Links Between the Mechanics of Ventilation and Spine Stability

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Spine stability is ensured through isometric coactivation of the torso muscles; however, these same muscles are used cyclically to assist ventilation. Our objective was to investigate this apparent paradoxical role (isometric contraction for stability or rhythmic contraction for ventilation) of some selected torso muscles that are involved in both ventilation and support of the spine. Eight, asymptomatic, male subjects provided data on low back moments, motion, muscle activation, and hand force. These data were input to an anatomically detailed, biologically driven model from which spine load and a lumbar spine stability index was obtained. Results revealed that subjects entrained their torso stabilization muscles to breathe during demanding ventilation tasks. Increases in lung volume and back extensor muscle activation coincided with increases in spine stability, whereas declines in spine stability were observed during periods of low lung inflation volume and simultaneously low levels of torso muscle activation. As a case study, aberrant ventilation motor patterns (poor muscle entrainment), seen in one subject, compromised spine stability. Those interested in rehabilitation of patients with lung compromise and concomitant back troubles would be assisted with knowledge of the mechanical links between ventilation during tasks that impose spine loading.

Keywords: challenged breathing, low back, motor patterns, EMG

Spine stability is ensured through continuous muscle activation of the surrounding trunk muscles (Cholewicki & McGill, 1996). Coincidently, those same muscles are contracted rhythmically to actively assist breathing during expiration. Specifically, ventilation during challenged breathing is linked to a cyclic pattern of abdominal recruitment during active expiration and relaxation during inspiration (Naus et al., 1990, and Henke et al., 1988). Because the same muscles are required to perform two different tasks, often simultaneously, a paradox is created: isometric contraction to develop the stiffness needed to ensure spine stability vs. rhythmic contraction linked to each ventilation cycle. The goal of this study was to explore this phenomenon and further understand the links between ventilation mechanics, spine mechanics, and stability, which may assist those dealing with rehabilitation of the compromised lung.

The close relationship between challenged ventilation and torso muscle activation has been seen previously in a number of studies (McGill et al., 1995; Hodges et al., 1997; Grenier et al., 2000; Hodges & Gandevia, 2000; Kang & Lee 2002; Hamaoui et al., 2002; Shirley et al., 2003; Saunders et al., 2004). The types of challenged breathing in these studies were as follows: breathing stimulated with a 10% carbon dioxide mixture, increased/decreased respiration volume and rate, breath holding, and locomotion. However, only two of these studies (the ones involving breath holding) attempted to draw mechanically based conclusions explaining the reported observations. One interesting study by Shirley et al. (2003), demonstrated how torso stiffness is influenced by increased torso muscle activation and intra-abdominal pressure during breath holding at different tidal volumes. The stiffness was directly measured from the application of an external force such that spine stiffness remains unknown. Nonetheless these thought-provoking studies have encouraged further

investigation on how torso muscles are engaged in both challenged ventilation and spine loading.

The purpose of this study was to obtain better understanding of the links between the muscular motor patterns (sequential and synergistic muscle entrainment) needed to breathe and subsequent effects on spine mechanics. The first hypothesis tested was that certain components of ventilation (i.e., flow rate, dynamic lung volume, direction, and change in direction, and level of ventilatory challenge) will affect trunk muscle activation in a way that influences spine stability. The second hypothesis tested in this study was that a combination of holding a load with a simultaneous breathing challenge would compromise spine stability for some individuals. The knowledge gained could be useful to enhance understanding of spine stability in individuals, enhance the training of breathing techniques for those involved in manual labor, exercise, and sport, and possibly in those with respiratory pathology populations.

Methods

Subjects, Data Collection, and Breathing and Spine Load Challenges

Eight male university students with an average age of 24.5 years (*SD* ±2.6) with a height of 179 cm (*SD* ±0.06) and a weight of 78 kg (*SD* ±12.4) volunteered after proper informed consent was obtained. All subjects were screened to ensure no known history of back pain, lung disease, heart disease, or any other health-related problem that may have negatively affected the results of this study.

Participants performed a series of three broad types of tasks/challenges, specifically: four breathing and spine challenges (precycle challenges), one cycle challenge with four incremental workloads (stationary cycle ergometer trial), and two breathing and spine challenges (postcycle challenges). The tasks within each of these task/challenges were conducted in a randomized fashion. Following are the instructions given to each subject for each challenge.

First, the instructions for the four precycle challenges:

1. Quiet breathing (QB): Stand, relax, and breathe normally.
2. Slow breathing: Take 5 s for inspiration, and 5 s for expiration (follow the metronome set at 1 Hz).
3. Slow breathing while holding a load: same as above while holding a 4.5-kg (10 lbs.) weight in front of their pelvis.
4. Maximal voluntary ventilation (MVV): Breathe in and out as fast as possible with deep breaths.

For the stationary cycle ergometer, each trial consisted of four bouts of cycling (labeled Bike 1, Bike 2, Bike 3, and Bike 4). After each bout, the instructions were to stop pedaling and to stay seated with hands on handlebars. Each bout was 1.5 min in length and progressively more difficult. Bike loads: There were four cycle ergometer bouts, each with increasing difficulty. Subjects were instructed to cycle at 50 rpm by following a metronome. All subjects performed the same work rates, which started at 150 W to 300 W, with 50-W increments for each bout.

There were two postcycle challenges:

1. Postcycle (following the fourth cycle bout): Stop pedaling, get off bike, stand with arms at your side and look forward. (This challenge was to measure breathing patterns after the last cycle bout).
2. Postcycle with load: as above with holding load in front of their pelvis.

Note that all tasks were done standing in neutral position with arms at the side, looking straight forward (except when on the bike).

Instrumentation

Fourteen channels of EMG were collected from the following muscles bilaterally: rectus abdominis (RA), oblique internus (OI), oblique externus (OE), latissimus dorsi (LD), thoracic erector spinae (T9), lower erector spinae (L3), and lower erector spinae (L5) according to the protocol of Cholewicki and McGill (1996). Pairs of Ag–AgCl surface electrodes were used with a center-to-center distance of approximately 3 cm. The EMG signals were amplified and then A/D converted with a 12-bit, 16-channel A/D converter at a rate of 1,024 Hz. Each subject performed a maximal voluntary contraction (MVC) for each measured muscle to provide a basis for normalization and subsequent interpretation. For the abdominal muscles, each subject, while in a sit-up position and manually braced, produced a maximal isometric flexor moment followed sequentially by a right and left lateral bend moment and then a right and left axial twist moment. These were restrained isometric efforts and little motion occurred. For the extensor muscles, a resisted maximum extension was performed in prone position with the trunk cantilevered over the end of a padded bench, while squeezing the shoulders back together and attempting to arch and extend the back upward.

For the internal and external oblique muscles, a resisted side bridge was used, where the lower arm, hip, and legs were in contact with the bench, and when ready, the subject would raise their hips to neutral position (so that the head, trunk, and legs were aligned) and so that only the arm and feet were left in contact with the bench. Resistance was provided at the hips and the subject was asked to brace the abdomen and push up on the resistance. The EMG signal was normalized to these maximal contractions, full wave rectified and low-pass filtered with a 2nd-order Butterworth filter (cut-off frequency of 2.5 Hz, after Cholewicki & McGill, 1996, and Brereton & McGill, 1998). Despite the use of high common mode rejection amplifiers (115 dB at 60 Hz), there was still some EKG contamination of the signals given the

closeness to the heart. This artifact was removed using a second filtering process (dual-pass 2nd-order Butterworth filter, cut-off frequency, 1.0 Hz). Given that the breathing tasks were not less than 1 cycle per second, this procedure had no visible effect on the amplitude of the time history of the EMG signals. This technique sufficiently decreased the signal-to-noise ratio of EMG:EKG amplitude.

Spine Motion

Three-dimensional lumbar spine kinematics were measured with a 3Space Inside TRAK electromagnetic tracking device (Polhemus Inc, Colchester, VT). The 3Space transmitter device was mounted via strapping over the sacrum and the receiver was mounted directly over the spinous process of T12 (see Figure 1). This arrangement provided the position of the ribcage relative to the pelvis or lumbar curvature, which was required as input into the stability model. The kinematic data were collected at a frequency of 64 Hz.

Ventilation Mechanics.   Airflow through a rubber mouthpiece (both flow volume and direction) was measured with an ultrasonic flowmeter (Model UF202, Kon and Associates, Redmond WA, USA) collected for each trial (Figure 1). Flow rate was obtained from the derivative of volume over time.

Data Analysis

For each trial, the L4-L5 reaction moments about each of the three axes of rotation was calculated using a link segment model (McGill & Norman, 1985) using the static posture body segment data collected from a test subject and anthropometric data of the actual subject together with measured hand forces as input. The reaction moments were then input to a 50th-percentile anatomically detailed spine model. Electromyographic data that was collected during each trial was used to activate the "modeled muscles" (after Cholewicki & McGill, 1996) to determine muscle force and subsequently muscle stiffness and stability.

Muscle Forces

The distribution-moment muscle model of Zahalak (1986) was used to estimate muscle force and convert the force to stiffness on order to estimate stability. Because moment equilibrium is desirable at each lumbar joint (see Cholewicki & McGill, 1995), the typical approach is to adjust muscle forces to match external measured moments using an EMG optimization program. Because this optimization process uses the minimal change of muscle force to match external moments, some muscles with low levels of activation are often shut off. This technique works well with isometric tasks, but in this case, since breathing is dynamic, the process needed to be sensitive to the small changes in muscles (i.e., internal and external obliques) associated with each breath. For
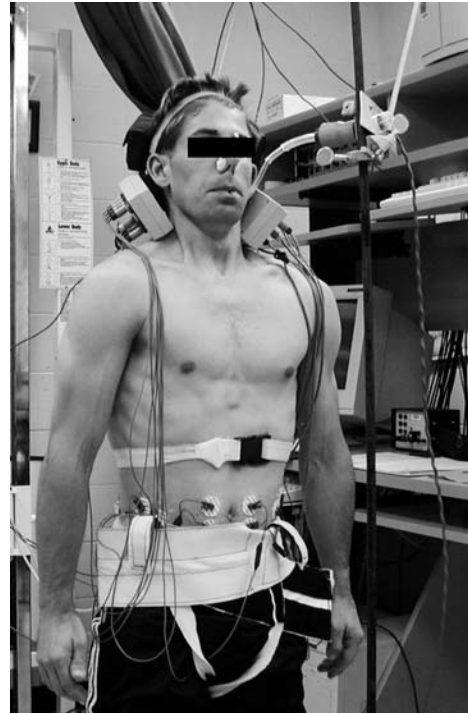

Figure 1 — Subject instrumented with the 3space to monitor three-dimensional lumbar motion (strapped to thorax and pelvis), EMG electrodes for torso muscle activity, mouthpiece to flowmeter to record flow volume for ventilation volume and rate, and a force transducer to measure hand load.

this reason, the optimization program was not run for the calculation of muscle forces.

Spine stability analysis was performed using the model developed by Cholewicki and McGill (1996). A global stability index for the entire lumbar spine is output as well as a measure of the critical stiffness at each instant in time. Ventilatory data were aligned in time and compared with the index to assess links between variations in stability with ventilatory mechanics.

Comparing Results

An unpaired t test was used to determine a difference in means for analysis involving muscle onset and peak timing, spine compression, and breathing measurements. Global results for the group are documented in the following parargraphs.

Results

Muscle Patterns

Only the right-side muscle activations have been shown in figures, owing to similar data from both sides, resulting from the sagittal symmetry of tasks. Four instants in time for each respiration cycle were analyzed:

1. Ab peak, when the abdominal activation was at its maximum (during expiration)

2. Ab onset, when the abdominal activation was starting to increase (lowest point, during inspiration)

3. Back peak, when back muscle activation was at its maximum (during inspiration)

4. Back onset, when back activation was starting to increase (lowest point, during expiration).

Muscle activation for all muscles during quiet breathing was below 5% MVC for all subjects (Figure 2). The highest level of activation among the abdominal muscles appeared to be the internal oblique during quiet breathing (2–5% MVC).

With increased ventilatory demand (postcycle challenges), the peak activation of the back muscles was during inspiration, which suggests that using back extensor activation was generally a common strategy to assist inspiration (Figure 3). Observations were made of a number of subjects extending their back during inspiration, "lifting up" the chest. Similar entrainment of the abdominal obliques to expiration were also seen during the postcycle challenges—specifically abdominal muscles contracted during expiration. An example is shown in Figure 4.
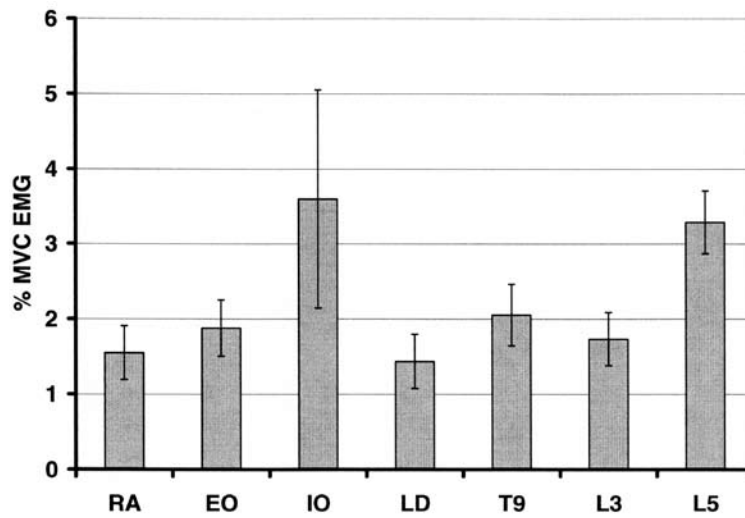

Figure 2 — Right-side mean trunk muscle activation during quiet breathing trial. Standard error of mean on each bar.
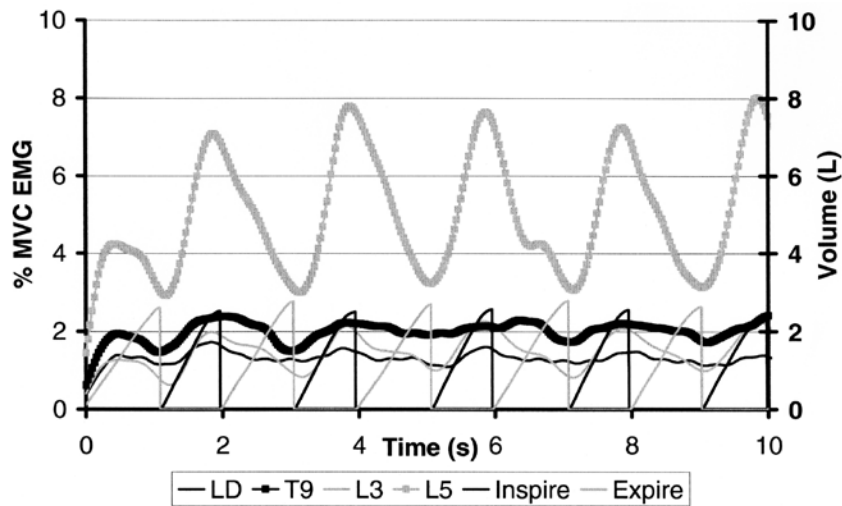

Figure 3 — Right latissimus dorsi, and erector spinae muscles at the levels of L3 and L5 are activated to assist in lifting up the rib cage to assist inspiration during postcycle challenge for one typical subject.
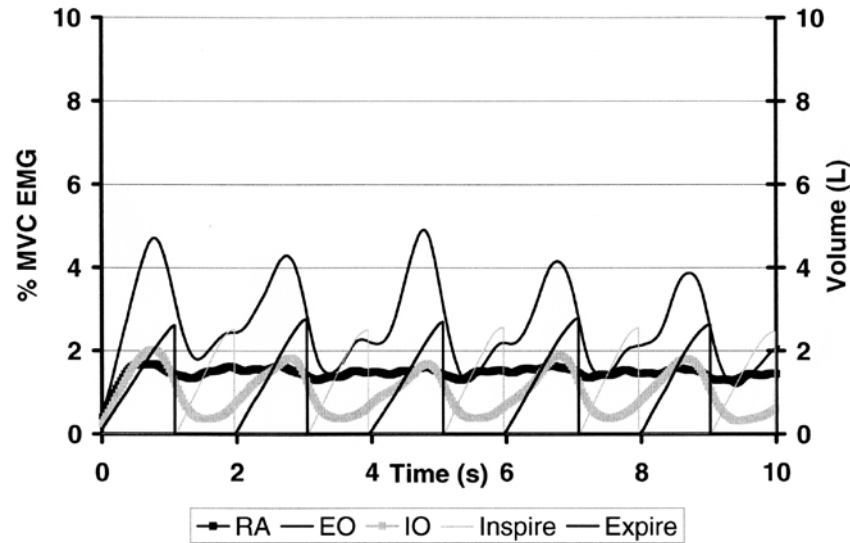
Figure 4 — Right abdominal muscle activation over time during postcycle challenge for one typical subject shows the link to the expiration of air from the lungs.

Spirometry

Increases in volume and rate were achieved via cycle ergometry. An average of a 469% volume increase and a 157% breathing rate increase was obtained in the postcycle challenges, compared with the resting trials (Tables 1 and 2).

Average spine compression means across quiet breathing and postcycle challenges showed compression to be well within NIOSH limits of 3,000 N (Table 3).

Lumbar Spine Stability

Although many trials demonstrated a notable EMG muscle pattern, not all patterns translated into changes in spine stability for two reasons: (1) Muscle activations, although displaying a pattern, were very low (i.e., <5% MVC) and were buried by other more active muscles and 2) many bike and postcycle challenges had alternating abdominal and back muscle activation during expiration and inspiration, respectively, which provided continuous muscle stiffness to support the spine. For each subject, spine stability was generally very stable during quiet breathing trials. The postcycle challenges showed an increasing stability during inspiration, with a maximum occurring just before expiration and a minimum toward the end of expiration (an example is shown in Figure 5).

Spine loading trials (holding hand loads) increased back EMG activation above any activation due to muscle breathing patterns, which lead to higher and steadier stability index values (Figure 6).

Table 1   Tidal Volumes During Resting and Postcycle Challenges

Subject	Rest TV (L)	Postcycle TV (L)	% change TV
1	0.43	2.01	468
2	0.47	2.67	570
3	1.49	3.86	259
4	0.57	2.78	484
5	0.51	2.65	523
6	0.48	2.54	525
7	0.40	2.27	565
8	0.54	1.90	354
Average	0.61	2.59	469

Table 2   Breathing Rates During Resting and Postcycle Challenge; Breaths per Minute (bpm)

Subject	Rest bpm	Postcycle bpm	% change
1	19	36	193
2	13	26	196
3	10	16	164
4	22	25	111
5	20	30	150
6	14	25	177
7	15	21	141
8	19	29	148
Average	17	26	157

Table 3   Group Mean and Standard Deviation of Spine Compression (newtons) for Quiet Breathing and Postcycle Challenges (Averaged Across Entire Trial)

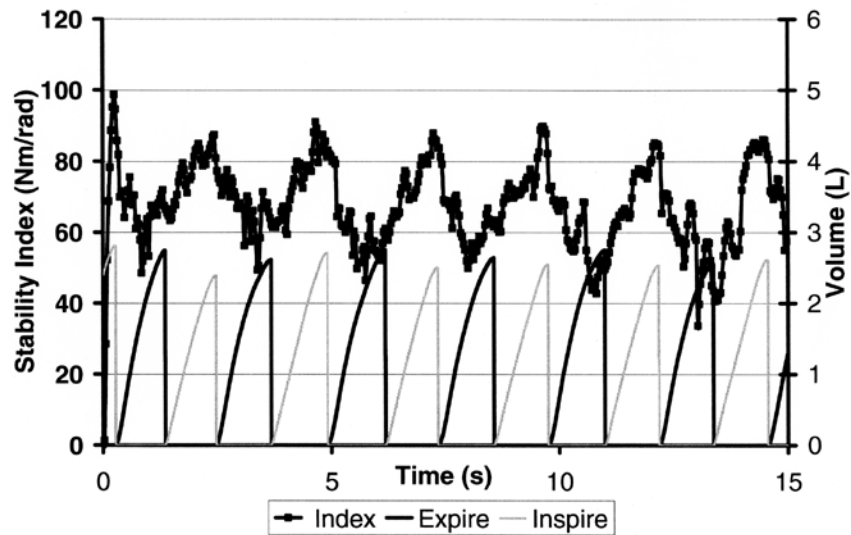| Trial	Group *M* (N)	Group *SD*
Quiet breathing	1,282	389
Postcycle challenge	1,461	248

Figure 5 — Spine stability over time during postcycle challenge for one typical subject showing maximal stability associated with full lungs and about to begin the expiration phase.
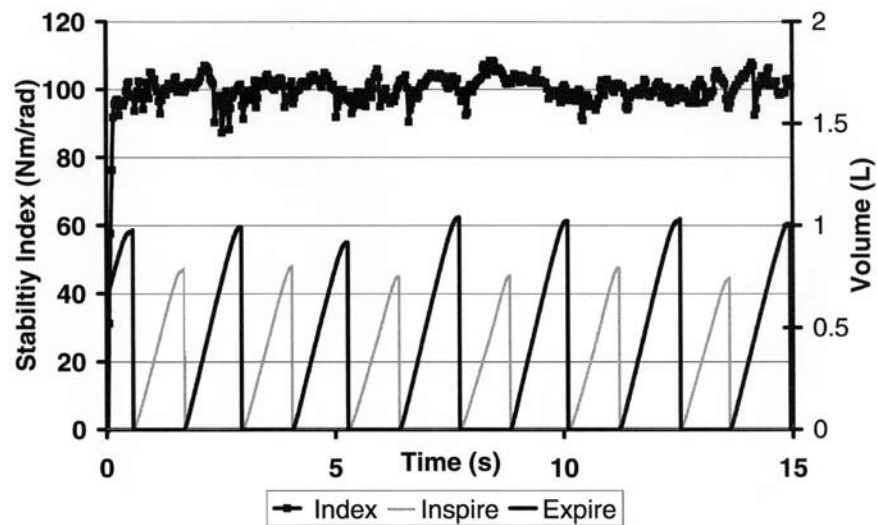

Figure 6 — Spine stability over time during postcycle challenge with hand load trial for one typical subject shows how the hand load increased and leveled off lumbar spine stability, and diminished the underlying effects of challenged breathing.

Interesting Case Results

During the postcycle challenge, one subject experienced a few asymptomatic, back motor pattern errors, in which asymmetric torso movement and muscle activation precipitated drops in spine stability (Figure 7). What appears to have happened was that, within a few seconds, the subject's back muscles turned almost completely off, one side before the other, causing a twist and flexion toward the activated side (Figure 8). This aberrant motor pattern caused an abrupt drop in the spine stability index. In addition, this one subject, who had no known ventilatory or back problems, had a distinguishing ratio of forced expiratory volume to forced vital capacity (FEV/FVC) of 71%, whereas the group mean was 84%.

Discussion

An analysis of ventilatory mechanics was conducted to address the hypothesis that certain components of ventilation (i.e., flow rate, lung volume, direction and change in direction, and level of ventilatory challenge) will affect the trunk musculature in a way in which spine stability becomes compromised. Results have shown that "involuntarily" increasing flow rate (bike ride trials) increased the activity in abdominal and back muscles, allowing them to maintain an increased spine stabilization level without aberrant drops during the trials. In addition, lung volume appeared to have an indirect effect on spine stability in that during slow breathing, MVV, and postcycle challenge, an increase in stability occurred with increasing lung volume and with the stability peak occurring close to peak inspiration. These stability and lung volume results agree with the torso "stiffness" data

obtained by Shirley et al. (2003), where indentation stiffness was higher at volumes above end tidal inspiration and were at their lowest at functional residual capacity (full expiration). Perhaps this is why those needing a very stable spine (e.g., weightlifters and sprinters) fill their lungs and cease breathing during their events.

Direction of airflow was an interesting variable in the context of breathing mechanics. Subjects (during post-cycle challenges especially) were observed to extend their backs during inspiration, while increasing back muscle activation. We can argue that this technique was adopted by these subjects to reconfigure the chest in a way to assist in creating the vacuum pressure to assist the lungs in filling with air given the increased ventilatory demand. Changes in direction of airflow, from the end of expiration to inspiration, during challenged breathing, was seen to be linked to decreased abdominal muscle activation and subsequently decreased spine stability. This result may seem counter to previously published work in which torso "stiffness" increased with expiration (Shirley et al. 2003), but those results were based on isometric breath holds in the expiratory phase, which increased both abdominal and erector spinae muscle recruitment, whereas these results are of very low torso muscle recruitment during the dynamic ventilation cycle.

In the case study, an interesting connection between direction of airflow and sagittal trunk movement may be linked to the isolated and transient drop in spine stability that was observed for an instant. Lewit (1980) described the association between inspiration and trunk extension and between expiration and trunk flexion. Analysis of the three-dimensional spine kinematic data showed that all observed drops in spine stability occurred when the subject transitioned from slight torso extension to
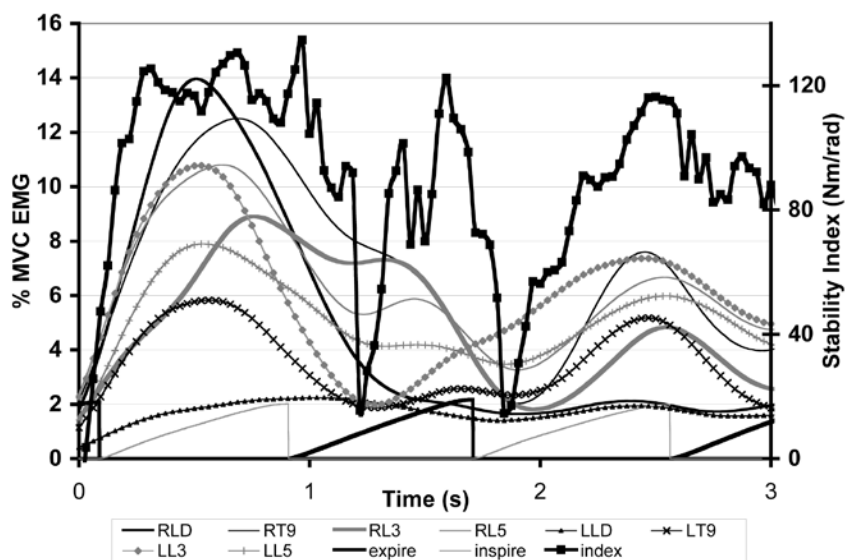

Figure 7 — Spine stability and back muscles over time during postcycle challenge for one subject in whom the compromise in spine stability was associated with a loss of muscle activation.
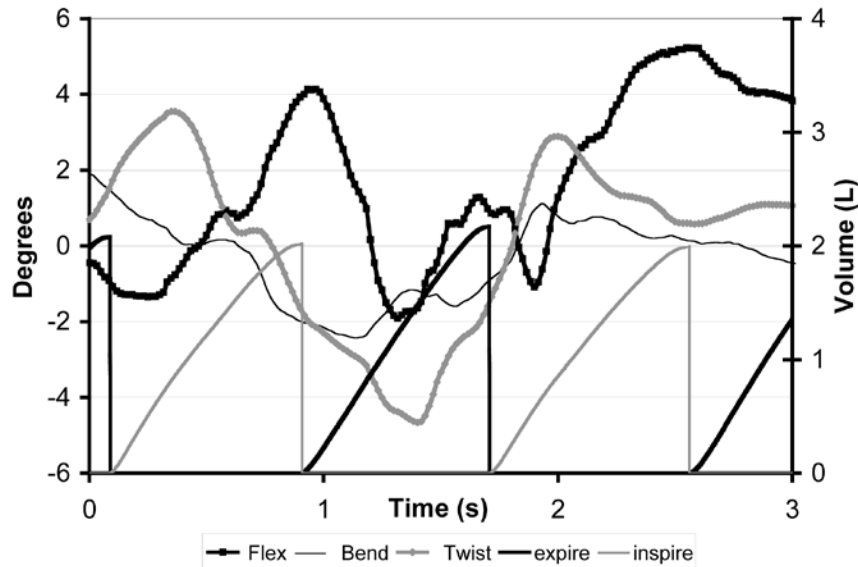
Figure 8 — Three-dimensional spine kinematic data over time during postcycle challenge for one subject (same trial as Figure 7).

slight torso flexion very quickly (e.g., Figure 8). This was coupled with a rapid decrease in back extensor activation. Although inspiration was seen to increase back muscle activation, with correspondingly increasing spine stability, it is difficult to provide suggestions for breathing in terms of stability. Inspiration would also recruit the diaphragm, which has been suggested to assist with spine stiffness (Shirley et al., 2003). However, for heavy loads, the breath must be held to optimize muscle stiffness and also intra-abdominal pressure via elastic distension (Cholewicki et al. 1999).

Two mechanisms may account for spine stability particularly in those with compromised ventilation mechanics or in those with uncompromised mechanics involved in challenged breathing. Changing airflow direction or specifically transitioning from inspiration to expiration can be associated with a slight back extension to flexion motion, with a corresponding quiescence of the back muscles. After a short delay, the abdominal muscles turn on to assist with active expiration. This period of little muscle stiffness is problematic. Secondly, as the end of expiration is approached, most of the air has been expelled and is not in the lungs to assist with intra-abdominal pressure, and torso stiffness (Cholewicki et al., 1999). Also, during expiration, the diaphragm is not recruited and therefore cannot help with spine stability via the crural attachment to the lumbar spine (Shirley et al., 2003). These two mechanisms work together to minimize torso stiffness, resulting in an elevated possibility of spine buckling should the individual experience an unfortunate, appropriate external perturbation.

The second hypothesis tested in this study was that a combination of holding a load with a simultaneous breathing challenge would compromise spine stability for some individuals. Results for all load trials showed an increase in muscle activation that buried or attenuated any muscle patterns caused by challenged breathing (also seen by Abe et al., 1996, and Abraham et al., 2002). This increased activation caused an increase, and a leveling effect, on the spine stability index. These results lead us to reject the second hypothesis and suggest that there may be a circumstance in which a load carried in the front might aid in spine stability. Further, there may be times when there is no load or when an attempt at a task as easy as picking up a pencil from the floor could involve an elevated risk of spine buckling and back injury could occur. This is consistent with previous work suggesting that the lowest indices for stability occur during events with minimal load (such as picking up a pencil) owing to minimal muscle activity, or with very high loads, in which a poorly coordinated pattern of stiffening muscle around the spine could cause insufficient potential energy at a nodal point (buckling location) (Cholewicki & McGill, 1996).

Giving context to the data in this study is influenced by some methological limitations. Not all muscles were monitored with EMG in the biomechanical model used to calculate spine stability. For example, muscles such as quadratus lumborum and transverse abdominis were not directly monitored with EMG in this study. Quadratus lumborum has been shown to be an important stabilizer in many tasks. Its activation in the model was driven from the EMG signal obtained from the lumbar erector spinae. However, carefully selected surface EMG electrode placements have been shown to reasonably represent the deep muscle activation of the trunk (McGill et al., 1996a, 1996b) in some activities—in this case driving quadratus in this way seems reasonable. When the relative contributions of different muscles are compared, the relative contribution of TvA, for example, appears to be relatively less important than many other muscles. Nonetheless, it was activated by the internal oblique signal as

it is an EMG "twin" in standing activities such as ones assessed in this study (Juker et al., 1998). Secondly, we did not directly measure intra-abdominal pressure in this study. Cholewicki and McGill (1996) suggested simply increasing the passive stiffness of the spinal column to account for the contributions of the intra-abdominal pressure. Nonetheless, our major findings in this study were associated with periods of low lung volume (near functional residual capacity) and low muscle activity, suggesting that intra-abdominal pressure would have been at a minimum. In the instances when intra-abdominal pressure was high (higher abdominal muscle activation, and diaphragm activation during inspiration), we know that spinal stability would be increased and would simply add to our stability estimates. Finally, we assumed EMG symmetry between right and left sides of the torso. Given that this was a symmetric task, this would appear to be a reasonable assumption. However, should some asymmetry occur, we would expect that it would result in a reduction in stability

In summary, at rest normal ventilation does not require much use of the trunk muscles and does not significantly affect lumbar spine stability. When the ventilatory system is put under stress, such as with exercise, stability increases with increasing lung volume and back muscle activation, with the stability peak occurring close to peak inspiration. This may explain why those needing a very stable spine (e.g., weightlifters and sprinters) fill their lungs during their events. In addition, aberrant ventilation motor patterns, particularly instances of low lung volume and simultaneous loss of torso muscle activation, results in compromised spine stability.

Acknowledgments

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