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Effects of external trunk loads on lumbar spine stability

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Abstract

Stability of the lumbar spine is an important factor in determining spinal response to sudden loading. Using two different methods, this study evaluated how various trunk load magnitudes and directions affect lumbar spine stability. The first method was a quick release procedure in which effective trunk stiffness and stability were calculated from trunk kinematic response to a resisted-force release. The second method combined trunk muscle EMG data with a biomechanical model to calculate lumbar spine stability. Twelve subjects were tested in trunk flexion, extension, and lateral bending under nine permutations of vertical and horizontal trunk loading. The vertical load values were set at 0, 20, and 40% of the subject's body weight (BW). The horizontal loads were 0, 10, and 20% of BW. Effective spine stability as obtained from quick release experimentation increased significantly ($p < 0.01$) with increased vertical and horizontal loading. It ranged from 785 (S.D. = 580) Nm/rad under no-load conditions to 2200 (S.D. = 1015) Nm/rad when the maximum horizontal and vertical loads were applied to the trunk simultaneously. Stability of the lumbar spine achieved prior to force release and estimated from the biomechanical model explained approximately 50% of variance in the effective spine stability obtained from quick release trials in extension and lateral bending ($0.53 < R^2 < 0.63$). There was no such correlation in flexion trials. It was concluded that lumbar spine stability increased with increased trunk load magnitude to the extent that this load brought about an increase in trunk muscle activation. Indirectly, our data suggest that muscle reflex response to sudden loading can augment the lumbar spine stability level achieved immediately prior to the sudden loading event.

Keywords: Lumbar spine; Stability; Sudden loading; Modeling

1. Introduction

Stability of the lumbar spine, provided by surrounding musculature and controlled by the central nervous system, is an important factor in determining trunk response to sudden loading (Cholewicki and McGill, 1996; Gardner-Morse et al., 1995; Panjabi, 1992). Low back injuries are frequently caused by slips and falls, which impose sudden loading/unloading on the lumbar spine (Bigos et al., 1986; Manning et al., 1984; Manning and Shannon, 1981; Omino and Hayashi, 1992; Troup et al., 1981). It has been hypothesized that either inappropriate motor control responses by the trunk musculature (Magnusson et al., 1996; Panjabi, 1992; Radebold et al., 2000) or inadequate stabilization of the lumbar spine prior to a sudden loading incident (Cholewicki and McGill, 1996; Panjabi, 1992) may be the cause of low back injuries. Most likely, both the mechanical stability level of the spine prior to loading and the reflex response of the muscles immediately after loading combine to determine the kinematic response of the trunk and subsequent likelihood of injury. Kearney et al. (1997) showed that, at least in the ankle joint, reflex mechanisms could generate torques of similar magnitude as those generated by the pre-set joint stiffness. It is presently not known how the motor control system regulates the stability of the lumbar spine under external loads of various magnitudes and directions.

Active control of spine stability is achieved through the regulation of force in the surrounding muscles. Muscle force, in turn, is approximately linearly proportional to muscle stiffness (Bergmark, 1989; Cholewicki and McGill, 1995; Crisco and Panjabi, 1991). Therefore, co-activation of agonistic and antagonistic trunk muscles stiffens the lumbar spine and increases its stability

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(Bergmark, 1989; Cholewicki and McGill, 1996; Cholewicki et al., 1997; Gardner-Morse et al., 1995). Cholewicki and McGill (1996) observed that the stability index (SI) of the lumbar spine increased with increased task demands, quantified as the spine compression force. They explained this increase in spine stability on the basis of increased muscle co-activation with increased muscular effort. Krajcarski et al. (1999) also observed that higher trunk pre-loads resulted in lower trunk rotations in response to a suddenly applied flexion load.

There are two general approaches for assessing the stability of the lumbar spine. One approach relies on mathematical modeling of the spine and quantifying its stability based on estimated muscle forces generated during a given task (Bergmark, 1989; Cholewicki and McGill, 1996; Gardner-Morse et al., 1995). The other approach is based on perturbation experiments and the calculation of the effective trunk stiffness and stability from kinematic data obtained after the perturbation. The quick release method, falls into the latter category and has been used to identify stiffness in various joints (Hogan, 1990; Hunter and Kearney, 1982; Lacquaniti et al., 1982; Tsuji et al., 1995; Winters et al., 1988). If the trunk kinematic response immediately after the force release is determined entirely by the stability of the lumbar spine achieved prior to the force release, then the estimates of stability obtained from the above two methods should correlate. However, in a quick release experiment, trunk kinematics may be modulated by the reflex response of agonistic and antagonistic muscles after the resisted force release. In this case, where both the stability of the lumbar spine prior to force release and the muscle reflex response after the release combine to determine the trunk kinematics, only a partial (or even non-existent) correlation will be found between these two estimates.

The purpose of the present study was to determine how various external load magnitudes and directions affect lumbar spine stability. Experimental quick release and analytical modeling estimates of spine stability were compared across 12 subjects. It was hypothesized that (1) lumbar spine stability will increase with increased external load magnitude and (2) the experimental and analytical estimates of spine stability will partially correlate.

2. Methods

A two-factor experimental design was used, in which spine stability was a dependent variable and the vertical and horizontal loads constituted the two factors. Stability was determined experimentally and analytically. The experimental assessment of lumbar spine stability was accomplished by calculating the instantaneous trunk stiffness and stability from the kinematic data of trunk response to a sudden force release (quick release method). Analytically, lumbar spine stability was estimated from a three-dimensional, 18 degrees-of-freedom biomechanical model, assisted with electromyographic data (EMG) recorded from 12 major trunk muscles immediately prior to the release. As a measure of stability, the curvature of the system's potential energy computed in the vicinity of the static equilibrium was used to compare the results between the two methods.

Twelve healthy subjects (6 males and 6 females; mean age 24, S.D. 6 years; mean height 1.74, S.D. 0.11 m; mean weight 65, S.D. 13 kg) with no previous history of low back pain volunteered for the experiment. Each volunteer signed the informed consent form outlining the protocol approved by the Yale Human Investigation Committee. Subjects were placed in a semi-seated position in a jig that restricted hip motion while leaving the upper torso free to move in all directions (Fig. 1). In the quick release method, the subjects exerted isometric trunk extension, flexion, and lateral bending to the left and right. Release occurred when subjects reached 35% of their maximum force, as measured for each direction at the beginning of the experiment. The release force averaged 172 (S.D. = 54) N. Resistance was provided by a cable attached to a chest harness at approximately the T9 level. This cable was held by an electromagnet, which was suddenly released by the researcher with a random delay after the target force level was reached and maintained. The force level was displayed for the researcher and subjects using an oscilloscope. The resulting trunk motion was measured at 80 Hz with an inductive sensor

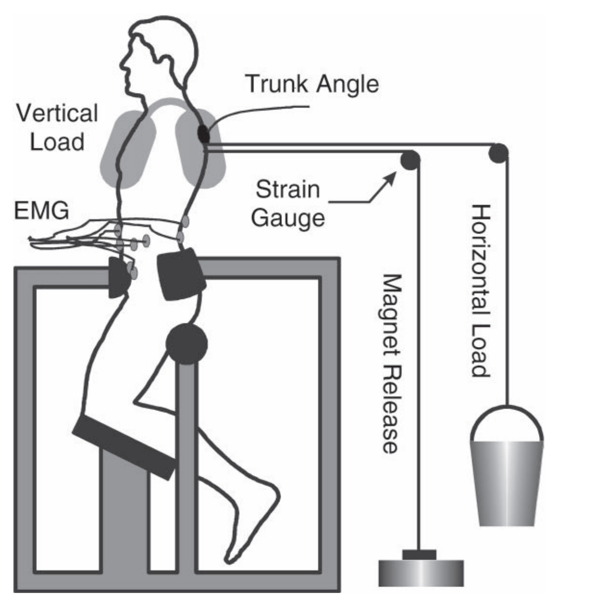

Fig. 1. A subject positioned in test apparatus configured for a trunk flexion trial. To change the trial direction, the entire load application assembly was detached from the apparatus and moved around the seated subject. The horizontal load and the force resisted for a quick release were always applied in parallel. The release force was measured with a strain gauge applied to the base of a pulley bracket.

(Flock of Birds, Ascension Technologies, VT) affixed to the back at the T9 level.

Three quick release trials were performed at each permutation of vertical and horizontal loading (applied in addition to the release force). The vertical load was applied with bagged lead shot evenly distributed between four pouches in a tight chest harness. The center of mass of the load was at approximately the T9 level. The horizontal load was applied through a cable passing over a low-friction pulley. On one end, this cable was attached to the chest harness at T9, while the other end supported a bucket filled with lead shot (Fig. 1). The horizontal load was always applied in addition to and in parallel with the quick release resistance force. The entire load-application assembly could be detached from the jig and moved around the seated subject to change the direction of load and release. All permutations of three magnitudes of vertical and horizontal loading were used. The vertical load values were set at 0, 20, and 40% of the subject's body weight (BW). The horizontal loads were 0, 10, and 20% of BW. These load values were selected during the earlier preliminary study to cover the largest possible range of loads without fatiguing the subjects. All trials (directions and load levels) were performed in random order.

In the first method, trunk stiffness was calculated from the trunk motion data in accordance with a quick release protocol (Hogan, 1990; Hunter and Kearney, 1982; Lacquaniti et al., 1982; Tsuji et al., 1995; Winters et al., 1988). The trunk was represented as a second-order system with viscoelastic properties, oscillating freely after the release of a moment that subjects had been resisting (Fig. 2). Amplitude and frequency of such oscillations measured immediately after the release, but before voluntary muscle intervention takes place, are determined by the trunk inertia (I), damping coefficient (B) and stiffness coefficient (K) established prior to the release. For small trunk angles (θ) (Fig. 2):

$$I\ddot{\theta} + B\dot{\theta} + K(\theta - \theta_o) = m_v g L \sin\theta - m_h g L \cos\theta, \qquad (1)$$

where $m_v g$ is trunk and vertical weight, $m_h g$ is horizontal weight, L is the height measured from the center of trunk mass assumed to be at T9 level to the L4/L5 joint, and θ_o is a hypothetical resting angle of spring K. Total trunk (including arms and head) mass and moment of inertia were calculated from the subjects' weight and height (Winter, 1990). The additional masses, used for generating the vertical and horizontal loads, were included in the calculation of total trunk inertia (I) (Eq.(1)). Coefficients B, K, and a constant C (encompassing θ_o and integration constants) were obtained with a curve-fitting algorithm designed to obtain the best match between the modeled and measured trunk rotation trajectories. This procedure was applied to a double integration of Eq. (1), because integration is numerically a more robust operation than differentiation (Tsuji et al., 1995):

$$I\theta + B\int\theta \, \mathrm{d}t + K\iint\theta \, \mathrm{d}t^2 + Ct^2$$

$$= gL\iint(m_v \sin\theta - m_h \cos\theta) \, \mathrm{d}t^2. \qquad (2)$$

A preliminary study indicated that the minimum length of data needed to identify the parameters in Eq. (2) accurately was equivalent to at least a quarter of the wavelength. Therefore, angular trunk motion data, taken from the time of magnet release to the point of maximum trunk deflection (average = 250 ms, S.D. = 112 ms), was used for a curve fit. This time interval was short enough to eliminate the voluntary intervention. However, involuntary muscle reflex responses can occur between 40 and 80 ms following the resisted force release (Radebold et al., 2000), in time to modify trunk kinematics (Fig. 3). Therefore, the measurement of trunk stiffness obtained from the above method was an effective stiffness, combining pre-set muscle stiffness and reflex response.

The effective stability of the lumbar spine was then calculated in the vicinity of the static equilibrium as a second derivative (curvature) of the potential energy, given the effective trunk stiffness, trunk mass, and the external loads. The potential energy change (V) of the system depicted in Fig. 2 is

$$V = \frac{1}{2}K\theta^2 - L(1 - \cos\theta)m_v g + L\sin\theta m_h g. \qquad (3)$$

Linearizing Eq. (3) for small θ by preserving only quadratic terms of θ, gives

$$V = \frac{1}{2}K\theta^2 - \frac{1}{2}L\theta^2 m_v g + L\theta m_h g. \qquad (4)$$

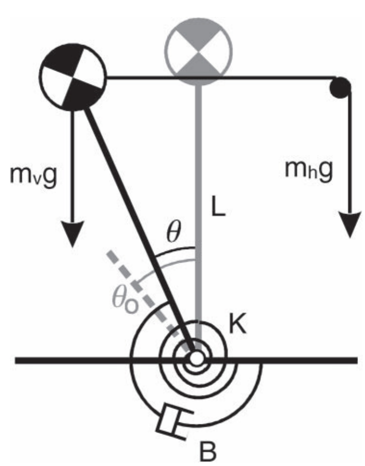

Fig. 2. A free body diagram of a trunk oscillating after the quick force release. Effective trunk stiffness (K) was calculated from trunk motion data using Eq. (2). $m_v g$ is the trunk and vertical weight, $m_h g$ is the horizontal weight, L is the height measured from the L4/L5 joint to the center of trunk mass assumed to be at the T9 level, and θ_o is a hypothetical resting angle of spring K.

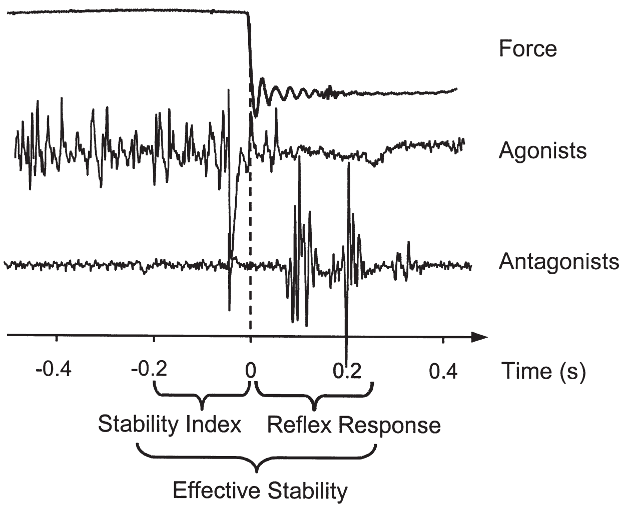

Fig. 3. An example of trunk muscle reflex response to quick force release that occurred at time zero. For a given force direction, agonists were the muscles that were active prior to the force release and switched off after the release. Antagonists were the muscles that were relatively inactive prior to the force release and switched on after the release. Output from the biomechanical model reflected the stability level of the lumbar spine, which was achieved prior to the force release, while the quick release method estimated effective spine stability, which included changes in muscle activation level after the release. Vertical axis units are arbitrary.

First and second derivatives of V are as follows:

$$\frac{dV}{d\theta} = K\theta - L\theta m_v g + Lm_h g, \quad (5)$$

$$\frac{d^2V}{d\theta^2} = K - Lm_v g. \quad (6)$$

Eq. (6) quantifies, under the static conditions, the effective stability of the lumbar spine (curvature of the potential energy) based on the effective trunk stiffness (K) obtained from the quick release experiment.

The second method for estimating stability of the lumbar spine with its surrounding musculature was based on an analytical model that had been developed previously (Cholewicki and McGill, 1996). Briefly, it consisted of a rigid pelvis and sacrum, five lumbar vertebrae separated by a lumped parameter, nonlinear disc and ligament equivalent for rotational joint stiffness about the three axes, rigid ribcage, and 90 muscle fascicles. Three axes of rotations were assigned to each intervertebral joint between T12 and S1, for a total of 18 DOF (6 joints $\times$ 3 DOF each). The moments and forces (after accounting for passive tissue contribution), necessary to balance the external load and upper body weight, were partitioned between all 90 muscle fascicles with the assistance of EMG. For that purpose, the cross-bridge bond distribution moment (DM) model for obtaining muscle force and stiffness simultaneously (Cholewicki and McGill, 1995) and the EMG-assisted optimization approach to balance the moment equations (Cholewicki and McGill, 1994; Cholewicki et al., 1995) were used. Stability analysis was performed in accordance with the minimum potential energy principle. Average curvature of the surface of the system's potential energy in the vicinity of the static equilibrium served as the relative stability index (SI).

The EMG signals were recorded at 1600 Hz from 12 muscles (left and right rectus abdominis, external and internal oblique, latissimus dorsi, thoracic and lumbar erector spinae) according to a previously established protocol (Cholewicki and McGill, 1996; Cholewicki et al. 1997). It was assumed that the muscle activation pattern established prior to a sudden trunk perturbation determines the spine stability and in turn the kinematics of the trunk response to that perturbation. Accordingly 200 ms of EMG data recorded immediately prior to the magnet release were digitally rectified and averaged. The baseline EMG values recorded when the subjects were lying completely relaxed were subtracted. Finally the data were normalized to the EMG activity recorded during maximum voluntary contractions. Combined with external load magnitudes these EMG data served as input for the model.

Both the experimental and the modeled estimate of lumbar spine stability were averaged over three trials. Two-factor (horizontal and vertical load) repeated measures ANOVA ($p < 0.01$) was used to statistically test the effects of external loads on spine stability.

3. Results

Eq. (2) fit very well the experimental data of trunk angular motion recorded after the quick force release (Fig. 4). The average root-mean-square error was 0.30° (S.D. = 0.34°).

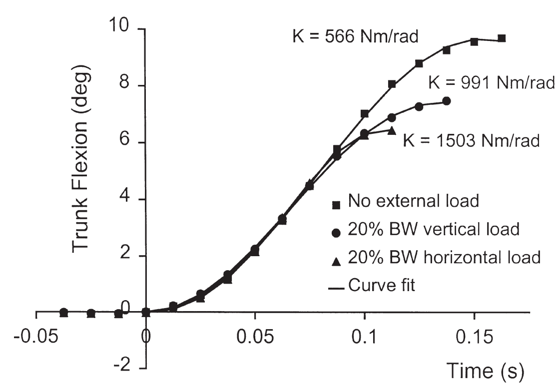

Fig. 4. An example of trunk motion response to quick force release that occurred at time zero. Data belong to one subject executing trials with no external load and with 20% of body weight (BW) applied to the trunk vertically and horizontally. Points represent experimental data and solid lines are curves calculated with the best-fit coefficients using Eq. (2). The overall average root-mean-square error of the curve fit was 0.30°. Corresponding effective trunk stiffness (K) is noted beside each curve.

Table 1
Main effects of load direction and magnitude on effective trunk stiffness (Nm/rad) estimated from quick release experiment

Horizontal load (%BW)	0	10	20
Extension[a]	1237 (698)	1839 (829)	2004 (1042)
Flexion[a]	1253 (760)	1707 (716)	1872 (816)
Left lateral bending[a]	1180 (722)	1512 (715)	1828 (743)
Right lateral bending[a]	1191 (685)	1816 (724)	2120 (849)
Vertical Load (%BW)	0	20	40
Extension[a]	1493 (616)	1606 (1030)	1980 (965)
Flexion[a]	1028 (688)	1586 (800)	2218 (865)
Left lateral bending[a]	1202 (662)	1514 (624)	1804 (891)
Right lateral bending[a]	1225 (603)	1819 (746)	2083 (764)

[a]The horizontal and vertical load applied to the trunk increased effective trunk stiffness significantly ($p < 0.01$). No significant interactions between these two loading conditions were found. Therefore, the values presented for one load direction were averaged across all levels of load applied in the other direction (standard deviations are in parenthesis).

Vertical and horizontal loads applied to the trunk both in combination and separately resulted in a significant increase in effective trunk stiffness and stability in the quick release experiment (Table 1 and Fig. 5). There was no interaction between these two load directions indicating that their effects on effective trunk stiffness and stability were additive. These results held true for all of the exertion directions considered in this study (extension, flexion, and lateral bending to the left and right). The increase in effective trunk stiffness and stability due to added horizontal load was approximately twice as large as the effective stiffness and stability increase due to vertical loading (Table 1 and Fig. 5). In other words, a load of 20% of BW applied to the trunk horizontally had a similar trunk-stiffening effect as a load of 40% of BW applied vertically (Table 1 and Fig. 5). Damping coefficients (B) generally increased with increased external trunk load, but overall their magnitudes were negligible (mean = 42, S.D. = 64 Nm s/rad).

The stability index (SI) of the lumbar spine prior to the force release, estimated from the biomechanical model, increased significantly with increased horizontal load (Table 2). However, in the case of vertical loading, the biomechanical model predicted no significant change in SI for all motion directions but flexion. In this direction, the SI paradoxically decreased with increased vertical load on the trunk (Table 2). However, because the vertical load was applied slightly anterior to the lumbar spine, the EMG of the abdominal musculature decreased with added vertical load in flexion. In general, the effects of trunk loads were greater on the effective spine stability obtained from a quick release experiment than on spine stability estimated from the biomechanical model.

Partial correlation existed between the estimates of lumbar spine stability calculated from the biomechanical

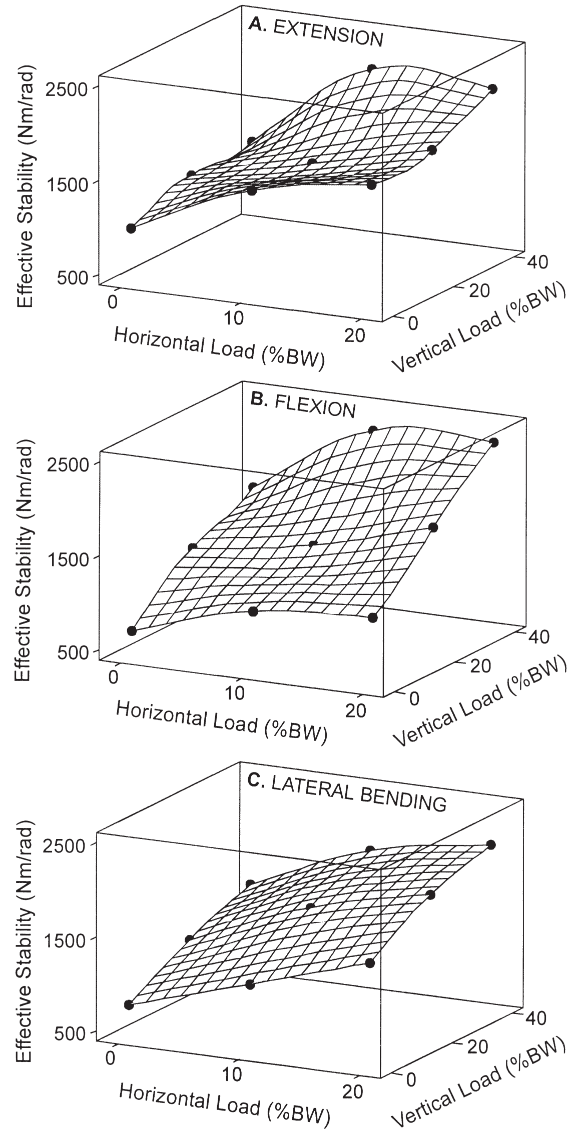

Fig. 5. The effects of trunk loads (direction and magnitude) on effective spine stability estimated from quick release experiment in extension (A), flexion (B), and lateral bending (C). Both horizontal and vertical loads applied to the trunk increased effective spine stability significantly ($p < 0.01$) and their effects were additive. Values marked with dark spheres were interpolated to create the three-dimensional surface graphs. Left and right lateral bending results were combined into one graph because there was no significant difference between these trials. (% BW)— percent body weight.

model and the effective spine stability obtained from quick release experiment in extension and lateral bending to the left and right (Fig. 6A, C, D). Squared correlation coefficients (R^2) ranged from 0.53 (Fig. 6D) to 0.63 (Fig. 6A) and indicated that the lumbar spine stability level achieved immediately prior to the force release could explain approximately 50% of variance in effective spine stability calculated from trunk kinematics obtained immediately after the force release. In flexion, there was no correlation between the estimates of spine stability obtained from the two methods (Fig. 6B). However, when both stability estimates were averaged across all vertical loads and the correlations recalculated, the following R^2 were obtained: 0.90, 1.00, 1.00, 0.98 for the extension, flexion, left, and right lateral bending trials, respectively. In other words, poor correlations between the stability estimates from the two methods were caused entirely by the discrepancies in the results obtained under the various vertical loading conditions.

Table 2
Main effects of load direction and magnitude on spine stability index (SI) (Nm/rad) achieved prior to force release and estimated from an EMG-assisted biomechanical model

Horizontal load (%BW)	0	10	20
Extension[a]	423 (85)	477 (94)	532 (102)
Flexion[a]	270 (46)	309 (59)	320 (52)
Left lateral bending[a]	335 (58)	380 (70)	425 (82)
Right lateral bending[a]	315 (57)	371 (77)	417 (84)
Vertical load (%BW)	0	20	40
Extension	473 (92)	474 (96)	486 (97)
Flexion[a]	322 (62)	291 (48)	285 (45)
Left lateral bending	382 (73)	376 (68)	382 (69)
Right lateral bending	374 (93)	360 (64)	369 (67)

[a]Only the horizontal load applied to the trunk increased the SI significantly ($p < 0.01$). No significant interactions between the horizontal and vertical loading conditions were found. Therefore, the values presented for one load direction were averaged across all levels of load applied in the other direction (standard deviations are in parenthesis).

4. Discussion

The effect of varying trunk loads on lumbar spine stability has been evaluated with two methods. The quick release method estimated spine stability via the calculation of trunk stiffness from trunk kinematic response after the force release. Because the involuntary muscle reflexes may have the potential to augment trunk kinematics, we referred to these estimates of stability as the "effective spine stability." The second method used trunk muscle EMG data in combination with a biomechanical model to calculate the lumbar spine stability level achieved immediately prior to the resisted force release.

The effective spine stability, obtained with the quick release method, increased with increased vertical or horizontal load on the trunk. In contrast, the lumbar spine stability level achieved immediately prior to the force release, increased significantly with added external load only to the extent that this load brought about an increase in trunk muscle activity. For example, no significant change in spine stability was found in extension and lateral bending with added vertical load, because it did not result in a significant increase of trunk muscle activity. On the other hand, the horizontal load created large bending moments about the lumbar spine and required significant counteractive muscular effort. Lumbar spine stability increased significantly in all directions with added horizontal load. The increased muscle activation lead to greater spine stability, because muscle stiffness increased in approximately linear proportion to muscle force. These results were consistent with previous

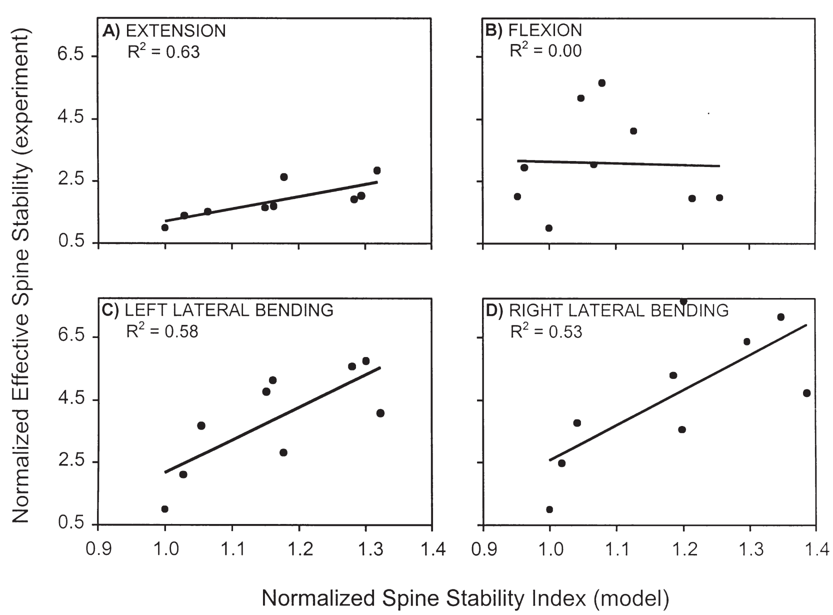

Fig. 6. Correlation between the estimates of lumbar spine stability level achieved prior to the force release and the effective spine stability calculated from trunk kinematics after the force release. Partial correlation existed for the trials in trunk extension (A) and lateral bending to the left (C) and right (D), but not for the trials in trunk flexion (B). All stability data were normalized to the values obtained from the trials where no external loads (vertical and horizontal) were applied to the trunk.

observations that the SI increased along with increased task demands, quantified as the lumbar spine compression force (Cholewicki and McGill, 1996).

Flexion trials provided another example of the close relationship between lumbar spine stability and trunk muscle activation level. In flexion, the estimated lumbar spine stability achieved prior to the force release decreased with increased vertical loading. Closer inspection of EMG data in flexion trials revealed that abdominal muscle activity became smaller with each increase of the vertical load added to the torso. The center of mass of the upper body and the external load was anterior to the lumbar spine. Therefore, the vertical external load, by creating a flexor moment, assisted subjects in reaching their target force in trunk flexion. The modeled spine stability in flexion trials was, therefore, lower with added vertical load, due to lower muscle activity.

The main limitation of the EMG-assisted biomechanical model was the lack of access to deep trunk muscles, in addition to the many other assumptions already discussed in an earlier publication (Cholewicki and McGill, 1996). Furthermore, there is currently no method available, which allows for direct validation of such a model. The quick release experiment assumed the trunk to be a simple one-degree-of-freedom system with the center of rotation located at L4/L5 joint. Stokes (1987) showed that L3 might be a better approximation of a physiological center of lumbar spine rotation. However, this change would only give results scaled by a ratio of the assumed spine lengths (i.e. T9-L4/L5 and T9-L3); a negligible percentage change. Stiffness and damping coefficients were also assumed to be constant over the duration of the data segment used to calculate these coefficients. In reality, reflex response changes the muscle activation and hence its stiffness and damping. Therefore, we termed these coefficients and measures as the "effective" stiffness, damping and stability. Similarly, the upper body (trunk, head, neck, and arms) was assumed to be rigid. To minimize errors stemming from this assumption, we asked the subjects to cross their arms tightly against their chest during the experiment. Notwithstanding these limitations, the derived equations of motion described experimental trunk rotation data with high accuracy (Fig. 4), indicating that the above assumptions were reasonable.

The major difference between the two methods for estimating stability of the lumbar spine used in this study was the way in which they accounted for trunk muscle reflex response. The EMG-assisted model simply returned estimates of spine stability achieved immediately prior to the resisting force release. On the other hand, the effective spine stability obtained from quick release experiments reflected the changes in muscle activation that occurred immediately after the force release, in combination with the spine stability level established prior to the force release. Because trunk kinematics recorded up to 250 ms after the release were used for estimating the effective spine stability, any changes in trunk muscle activation could only be due to involuntary muscle reflex response. Indeed, we found trunk muscle reaction times under similar force-release conditions average between 40 and 80 ms (Radebold et al., 2000). Hence, the differences between the estimates of spine stability obtained from the above two methods, apart from modeling and experimental limitations, reflect the effect of trunk muscle reflex response to sudden unloading.

Because of the above differences, no attempt was made to compare the absolute values of spine stability estimated by the two methods. Instead, the normalized or relative magnitudes of the stability index (SI) and the effective spine stability were correlated. An excellent agreement between the results of these two methods existed for horizontal loading conditions ($0.90 < R^2 < 1.00$). Therefore, the poor correlation obtained under the vertical loading scheme could be deduced, with a high level of confidence, to be caused by the fact that the experimental estimates of effective spine stability included the effect of muscle reflex response whereas the modeled estimates did not. This could also be the reason why even the normalized estimates of effective spine stability were several times greater than the modeled estimates. Also consistent with the above explanation is the observation that the effective spine stability for horizontal loading was approximately twice as large as the effective stability for the vertical loading scheme. The effective spine stability was composed of the stability achieved immediately prior to the force release and the muscle reflex response after the release. As indicated by the modeling results, there was no increase in spine stability achieved prior to the force release under the vertical loading scheme, while under the horizontal loading there was. Hence, the effective spine stability under the horizontal loading conditions was greater than the vertical.

Comparison of results from the two methods suggested that both the mechanical stability level of the spine prior to the force release, and the reflex response of the trunk muscles after the force release, combine to determine the kinematic response of the trunk. Subsequently, the kinematics of the lumbar spine determine the likelihood of injury following a sudden loading or unloading incident. These results are consistent with Kearney et al. (1997) who demonstrated that reflex contribution to ankle stiffness could be as large as the pre-set muscle stiffness surrounding the joint. More specifically, it appears that under most circumstances muscle reflex response can compensate for insufficient initial stability of the spine to constrain trunk motion within a safe boundary. If the spine stability level is sufficiently high for a given external load carried prior to a sudden loading incident, minimal adjustment by muscle reflex response may be necessary. In our study, this was the case for horizontal trunk loading, where spine stability assessed before the quick release correlated almost perfectly with the effective trunk stability estimated from the quick release experiment. However, if the initial spine stability is insufficient in relation to the external load carried, a fast and strong reflex response may be crucial in preventing large intervertebral displacements or buckling of the spine, and subsequent damage of soft tissues under sudden loading conditions. In this study, no increase in spine stability prior to the quick release was found when the vertical load on the trunk was increased. From that perspective, it seems that vertical trunk loading may be potentially more dangerous if a sudden slip or fall occurs, because it generally does not generate as much muscle tension and initial spine stability as does horizontal-type loading. These findings are also supported by Stokes et al. (1999) who found that trunk muscle response to perturbation is less likely to occur when the trunk is preloaded resulting in a higher stability of the lumbar spine, than it is without a trunk preload.

Some sudden loading/unloading situations may arise in which trunk muscle reflex response may not be effective enough in augmenting the lumbar spine stability. There exists a delay in the reflex response taking time before sufficient muscle force is generated. Some unexpected loading scenarios may be too fast and with too high of a magnitude for the reflex response to control the ensuing trunk displacement effectively and safely. For example, individuals unconsciously increase coactivation of trunk musculature prior to sudden loading of unknown magnitude in an attempt to maximize their spine stability (Lavender et al., 1989; Marras et al., 1987). This preparatory strategy may originate from the threat that the load magnitude may be too high to avoid pain or injury when relying solely on a muscle reflex response. In fact, Omino and Hayashi (1992) found that lack of preparation for dynamic posture perturbations lead to a higher occurrence of low back pain among airline attendants. There also exists evidence that individuals with chronic low back pain exhibit delayed trunk muscle response to sudden loading (Magnusson et al., 1996; Radebold et al., 2000), which may constitute a predisposing risk factor to sustaining a low back injury under such circumstances.

The following conclusions were reached: (1) Lumbar spine stability increased with the increased trunk load magnitude to the extent that this load brought about an increase in trunk muscle activation. (2) Trunk muscle reflex response to sudden loading can augment the stability level of the lumbar spine achieved immediately prior to a sudden loading event. This latter conclusion was deduced indirectly from our data and future experimental studies should address this issue directly.

Acknowledgements

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