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Effects of Abdominal Muscle Coactivation on the Externally Preloaded Trunk: Variations in Motor Control and Its Effect on Spine Stability

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Study Design. A repeated measures biomechanical analysis of the effects of abdominal bracing in preparation for a quick release of the loaded trunk.

Objectives. To quantify the ability of individuals to abdominally brace the externally loaded trunk, and assess their success in achieving and enhancing appropriate spine stability.

Summary of Background Data. Spine stability requires trunk muscle coactivation, which demands motor control skill that differs across people and situations. The quick release protocol may offer insight into the motor control scheme and subsequent effect on spine stability.

Methods. There were 10 individuals who sat, torso upright, in an apparatus designed to foster a neutral spine position. They were instructed to support a posteriorly directed load to the trunk in either their naturally chosen manner, or by activating the abdominal muscles to 10%, 20%, or 30% of maximum ability. The externally applied load was then quickly released, thereby unloading the participant. Muscle pre-activation patterns, spine stability, and kinematic measures of trunk stiffness were quantified.

Results. Participants were able to stabilize their spine effectively by supporting the load in a naturally selected manner. Conscious, voluntary overdriving of this natural pattern often resulted in unbalanced muscular activation schemes and corresponding decreases in stability levels.

Conclusions. Individuals in an externally loaded state appear to select a natural muscular activation pattern appropriate to maintain spine stability sufficiently. Conscious adjustments in individual muscles around this natural level may actually decrease the stability margin of safety.

Key words: quick release, biomechanical model, spine stability, stiffness, abdominal brace, co-contraction. **Spine 2006;31:E387–E393**

Spine stability can be maintained with moderate levels of trunk muscle coactivation.[1–3] The amount of coactivity necessary to ensure a stable state increases with the load demand of the task.[4,5] The musculature surrounding the spine acts to provide a stiffening mechanism to the vertebral joints, thereby reducing the likelihood of a net energy loss (potential for buckling, tissue damage) in response to an applied perturbation. Therefore, increasing the muscular coactivation around an equilibrium state should theoretically increase the stiffness and stability of the trunk. However, similar to the guy wire system on a ship's mast, the muscle tensions and stiffness must be balanced and tuned to one another. Uncoordinated muscle coactivation has the potential to destabilize the spine by creating force, and thus, torque, imbalances between the muscles supporting the spine. Maintenance of the spine in static equilibrium is essential in assessing the stability of the spine; however, this equilibrium can be achieved through a variety of muscular activation patterns. Changing these patterns will change the ability of the active muscles to stiffen and stabilize the spinal column.

Numerous studies have shown the beneficial effects of preloading the trunk in preparation for suddenly applied loads[6–9] and quick load releases.[4] Each of these studies measured trunk muscle activation patterns during the preload state as well as the kinematic response of the trunk after perturbation. Those studies using the "sudden load" approach performed such measures to infer the stability of the system, indicating that stability increased as a result of the increased muscular activation levels. Cholewicki *et al*[4] went a step further and mathematically quantified the level of spine stability immediately before the load release, and found it to increase with increased preloads.

Bracing the trunk by consciously activating all the abdominal musculature (abdominal bracing) is another method of increasing the coactivation patterns of the trunk musculature. As the abdominal muscles increase their activity, and thus, their force and torque output, back extensor muscles must act in kind to maintain an equilibrium state. Therefore, ideally, all surrounding muscles would increase their activation levels in concert to ensure a coordinated bracing effort. Abdominal bracing has been studied as a method of retraining abdominal muscles[10–12] and, recently, as a method of increasing spine stability in preparation for loads rapidly applied to the spine.[13] The first 3 articles all found the oblique abdominal musculature to activate to higher levels during neutral posture abdominal bracing maneuvers, with Kavcic *et al*[11] showing commensurate activation across the extensor musculature during the brace as well. Meanwhile, Vera-Garcia *et al*[13] established that abdominally bracing in an unloaded state significantly increased the stability and stiffness of the lumbar spine in preparation for sudden loading.

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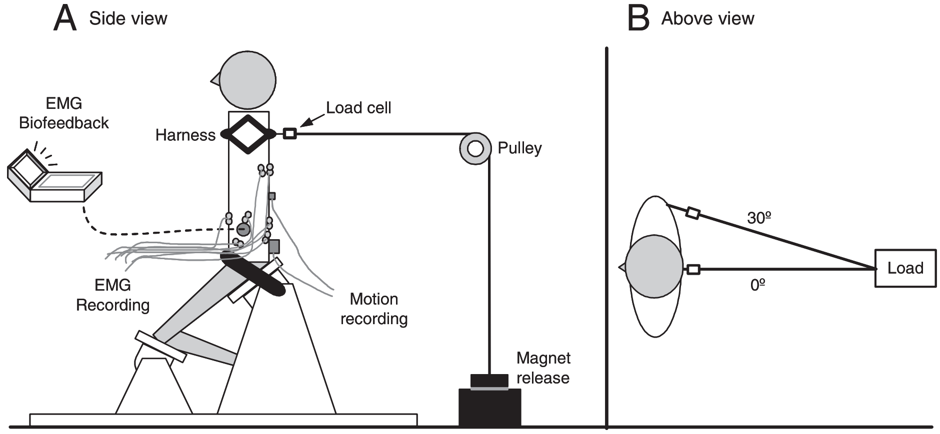

Figure 1. Sagittal view of the experimental setup for generating the quick release (**A**). Top view of the 2 loading directions (*i.e.,* 0° and 30°) from sagittal plane (**B**).

To our knowledge, no study to date has examined the effects of abdominally bracing the trunk in an externally loaded situation, and, therefore, how this bracing alters the previously indicated benefits of preloading the trunk in preparation for an external perturbation is unknown. The purpose of this study was to quantify the muscular activation patterns, and subsequent spine stability levels, during various magnitudes of abdominally bracing a preloaded trunk. In addition, trunk angular displacements were quantified in response to a quick release of the trunk preload to assess the global kinematic effects of abdominal bracing.

Materials and Methods

Participants. There were 14 male volunteers, who had not had back pain in the previous year, recruited from the university population and who participated in the study. However, because of the difficulty in obtaining appropriately measured signals in all components of the instrumentation, only 10 of the subjects were used in the modeling analyses. This article focuses mainly on the results and implications stemming from the modeling analysis, and, therefore, only these 10 subjects will be included. The 10 participants had a mean age of 27.1 years (standard deviation [SD] 7.6), height of 1.77 m (SD 0.04), and mass of 76.6 kg (SD 9.4). Participants completed an informed consent form approved by the University Office for Research Ethics.

Instrumentation and Data Collection

Electromyogram Biofeedback. While maintaining the lumbar spine in a neutral position, participants were instructed to coactivate isometrically the abdominal muscles ("abdominal bracing") at 4 different levels. The MyoTrac™ (Thought Technology, Inc., Montreal, Quebec, Canada) electromyogram (EMG) biofeedback system was used to control and provide feedback concerning the intensity of the abdominal brace. An EMG sensor was placed over the lower region of the right external oblique using a disposable triode electrode (Ag-AgCl). The participants used the biofeedback to achieve the desired target levels of activation normalized to maximal levels established during the previously conducted maximum voluntary contractions (MVC). Participants were instructed to "try to attain the EMG activation target and to maintain it." The target was programmed at 0%, 10%, 20%, and 30% of the maximal voluntary isometric contraction (MVC) amplitude at the right external oblique site. The MVC amplitude was obtained in resisted maximal twist and lateral bend efforts while restrained in a sit-up posture. Participants were allowed to practice activating to the target levels until both they and the experimenter were satisfied with their ability.

Quick Release. Participants were placed in a semi-seated position in a wooden apparatus that restricted hip motion while leaving the trunk free to move in all directions (Figure 1), a position that has fostered a neutral spine posture.[14] Subjects were statically loaded *via* a cable aligned approximately with the T7 level, with either a 8.0 or 10.3-kg load. The cable was oriented horizontally through a pulley and attached to the load. The cable load was delivered either in the sagittal direction (0° sagittal condition) or in an oblique direction from the sagittal plane (30° sagittal condition). Participants were then instructed either to hold the load in a natural manner, or brace by activating the biofeedback monitored muscle to either 10%, 20%, or 30% of MVC. At each direction, the load being maintained was released *via* a magnet by the investigators without warning within a 15-second window. Each participant performed 1 trial at each of the 16 test conditions (4 pre-activation levels, 2 loading directions, and 2 loading amplitudes). All conditions were presented randomly.

External Force Measures. The magnitude and timing of the force perturbation produced by releasing the load were measured using a load-cell force transducer located in series between the cable and harness. The force signals were amplified and A/D converted (12 bit resolution over ±10 V) at 1024 Hz.

Trunk Kinematics. Lumbar spine kinematics were measured about 3 orthogonal axes (flexion-extension, lateral bend, and twist) using an electromagnetic tracking instrument (3Space ISOTRAK; Polhemus Inc., Colchester, VT), sampled at a frequency of 32 Hz. The source was strapped to the pelvis over the sacrum and the receiver on the rib cage, over the T12 spinous process. Thus, the 3-dimensional (3-D) angular displacements of the rib cage relative to the sacrum were measured.

EMG Recording. Surface electromyographic signals were collected bilaterally (right, left) from the following trunk muscles and locations: rectus abdominis, approximately 3 cm lateral to the umbilicus; external oblique, approximately 15 cm lateral to the umbilicus and just superior to the biofeedback sensor site; internal oblique, halfway between the anterior superior iliac spine of the pelvis and the midline; latissimus dorsi, lateral to T9 over the muscle belly; and erector spinae at T9, L3, and L5 (considered thoracic, lumbar, and multifidus levels, respectively), located approximately 5, 3, and 1 cm lateral to each spinous process. Ag-AgCl surface electrodes were positioned with an interelectrode distance of 3 cm. The electromyographic recording was synchronized to the ISOTRAK and load cell data with a common trigger. The EMG signals were amplified (±2.5 V), A/D converted (12 bit resolution) at 1024 Hz, and full wave rectified. For the modeling purposes to be described later, EMG was low-pass filtered (second order single pass Butterworth) at 2.5 Hz and normalized to MVC amplitudes. The MVCs were obtained in isometric maximal exertion tasks performed before the quick release trials.[11]

Data Reduction. The onset of the force perturbation was detected from the load-cell signal by visually identifying when the force-time slope changed significantly. Time windows of 200 milliseconds before and 250 milliseconds after the force perturbation were selected for subsequent analyses.

External Force and Kinematics. The force of release and the peak angular displacement of the lumbar spine (extension, bend, and twist) in the 250 milliseconds after sudden loading were recorded in every trial. To incorporate the effects of the force release on the subsequent trunk angular perturbation, a gross stiffness measure was obtained. The release moments (Nm) were calculated as the products of the release force (N) and the moment arms representing the point of application of the release force in either the frontal (flexion moment) or transverse (twist moment) planes. A gross lumbar measure of stiffness (Nm/degree) was then obtained from the following equations:

$$k_{flex} = \frac{M_{flex}}{\theta_{flex}} \quad (1)$$

$$k_{twist} = \frac{M_{twist}}{\theta_{twist}} \quad (2)$$

where k_{flex} and k_{twist} = stiffness about the flexion and twist axes, respectively (Nm/degree).

M_{flex} and M_{twist} = moments about the flexion and twist axes, respectively (Nm).

θ_{flex} and θ_{twist} = angular displacement about the flexion and twist axes, respectively (degrees).

Pre-Activation. For each muscle site, the average normalized EMG for the 50 milliseconds before the perturbation was used to evaluate the amplitude of the muscle pre-activation at each of the 4 pre-activation levels (no brace, 10%, 20%, and 30% MVC).

Stability and Compression. First, static whole-body postures were hand digitized from a single digital video image and entered into a full-body linked segment model to determine the 3-D reaction forces and moments at the L4–L5 joint. Next, 14 channels of EMG and 3-D lumbar spine angles acquired from the 3-Space were entered into an anatomically detailed spine model representing 118 muscle elements as well as lumped passive tissues, spanning the 6 lumbar joints (T12–L1 through L5–S1). This model has been comprehensively reported previously.[2] Muscle stiffness and force were calculated as the first and second moments, respectively, of a distribution-moment model[15] representing the instantaneous number of attached cross-bridges in a given muscle, dependent on muscle cross-sectional area, activation, length, and velocity.

To quantify spine stability, an 18 × 18 (6 joints by 3 anatomic axes) Hessian matrix of the second partial derivatives of the potential energy of the entire lumbar spine system was calculated and diagonalized to obtain its eigenvalues. The potential energy theory states that each eigenvalue of the matrix must be positive definite for the system to be stable. Therefore, both the low eigenvalue and the stability index (an average of the 18 eigenvalues[16]) were used as measures of spine stability. Specifically, the low eigenvalue indicates the absolute stability of the system ("weakest link"), while the stability index provides a solution more sensitive to all joints and potential modes of buckling. L4–L5 compressive force and the 2 measures of spine stability were analyzed as the average over the 50 milliseconds before the sudden load.

Statistical Analysis. A 3-way repeated measures analysis of variance was used to compare the EMG preperturbation amplitudes between pre-activation levels, load direction, and load mass for each muscle, as well as to compare differences between pre-activation conditions for the low eigenvalue, stability index, compressive force, and flexion and twist stiffness variables. Where applicable, post hoc analyses were performed using the Tukey HSD test. Correlations were also calculated between the measures of stability (stability index and low eigenvalue) and measures of stiffness (flexion and twist). Significance levels were set at $\alpha = 0.05$ for all tests.

Results

Pre-Activation

Brace level affected the pre-activation level of all muscles except the left multifidus ($P < 0.0001$ for all other muscles except the left rectus abdominis $P = 0.0009$) (Figure 2). The general trend was that all muscles, except the left multifidus, showed significantly higher pre-activation in the 30% brace level as compared to at least the natural and 10% brace levels. Average abdominal pre-activation taken across the 6 monitored muscles was 7.4%, 11.1%, 18.9%, and 26.5% for the natural, 10%, 20%, and 30% brace levels, respectively.

A main effect of load direction was also found for the right latissimus dorsi ($P = 0.0015$) and right thoracic erector spinae ($P < 0.0001$). For both muscles, pre-activation levels were higher in the 0° sagittal condition (13.9 and 7.4% for right latissimus dorsi and right thoracic erector spinae, respectively) as compared to the 30° sagittal condition (7.7 and 4.7% for right latissimus dorsi and right thoracic erector spinae, respectively). On closer inspection of the data, it was revealed that although the global activation means displayed general trends of increasing as brace level increased, it was found that many individuals had difficulty in achieving the appropriate brace levels. Participants were able to increase activation across all 14 monitored muscles in only 12.8% of the 10% brace trials as compared to the natural brace trials, 37.5% of 20% as compared to the 10% brace trials, and 53.6% of the 30% as compared to the 20% brace trials.

Figure 3 displays mean activation levels across all 14 muscles for 4 separate trials for 2 subjects. The trials

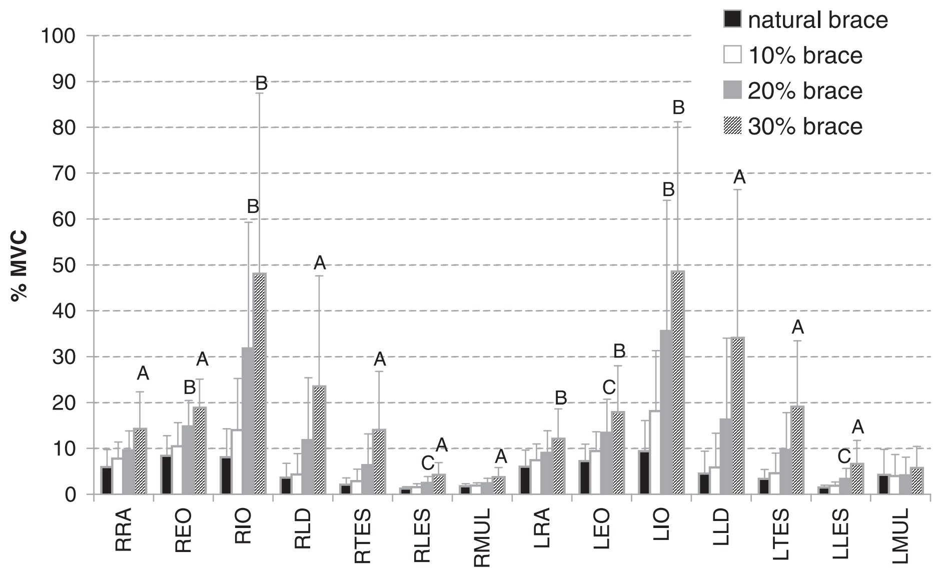

Figure 2. Muscle activation levels (percent MVC) averaged over the 50 milliseconds before the load release. SD bars are shown. Indications of statistical significance ($P < 0.05$): different from natural, 10%, and 20% brace levels (A); different from natural and 10% brace levels (B); and different from the natural brace level (C). Although these means display a general trend of increasing as brace level increased, individuals were only able to increase activation across all muscles in 12.8% of trials between the natural and 10% brace levels, 37.5% of trials between 10% and 20% brace levels, and 53.6% of trials between 20% and 30% brace levels. EO indicates external oblique; IO, internal oblique; L, left; LD, latissimus dorsi; LES, lumbar erector spinae; MUL, multifidus; R, right; RA, rectus abdominis; TES, thoracic erector spinae.

represent the natural brace and 10% brace levels for the 0° sagittal 10.3 kg condition. Note the differences in motor strategies used by the 2 individuals to achieve the brace and how they lead to differing levels of stability. The more unbalanced motor pattern (subject A) leads to a decrease in the low eigenvalue in the 10% brace as compared to the natural brace trial. In this case, the abnormally high internal oblique activity in the absence of corresponding extensor muscle activity appears to be the culprit.

Model Results

A significant interaction ($P = 0.0277$) was found between brace level and load direction for the stability index. The stability index displayed a trend of more significant increases occurring between bracing levels in the 0° sagittal condition as compared to the 30° sagittal condition. A significant interaction ($P = 0.0336$) between brace level and load was found for the low eigenvalue (Figure 4). Post hoc testing revealed that significant differences between bracing levels occurred only in the 8.0 kg as opposed to the 10.3-kg condition. Furthermore, the low eigenvalue actually decreased in 59% of the 10% brace trials as compared to the natural brace trials, in 35% of the 20% as compared to the 10% brace trials, and 17.5% of the 30% as compared to the 20% brace trials. The stability index did not display this same trend because individuals had success in increasing this measure in approximately 85% of trials between each of the brace levels.

Examining the compressive force, a significant interaction ($P = 0.0157$) was found between brace level and load direction. Similar to the stability index, a higher number of significant differences were found between bracing levels for the 0° sagittal condition as compared to the 30° sagittal condition. Across all trials, no significant findings were found for flexion stiffness. A significant main effect ($P = 0.004$) of brace level was found for twist stiffness (Figure 5). Furthermore, in examining the 30° sagittal trials alone (*i.e.*, trials that produced external moments about both axes), flexion stiffness was significantly higher than twist stiffness ($P = 0.0135$).

Correlations Between Stability and Stiffness Measures

When measured across all trials, flexion stiffness correlated significantly with the stability index (r = 0.31; $P = 0.0003$), but not significantly with the low eigenvalue. When only analyzed across the 0° sagittal trials (*i.e.*, trials in which the external moment produced isolated trunk flexion), the significant correlation between flexion stiffness and the stability index increased to r = 0.48 ($P < 0.0001$) and approached significance with the low eigenvalue (r = 0.29; $P = 0.05$). Twist stiffness, measured only in the 30° sagittal trials, also showed a significant correlation with the stability index (r = 0.57; $P < 0.0001$), while approaching a significant correlation with the low eigenvalue (r = 0.25; $P = 0.05$).

■ Discussion

The major finding of this study is that, although the general trends show muscle activation patterns and stability increasing as subjects attempted to brace abdominally, individuals were often unsuccessful in performing this coactivation procedure in a balanced way. In fact, no subject was able to show increasing activity across all

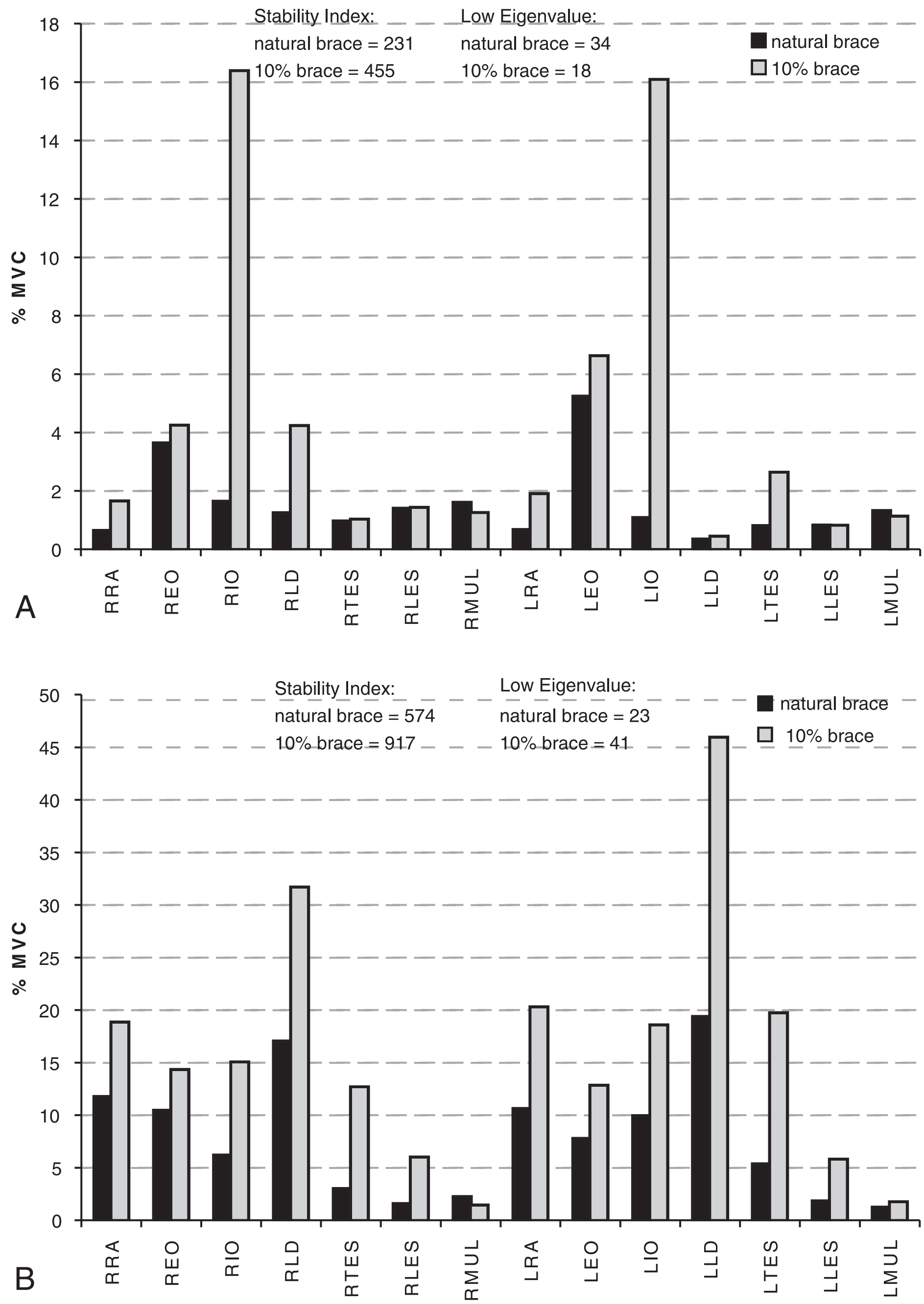

Figure 3. Example of muscle activation patterns averaged over the 50 milliseconds before the load release for 2 participants (**A** and **B**) in the 0° sagittal 10.3-kg natural brace and 10% brace trials. The stability index and low eigenvalue (Nm/rad/rad) are also indicated for each trial. Note that participant A has a poorly balanced coactivation adjustment from the natural to the 10% brace condition and, subsequently, has a decrease in the low eigenvalue. The abnormally high internal oblique activity in the absence of corresponding extensor muscle activity appears to be the cause of this reduction in stability. Participant B achieves an increased stability level with the 10% brace by involving more muscles. EO indicates external oblique; IO, internal oblique; L, left; LD, latissimus dorsi; LES, lumbar erector spinae; MUL, multifidus; R, right; RA, rectus abdominis; TES, thoracic erector spinae.

trunk muscles in each trial with increased bracing, at least in this task in which the applied load resulted in spine shearing together with a challenge to the flexion and twist moments. This result shows that in situations in which the trunk is loaded, individuals often find a natural bracing level, which, if altered by a bracing technique, can actually decrease the stability of the spine.

Participants often showed what can be considered odd ratios of muscular coactivation. For example, 2 participants achieved differing levels of stabilizing success using different coactivation patterns (Figure 3). Participant A braces by highly activating the internal oblique muscles without corresponding increases from the monitored extensor musculature. This resulted in a decrease in the low eigenvalue and, therefore, a compromise to spine stability. In contrast, participant B achieved a similar brace by increasing activation in a more even manner across the abdominal and extensor musculature, thereby increasing spine stability. It is noteworthy that not a single subject showed the motor control skill necessary to increase spine stability consistently through the desired increase in trunk muscle coactivation.

The stability index and low eigenvalue gave different indications as to the stability of the spine and are worthy of a short discussion. In the 0° sagittal trials, the stability index showed significant increases in the 30% brace level as compared to each of the natural, 10% and 20% brace levels, and significant increases between natural and both 20% and 30% brace levels in the 30° sagittal trials. On the other hand, the low

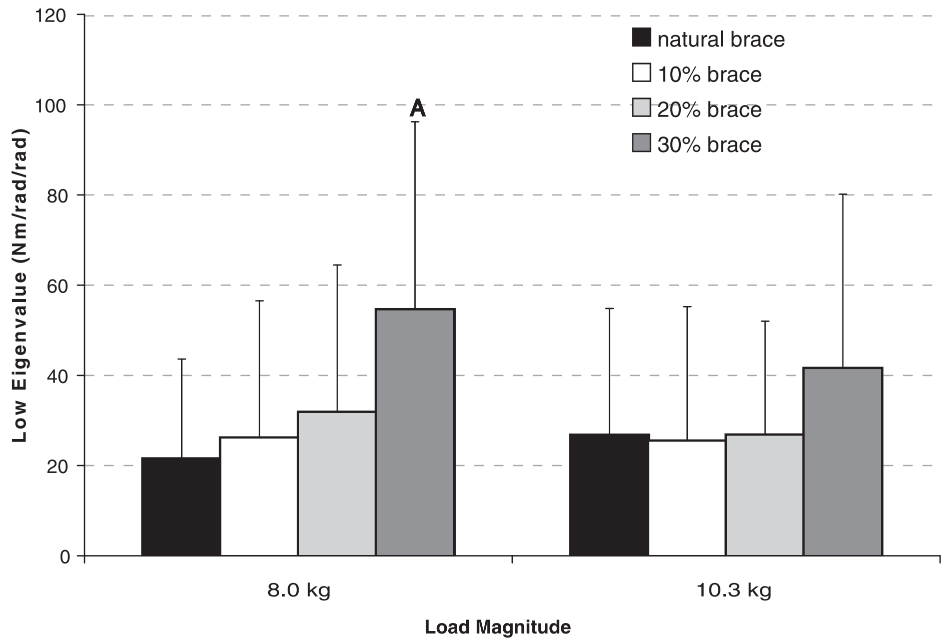

Figure 4. Low eigenvalue averaged over the 50 milliseconds before the load release. SD bars are shown. Indications of statistical significance ($P < 0.05$): different from natural, 10%, and 20% brace levels (A). Across all subjects, the low eigenvalue was actually decreased in 59% of the 10% brace trials as compared to the natural brace trials, 35% of the 20% brace trials as compared to the 10% brace trials, and 17.5% of the 30% brace trials as compared to the 20% brace trials.

eigenvalue was only significantly higher in the 30% brace as compared to each of the natural, 10% and 20% brace levels for the 8.0-kg trials. Furthermore, the stability index significantly correlated with gross measures of trunk stiffness about each of the flexion/extension (r = 0.31) and twist (r = 0.48) axes, whereas the low eigenvalue did not. This result is to be expected because the stability index provides an indication of the global state of stability in the 18 degrees of freedom spinal column, and the effect of single but multiarticular muscles stiffening many joints, whereas the low eigenvalue provides the ultimate determination of stability at a single degree of freedom.[16]

Most interestingly, it was found that the 2 measures of stability did not show agreement in 26% of all trials. This agreement was determined as the number of trials in which both measures did not provide the same indication of increasing or decreasing stability between consecutive trials. The low eigenvalue appears to be much more sensitive to muscle coordination patterns when compared to the stability index. Trials in which the 2 measures did not show agreement were often marked by the previously detailed unbalanced increases in activity across a single, or often multiple, muscle bilaterally. These trials tended to result in an increased stability index, but a decreased low eigenvalue, when compared to the trial of the next lowest brace level. The many nonlinearities in the variables that affect stability preclude the interpretation of either the low eigenvalue or the stability index to predict the location or mode of instability. Rather, one may only assess the ability of a particular coactivation pattern to stabilize.

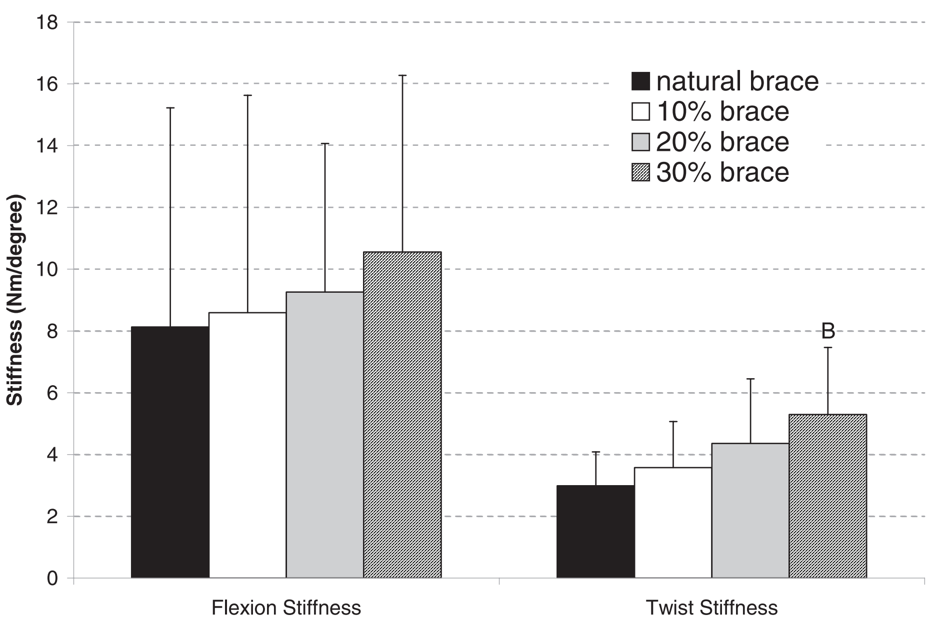

Figure 5. Trunk flexion and twist stiffness calculated over the 250 milliseconds after load release. SD bars are shown. Indications of statistical significance ($P < 0.05$): different from natural and 10% brace levels (B).

This study differs from previous studies that have examined both the effect of preloading the trunk and abdominally bracing on sudden loads[6–9] and quick releases[4] applied to the trunk. In the current study, abdominal bracing is applied to an externally preloaded trunk in preparation for a quick load release. Thus, subjects find a natural muscular activation pattern sufficient to maintain the external load and then adjust this pattern to meet predetermined criteria for an abdominal brace. Based on the results of this study, it appears that individuals often find an optimal natural brace level for the given load demand. Altering this pattern often changes the stability and stiffness of the trunk. Whether this change enhances or diminishes stability levels depends on whether the individual moves toward a balance between all muscles or simply alters a single muscle so that a "weak link" in terms of stability is created. The exception appears to be bracing at very high levels (30% brace in this study), which seems to activate the full musculature, thereby ensuring an enhanced stability level. It is noteworthy that this high bracing level carries with it the additional penalty of significantly higher compressive loads acting on the spine, which have been linked to low back pain and injury.[17]

There are certain limitations inherent in this study. Only 1 repetition of each trial was performed on each subject. With the high number of trials already included in the data collection, this was deemed necessary to prevent both physical and mental fatigue. Repeated trials would have no doubt reduced the variability in the measures taken; however, the main conclusions reached here would not be affected. Also, the measure of trunk stiffness used here was by no means a robust measure of the true dynamic stiffness of the trunk. Inertial and damping terms were not considered; however, the stiffness calculations do provide an indication of the interaction between the measures of the released load and ensuing trunk angular displacements, measures that if considered alone could have led to erroneous conclusions. Finally, the participants in the current study were relatively untrained in performing abdominal brace maneuvers. It is possible that additional training in proper bracing technique would improve the muscular coordination patterns and enhance the stabilizing ability of such maneuvers at lower bracing levels. It has been indicated that such training is often necessary for individuals performing specific abdominal muscle activation exercises.[18]

■ Conclusions

Individuals appear to respond to the type of loads reported here by bracing the trunk musculature to ensure stability. Changing the intensity of contraction, or focusing on activating a single muscle, may compromise stability in certain situations. More robust coactivation levels (30% MVC) appear to create a more balanced contraction across all muscles to ensure enhanced stability.

■ Key Points

- Externally loaded individuals often select trunk muscular activation patterns that successfully achieve spine stability and stiffness appropriate for the loading demand.
- Conscious alterations of these patterns can reduce the margin of safety surrounding the trunk, producing an imbalance in support and stiffness around the spine, as indicated by reductions in spine stability.
- Motor strategy was determined to be the most important factor in successfully stabilizing the spine through muscular coactivation.

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