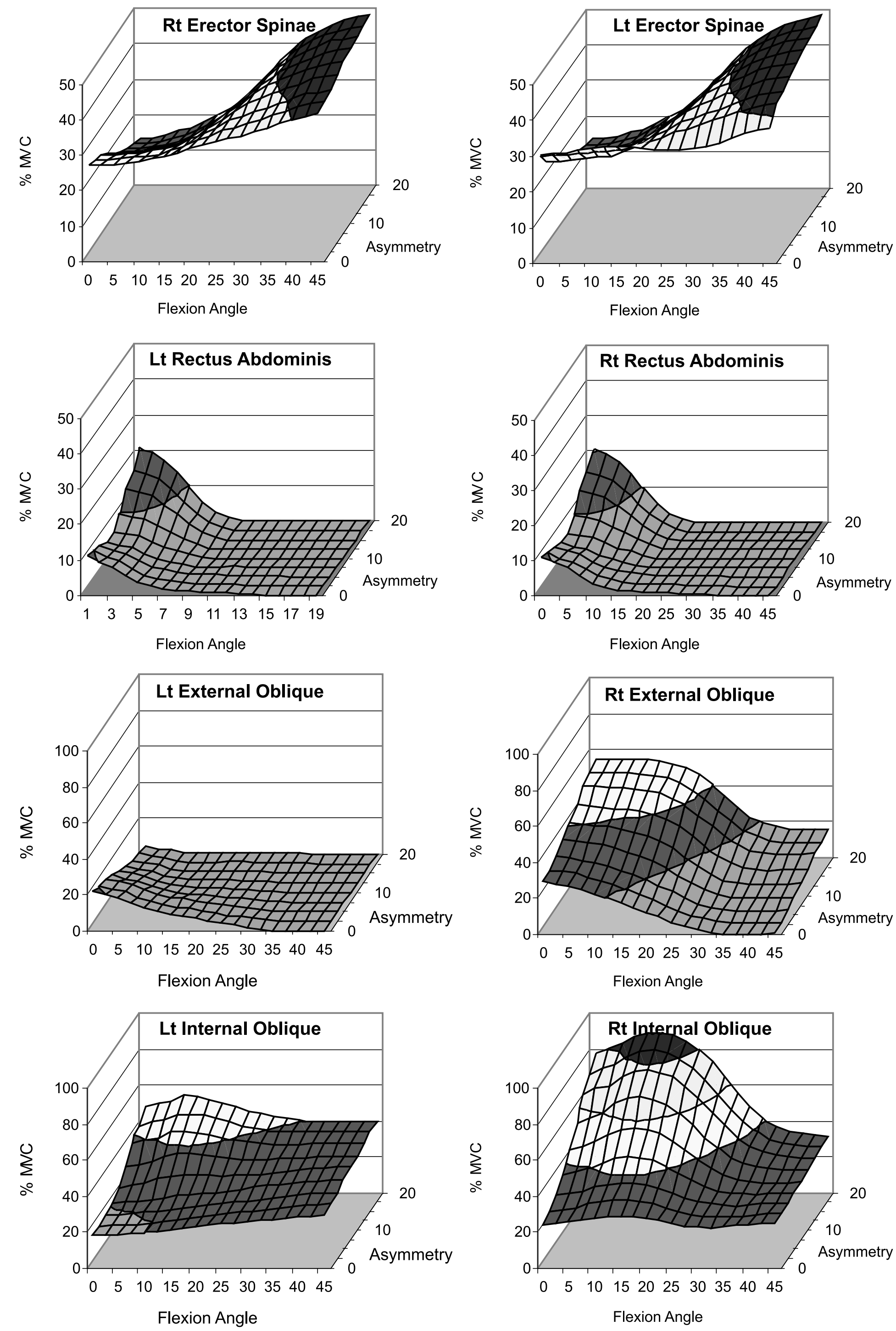

Fig. 3. Predicted muscle recruitment from model satisfying both equilibrium and stability.

change in ES recruitment with asymmetry, although in upright postures the stability model reduced right ES activity by 13% MVC, and increased left ES levels by 11%. Similarly, measured activity in asymmetric lifting postures were associated with trends that reduced right ES EMG slightly, less than 2%, and increased left ES activity by 4.5%. These changes failed to reach levels of a priori statistical significance ($P = 0.055$ and $P = 0.169$ 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

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

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

Table 3
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 × angle	**0.000**	**0.004**	0.091	0.160	**0.032**	**0.000**	0.053	0.068
Gender × asymmetry	**0.002**	0.341	0.599	0.270	0.795	0.150	0.137	0.835
Angle × asymmetry	0.968	0.866	0.340	0.267	0.652	0.808	0.647	0.394

RES: right erector spinae, LES : left erector spinae; RRA: right rectus abdominis, LRA: left rectus abdominis; REO: right external oblique, LEO: left external oblique; RIO: right internal oblique, LIO: left internal oblique; Bold values highlight significant effect ($P < 0.05$).

recruitment from the two models were identical, ranging from mean levels of 27% and 35% depending upon asymmetry. Measured IO activation in these same postures ranged from 15% to 23% MVC. In the upright postures, the stability model predicted greater IO recruitment than the equilibrium analyses, particularly in 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 from stability requirements.

Gender influenced muscle recruitment levels in specific muscles (Table 3). In general, female subjects recruited greater activity than males in the ES and EO muscle groups during flexed postures. Conversely, the female subjects recruited lower activity from the RA and IO muscles than males, although this difference reached statistical significance only for the right RA. This gender difference in recruitment patterns has been noted in previous efforts [28]. Whether it influences spinal stability must be examined in future research.

Spinal load was markedly increased when stability constraints were included in the model (Fig. 4). The

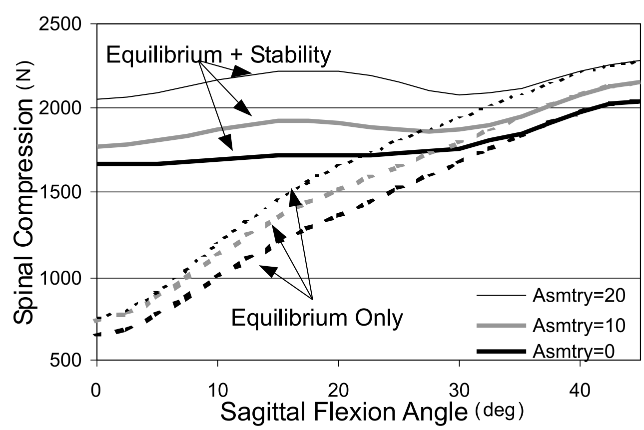

Fig. 4. Stability constraints resulted in greater spinal compression than equilibirum analyses without consideration of stability. Greater neuromuscular control is necessary to maintain stability in asymmetric lifting postures resulting in increased stability spinal load.

external load was small so mean spinal compression was only 720 N in the upright posture according to the equilibrium model. Conversely, in upright postures additional muscle recruitment including 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 than the equilibrium model, with a mean level of 1815 N. Equilibrium spinal 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. 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° of trunk asymmetry. Thus, high-risk postures were associated with reduced spinal stability and increased spinal load.

4. Discussion

It was proposed that LBD risk associated with asymmetric and forward flexed lifting exertions may be related in part to reduce spinal stability in these postures. To test these hypotheses a model of spinal stability was developed and exercised in various static trunk postures. 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% of the equilibrium load in upright postures. The spinal load associated with stability increased with asymmetry 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 [29–31]. It has been noted elsewhere that antagonistic co-activity is increased in asymmetric lifting tasks [12] causing greater spinal compression in these postures [32,33]. During asymmetric lifting modified muscle recruitment patterns are necessary to offset lateral components of trunk moment and establish equilibrium [27,34,35]. The current results demonstrate these equilibrium recruitment patterns during 20° 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 co-activation in these postures has until now never been described. Antagonistic co-contraction has recently been identified as a necessary component in the maintenance of stability [14]. 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.

Physiologic 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 [15]. However, measured antagonistic muscle activity did not approach the extraordinary 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 [36–38]. These myoelectric responses represent feedback control for the biomechanical system. 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 overestimate the required magnitude of recruitment co-activation. Analyses by Granata et al. [28] and Cholewicki et al. [39] agree that neuromuscular response dynamics play a significant role in stabilizing the spine. The difference between predicted and measured muscle recruitment suggesting greater reliance upon the dynamic neuromuscular response in asymmetric postures. Published measurements illustrate that muscle responses including feedback gain and delay are limited by various factors including posture and fatigue [40]. Failure to recruit an appropriate and timely activation response will risk instability injury. Future efforts must incorporate reflex and response mechanics in biomechanical assessment of spinal stability.

Although spinal stability may be improved with trunk flexion the risk of injury from spinal overload is enhanced in these postures. Muscle stiffness provides the primary mechanism of static stability in analyses of the spine [24]. Active muscle stiffness increases with contractile force [41], therefore large muscle forces associated with flexed postures [42] serves to increase stiffness and stability. This is observable in the results wherein the compressive load from equilibrium analyses and stability converge with trunk flexion. Unfortunately, spinal load is dramatically increased in these postures so the risk of overload injury is manifest. However, the suggestion of improved stability in flexed postures must be tempered by model design limitations. Trunk flexion may imposes large translation forces on the spine requiring more advanced analyses including 6-DoF per joint for the evaluation of stability [24] rather than the pure rotational behavior examined in the current analyses. Passive stability the osteoligamentous spine [43] and modified neuromotor response dynamics [36] may reduce stability in flexed postures. The role of stability and its interaction with spinal load as etiologic factors describing LBD risk in trunk flexion requires further research.

The model performed well, predicting antagonistic co-contraction, correctly predicting general trends in extensor muscle activation and typically predicting appropriate muscle activation. The two-segment inverted-pendulum representation of the spine was unique, permitting analyses to predict activity in the major trunk muscles because the six equilibrium and six stability constraints matched the number of muscles. However, discrepancies between predicted and measured muscle recruitment 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. [14] wherein similar reduction in flexor activity was predicted from their stability model but contrary to our EMG measurements they reported a significant trend towards 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. [24]. Conversely, Cholewicki et al. [14] 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 is influenced by postural asymmetry and trunk flexion regardless of this design parameter. Thus, results suggest spinal stability is reduced in lifting postures traditionally associated with high-risk for LBDs.

5. Conclusion

This research has demonstrated that lifting posture can influence musculoskeletal stability of the spine. Analyses were based upon a biomechanical model that predicted recruitment patterns in 12 muscle equivalents necessary to stabilize a two-segment, 6-DoF spine. Measured muscle activity agreed with model predicted recruitment. Results demonstrate antagonistic co-contraction is necessary to achieve stability and greater neuromuscular control is necessary to maintain stability in asymmetric lifting postures. Although estimated stability can be maintained in flexed postures spinal load increases significantly. Risk of occupational LBDs associated with lifting posture may be partially influenced by musculoskeletal stability of the spine.

Acknowledgements

This research was supported by grants K01 OH00158-03 from NIOSH of the Centers of Disease Control and R01 AR46000-03 from NIAMS of the National Institutes of Health.

References

[1] Waters TR, Putz-Anderson V, Garg A, et al. Revised NIOSH equation for the design and evaluation of manual lifting tasks. Ergonomics 1993;36:749–76.
[2] Konz S. NIOSH lifting guidelines. Am Ind Hyg Assoc J 1982;43:931–3.
[3] Crisco JJ, Panjabi MM. Euler stability of the human ligamentous lumbar spine: Part I Theory. Clin Biomech 1992:719–26.
[4] Punnet L, Fine LJ, Keyserling WM, et al. Back disorders and non-neutral trunk postures of automobile assembly workers. Scand J Work Environ Health 1991:17337–46.
[5] Granata KP, Marras WS, Ferguson SA. Relation between biomechanical spinal load factors and risk of occupational low-back disorders. Proc Human Factors Ergon Soc 1996:656–60.
[6] Wilke HJ, Wolf S, Claes LE, et al. Stability increase of the lumbar spine with different muscle groups. Spine 1995;20(2):192–8.
[7] Quint U, Wilke HJ, Shirazi-Adl A, et al. Importance of intersegmental trunk muscles for the stability of the lumbar spine. A biomechanical study in vitro. Spine 1998;23(18):1937–45.
[8] Marras WS, Lavender SA, Leurgans S, et al. Biomechanical risk factors for occupationally related low back disorder risk. Ergonomics 1995;38(2):377–410.
[9] Kelsey KL, Githens PB, White III AA, et al. An epidemiologic study of lifting and twisting on the job and risk for acute prolapsed lumbar intervertebral disc. J Orthop Res 1984;2(1):61–6.
[10] Marras WS, Lavender SA, Leurgans SE, et al. The role of dynamic three-dimensional trunk motion in occupationally related low back disorders: the effects of workplace factors, trunk position and trunk motion characteristics on risk of injury. Spine 1993;18(5):617–28.
[11] Marras WS, Mirka GA. A comprehensive evaluation of trunk response to asymmetric trunk motion. Spine 1992;17(3):318–26.
[12] Marras WS, Mirka GA. Muscle activities during asymmetric trunk angular accelerations. J Orthop Res 1990;8(6):824–32.
[13] Pope MH, Svensson MS, Andersson GBJ, et al. The role of prerotation of the trunk in axial twisting efforts. Spine 1987;12(10):1041–5.
[14] Cholewicki J, Panjabi M, Khachatryan A. Stabilizing function of trunk flexor-extensor muscles around a neutral spine posture. Spine 1998;22(19):2207–12.

[15] Orishimo K, Granata KP. Co-contraction in biomechanical stability of the trunk. PhD dissertation, University of Virginia, 2000.
[16] Cholewicki J, Panjabi MM, Radebold A, McGill SM. Can lumbar spine stability be augmented with an abdominal belt and/or increased intra-abdominal pressure? In: Proceedings of the International Society for the Study of the Lumbar Spine; 1999.
[17] Granata KP, Marras WS. Cost-benefit of muscle co-contraction in protecting against spinal instability. Spine 2000;25(11):1398–404.
[18] Cholewicki J, McGill SM. Mechanical stability on the in vivo lumbar spine: implications for injury and chronic low back pain. Clin Biomech 1996;11(1):1–15.
[19] Gardner-Morse M, Stokes IAF, Laible JP. Role of muscles in lumbar stability in maximum extension efforts. J Orthop Res 1995;13(5):802–8.
[20] Stokes IA, Gardner-Morse M. Lumbar spine maximum efforts and muscle recruitment patterns predicted by a model with multijoint muscles and joints with stiffness. Spine 1995;28(2):173–86.
[21] Bergmark A. Stability of the lumbar spine: a study in mechanical engineering. Acta Orthop Scand Suppl 1989:2301–54.
[22] Jorgenson MJ, Marras WS, Granata KP, Wiand B. MRI-derived moment-arms of the female and male spinal loading muscles. Clin Biomech 2001;16(3):182–93.
[23] Marras WS, Jorgenson MJ, Granata KP, Wiand B. Female and male trunk geometry: size and prediction of the spine loading trunk muscles derived from MRI. Clin Biomech 2001;16(1):38–46.
[24] Gardner-Morse M, Stokes IA. The effects of abdominal muscle coactivation on lumbar spine stability. Spine 1998;23(1):86–92.
[25] Granata KP, Marras WS. An EMG assisted model of biomechanical trunk loading during free-dynamic lifting. J Biomech 1995;28(11):1309–17.
[26] Mirka GA. The quantification of EMG normalization error. Ergonomics 1991;34(3):343–52.
[27] Schultz AB, Anderson GBJ. Analysis of loads on the lumbar spine. Spine 1981;6(1):76–82.
[28] Granata KP, Orishimo K, Sanford AH. Trunk muscle coactivation in preparation for sudden load. J Electromyo Kinesiol 2000 [in press].
[29] Chaffin DB, Park KS. A longitudinal study of low-back pain as associated with weight lifting factors. Am Ind Hyg Assoc J 1973:34513–25.
[30] Norman RW, Wells R, Neumann P, et al. A comparison of peak vs cumulative physical work exposure risk factors for the reporting of low back pain in the automotive industry. Clin Biomech 1998:13561–73.
[31] Herrin GA, Jaraiedi M, Anderson CK. Prediction of overexertion injuries using biomechanical and psychophysical models. Am Ind Hyg Assoc J 1986;47(6):322–30.
[32] Marras WS, Sommerich CM. A three-dimensional motion model of loads on the lumbar spine: I model structure. Human Factors 1991;33(2):123–37.
[33] Granata KP, Marras WS. An EMG-assisted model of loads on the lumbar spine during asymmetric trunk extensions. J Biomech 1993;26(12):1429–38.
[34] Schultz AB, Andersson GBJ, Ortengren R, et al. Loads on the lumbar spine. J Bone Joint Surg 1982;64(5):713–20.
[35] Schultz AB, Andersson GBJ, Haderspeck K, et al. Analysis and measurement of the lumbar trunk loads in tasks involving bends and twists. J Biomech 1982:15669–75.
[36] Krajcarski SR, Potvin JR, Chiang J. The in vivo dynamic response of the spine to perturbations causing rapid flexion: effects of pre-load and step input magnitude. Clin Biomech 1999;14(1):54–62.
[37] Lavender SA, Mirka GA, Schoenmarklin RW, et al. The effects of preview and task symmetry on trunk muscle response to sudden loading. Human Factors 1989;31(1):101–15.
[38] Thomas JS, Lavender SA, Corcos DM, et al. Trunk kinematics and trunk muscle activity during a rapidly applied load. J Electromyo Kinesiol 1998;8(4):215–25.
[39] Cholewicki J, Simons APD, Radebold A. Effects of external trunk loads on lumbar spine stability. J Biomech 2000;331377–85.
[40] Magnusson ML, Aleksiev A, Wilder DG, et al. Unexpected load and asymmetric posture as etiologic factors in low back pain. Eur Spine J 1996;5(1):23–35.
[41] Weiss PI, Hunter IW, Kearney RE. Human ankle joint stiffness over the full range of muscle activation levels. J Biomech 1988;21539–44.
[42] Chaffin DB. A computerized biomechanical model-development of and use in studying gross body actions. J Biomech 1969:2429–41.
[43] Posner I, White AA, Edwards WT, et al. A biomechanical analysis of the clinical stability of the lumbar and lumbosacral spine. Spine 1998;7(4):374–89.