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Cost–Benefit of Muscle Cocontraction in Protecting Against Spinal Instability

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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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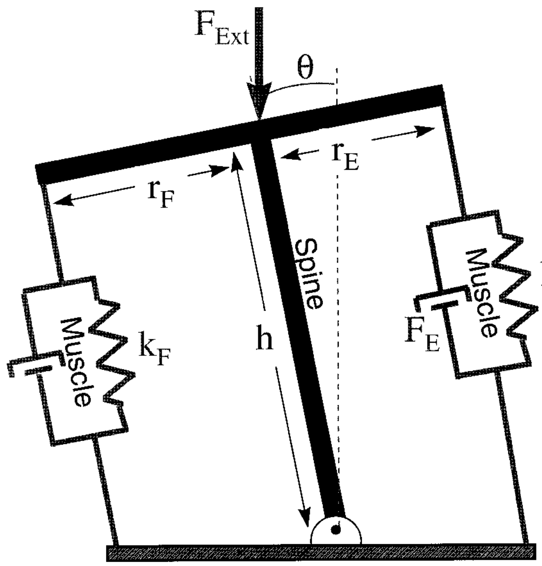


Figure 1. Simple model of spinal stability including trunk flexor and extensor muscle equivalents.

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),

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

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.[46] 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:

$$k = q\frac{F}{L} \tag{2}$$

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

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

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

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

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

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

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 simultaneously distributing the external trunk moments in relation to the conditioned EMG signals.

The model of biomechanical stability input measured external trunk kinetics as well as muscle force and length from the equilibrium model. Equivalent, instantaneous moment arm vectors associated with the external load were derived from the measured external forces and moments to determine changes in external moments from partial derivatives of the moment arm vector. Critical stability load was calculated from

$$\sum_{m} (r_m \times \hat{f}_m) \left(q \left| F_m \right| \frac{1}{L_m} \frac{\delta L_m}{\delta \theta} \right) - F_{Ext} \left(\frac{\delta r_{Ext}}{\delta \theta} \times \hat{f}_{Ext} \right) = 0 \tag{6}$$

where $\hat{f}_m$ represents the unit force vector for each muscle, F_m the equilibrium muscle force, r_m the muscle moment arm vector, and L_m the muscle length. The solution for external force satisfying critical stability, F_{Ext}, was a 3 by 3 matrix describing the three dimensions of load associated with the three dimensions of perturbation, which represented the maximum external force that could be stabilized in the current equilibrium state. Passive contributions from disc and ligamentous stiffnesses were considered negligible in the measured postures as compared with the stability generated by the muscles.[13] To determine maximum stable (spinal) load, $F^M{}_Z$, the critical force F_{Ext} was resubstituted into the equilibrium model. Maximum stable load describes the spinal load associated with the critical stability force.

Data. Motion, EMG, external force, and external moment about the lumbosacral junction were collected from 10 healthy men with no history of low back pain. These participants, ages 21 to 35 years, had a mean (standard deviation) weight of 72.7 (6.6) kg and height of 176.7 (4.2) cm.

The participants lifted a 22.7-kg box in a sagittally deviation symmetric motion from a platform 52 cm from the floor and 51 cm anterior to the ankles and placed it on a second platform 107 cm from the floor and 25 cm anterior to the ankles. These exertions were performed at freely selected lift rates while three-dimensional, dynamic trunk position and orientation were recorded from an electrogoniometer.[11,26] External forces and moments applied to the lumbosacral junction were determined from inverse dynamic analyses using force plate and electrogonimeter data.[11,19] Electromyographic activity was collected from bipolar surface electrodes over the modeled muscles. Signals were amplified, band-pass filtered (30–1000 Hz) and rectified in hardware. Kinetic, kinematic, and EMG data were identically filtered using a 10-Hz Hanning weighted low-pass filter before biomechanical modeling. The EMG electrode placement and data processing were described previously.[34]

Analyses. Analyses were performed to examine the relation between trunk muscle coactivity and the increase in biomechanical stability *versus* spinal load. The influence of coactivation was demonstrated by running the equilibrium/stability model once using the full set of EMG data and comparing the results with analyses wherein the antagonistic coactivity were eliminated from the model by zeroing the EMG activity of the rectus abdominis and external oblique.[17] When the flexor muscles were eliminated, the solution of the remaining muscle forces, spinal load, and maximum stable load necessarily adjusted according to the equilibrium constraints. The muscle

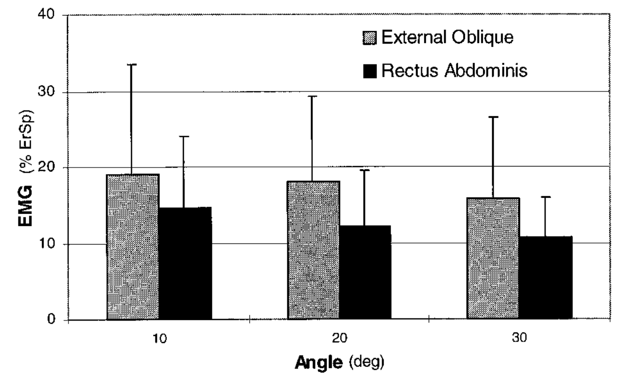


Figure 2. Coactivity of the (antagonistic) abdominal trunk muscles during dynamic extension tasks. Upright posture is defined as an angle of 0. Note that flexor activities were significantly greater than 0 and increased in more upright postures.

stiffness proportionality constant, q, was set equal to 10 for all analyses. Model output included spinal load from equilibrium mechanics and the maximum stable load associated with critical stability. *Stability margin,* defined as the difference between maximum stable load and applied spinal load, $F^M{}_Z - F_Z$, was computed also.

Independent variables included trunk flexion angle and coactivation level. Dependent variables included spinal load, maximum stable load, and stability margin. Analysis of variance (ANOVA) procedures were performed to assess statistical significance, with *post hoc* analyses of significant ($P < 0.05$) relations.

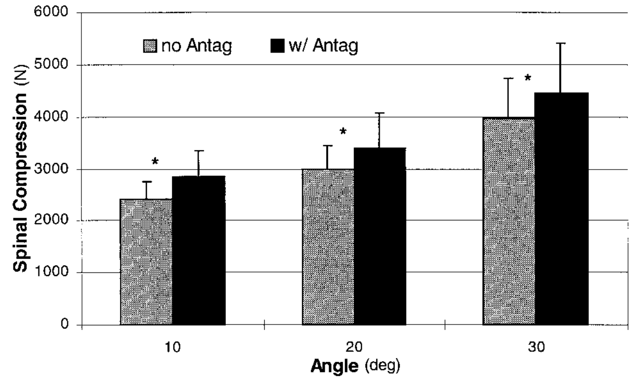
■ Results

No significant differences were found between the right- and left-side EMG activity in the sagittally symmetric lifting exertions, so the two sides were averaged in subsequent statistical analyses of EMG. Antagonistic EMG was significantly greater than 0 in all conditions ($P < 0.05$). To quantify coactivity, EMG magnitudes of the external oblique and rectus abdominis muscles were expressed as a percentage of erector spinae activity (Figure 2). Although the rectus abdominis EMG activity decreased significantly with trunk flexion angle, when

Table 1. ANOVA Results of EMG, Coactivity, Spinal Load, and Stability Margin

	Angle	Coactivity	Angle × Coactivity
Electromyogram			
Erector spinae	$P<0.94$		—
Internal oblique	$P<0.86$	—	—
Rectus abdominis	***P<0.01***	—	—
External oblique	$P<0.18$	—	—
Coactivity			
Rectus abdominis/erector spinae	$P<0.06$	—	—
External oblique/erector spinae	***P<0.04***	—	—
Moment (sagittal)	***P<0.01***	—	—
Spine load (F_Z)	***P<0.01***	***P<0.01***	$P<0.93$
Maximum stable load ($F^M{}_Z$)	***P<0.01***	***P<0.01***	$P<0.89$
Stability margin ($F^M{}_Z - F_Z$)	$P<0.07$	***P<0.01***	$P<0.62$

Bold values highlight significant effects for P < 0.05.
ANOVA = analysis of variance; EMG = electromyography.



* 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 margin 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.

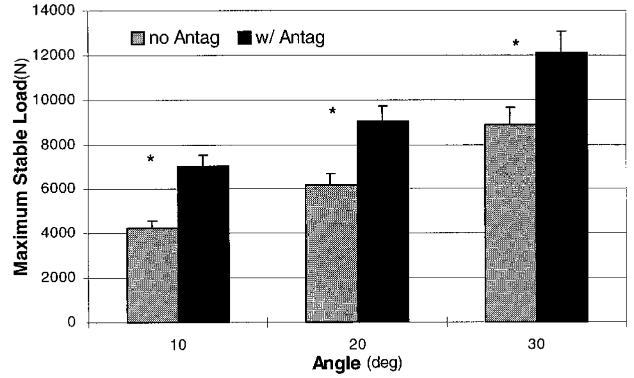


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Figure 4. Spinal stability, defined as the maximum compressive load that could be stabilized, increased with trunk flexion angle and antagonistic cocontraction.

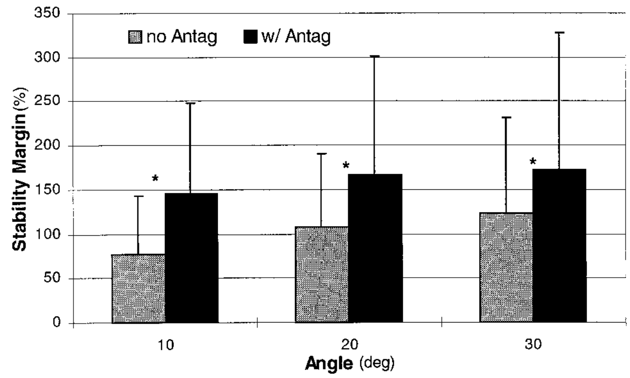

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Figure 5. Stability margin (*i.e.,* the difference between maximum stable load and spinal compression (F^{Cz}–F_Z) was expressed as a percentage of the spinal compression. Stability margin increased significantly with antagonistic muscle cocontraction.

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 was added. In contrast to these efforts, the current authors chose to quantify stability in terms of spinal load tolerance at critical stability determined from measured EMG coactivity. This permitted direct comparison of stability tolerance and spinal load as a function of antagonistic cocontraction in dynamic lifting.

Spinal stability increased 36% to 64% with antagonistic cocontraction. This represents a mean increase of 2925 N in the compression tolerance. Stability also improved with increased flexion angle. Both trunk flexion and antagonistic cocontraction work in a similar manner to stabilize the system. Increased flexion moments from cocontraction and trunk flexion require increased extensor muscle force. Inasmuch as active muscle stiffness is proportional to contractile force,[22,35,43] the increased extensor muscle force stiffened and stabilized the biomechanical structure.

Mean stability margin, expressed as a percentage of the applied spinal load, was 103% without antagonism and 161% when cocontraction was included. If risk of injury is associated with the relation between tolerance and load, then cocontraction reduced the risk of stability failure despite the fact that the load was increased. It is interesting to note that without cocontraction, the biomechanical system was unstable in the nearly upright postures (*i.e.*, a stability margin less than 100%). This indicates that flexor antagonism is necessary to maintain stability in the upright postures, as demonstrated by Cholewicki et al.[6] Accordingly, the EMG of the rectus abdominis was significantly greater in upright postures than in flexed postures. Hence, antagonistic cocontraction may be recruited to complement biomechanical need (*i.e.*, increased activity in the upright posture to stabilize the system and reduced activity in flexed postures to reduce spinal load).

The benefit of antagonistic cocontraction in terms of stability must be balanced against the risks associated with increased spinal load. *In vitro* experiments have demonstrated increased intervertebral passive stability with cocontraction[42,48] in addition to the stability contributions from active muscle stiffness. However, there is little evidence to suggest that antagonistic activity will improve the tissue overload tolerance of the spine. The requirement for biomechanical stability must be balanced against the risk of tissue overload.[4] Reduced coactivity in the trunk flexors with increased flexion angle suggests that the motor control system attempts to achieve a balance between stability and spinal load.

The magnitude of the muscle stiffness proportionality constant, q, influenced the predicted stability but not the trends of a stable system. Muscle stiffness rate, q, has been reported[6,13] in a range from 4.5 to 30. The current results were based on a value of 10 for q. *Post hoc* sensitivity analyses suggest that maximum stable load increased with q (Figure 6). The influence of antagonistic activity on stabilizing the system was amplified at greater values of q, but spinal load was unaffected. It should be noted that very low values of q result in nearly unstable

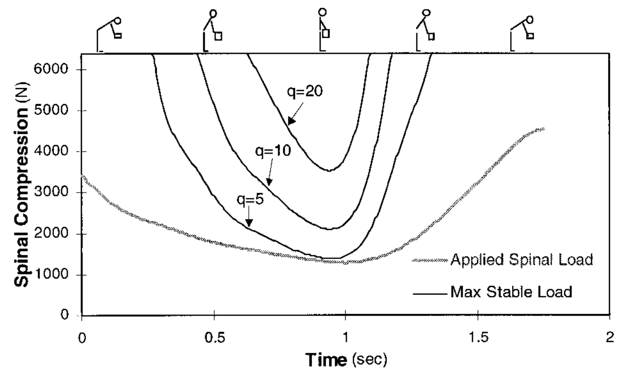

Figure 6. Muscle stiffness coefficient, q, influences the magnitude but not the trend of spinal stability. The maximum stable load was limited to less than 6400 N in this plot of a typical lifting exertion to illustrate that both tissue and stability tolerance must be considered.

behavior in the upright postures. A relatively low value was chosen for the stiffness coefficient to represent a conservative estimate of stability and stability margin. Therefore, the cost–benefit of antagonistic cocontraction depends on the physiologic relation between muscle stiffness and active muscle force.

Caution must be exercised in interpreting the biomechanical meaning of the maximum stable load. Maximum stable load represented the equivalent spinal compression resulting from the maximum external load the system could stabilize in its current equilibrium condition. Stability was dependent on equilibrium conditions because muscle stiffness was related directly to muscle force, which in turn was related to trunk moment. If external load were increased, biomechanical equilibrium would change and a new stability limit would be formed. Therefore, the maximum stable load was artificial in that it represented a virtual equilibrium solution to the stability limit. However, if the construct of maximum stable load is understood, this is an informative method by which stability tolerance can be evaluated. Moreover, this method allows the relative biomechanical risk in a dynamic lifting environment to be assessed.

These analyses were limited by the anatomic and biomechanical representations of the spine and muscle geometry. The lumbar spine was modeled as a single inverted pendulum, and the muscular anatomy of the trunk was represented with eight muscle vectors. The results represent global stability as opposed to multisegment buckling behavior.[7,9] Others have presented a more detailed representation of multivertebral spines,[2,5,12] including as many as 132 muscle elements.[13] However, more complex models, limited by increased assumptions regarding coordination of vertebral motion, are required to estimate muscle cocontraction in the overdetermined system. Conversely, less complex models require fewer assumptions, allow muscle forces to be distributed empirically,[16] and have produced accurate results.[6,14,15,27,28] An empirically based EMG-assisted model was considered the most appropriate method for investigating the influence 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.

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■ 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.

References

1. Agarwal GC, Gottlieb CL. Compliance of the human ankle joint. J Biomech Eng 1977;99:166–70.
2. Bergmark A. Stability of the lumbar spine: A study in mechanical engineering. Acta Orthop Scand Suppl 1989;230:1–54.
3. Chaffin DB, Page GB. Postural effects on biomechanical and psychophysical weight lifting limits. Ergonomics 1994;37:663–76.
4. Cholewicki J, McGill SM. EMG-assisted optimization: A hybrid approach for estimating muscle forces in an interdeterminate biomechanical model. J Biomech 1994;27:1287–9.
5. Cholewicki J, McGill SM. Mechanical stability on the *in vivo* lumbar spine: Implications for injury and chronic low back pain. Clin Biomech 1996;11:1–15.
6. Cholewicki J, Panjabi M, Khachatryan A. Stabilizing function of trunk flexor-extensor muscles around a neutral spine posture. Spine 1998;22:2207–12.
7. Crisco JJ, Panjabi MM. Euler stability of the human ligamentous lumbar spine: Part I. Theory. Clin Biomech 1992;7:19–26.
8. Crisco JJ, Panjabi MM. The intersegmental and multisegmental muscles of the lumbar spine: A biomechanical model comparing lateral stabilizing potential. Spine 1991;16:793–9.
9. Crisco JJ, Panjabi MM, Yamamoto I, Oxland TR. Euler stability of the human ligamentous lumbar spine: Part II. Experiment. Clin Biomech 1992;7:27–32.
10. deLooze MP, Groen H, Horemans H, Kingma I, Van Dieen JH. Abdominal muscles contribute in a minor way to peak spinal compression in lifting. J Biomech 1999;32:655–62.
11. Fathallah FA, Marras WS, Parnianpour M, Granata KP. A method for measuring external spinal loads during unconstrained free-dynamic lifting. J Biomech 1997;30:975–8.
12. Gardner-Morse M, Stokes IA. The effects of abdominal muscle coactivation on lumbar spine stability. Spine 1998;23:86–92.
13. Gardner-Morse M, Stokes IAF, Laible JP. Role of muscles in lumbar stability in maximum extension efforts. J Orthop Res 1995;13:802–8.
14. Granata KP, Marras WS. An EMG-assisted model of biomechanical trunk loading during free-dynamic lifting. J Biomech 1995a;28:1309–17.
15. Granata KP, Marras WS. An EMG-assisted model of loads on the lumbar spine during asymmetric trunk extensions. J Biomech 1993;26:1429–38.
16. Granata KP, Marras WS. Biomechanical Models in Ergonomics. In: Bhattacharya A, McGlothlin JD, eds. Handbook of Occupational Ergonomics. New York: Mercel Dekker, 1996.
17. Granata KP, Marras WS. The influence of trunk muscle coactivity upon dynamic spinal loads. Spine 1995;20:913–9.
18. Granata KP, Marras WS, Davis KG. Biomechanical assessment of lifting dynamics, muscle activity, and spinal loads while using three different style lifting belts. Clin Biomech 1997;12:107–15.
19. Granata KP, Marras WS, Fathallah FA. A method for measuring external trunk loads during dynamic lifting exertions. J Biomech 1995;29:1219–22.
20. Herrin GA, Jaraiedi M, Anderson CK. Prediction of overexertion injuries using biomechanical and psychophysical models. Am Ind Hyg Assoc J 1986;47:322–30.
21. Hodges PW, Richardson CA. Inefficient muscular stabilization of the lumbar spine associated with low back pain: A motor control evaluation of transversus abdominis. Spine 1996;21:2640–50.
22. Hoffer JA, Andreassen S. Regulation of soleus muscle stiffness in premammillary cats: Intrinsic and reflex components. J Neurophysiol 1981;45:267–85.
23. Hughes RE, Bean JC, Chaffin DB. Evaluating the effect of cocontraction in optimization models. J Biomech 1995;28:875–8.
24. Hunter IW, Kearney RE. Dynamics of human ankle stiffness: Variation with mean ankle torque. J Biomech 1982;15:747–52.
25. Kelsey KL, Githens PB, White AA III, et al. An epidemiologic study of lifting and twisting on the job and risk for acute prolapsed lumbar intervertebral disc. J Orthop Res 1984;2:61–6.
26. Marras WS, Fathallah FA, Miller RJ, Davis SW, Mirka GA. Accuracy of a three-dimensional lumbar motion monitor for recording dynamic trunk motion characteristics. Int J Ind Ergonomics 1992;9:75–87.
27. Marras WS, Granata KP. A biomechanical assessment and model of axial twisting in the thoracolumbar spine. Spine 1995;20:1440–51.
28. Marras WS, Granata KP. Spine loading during trunk lateral bending motions. J Biomech 1996;30:697–703.
29. Marras WS, Lavender SA, Leurgans S, et al. Biomechanical risk factors for occupationally related low back disorder risk. Ergonomics 1995;38:377–410.
30. Marras WS, Mirka GA. A comprehensive evaluation of trunk response to asymmetric trunk motion. Spine 1992;17:318–26.
31. Marras WS, Mirka GA. Muscle activities during asymmetric trunk angular accelerations. J Orthop Res 1990;8:824–32.
32. Marras WS, Sommerich CM. A three-dimensional motion model of loads on the lumbar spine: II. Model validation. Hum Factors 1991;33:139–49.
33. McGill SM. The biomechanics of low back injury: Implications on current practices in industry and the clinic. J Biomech 1999;30:465–75.
34. Mirka GA, Marras WS. A stochastic model of trunk muscle coactivation during trunk bending. Spine 1993;18:1396–409.
35. Nichols TR, Houk JC. Improvement of linearity and regulation of stiffness that results from the actions of the stretch reflex. J Neurophysiol 1976;39:119–42.
36. NIOSH. A Work Practices Guide for Manual Lifting. Technical Report No. 81–122. Cincinnati, OH: U.S. Dept. of Health and Human Services (NIOSH), 1981.
37. Norman RW, Wells R, Neumann P, et al. A comparison of peak *vs* cumula-

tive physical work exposure risk factors for the reporting of low back pain in the automotive industry. Clin Biomech 1998;13:561–73.
38. Panjabi MM. The stabilizing system of the spine: Part I. Function, dysfunction, adaptation, and enhancement. J Spinal Disord 1992;5:383–9.
39. Panjabi MM. The stabilizing system of the spine: Part II. Neutral zone and instability hypothesis. J Spinal Disord 1992;5:390–7.
40. Pope MH, Panjabi M. Biomechanical definitions of spinal stability. Spine 1985;10:255–6.
41. Punnet L, Fine LJ, Keyserling WM, Herrin GD, Chaffin DB. Back disorders and nonneutral trunk postures of automobile assembly workers. Scand J Work Environ Health 1991;17:337–46.
42. Quint U, Wilke HJ, Shirazi-Adl A, Parnianpour M, Loer F. Importance of intersegmental trunk muscles for the stability of the lumbar spine: A biomechanical study *in vitro*. Spine 1998;23:1937–45.
43. Rosenthal NP, McKean TA, Roberts WJ, Terzuolo CA. Frequency analysis of stretch reflex and its main subsystems in triceps surae of the cat. J Neurophysiol 1970;33:713–49.
44. Thelen DG, Ashton-Miller JA, Schultz AB. Cocontraction of lumbar muscles during the development of time-varying triaxial moments. J Orthop Res 1995; 13:390–8.
45. Thelen DG, Ashton-Miller JA, Schultz AB. Lumbar muscle activities in rapid three-dimensional pulling tasks. Spine 1996;21:605–13.
46. Thompson JMT, Hunt GW, eds. The general conservative theory. In: Elastic Instability Phenomena. New York: John Wiley, 1984:1–26.
47. Wilder DG, Pope MH, Frymoyer JW. The biomechanics of lumbar disc herniation and the effect of overload and instability. J Spinal Disord 1988;1:16–32.
48. Wilke HJ, Wolf S, Claes LE, Arand M, Weisend A. Stability increase of the lumbar spine with different muscle groups. Spine 1995;20:192–8.
49. Zetterberg C, Andersson GB, Schultz AB. The activity of individual trunk muscles during heavy physical loading. Spine 1987;12:1035–40.

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